PET Developments and Progress

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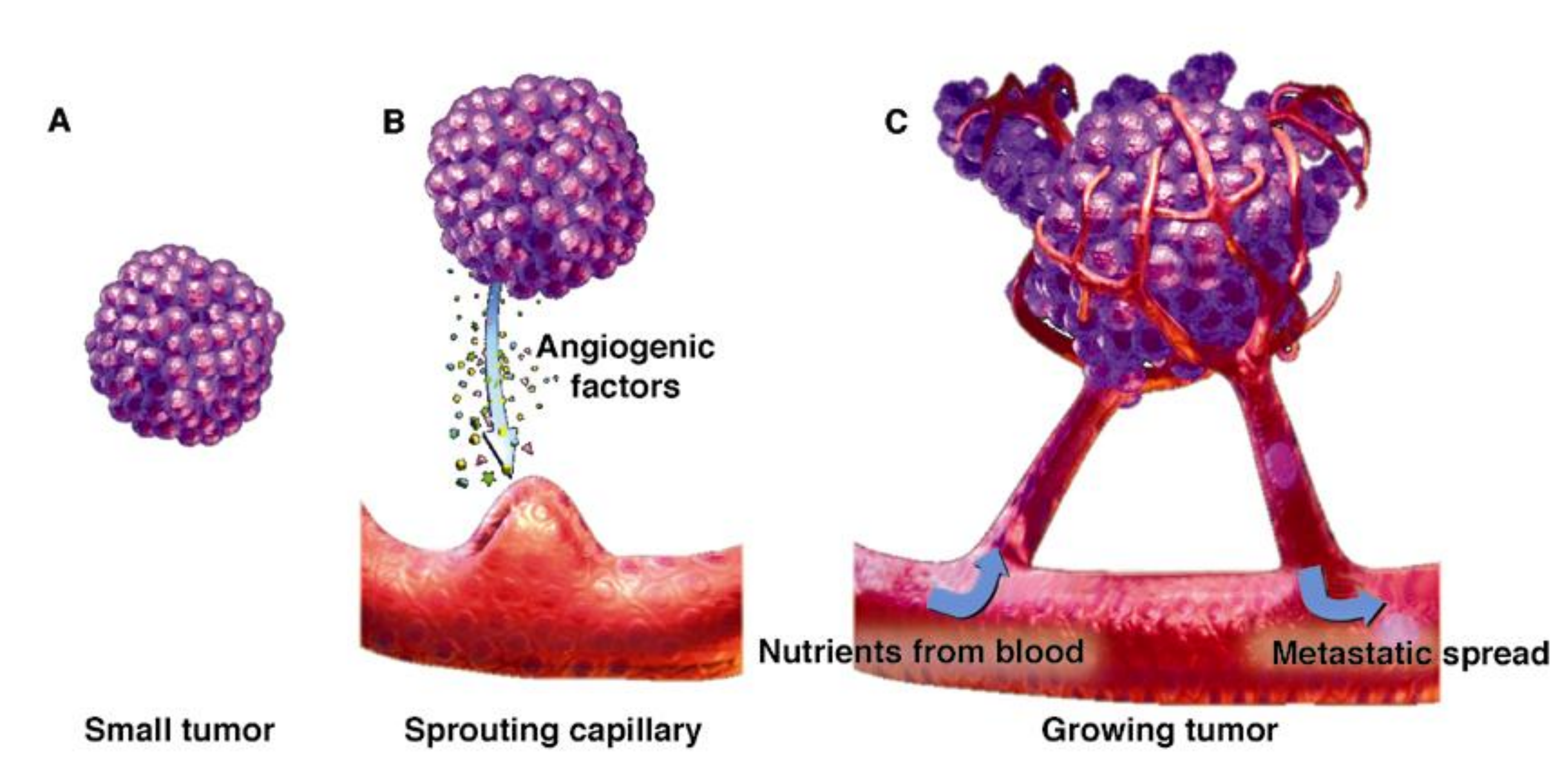
Basics
on
Positron Emission Tomography

- Metabolism imaging
- Sinogram
- Correcting for detector effects
- Attenuation map
- Quantification

Metabolism imaging

On the metabolism side, cancer cells are very different from normal cells.

They divide much faster and need large amounts of energy to do so.

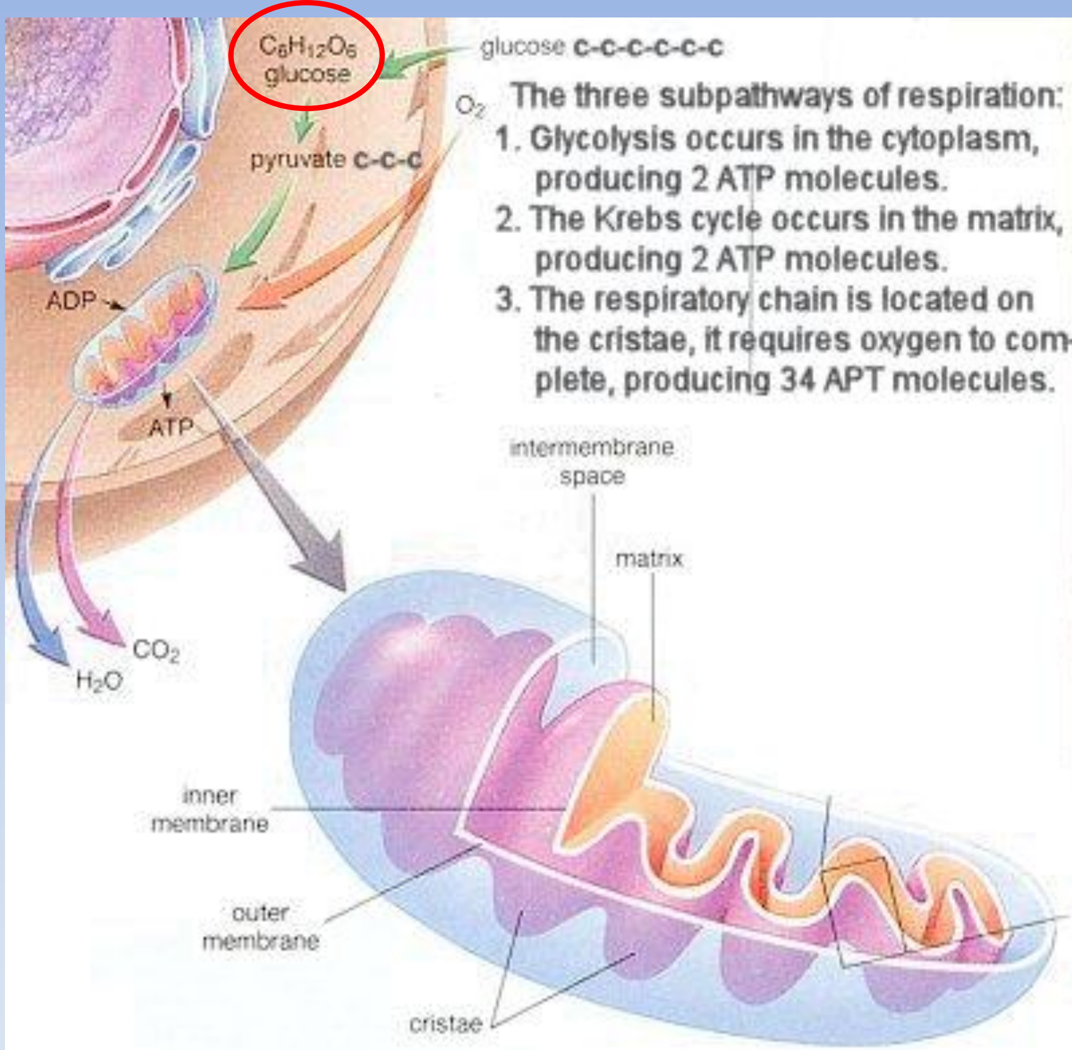


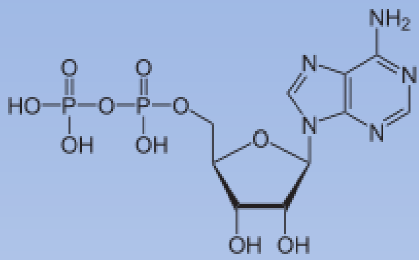
Active tumor = need for energy

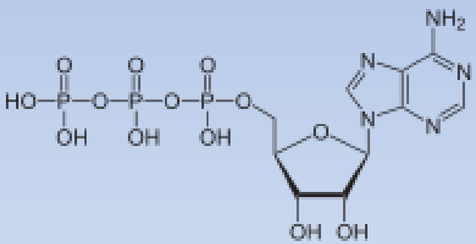
Energy comes from

ADP is “recharged” to ATP in the mitochondria

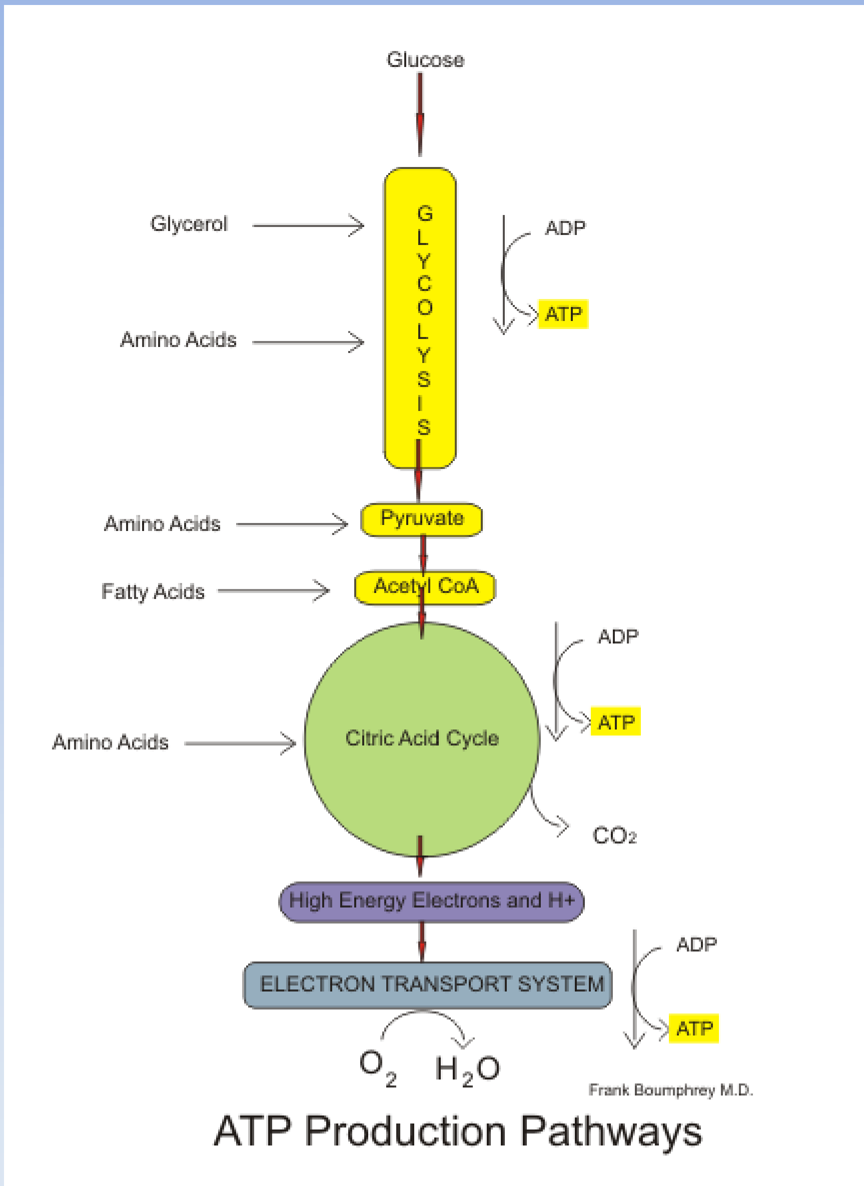
Metabolism imaging

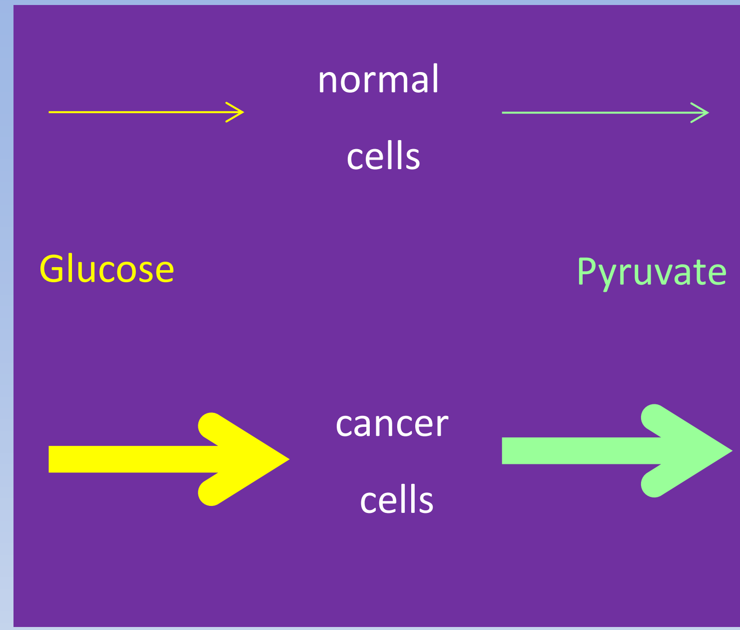




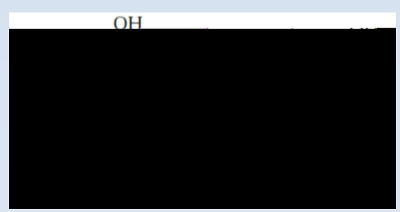


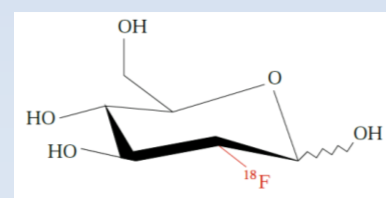
Metabolism imaging



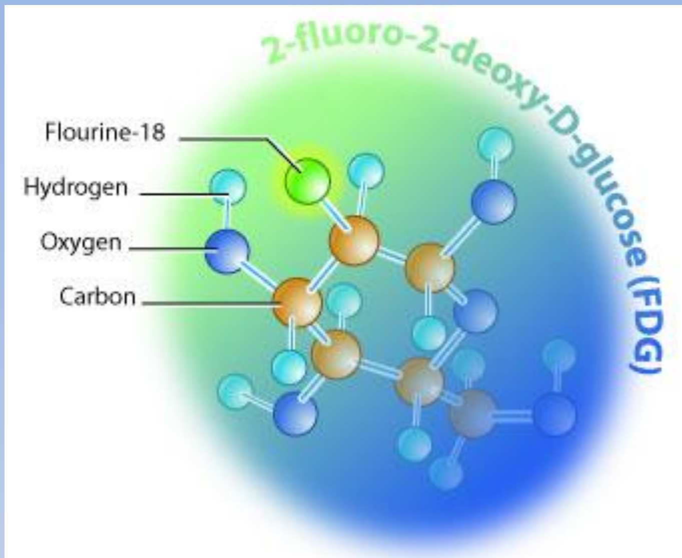


The tracer consists of a molecule analog to glucose associated with a radioactive beta+ emitter.





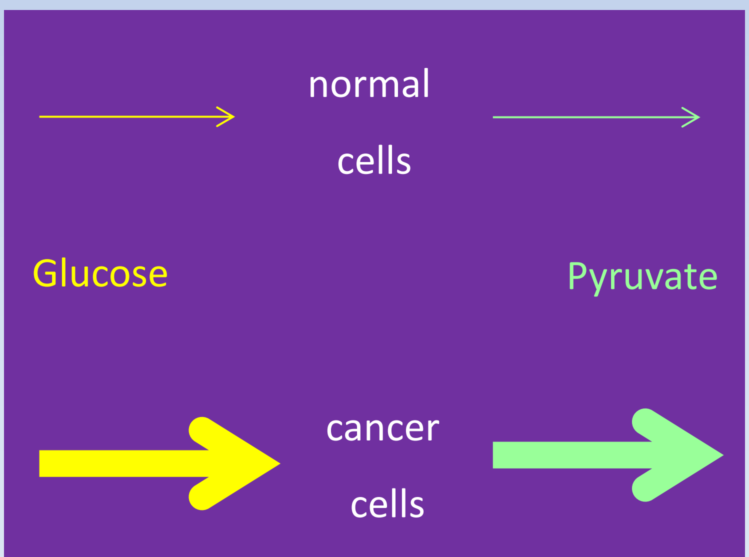
A radiotracer for tumor cells : the 18F-FDG



Because of its slightly different composition, the FDG cannot be metabolized into pyruvate.

The FDG is absorbed by the cells needing energy as considered as gluco,se but **remains trapped until fluor turns to oxygen**, after the beta decay.

The physiological properties are restored and the FDG can be metabolized. The concentration of FDG is an image of the glucose uptake.



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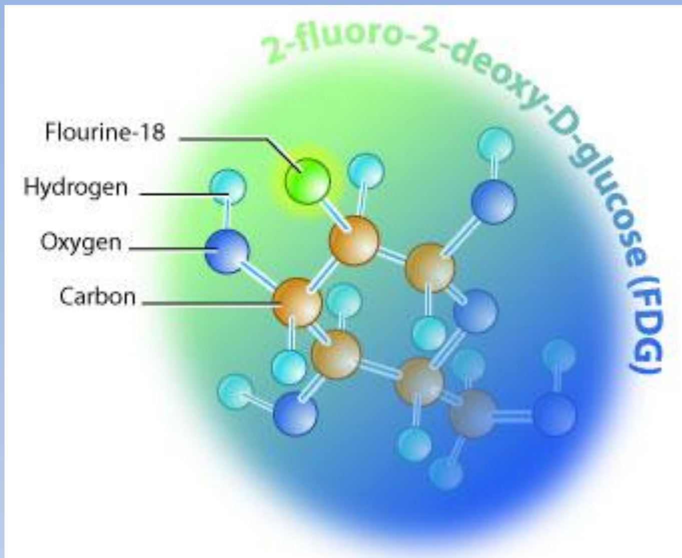

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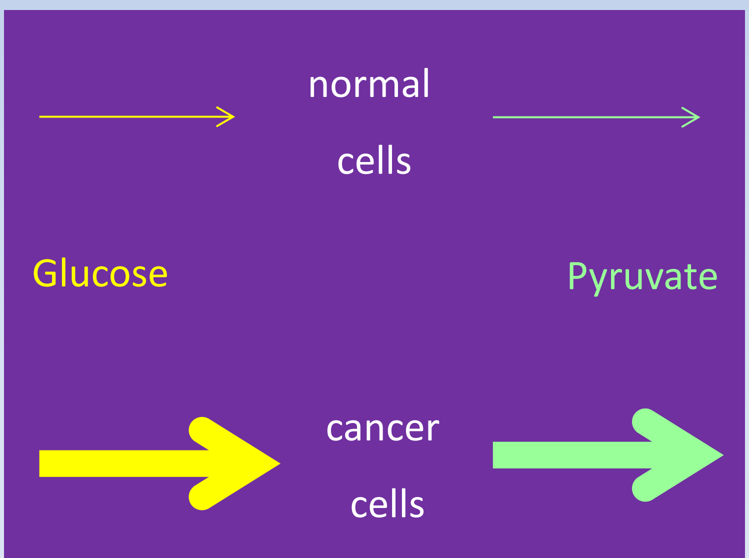

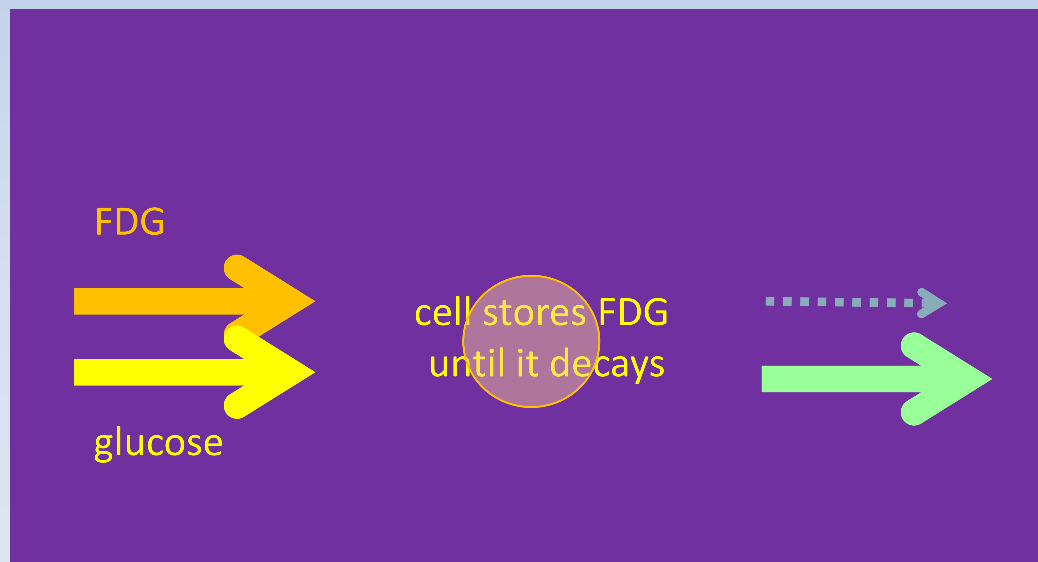

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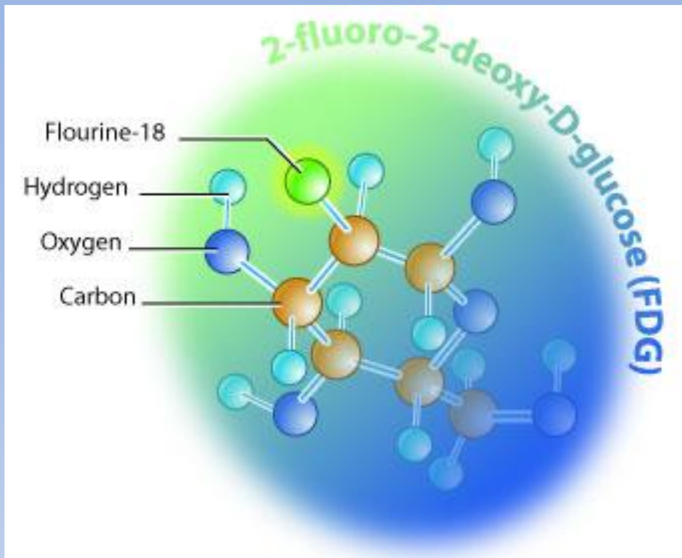

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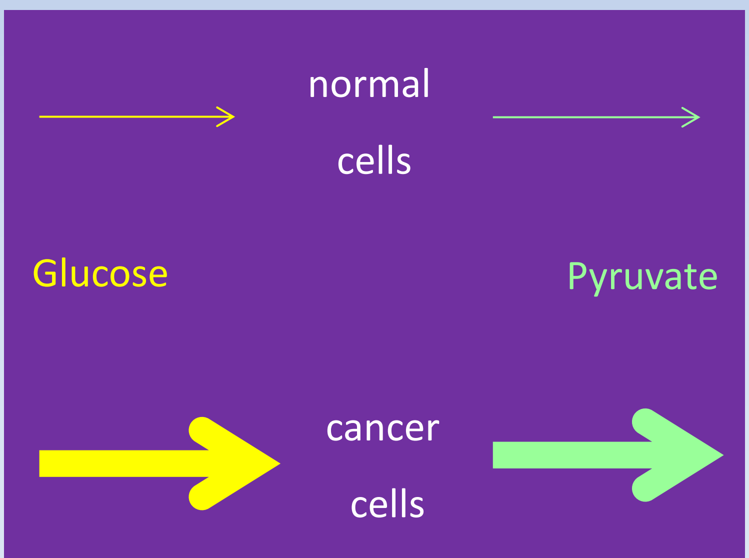
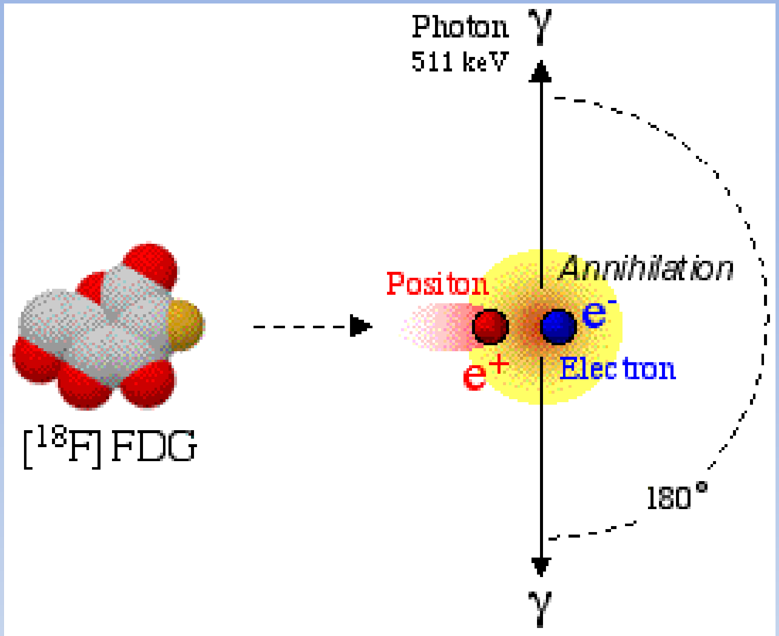
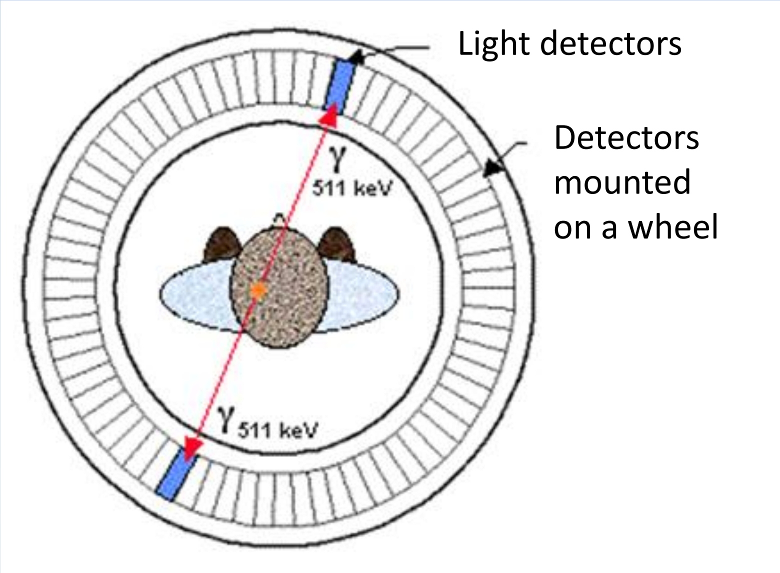


The Positron Emission Tomography

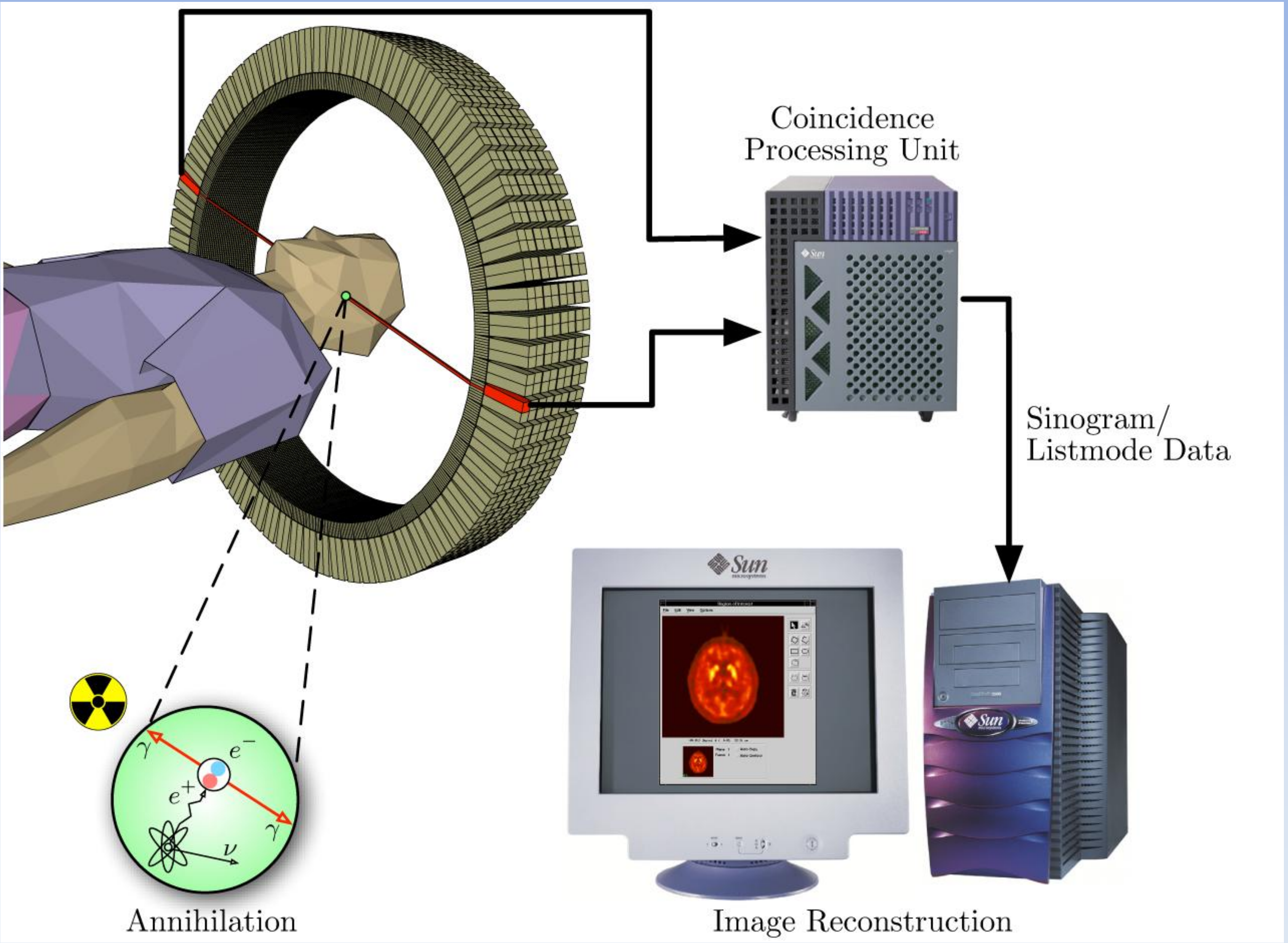


Positron from the fluor 18 nucleus decay losses its energy on a few millimeters in water, then annihilates with a nearby electron (97% of e^+ annihilations happens at rest). Two 511 keV photons are emitted back to back.

Photon detectors surrounding the patient detects pairs of 511 keV annihilation photons. Data is analyzed and allows to locate the beta+ emitters, where FDG has accumulated.



PET workflow

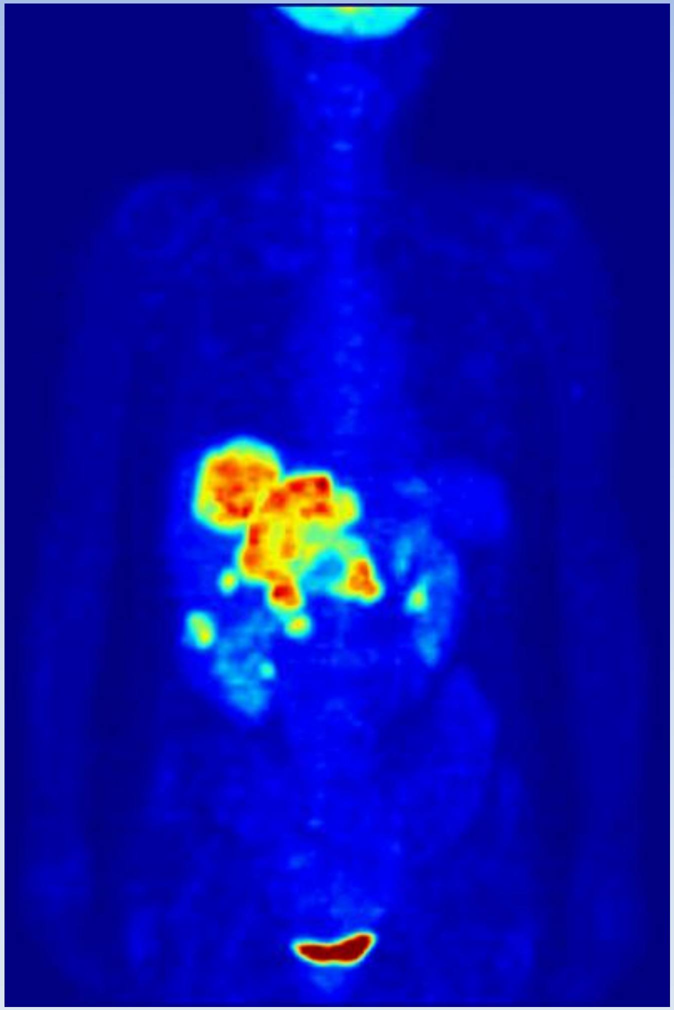


The Positron Emission Tomography

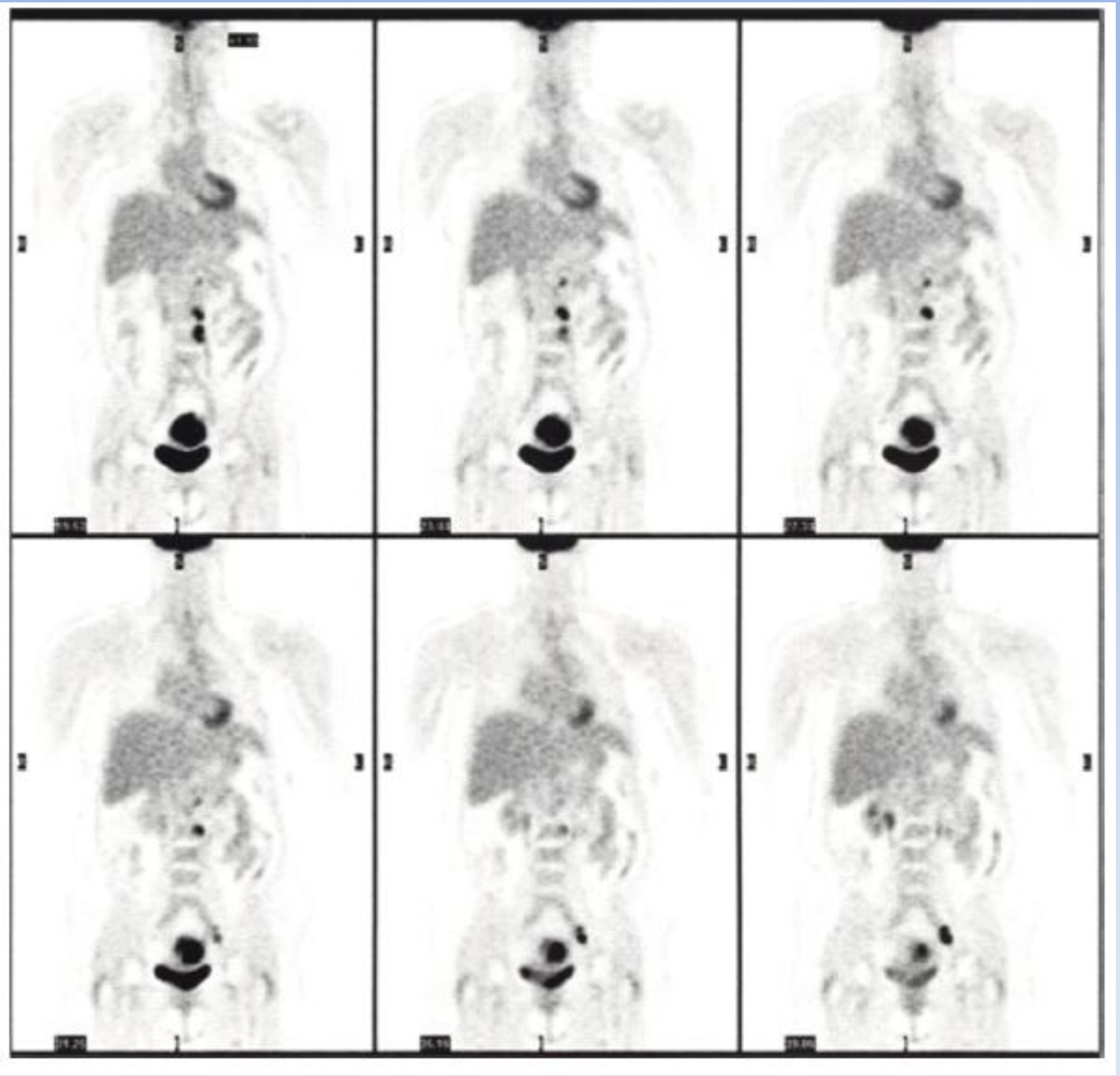
Naïve approach :

Inject patient with specific marker (pure beta+ emitter) and reconstruct images to sign tumors.

Radiotracers & radiopharmaceuticals	Examples of biomedical applications
$[^{15}O]$oxygen	oxygen metabolism
$[^{15}O]$carbon monoxide	blood volume
$[^{15}O]$carbon dioxide	blood flow
$[^{15}O]$water	blood flow
$[^{13}N]$ammonia	blood flow
$[^{18}F]$FDG	glucose metabolism
$[^{18}F]$FMISO	hypoxic tissue
$[^{18}F]$MPPF	serotonin $5HT_{1A}$ receptors
$[^{18}F]$A85380	nicotinic acetylcholine receptors
$[^{18}F]$FLT	DNA proliferation
$[^{11}C]$SCH23390	dopamine DI receptor
$[^{11}C]$Ro151788	central benzodiazepine receptor
$[^{11}C]$PK11195	peripheral benzodiazepine receptor
$[^{11}C]$PIB	amyloid plaque: Alzheimer's disease
$[^{11}C]$AG1478	EGF receptors
$[^{11}C]$choline	biosynthesis of phospholipids



FDG uptake after an injection



Time-lapse Images of 18-FDG Uptake.

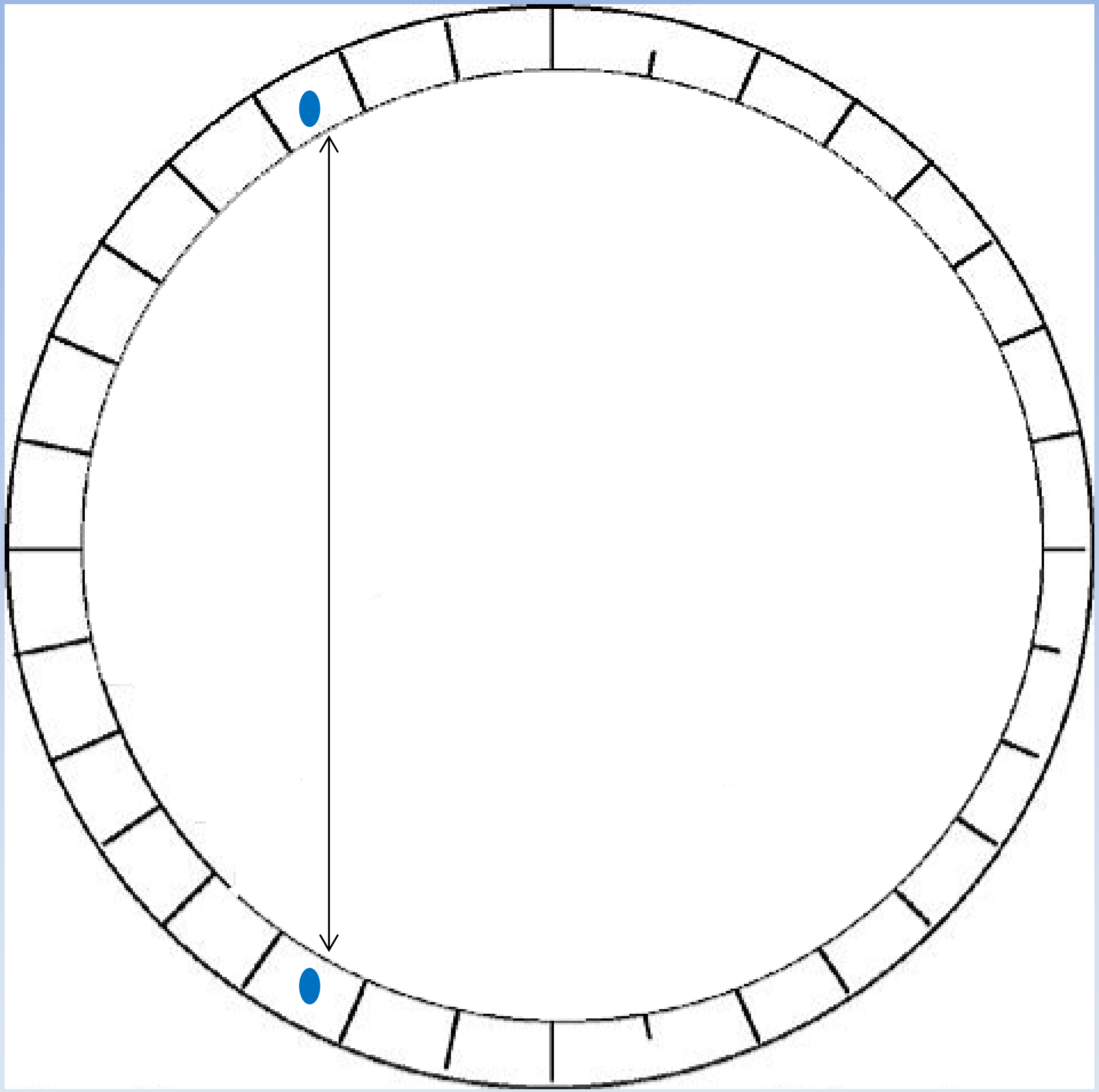
Initially, the tracer has high uptake in the cardiac and renal systems.
As time goes by, 18-FDG accumulates at the site of the tumor in the lower abdomen.
In addition, metabolic imaging detected multiple lymph node metastases along the left iliac and left para-aortic chains. (Driedger)

Functional imaging :

Huge amount of data (>150 Mb per second!!!) : can be “projected” on a sinogram or recorded in *listmode*.

- accumulate data (static image)
- dynamic mode (dynamic images)

Building a Sinogram



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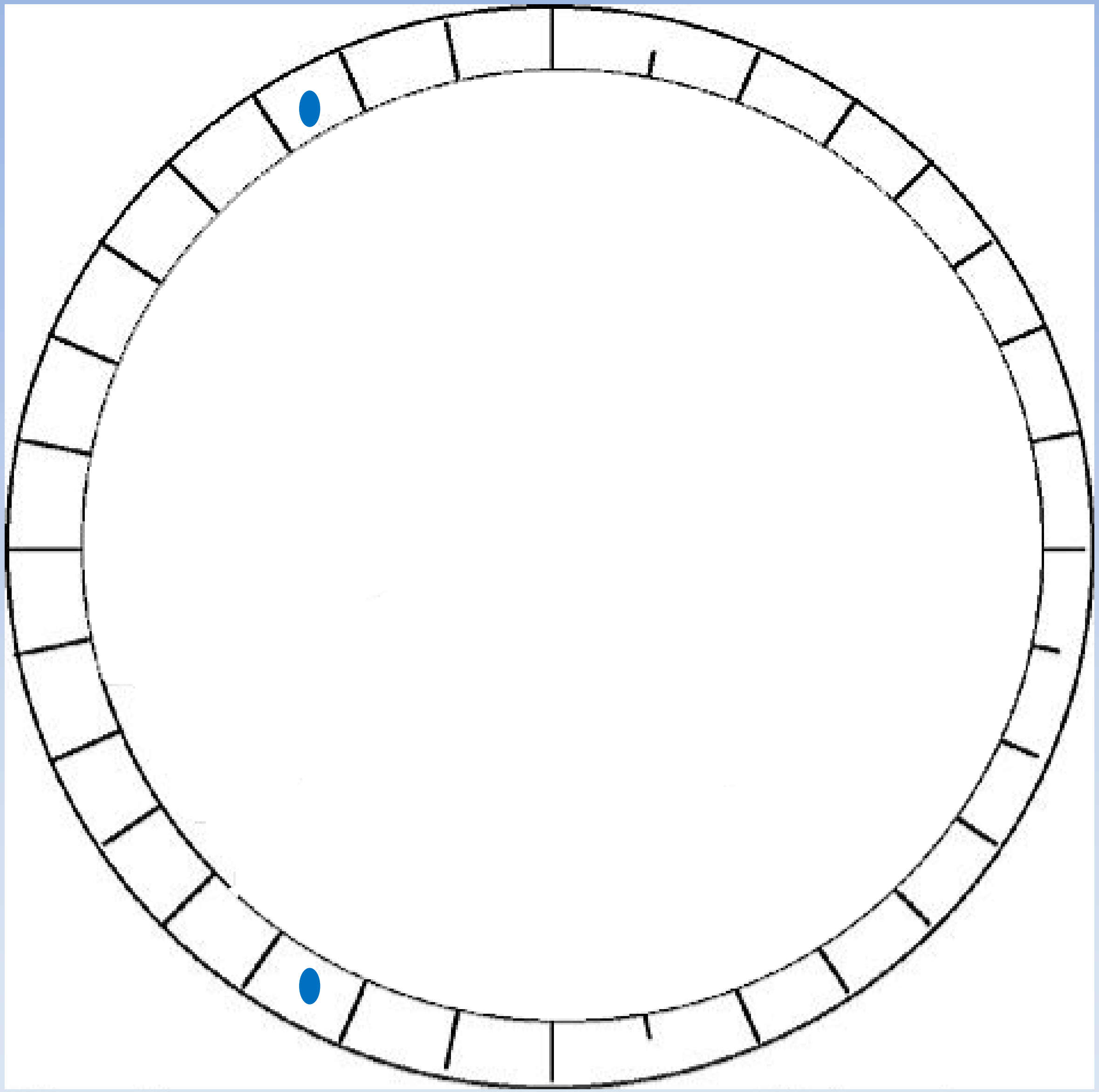

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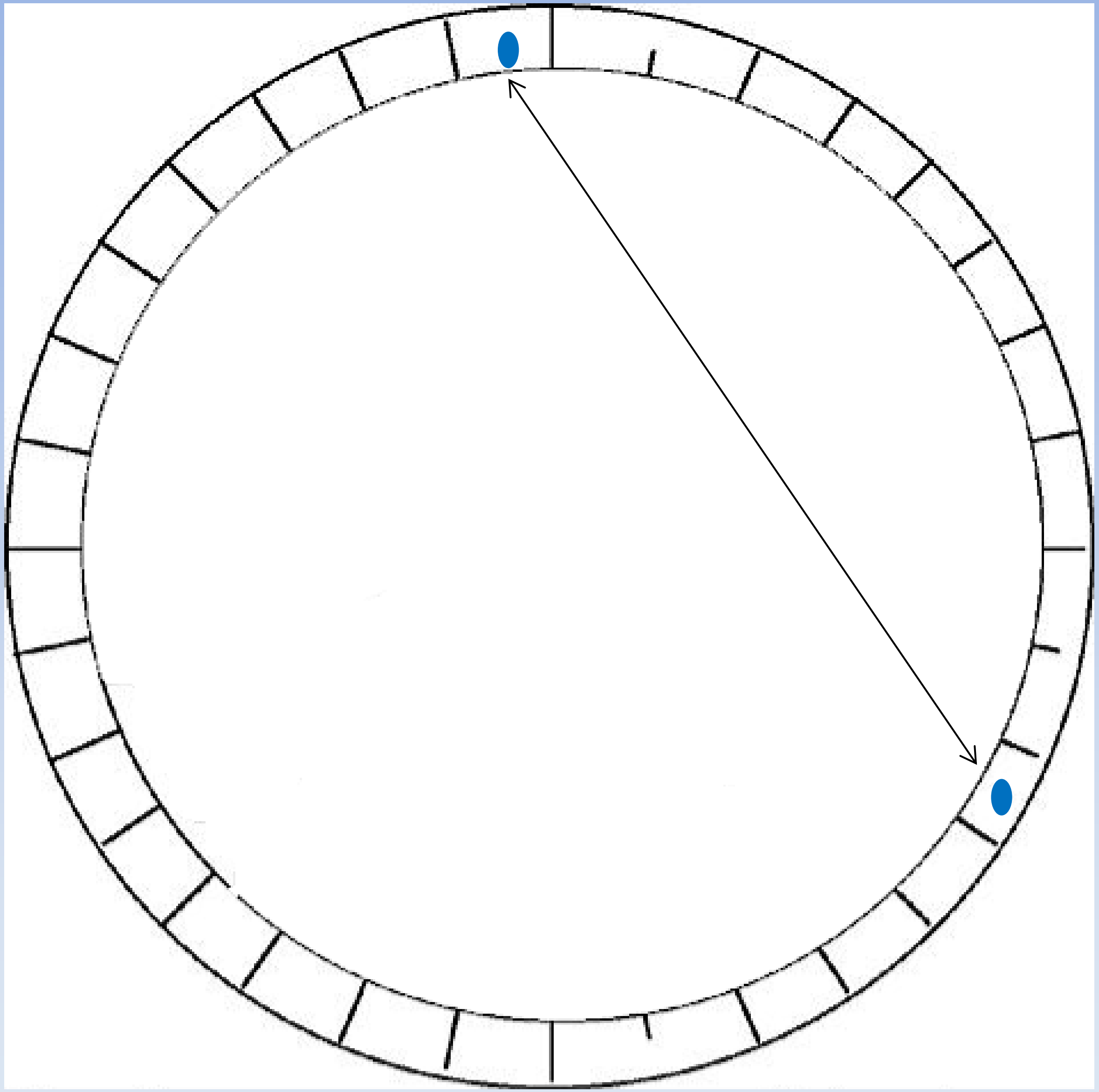

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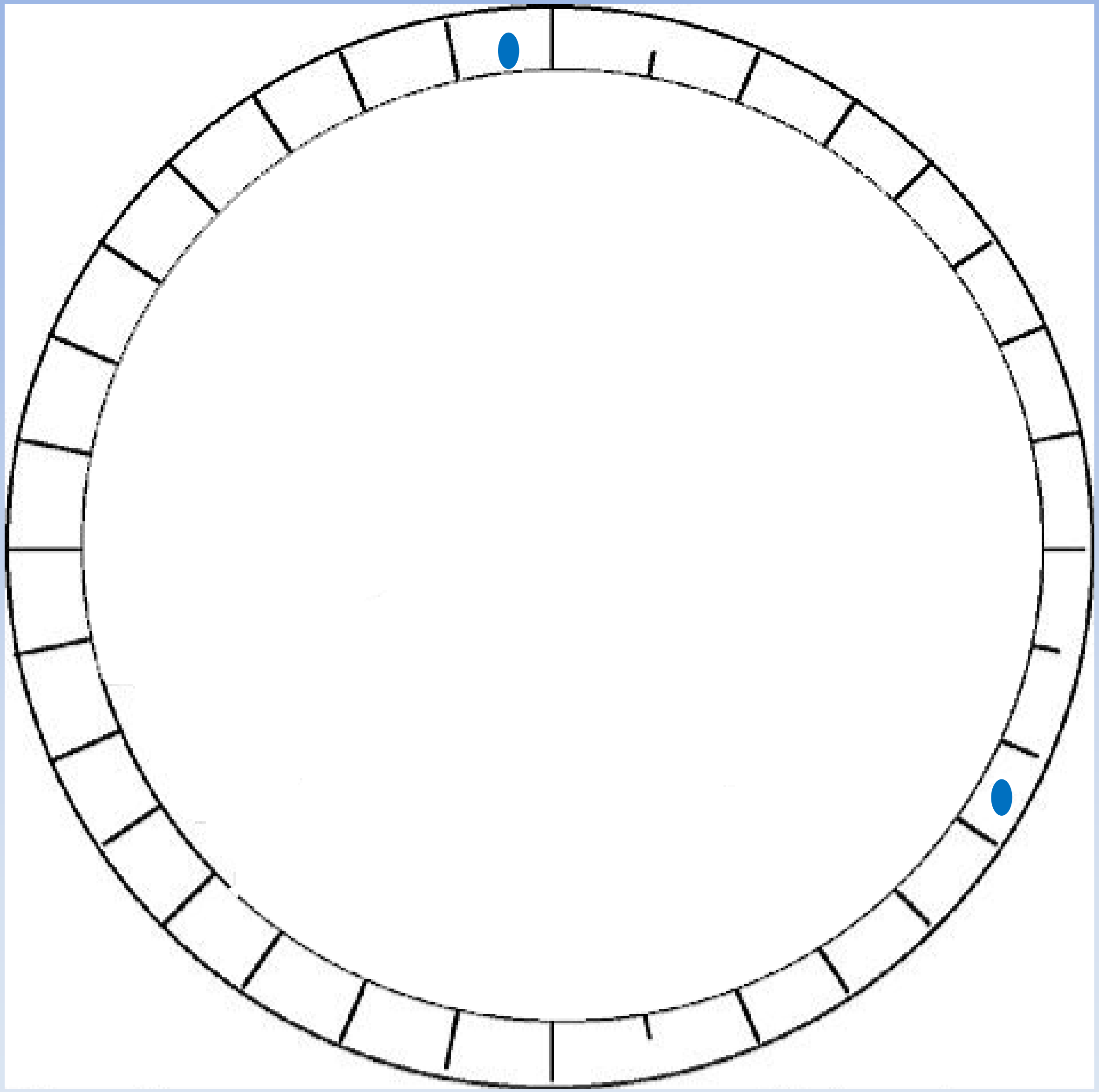
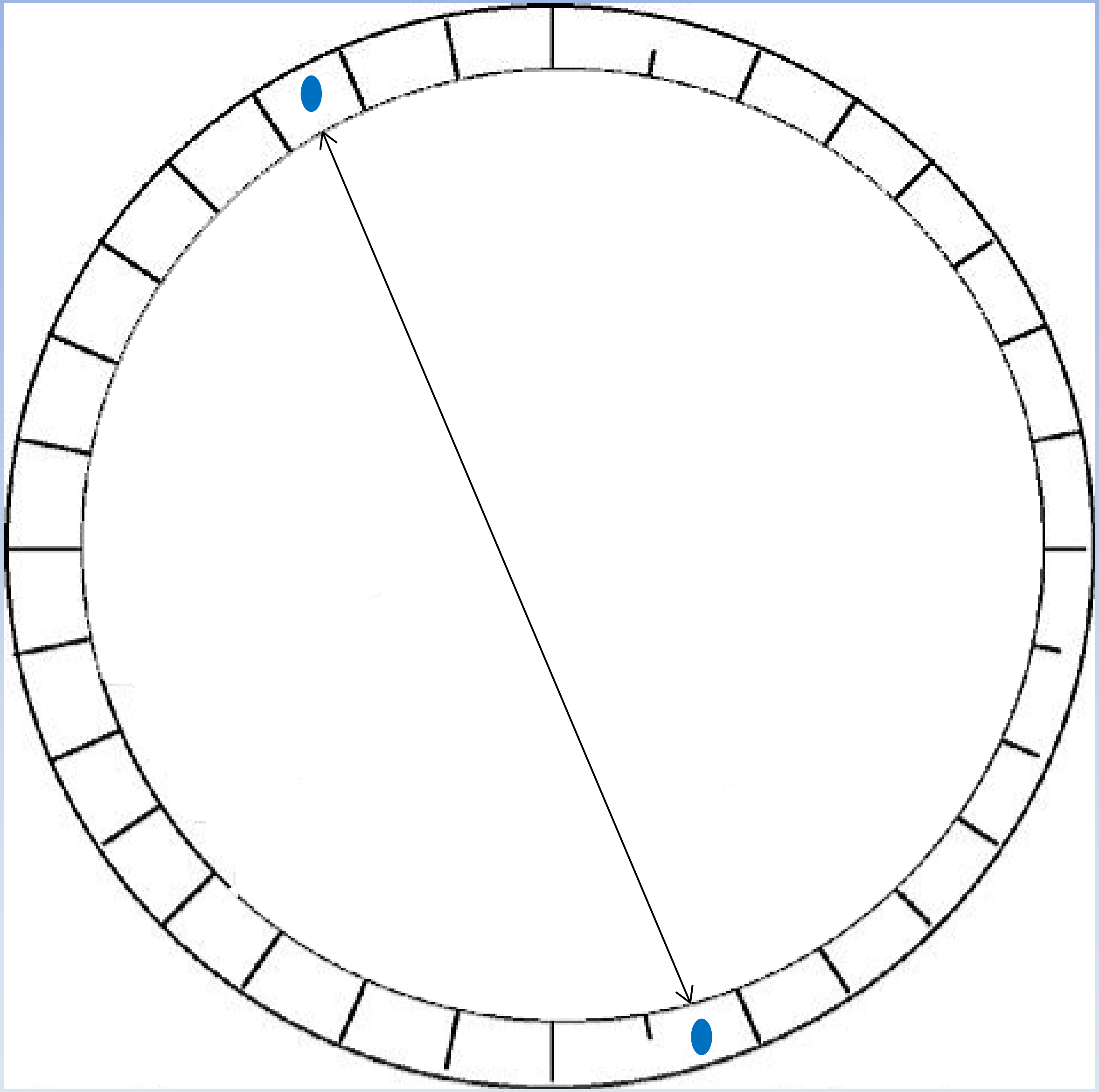


Building a Sinogram



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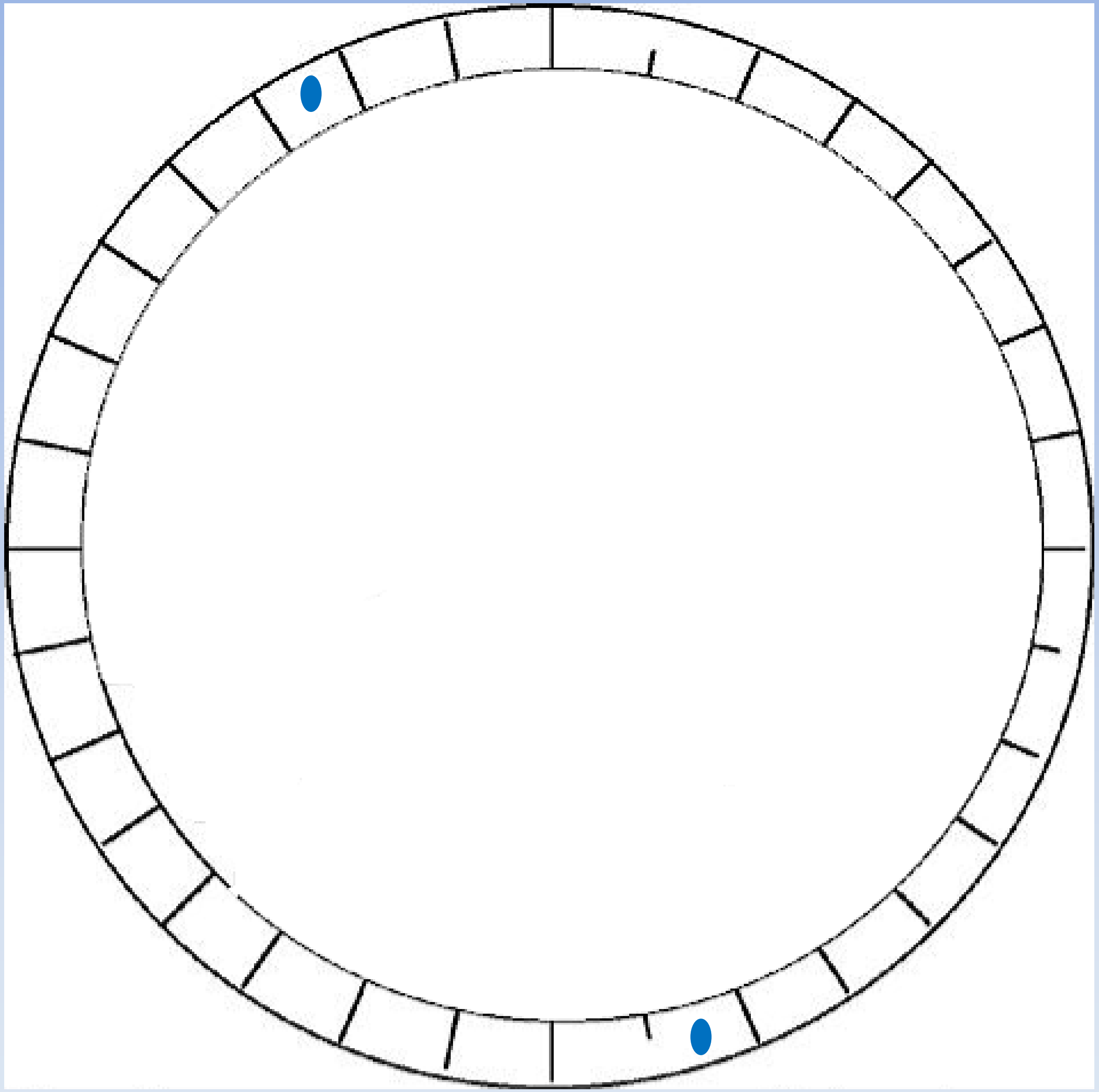

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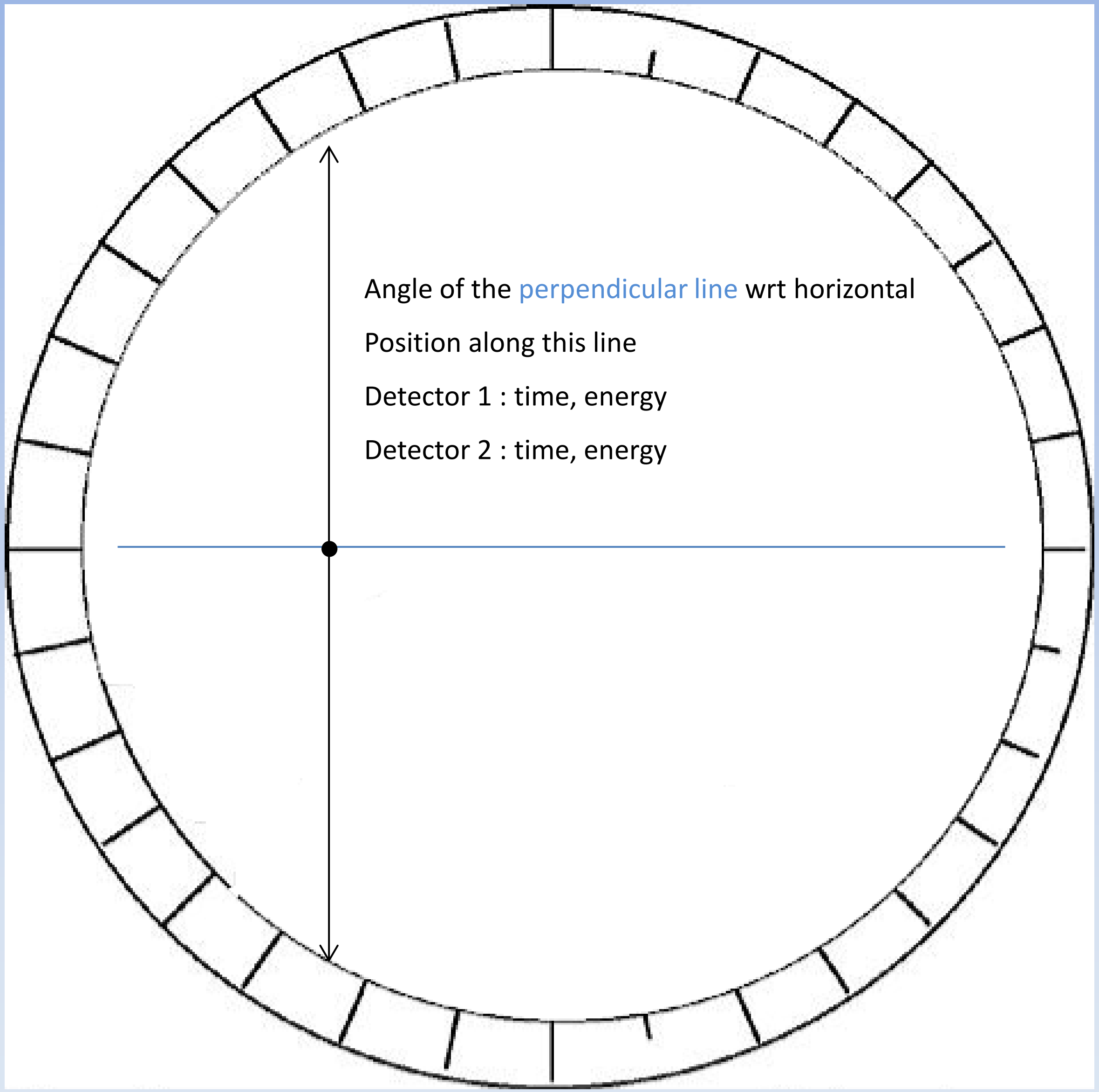

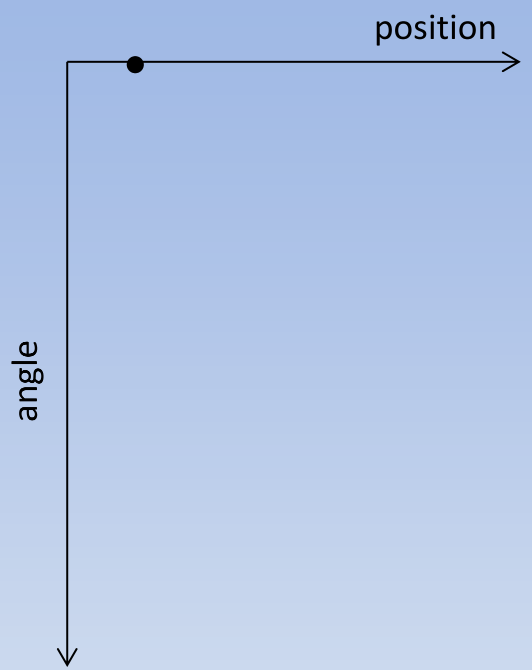

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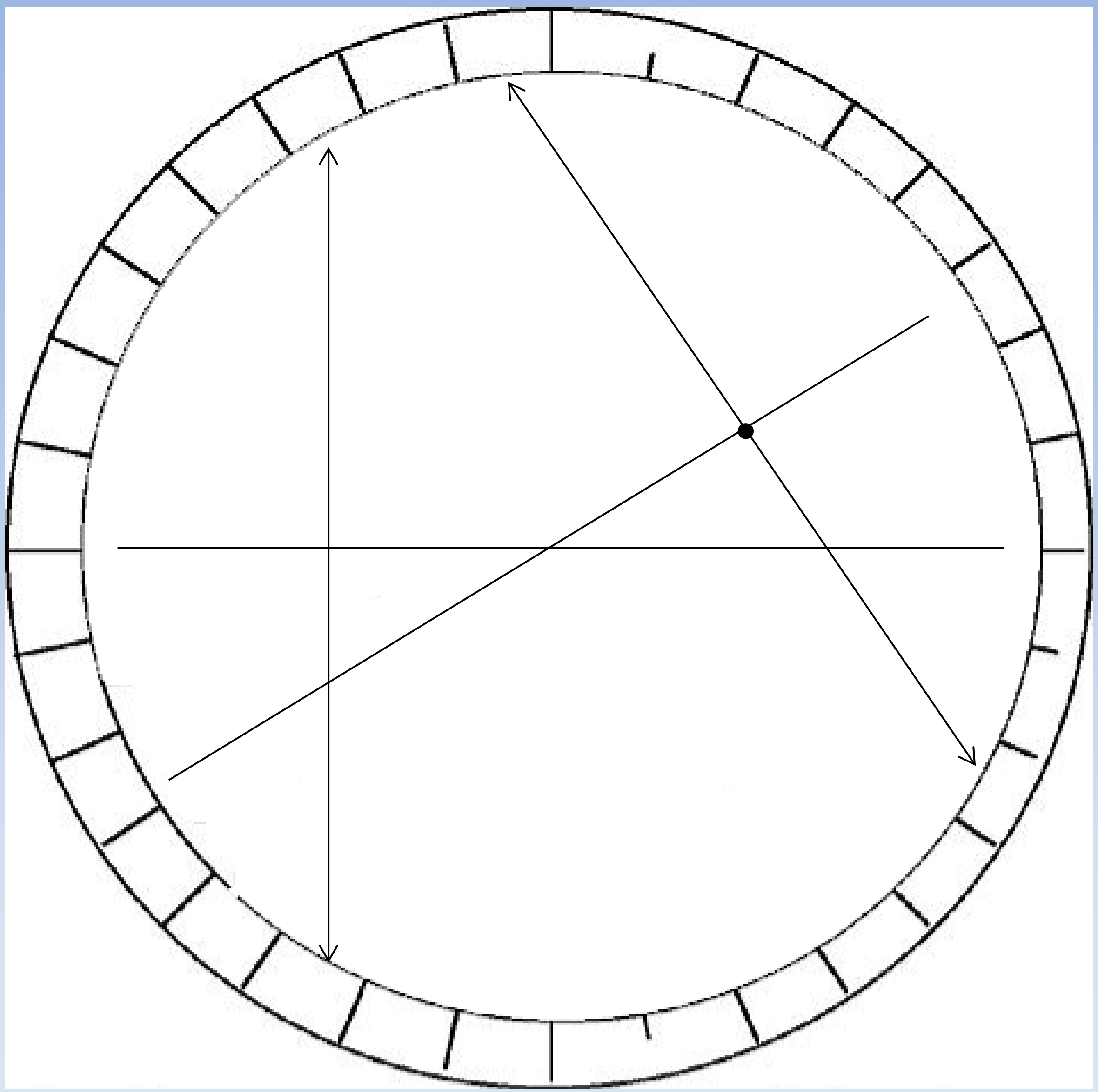

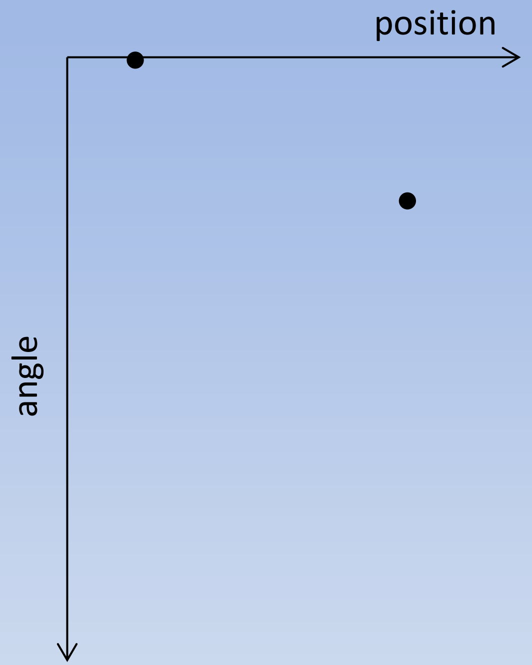

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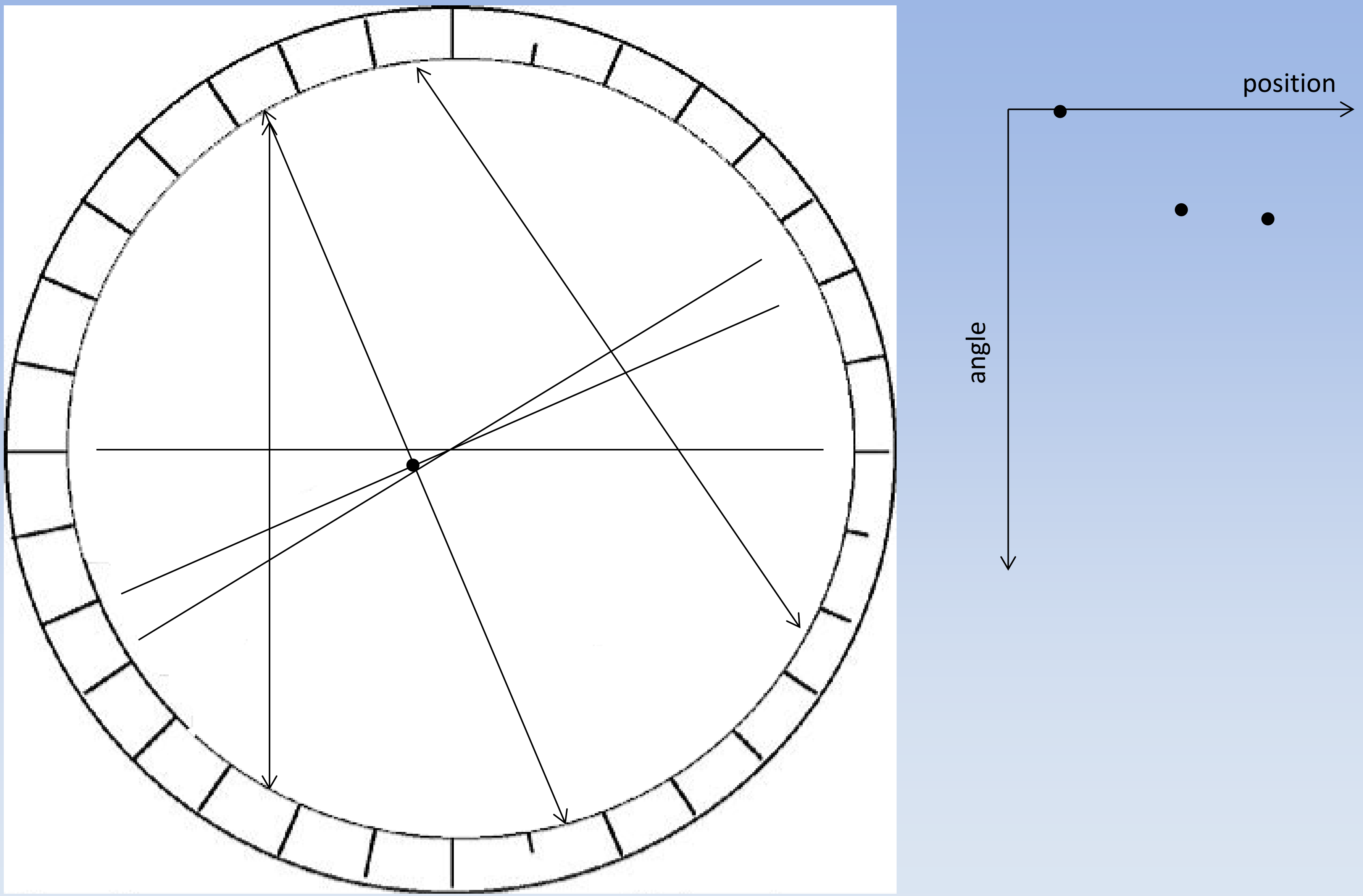

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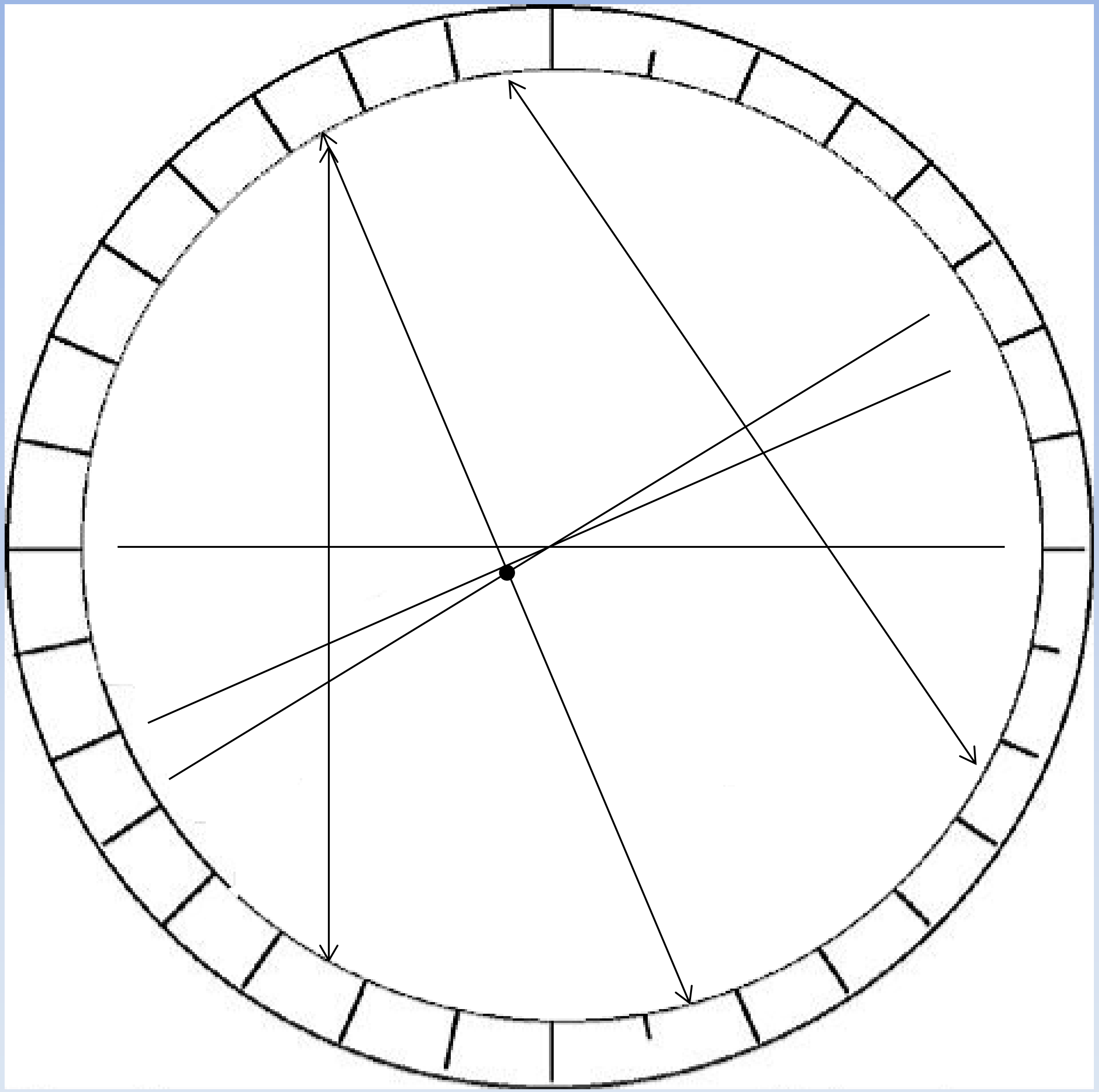
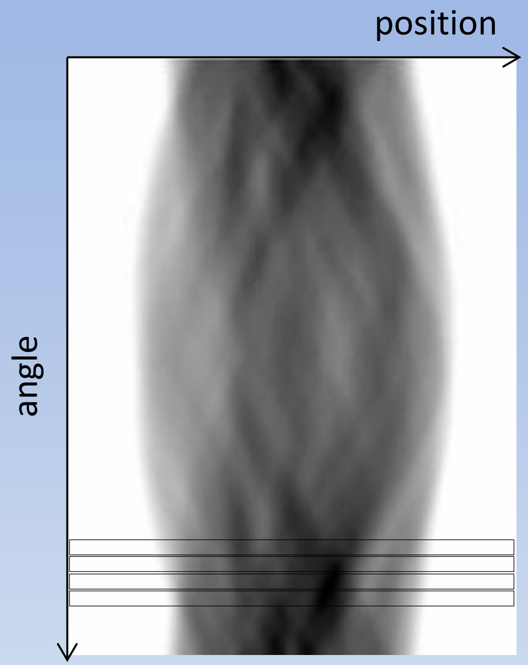


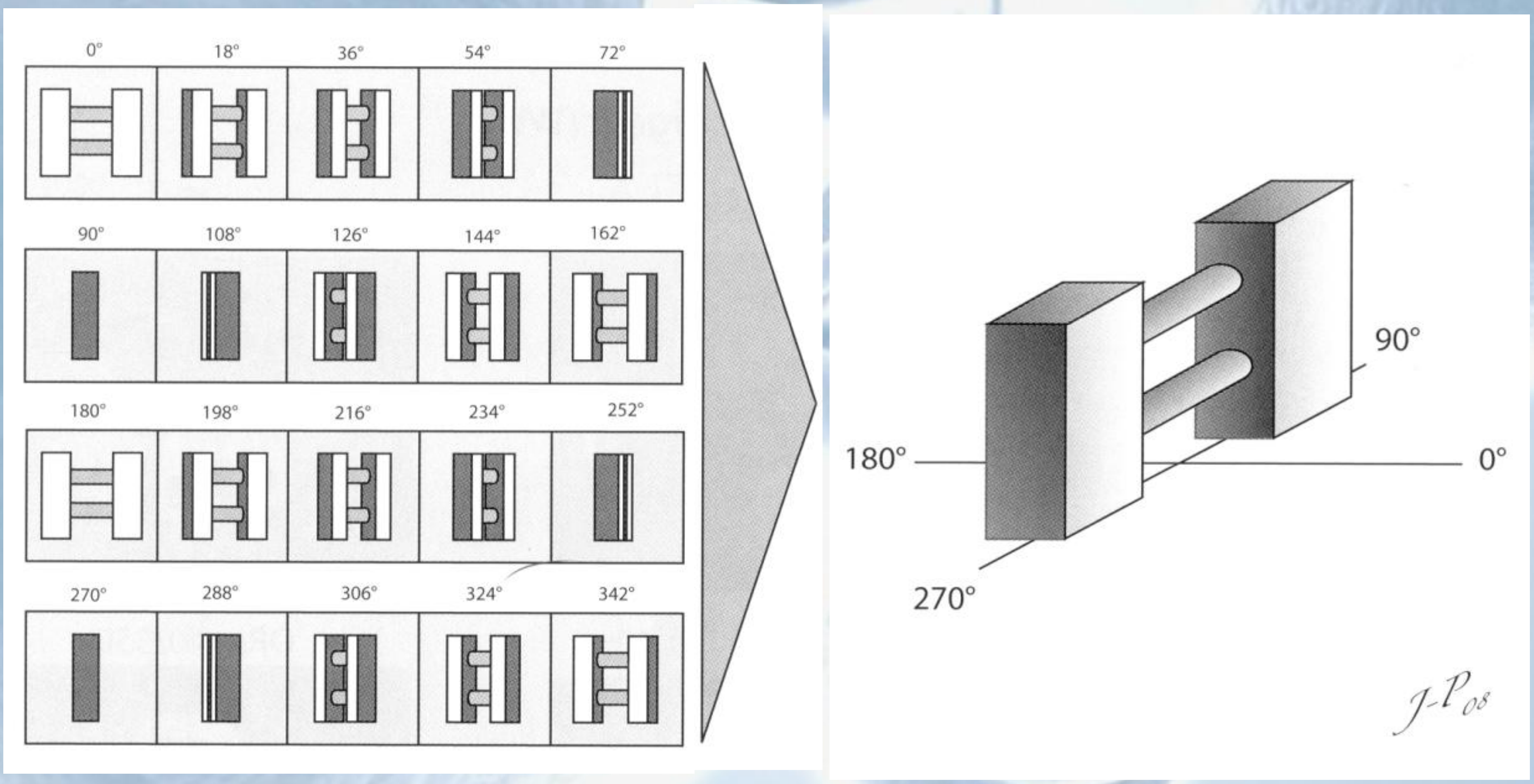


Set of 1D intensities, one for each projection angle

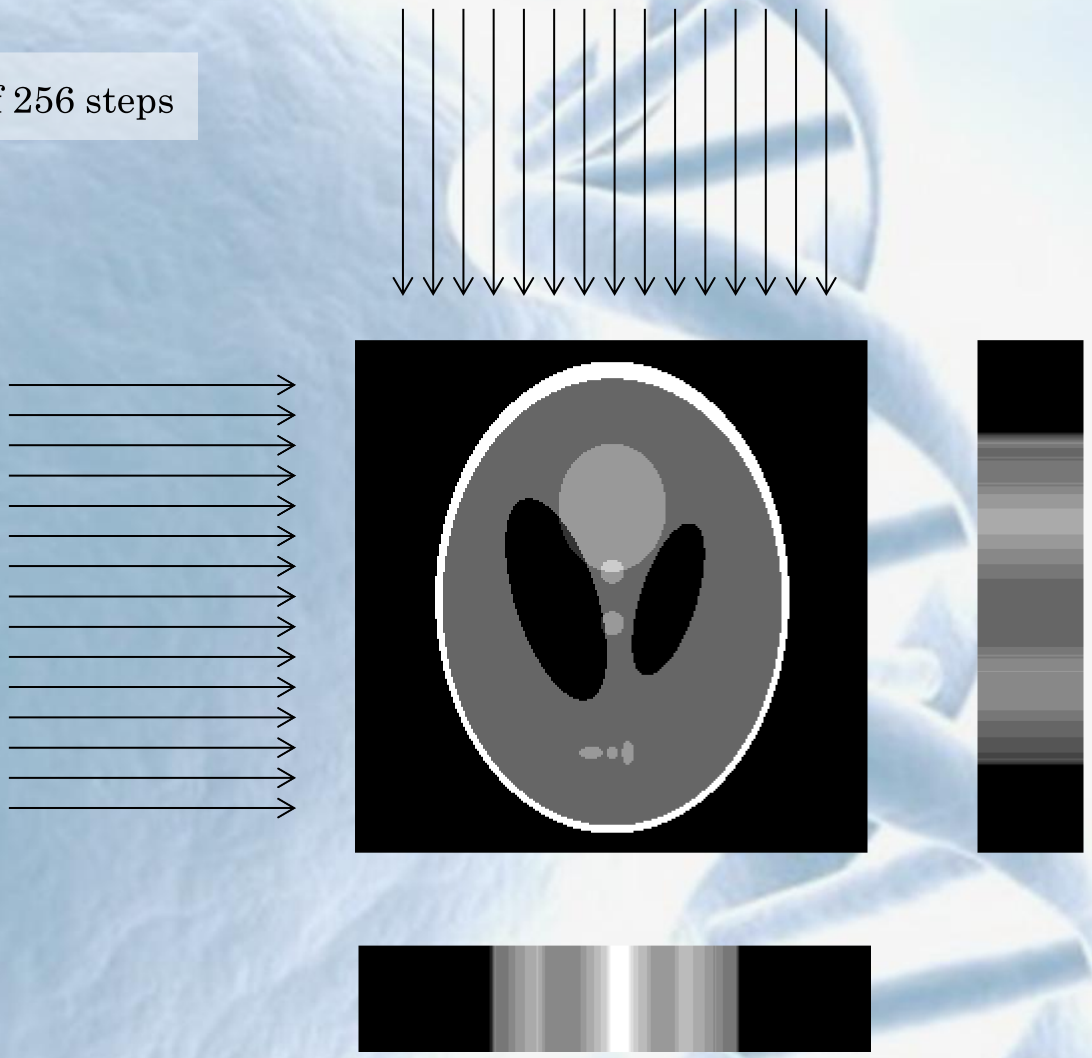
⇩

Location of F-18, in 2D

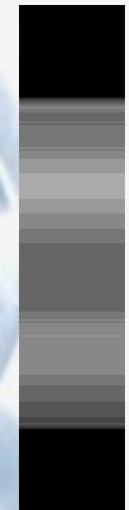
Tomography : how to get an extra dimension (eg : 2D → 3D) ?



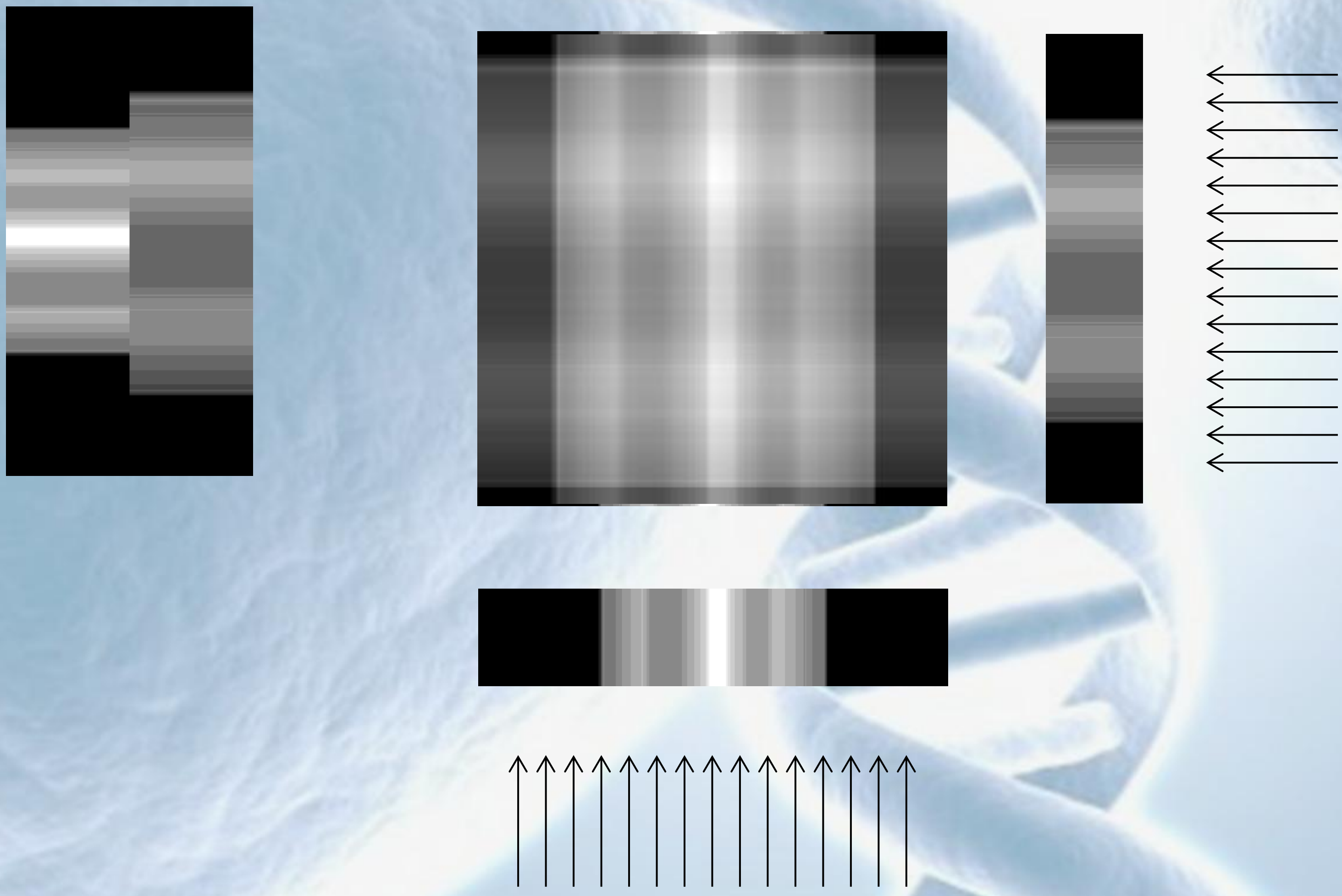
2 projections of 256 steps



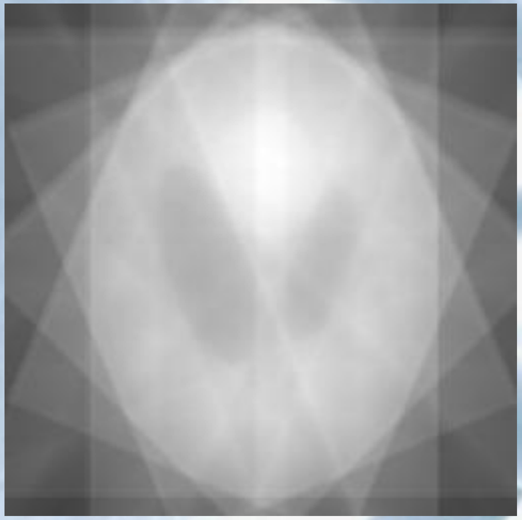
From 2 projections of 256 steps to the sinogram



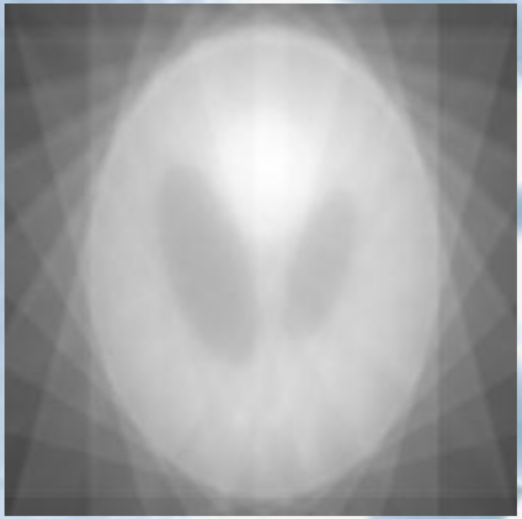
Back propagation of the sinogram with 2 projections



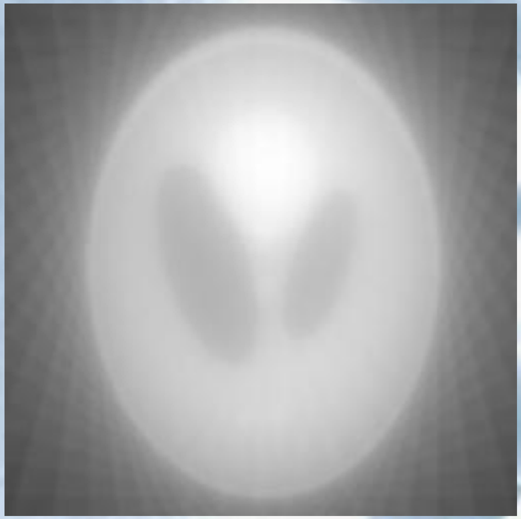
Back propagation of the sinogram with 8 projections



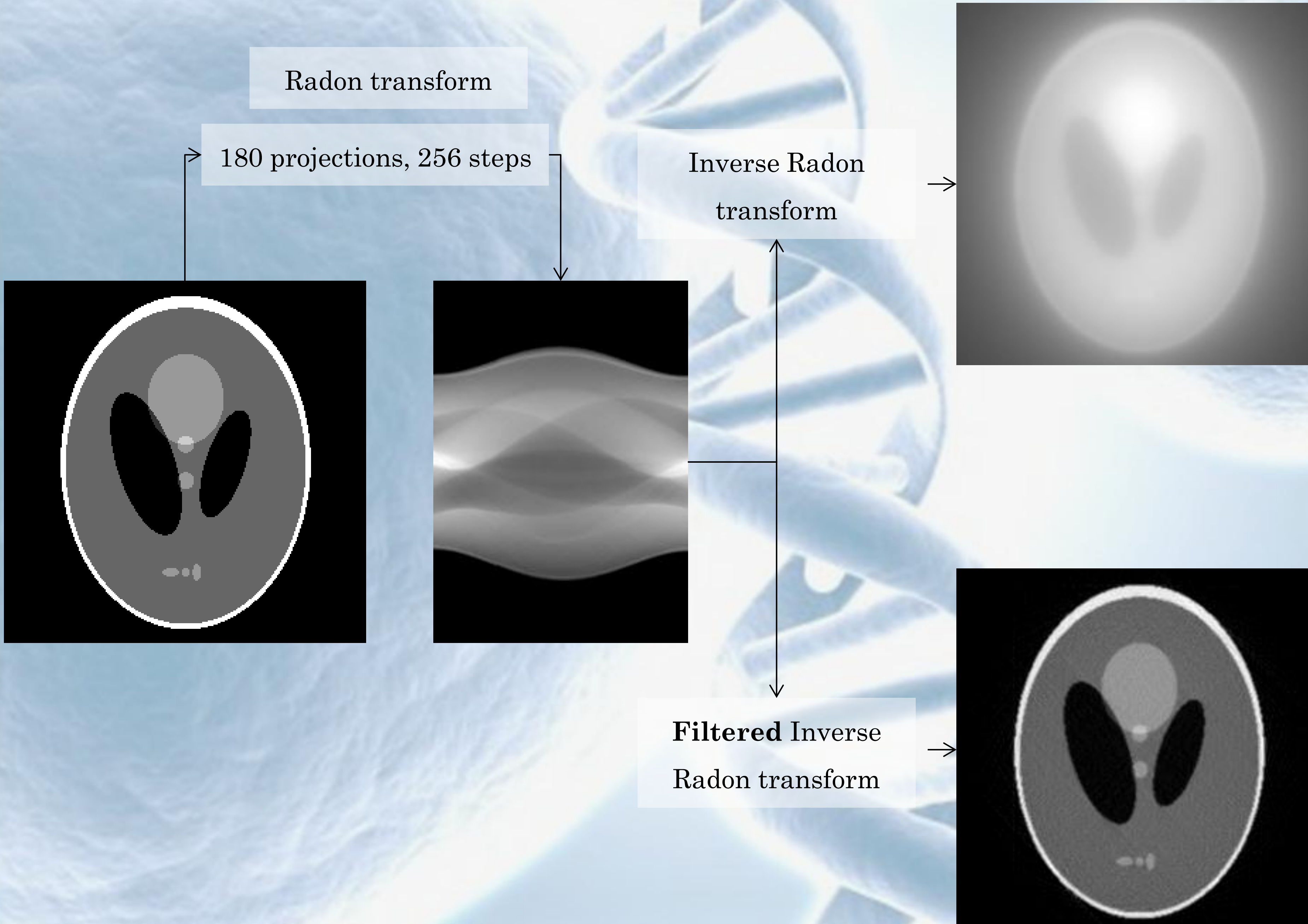
Back propagation of the sinogram with 12 projections

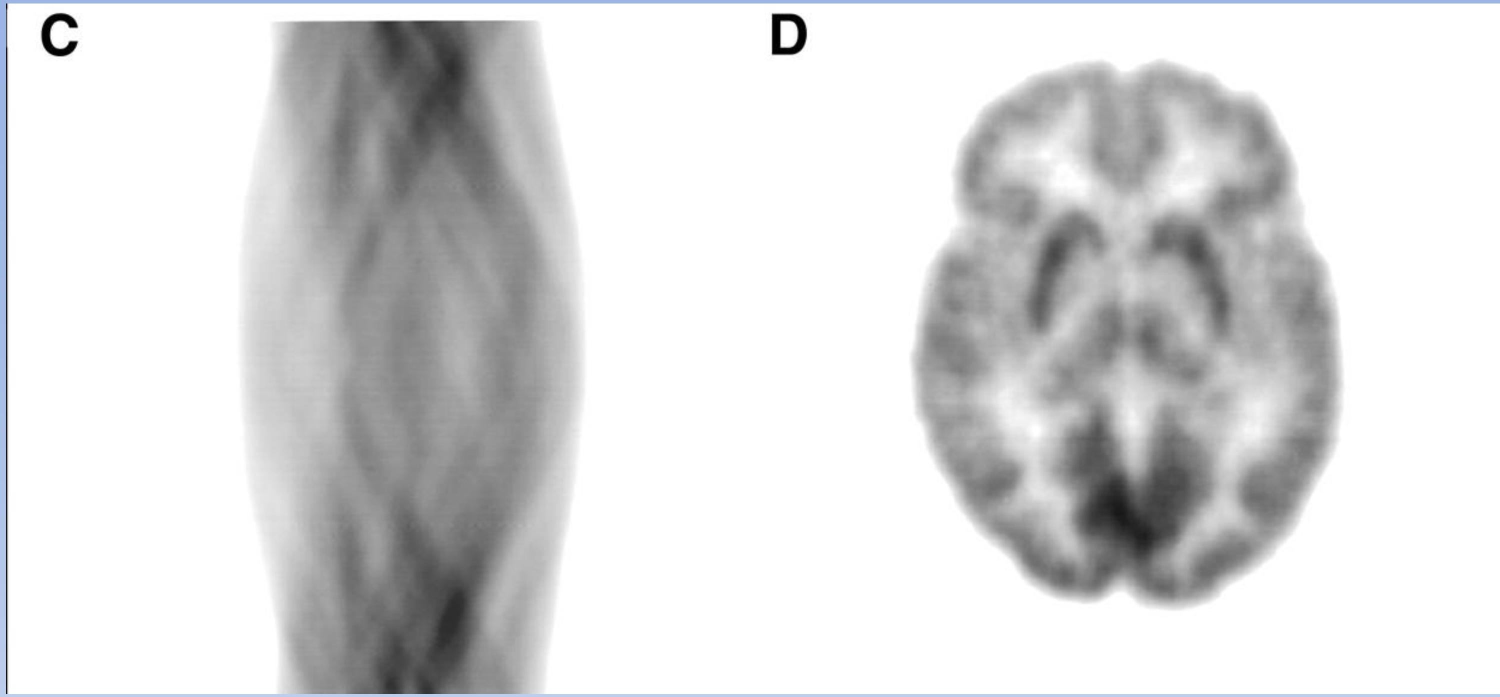


Back propagation of the sinogram with 24 projections



Back propagation of the sinogram with 180 projections

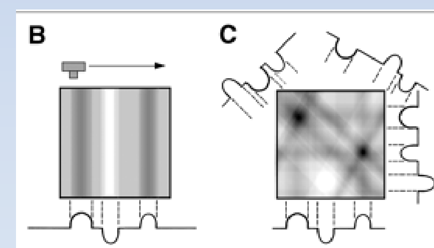

Radon transform
180 projections, 256 steps
Inverse Radon transform
Filtered Inverse Radon transform



Sinogram

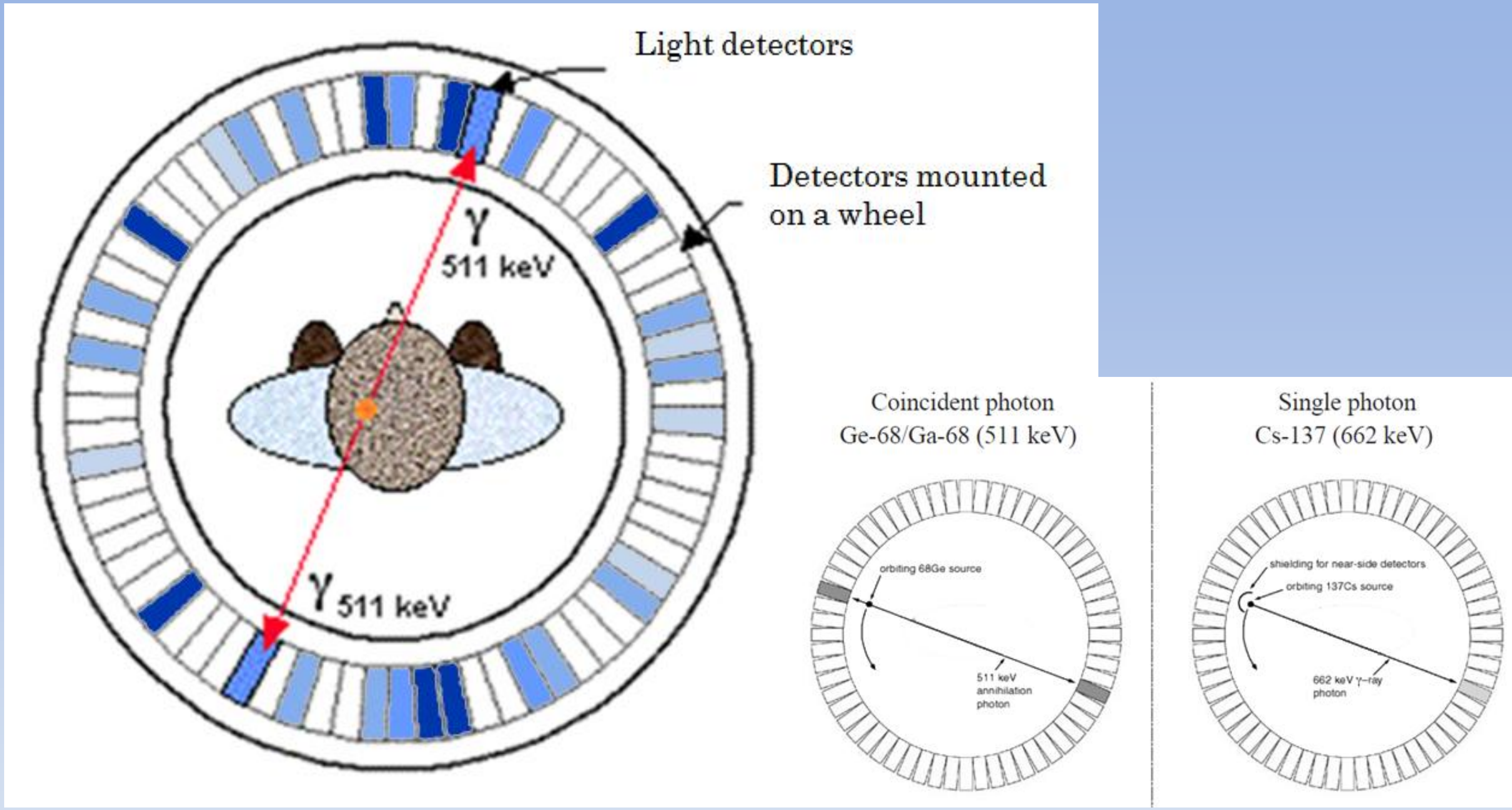
(or data in listmode)

Reconstructed image



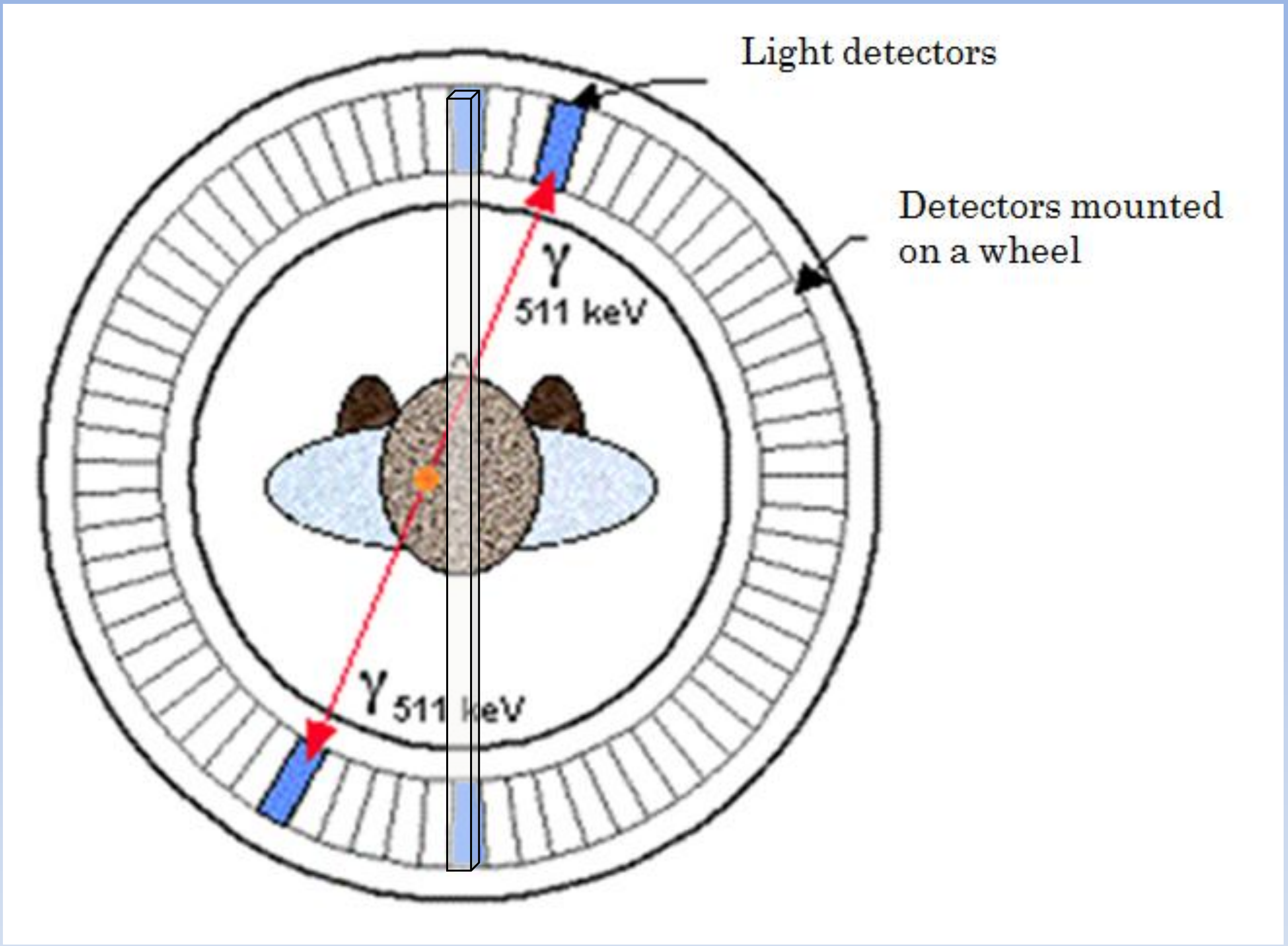
Corrections + Radon inverse transform + filtering + …

or corrections + OSEM 3D (ordered subset expectation maximization)

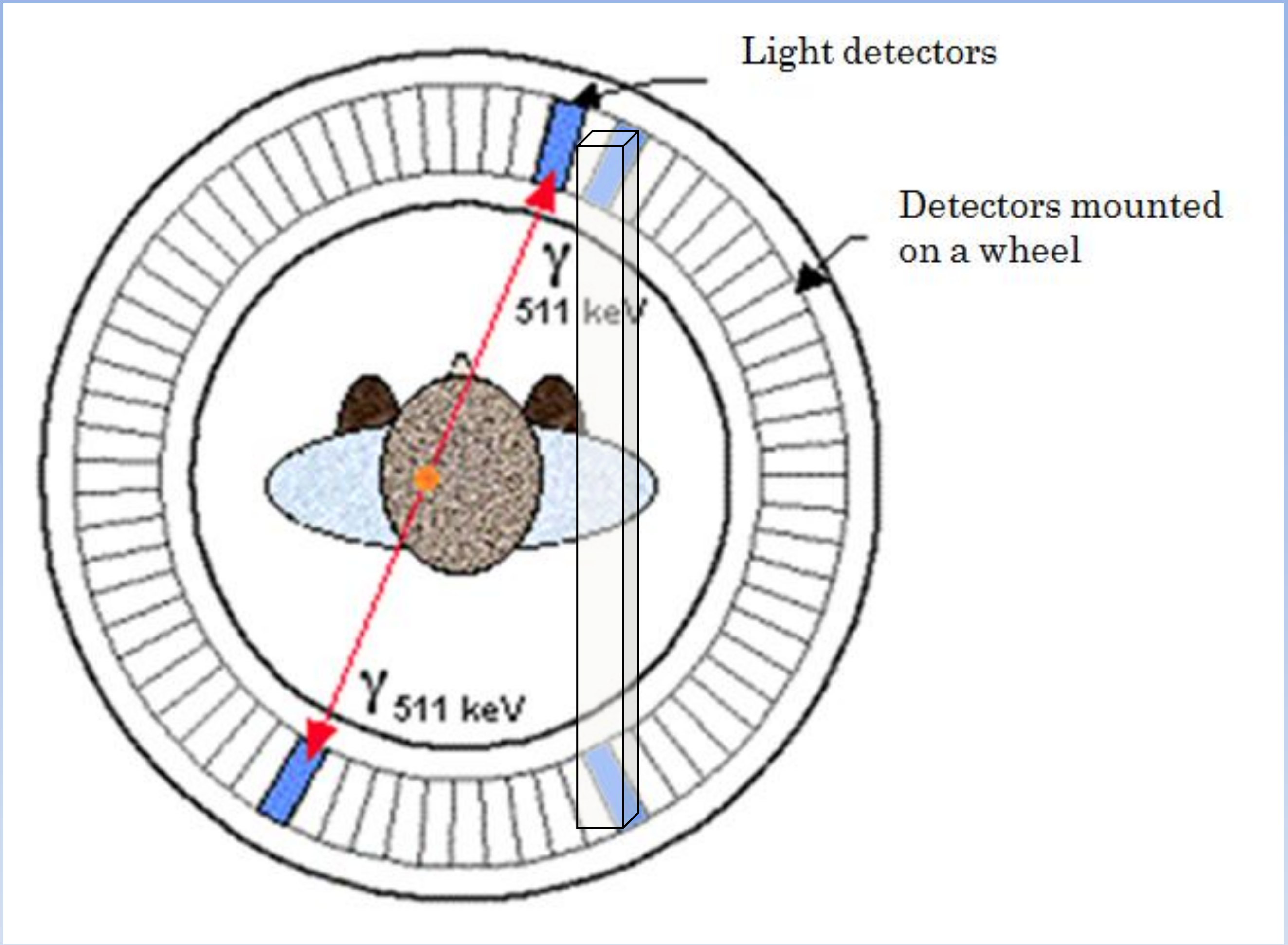


Overall detector response will depend on individual channel gain : need for calibration.

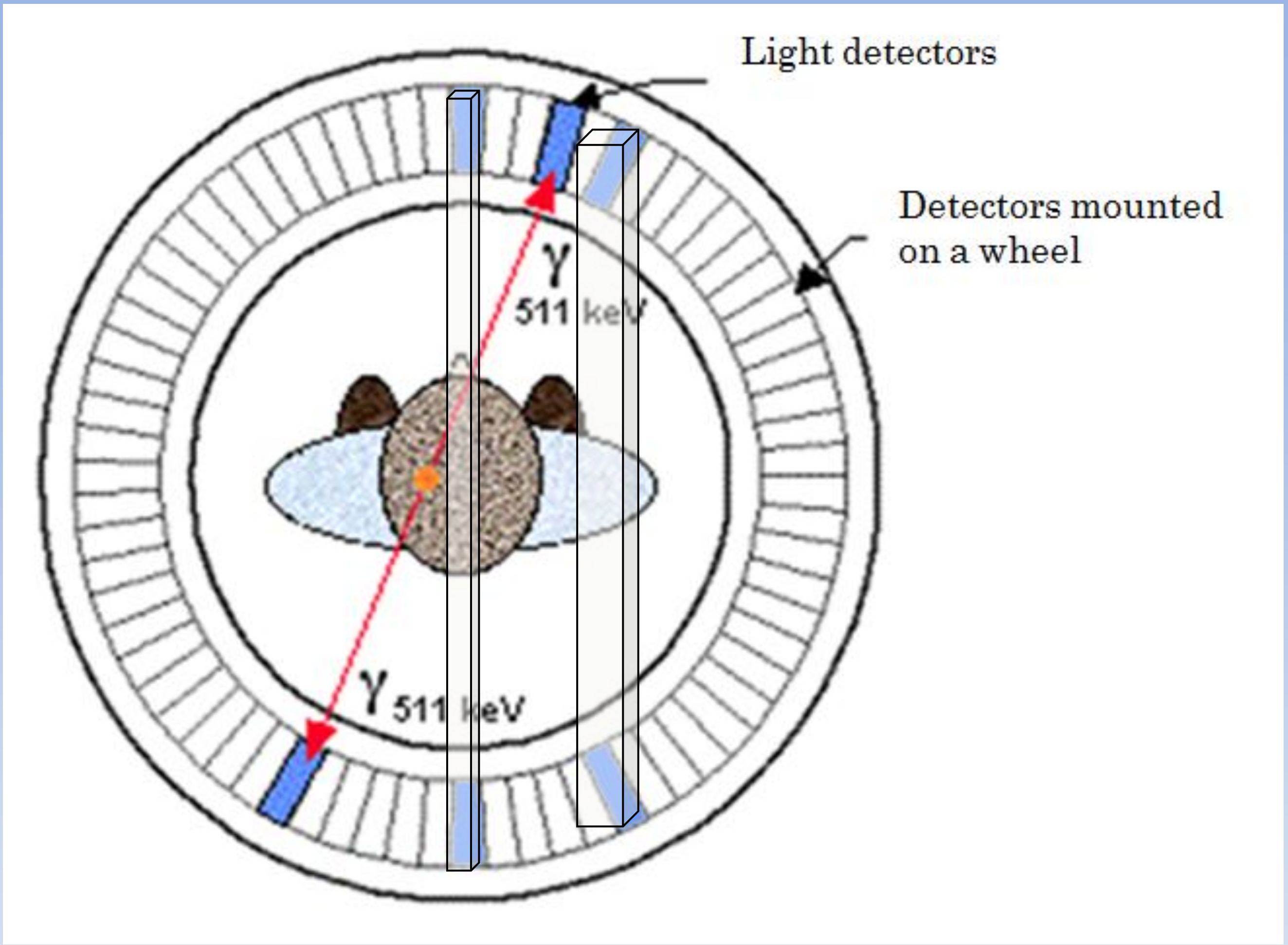
- Blank acquisition (noise)
- Calibration with a gamma emitter (energy close to 511 keV) rotating all around the detector ring



Line of response is indeed a volume of response

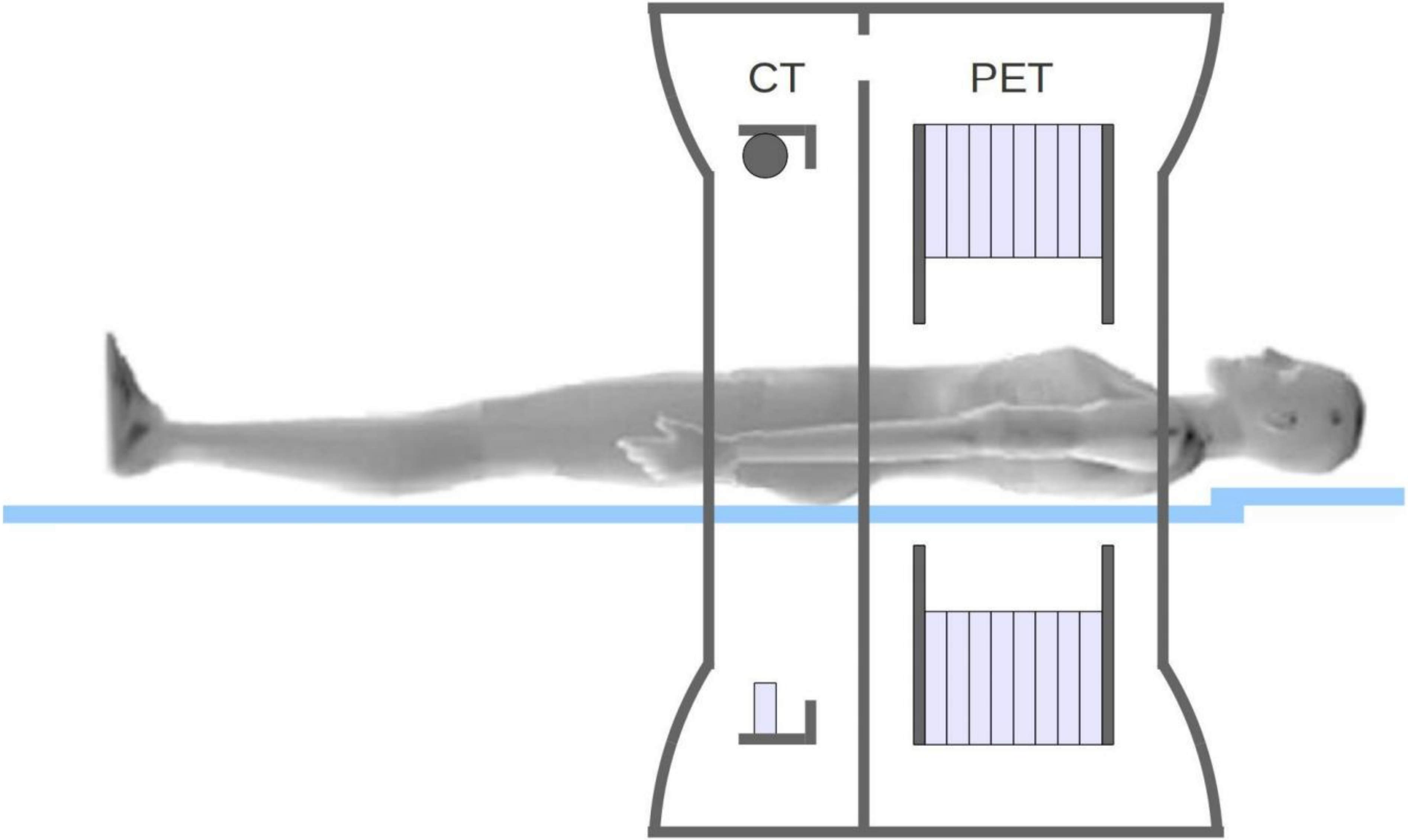


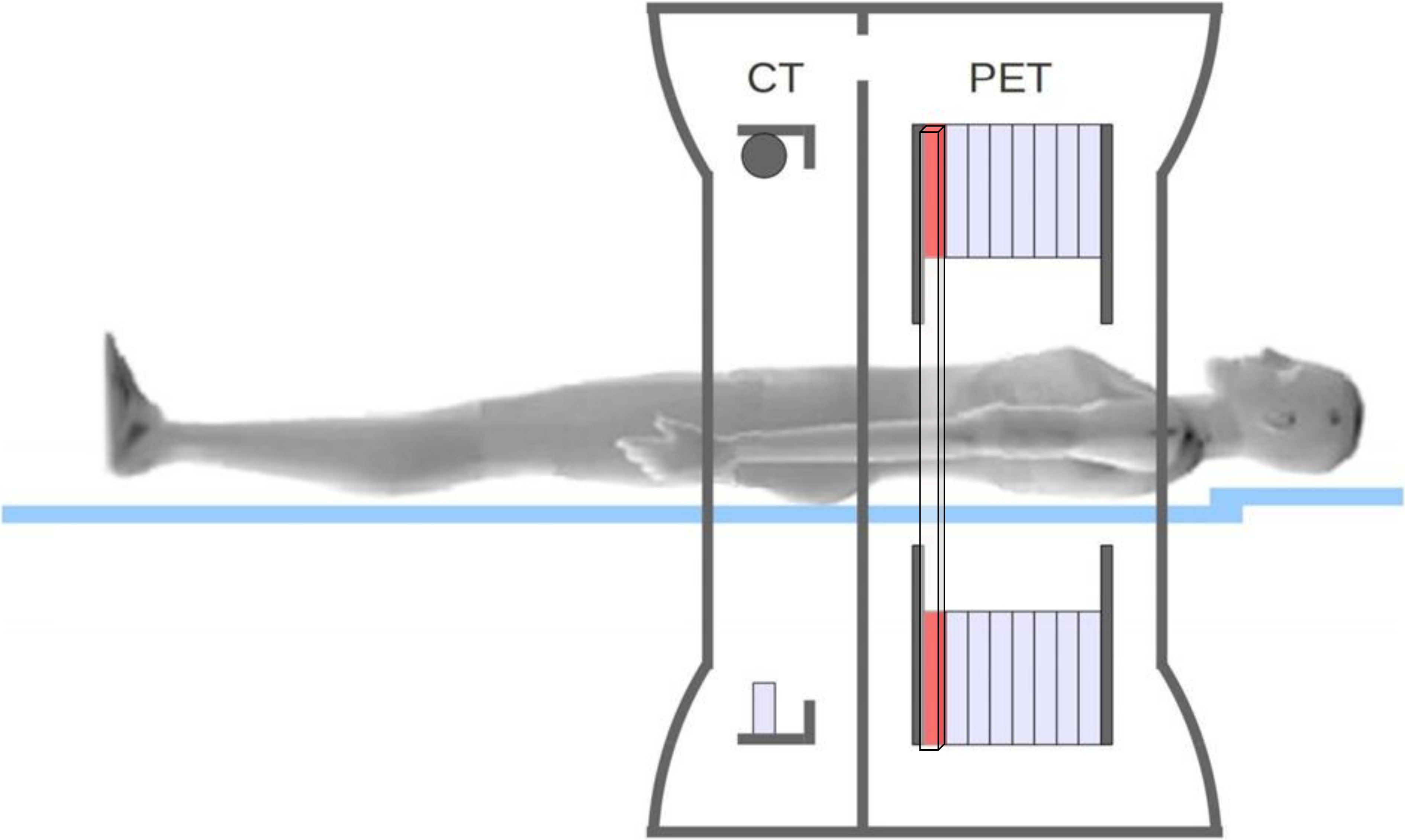
Volume of response depends on the source location w.r.t. the arc structure



Intensity of response and spatial accuracy will depend on the off-axis position of the emitter

- Geometrical corrections : use Ge-68 to activate all possible LOR situations
- Spatial resolution could be improved by measuring the Depth of Interaction (DOI)


CT
PET



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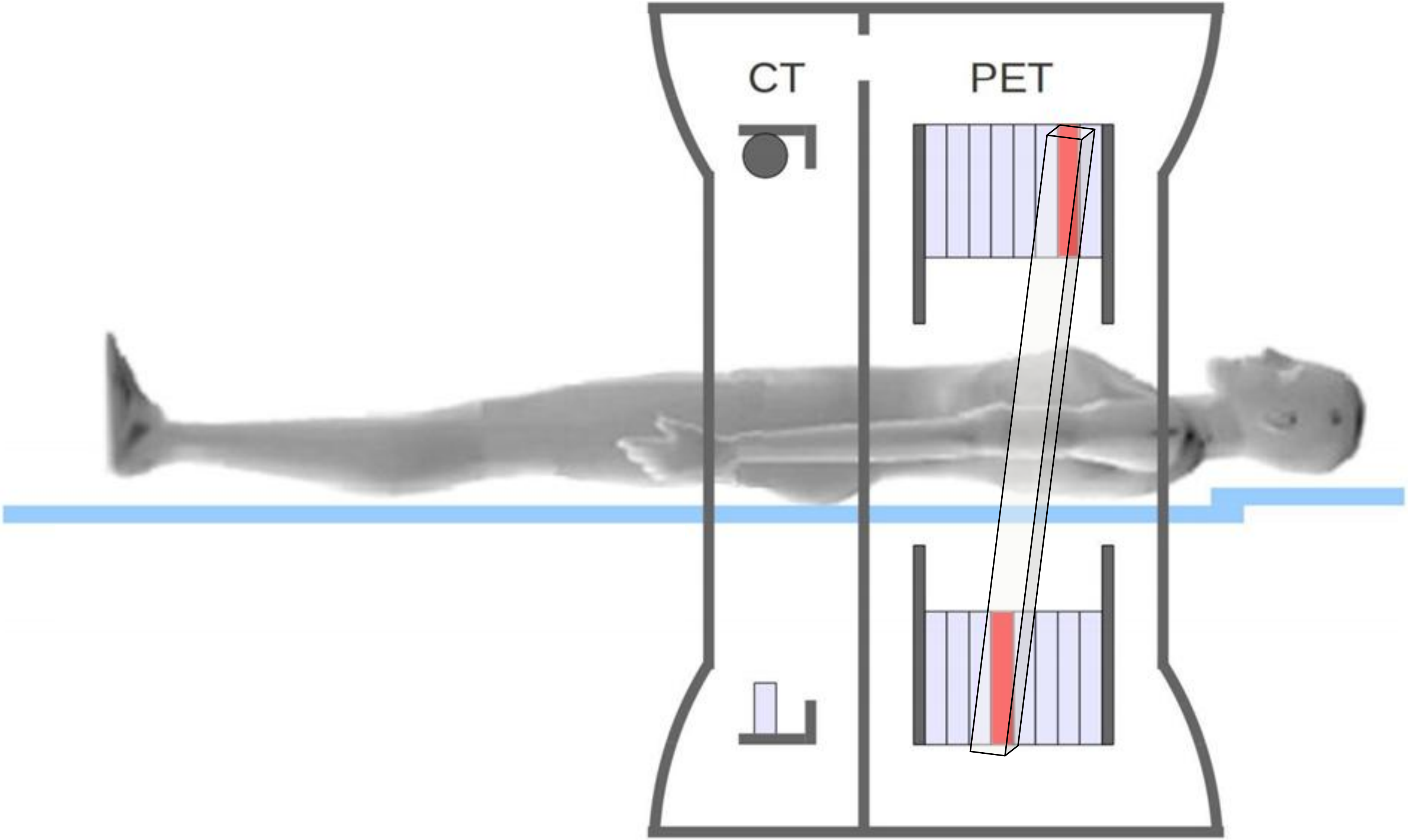
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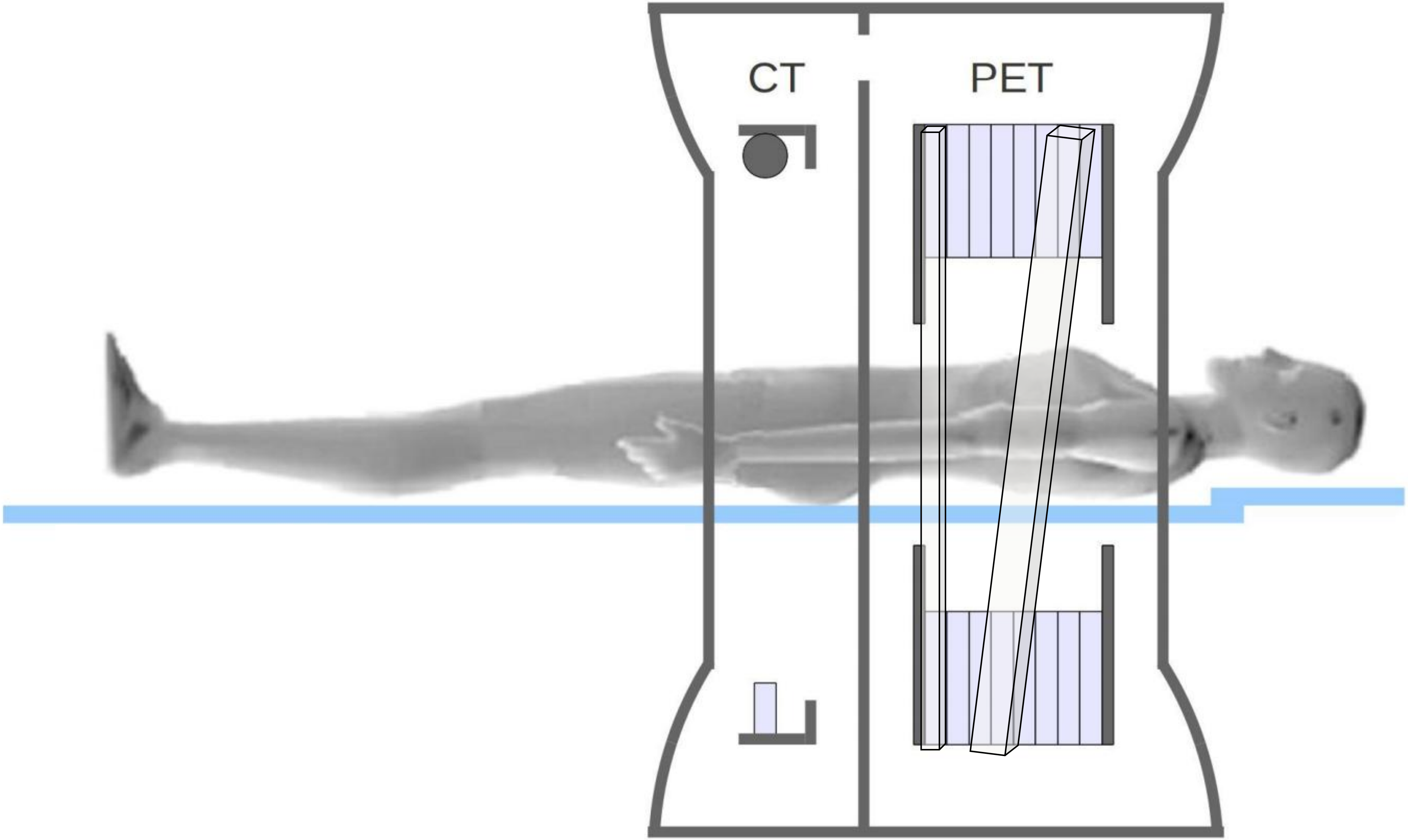
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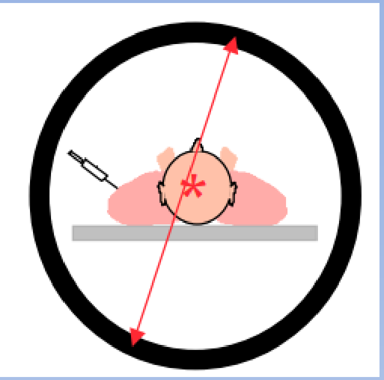
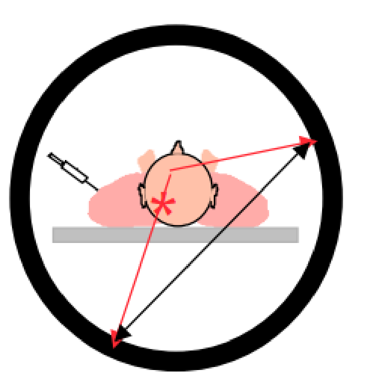


Real coincidence : useful information



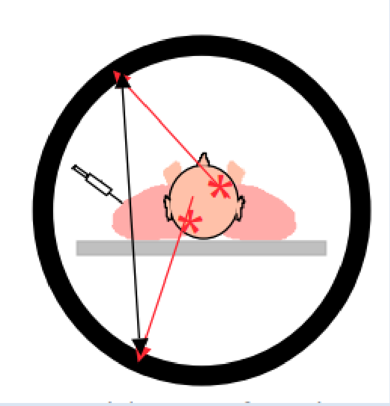
Scattered coincidence : useless information

Bad localization

Decreases contrast

Bias in quantification

Reduced by measuring the photon energy



Random coincidence : useless information

Bad localization

Decreases contrast

Dead time

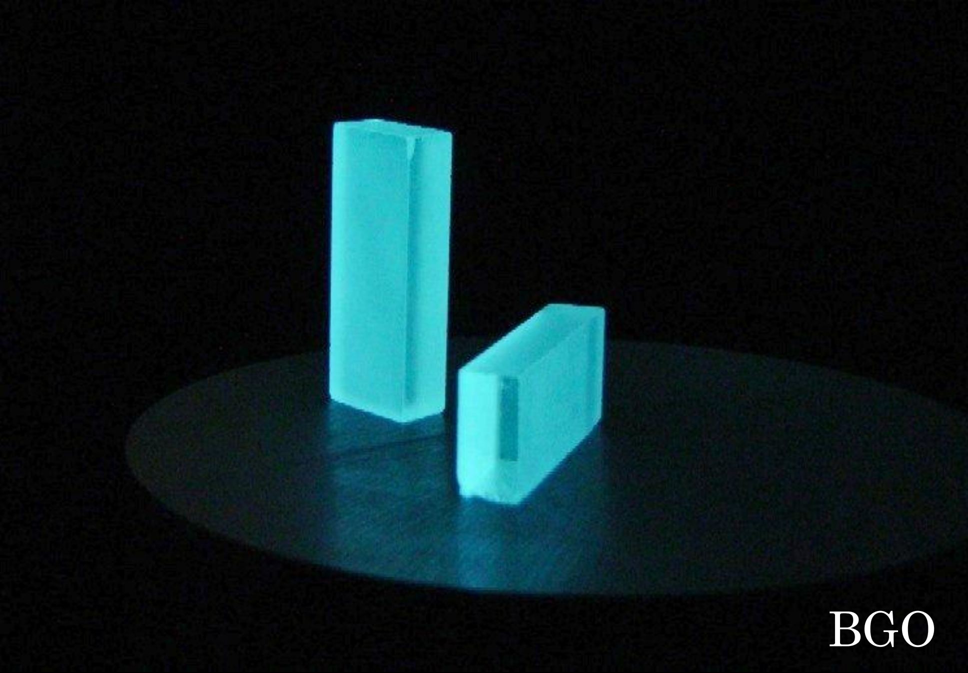
Bias in quantification

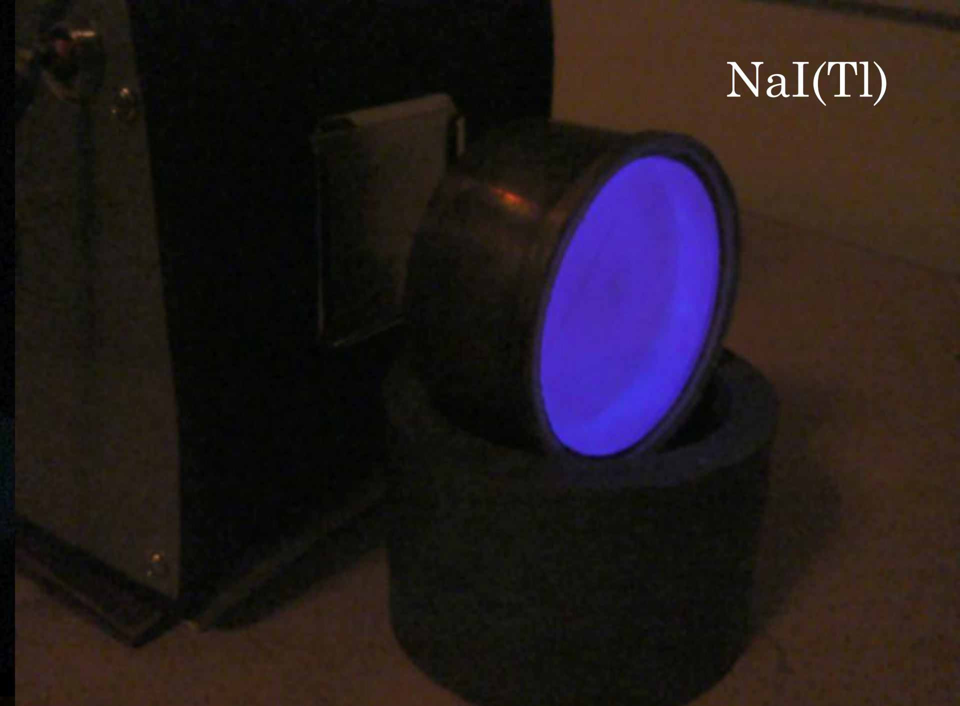
Reduced by accurate timing measurement

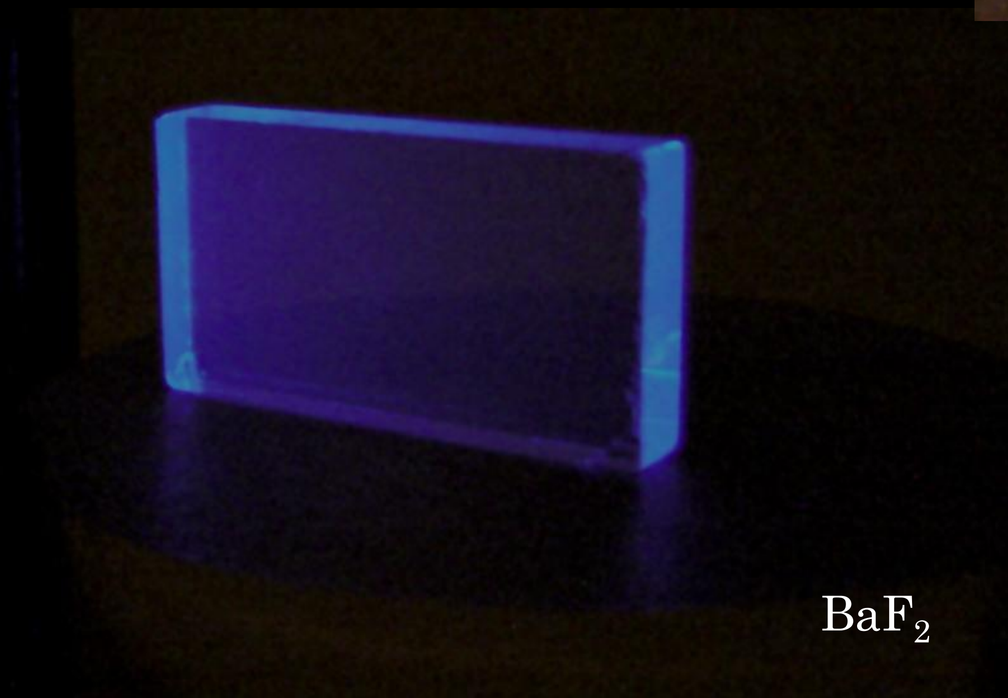
How to get signal from 511 keV photon interactions ?

X photons cannot be directly detected : first need an interaction medium

Inorganic scintillators provide visible light that can be detected







Inorganic crystals irradiated with a X-ray tube (dose rate in the air up to 2,5 Gy/h)

Visible light is produced

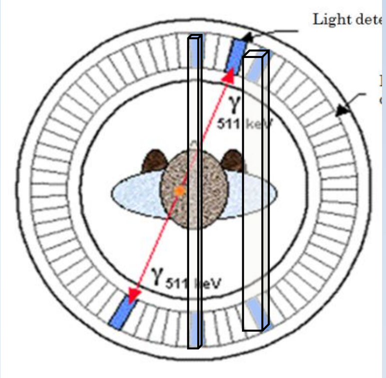
Exposure time from 5 to 15 seconds.

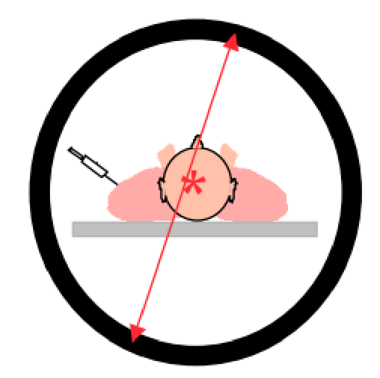
Scintillator : Light production from 511 keV photon interaction

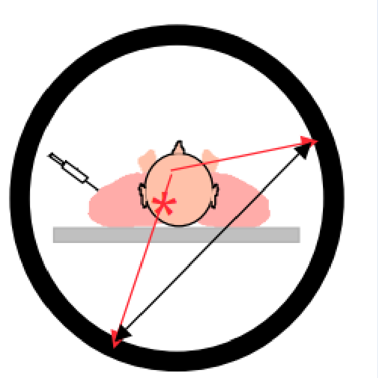
X photons cannot be detected : first need an interaction medium to generate visible photons

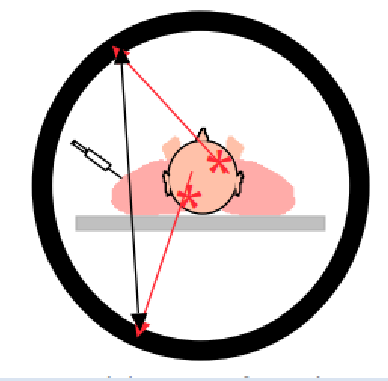
Specification :

- Photon energy converted in a tiny volume : high stopping power (small X_0)
- Short light pulse : remove random coincidences and decrease dead time
- Good energy resolution : remove scattered events
- Light yield and wavelength matching with visible light detector sensitivity
- Stability versus accumulated dose
- Reasonable price









Scintillator : Light production from 511 keV photon interaction

	NaI(Tl)	BaF_2	BGO	LSO	GSO	LYSO	$LaBr_3$
Effective atomic no. (Z)	51	54	74	66	59	60	47
Linear attenuation coeff. (cm^{-1})	0.34	0.44	0.92	0.87	0.62	0.86	0.47
Density (gm cm^{-3})	3.67	4.89	7.13	7.4	6.7	7.1	5.3
Index of refraction	1.85	–	2.15	1.82	1.85	1.81	1.88
Light yield (% NaI(Tl))	100	5	15	75	30	80	160
Peak wavelength (nm)	410	220	480	420	430	420	370
Decay constant (ns)	230	0.8	300	40	65	41	25
Hydroscopic	Yes	Slight	No	No	No	No	No

LYSO(Ce) is very similar to LSO(Ce) except that the crystal growing process is slightly different and with less impurities at the end. LYSO properties drift w.r.t. accumulated dose is reduced.

Materials **2011**, *4*, 1224-1237; doi:10.3390/ma4071224
Defects Identification and Effects of Annealing on Lu2(1-x)Y2xSiO5 (LYSO) Single Crystals for Scintillation Application
Samuel Blahuta et al.

Scintillator : energy resolution

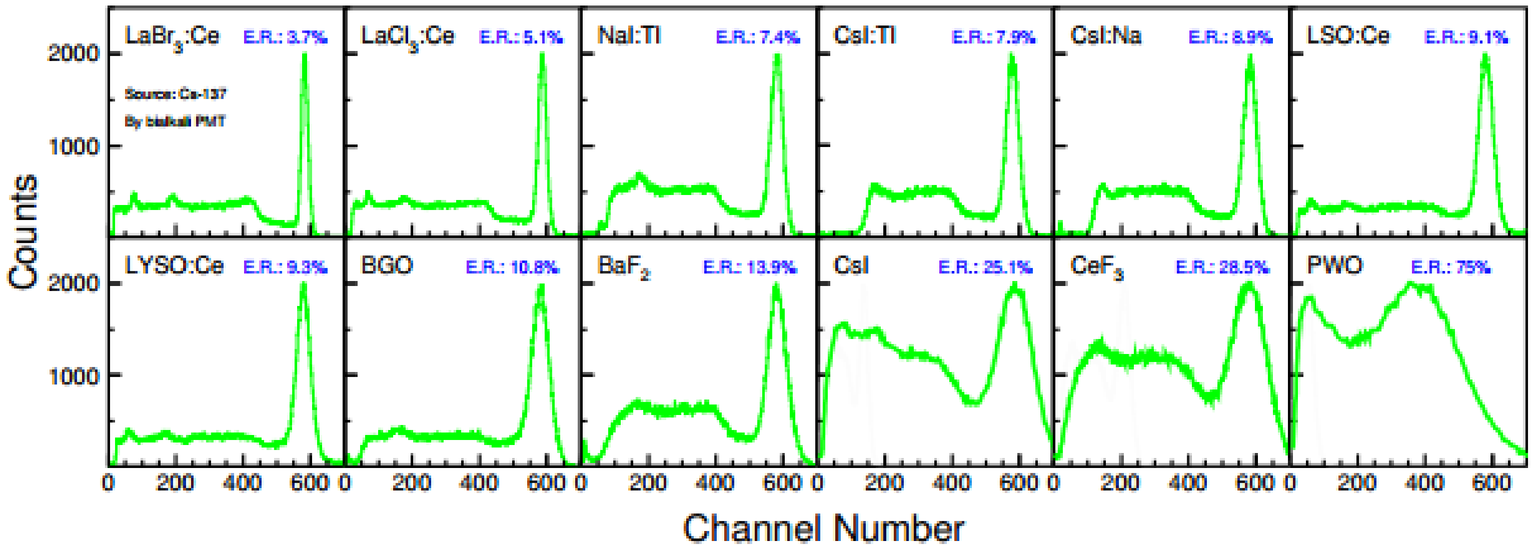


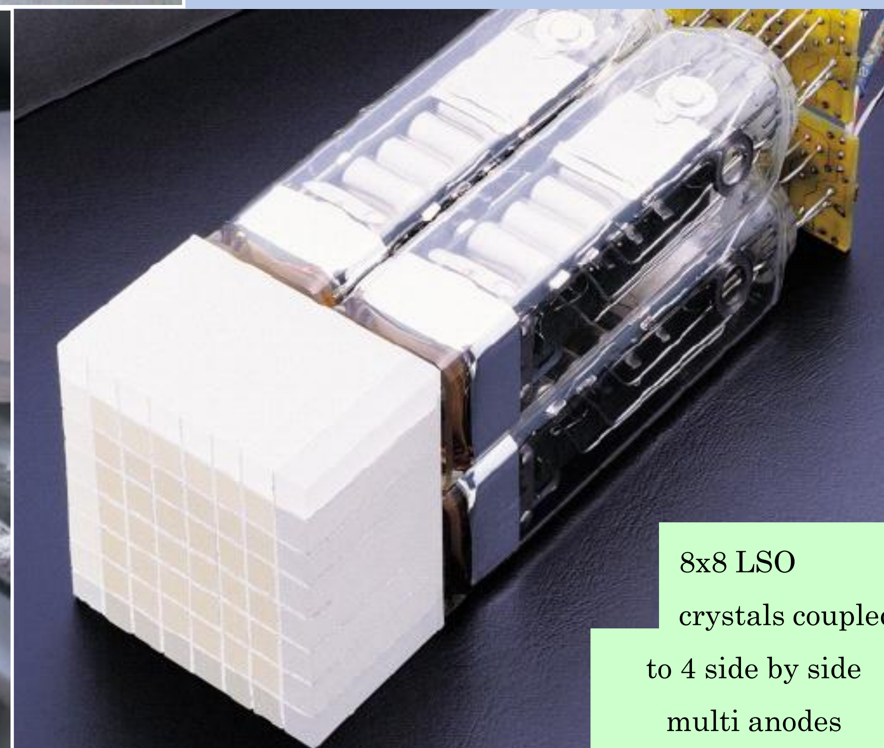
Figure 3. ^{137}Cs γ-ray pulse hight spectra measured by a Hamamatsu R1306 PMT are shown for twelve crystal samples. The numerical values of the FWHM resolution (E.R.) are also shown in the figure.

LSO/LYSO Crystals for Future HEP Experiments
2011 J. Phys.: Conf. Ser. 293 012004
(http://iopscience.iop.org/1742-6596/293/1/012004)

Growth of LSO crystal

LSO crystals

Slicing and polishing

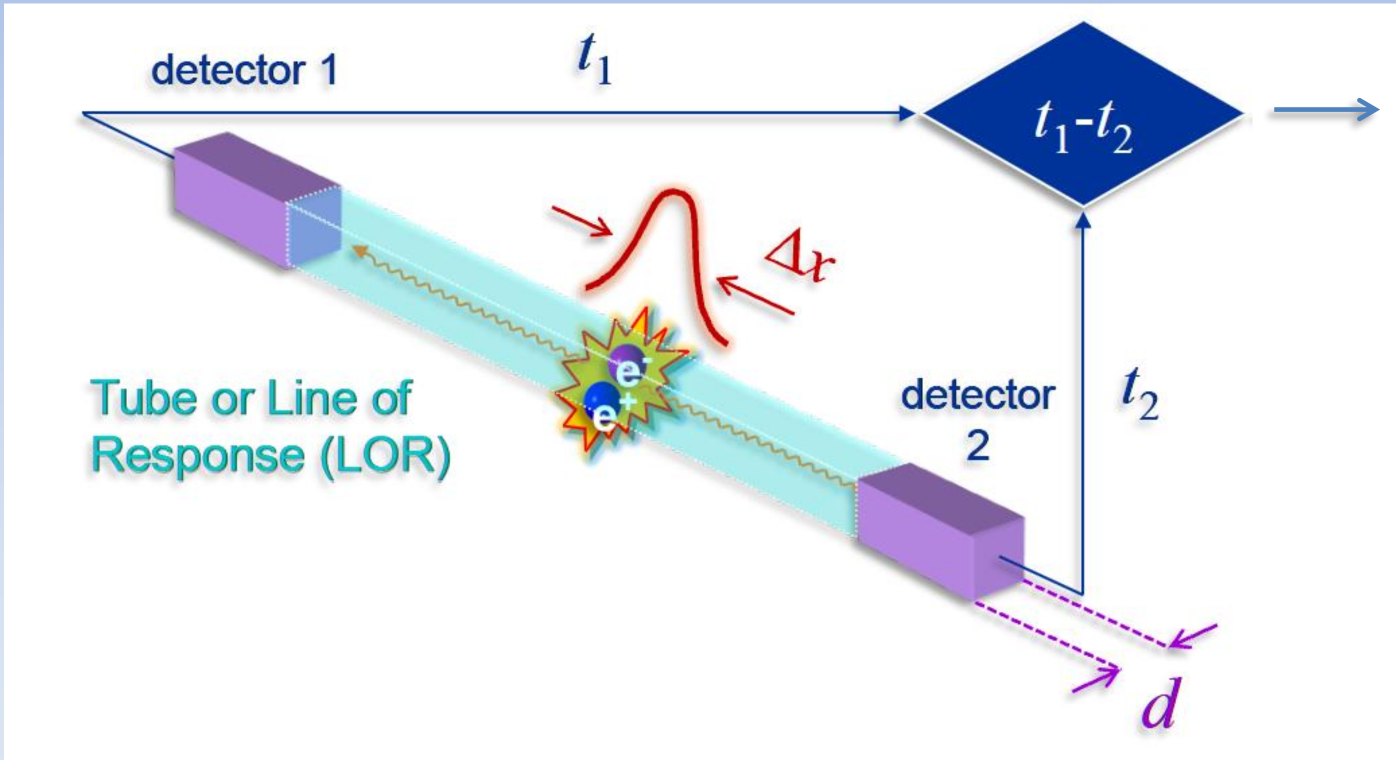


8x8 LSO crystals coupled to 4 side by side multi anodes photomultipliers

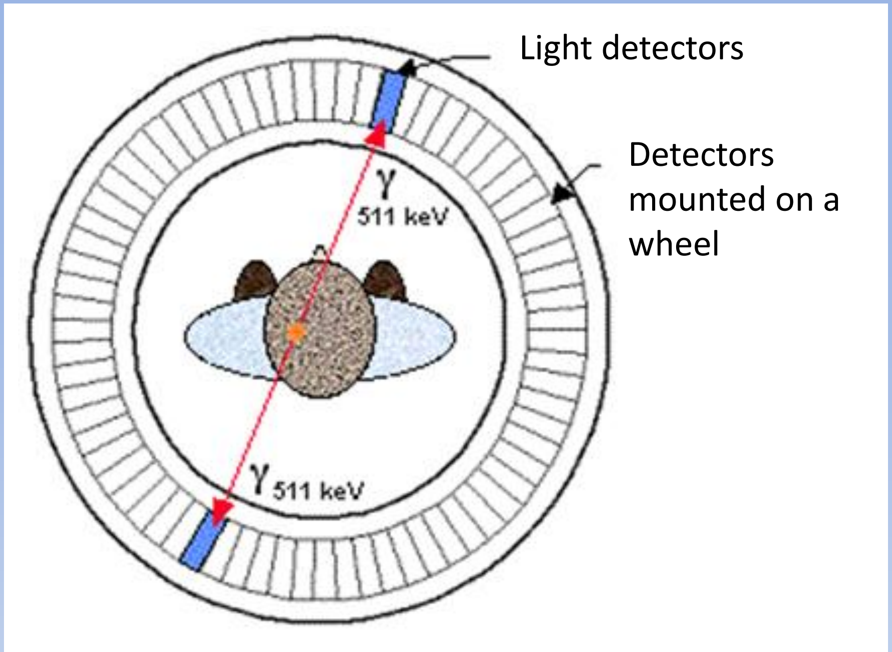
Energy resolution to reject scattered coincidences, timing to reject random coincidences

Could timing also help in image reconstruction ?

Introducing Time of Flight



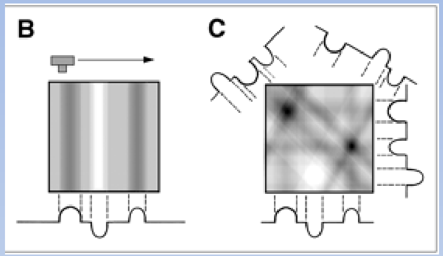
Position x along line of response

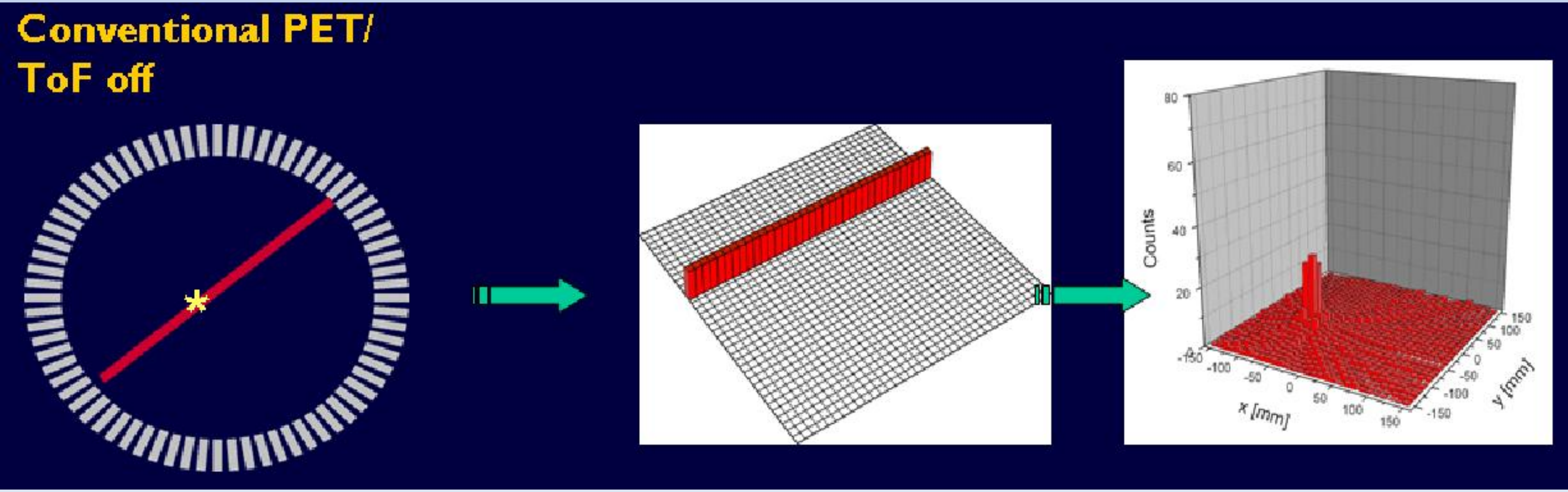


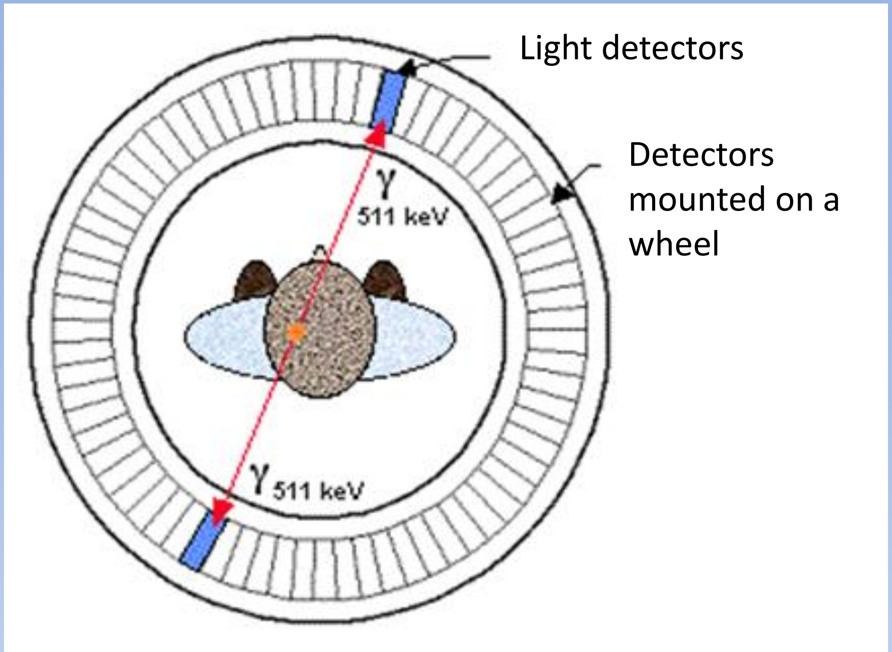
Light detectors are not (yet) fast enough :

the annihilation point cannot be reconstructed…

… only lines of response are

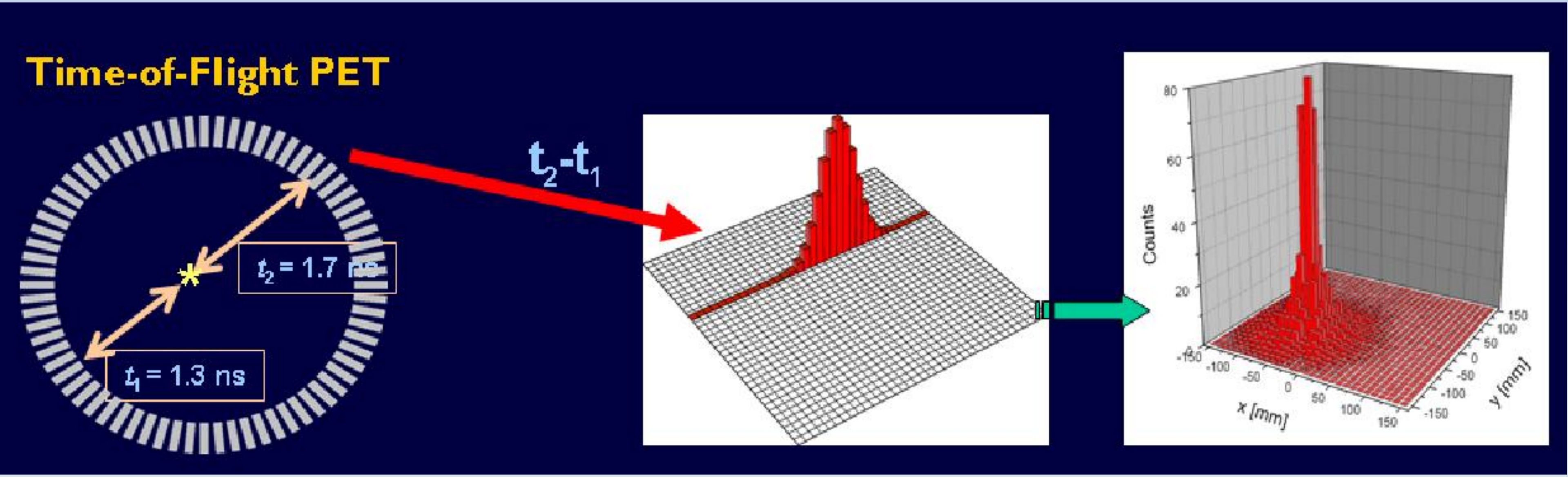






State of the art : resolution down to 500 ps

Δx = 7.5 cm



Visible light photodetectors

Requirements :

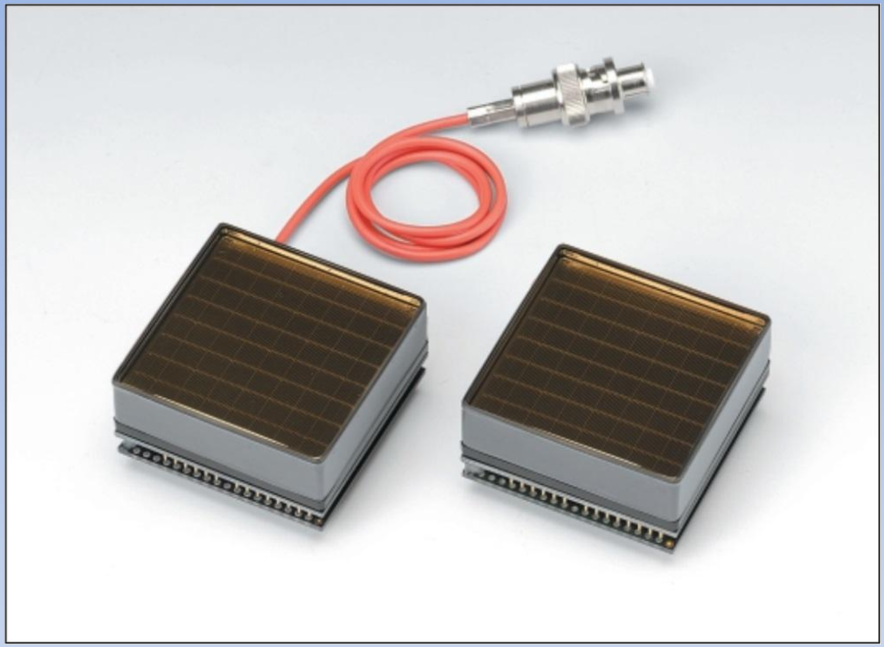
- Wavelength sensitivity in accordance with light from crystals
- High gain & quantum efficiency
- Small rise time
- Low noise
- behavior in magnetic field if used near magnetic devices (MRI)

	PM	APD	SiPM*
Gain	10^4 - 10^6	10^2 - 10^3	10^4 - 10^6
Additional preamplifier needed	no	yes	no
Multi channels	yes	yes	yes
Timing resolution	300 ps	1 ns	190 ps
Works in B	no	yes	yes
Gain	stable	Gain	dark current, Gain
Cost			$$$

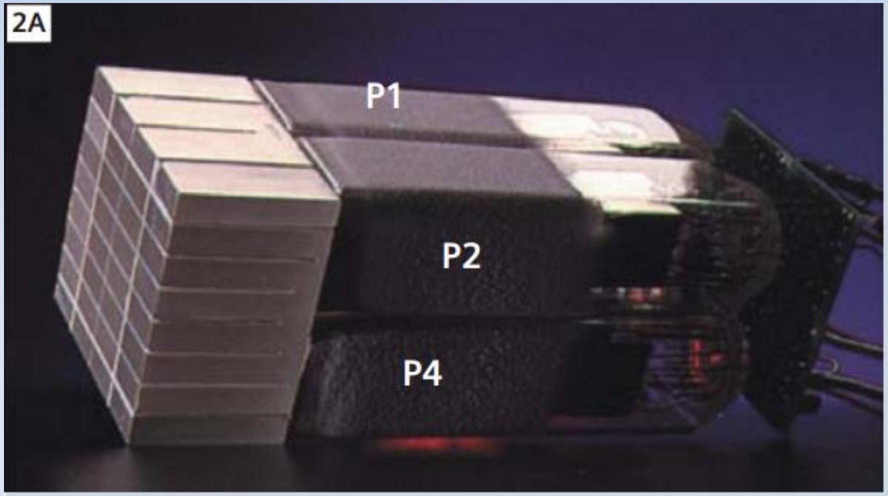
*Geiger Mode APD = Silicon PhotoMultiplier. Large gain, needs quenching after initiating the avalanche

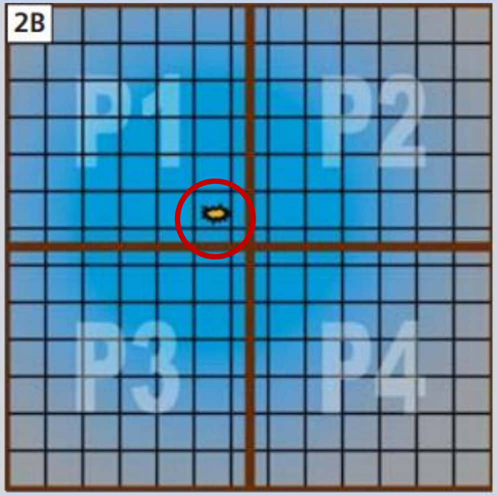
Thomas K. Lewellen The Challenge of Detector, Designs for PET Nuclear Medicine and Molecular Imaging, Review Article

Multi anodes photomultipliers

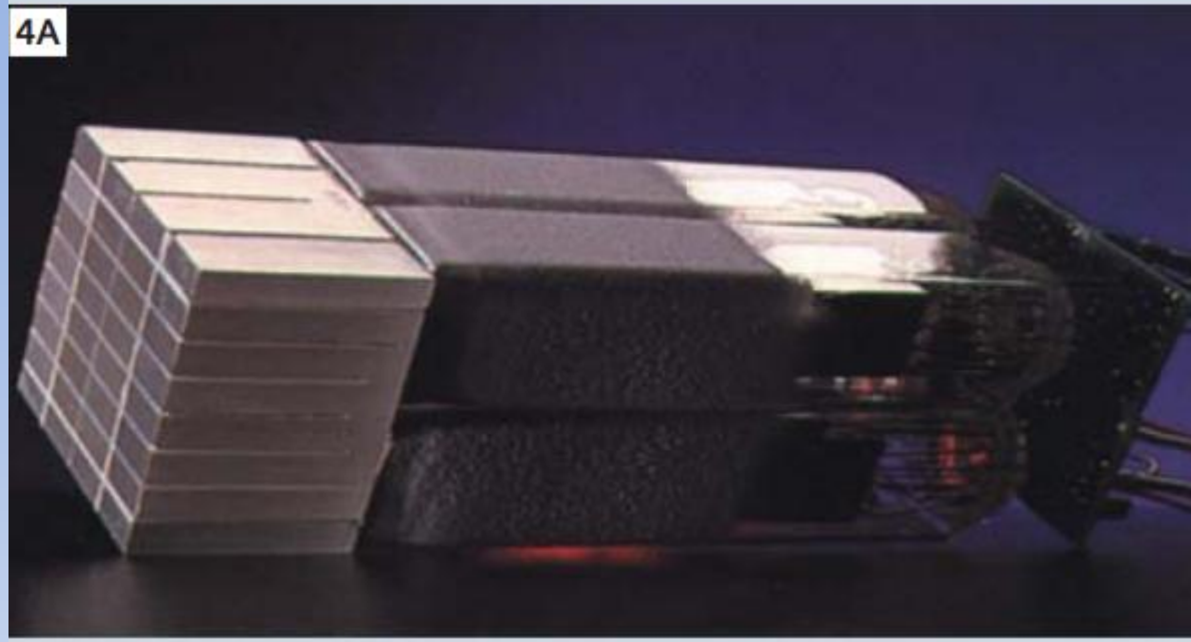


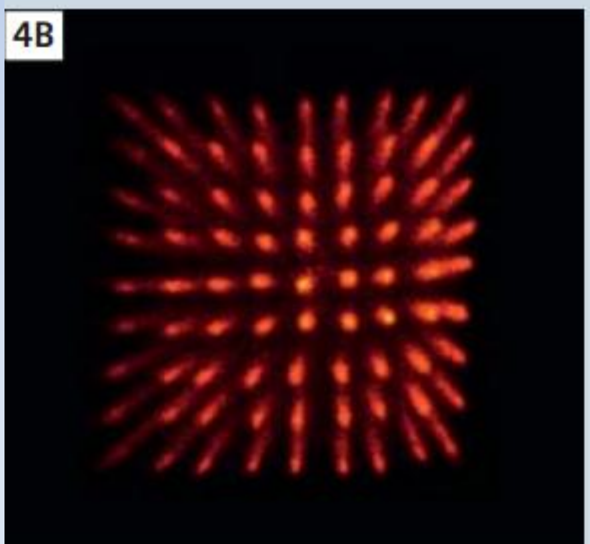
Use *Anger weighting algorithm* on measured signals : position of interaction of the incoming annihilation photon





Conventional PET detectors

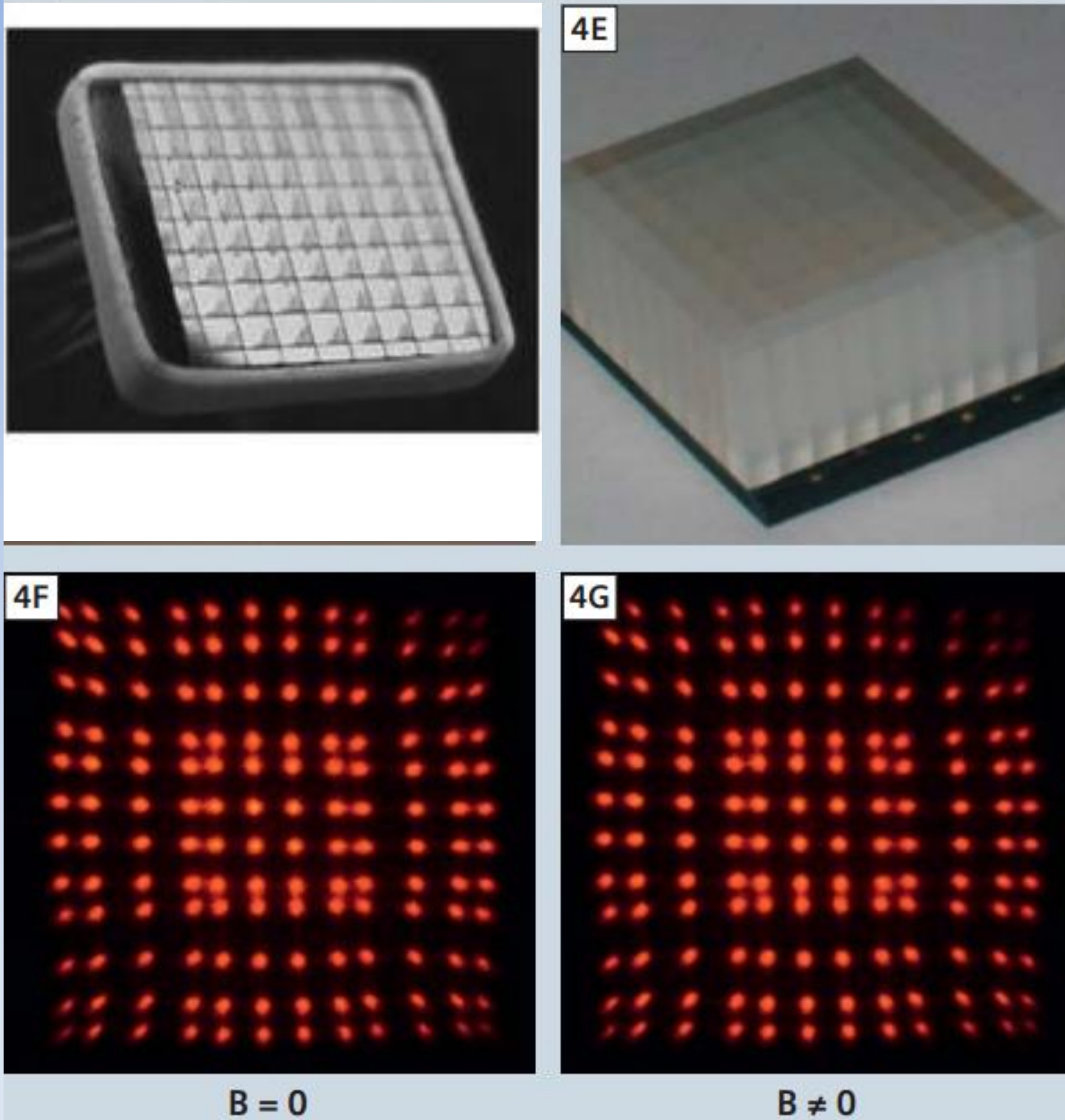




B = 0

B ≠ 0

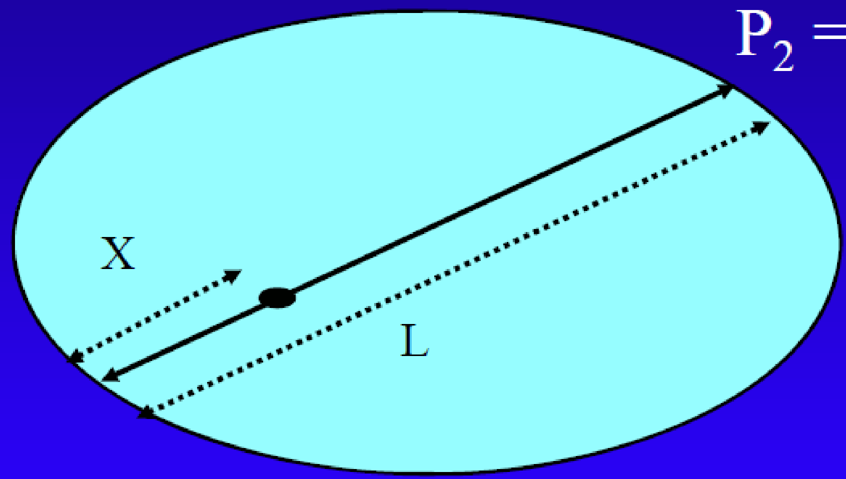
APD-based PET detectors



4 Conventional PET detectors **(A)** are based on scintillation crystals and Anger-logics readout. They work only outside magnetic fields **(B)**. If a PMT is operated inside a magnetic field (B ≠ 0) then the multiplier step is construed and the readout map severely distorted **(C)**. Avalanche photodiodes (APD) based detectors **(D, E)** are semiconductors that can be operated in magnetic fields, even at ultra-high field strengths **(F, G)**. *(Image courtesy of Prof. B. Pichler, University of Tübingen, Germany.)*

PET Attenuation Correction

$P_2 = e^{-\mu(L-x)}$



$P_1 = e^{-\mu x}$

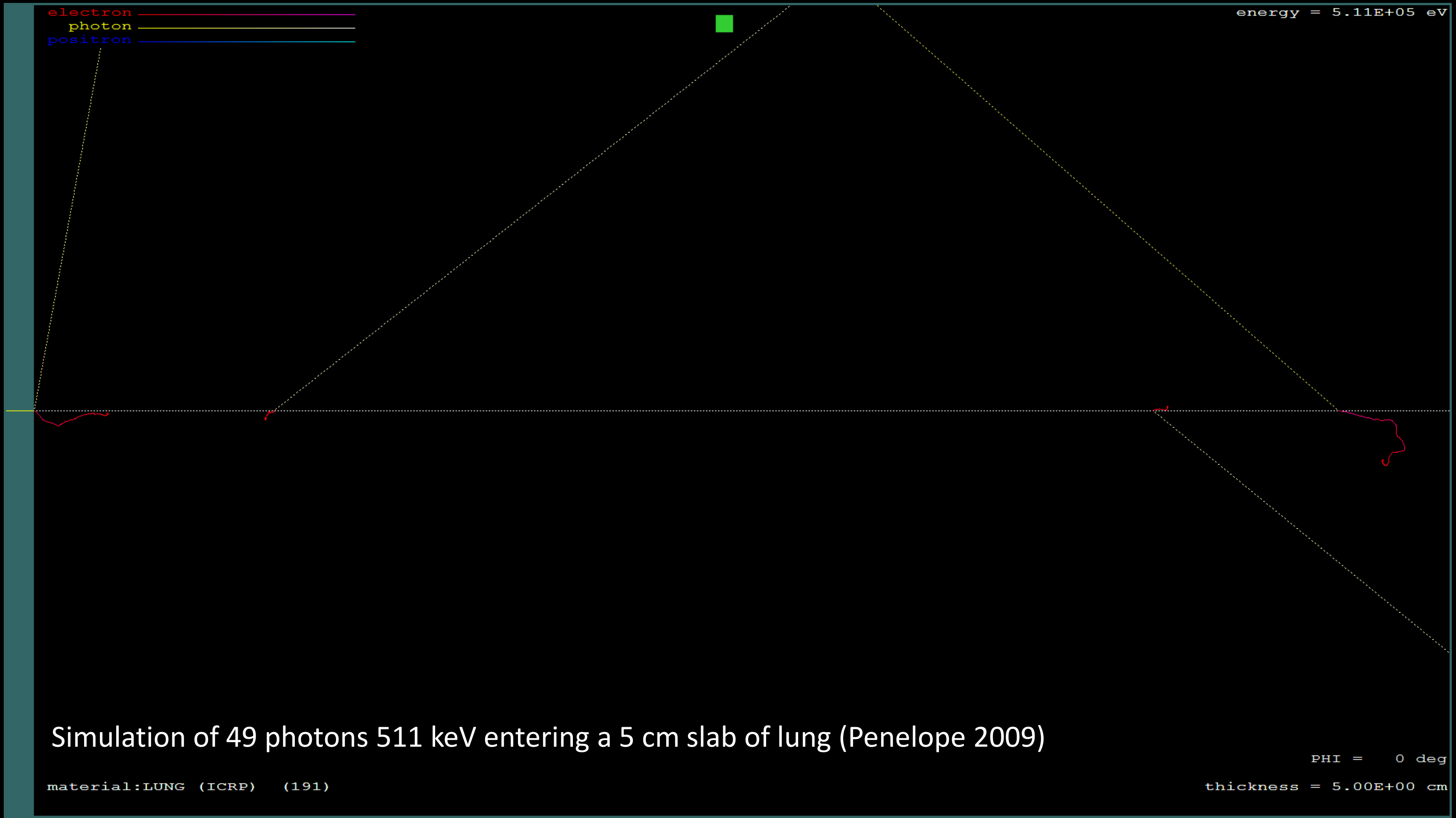
$$P_{TOT} = P_1 \times P_2 = e^{-\mu L}$$

Reconstruction algorithm needs

integrated attenuation for each LOR

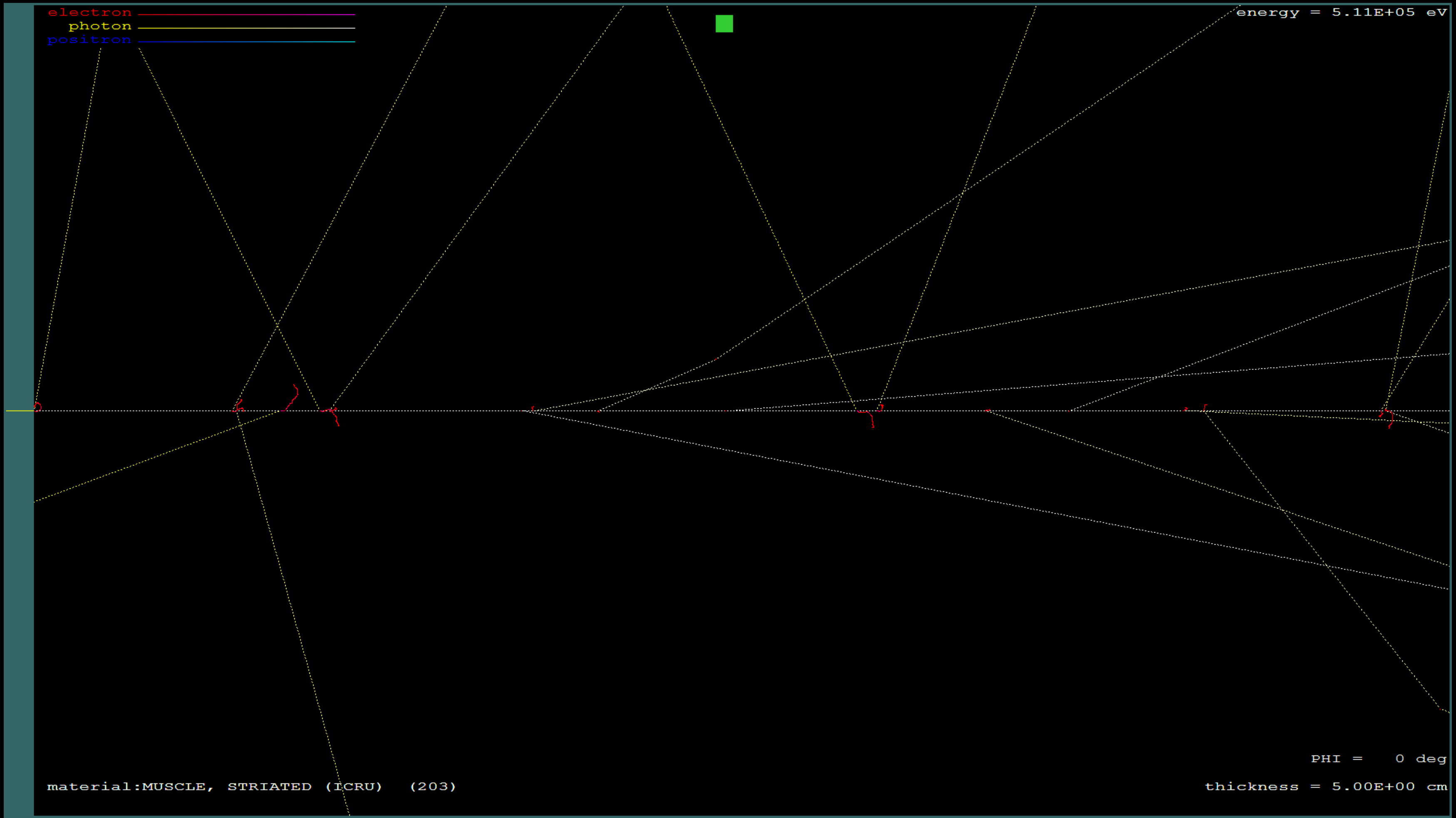
Photons do interact with matter : mostly with atomic electrons

Atomic electronic concentration depends on tissue (lung)



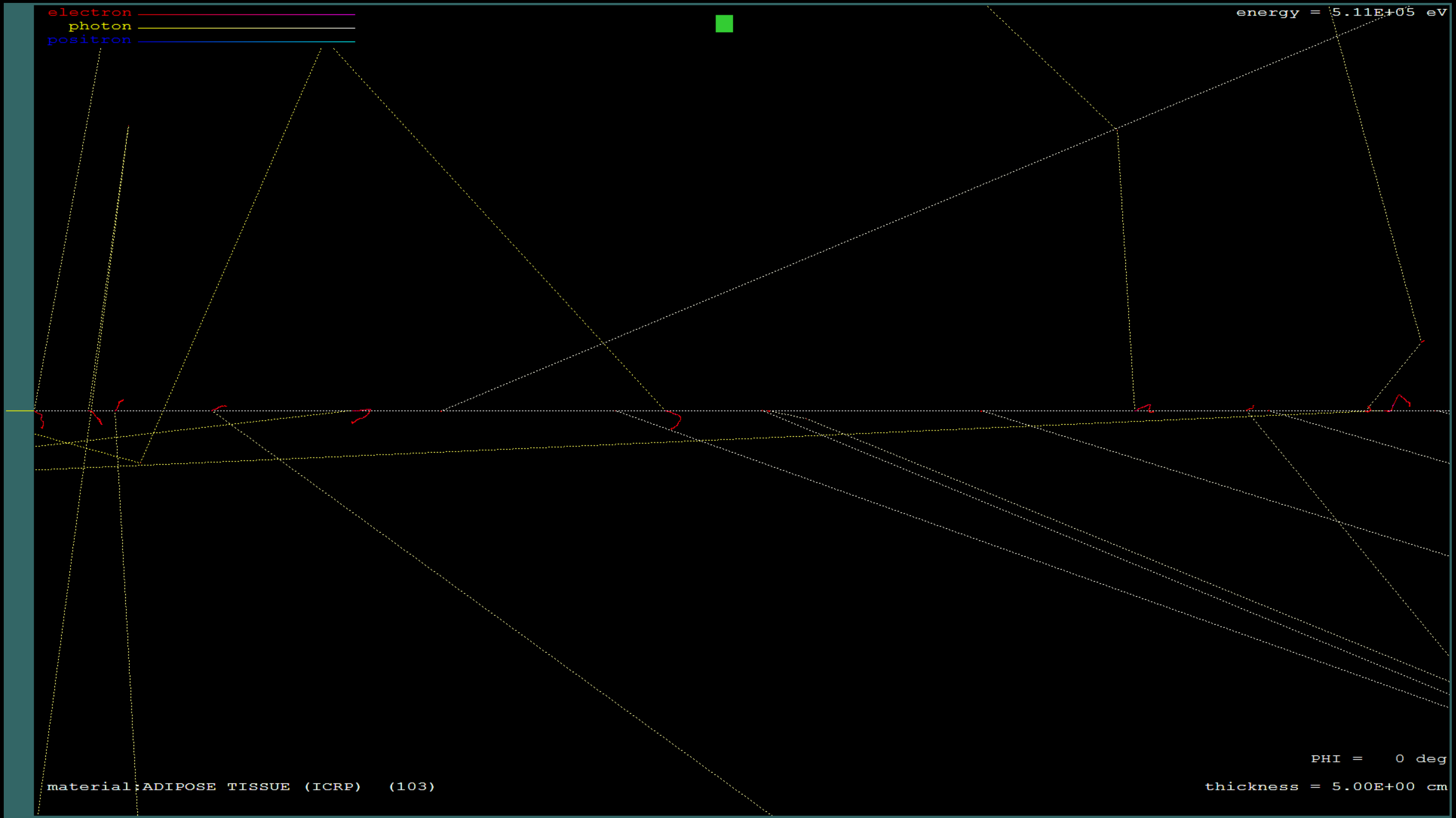
Photons do interact with matter : mostly with atomic electrons

Atomic electronic concentration depends on tissue (muscle)



Photons do interact with matter : mostly with atomic electrons

Atomic electronic concentration depends on tissue (fat)

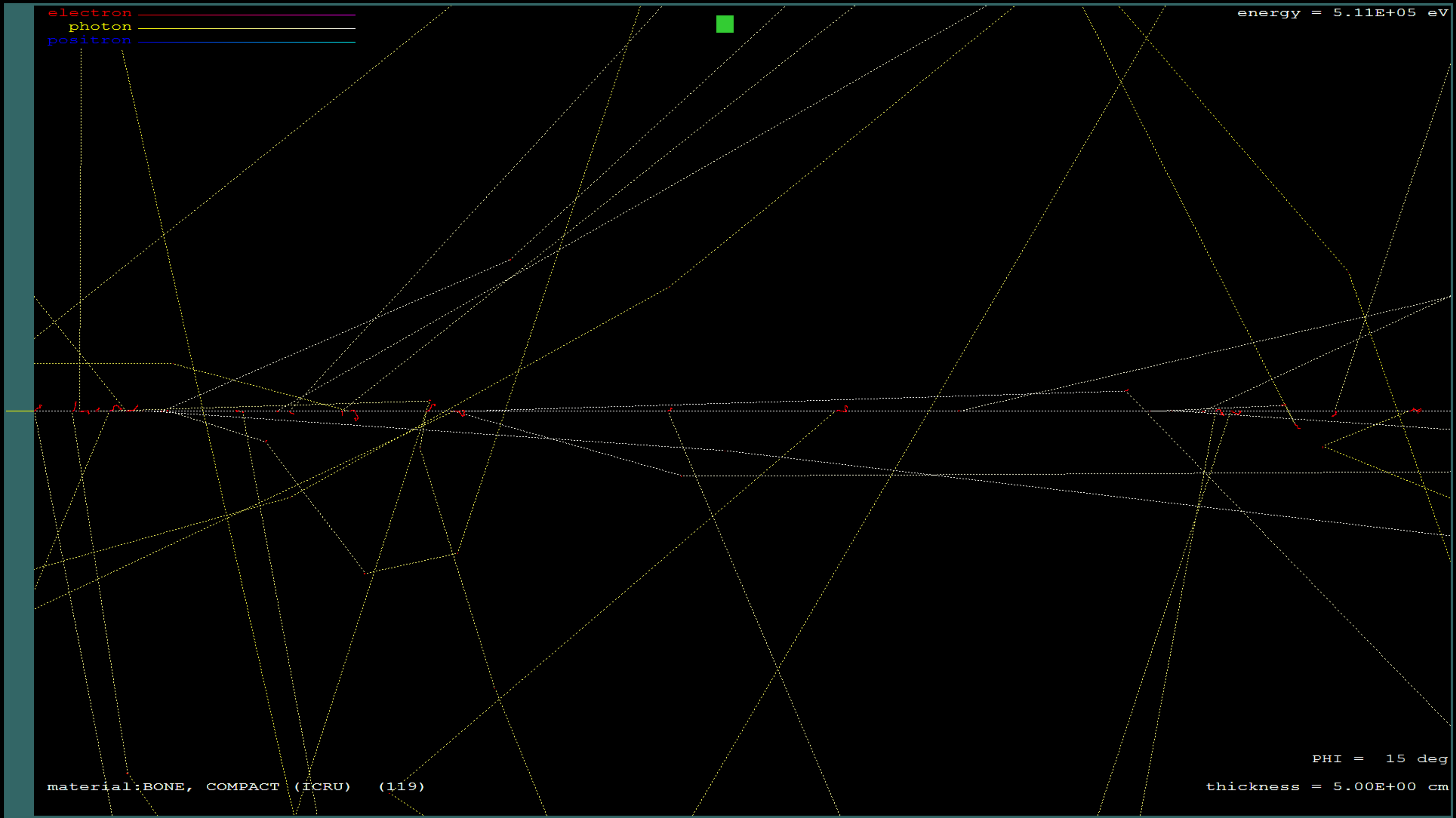


Photons do interact with matter : mostly with atomic electrons

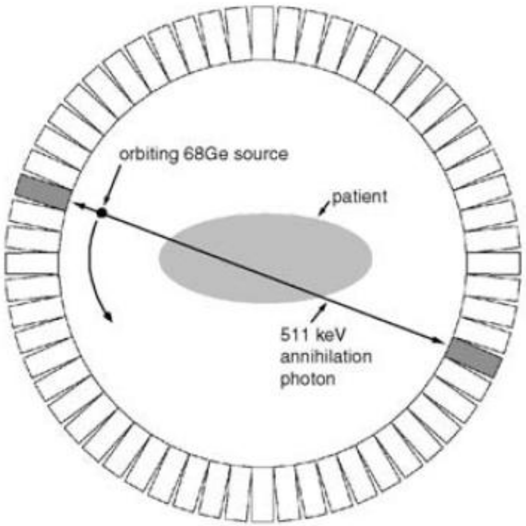
Atomic electronic concentration depends on tissue (bone / fat / muscle / lung, …)

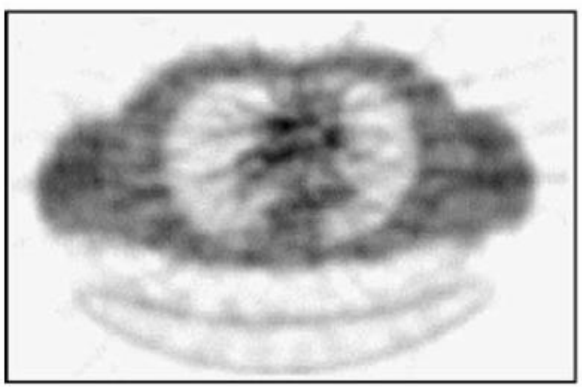
Atomic electronic concentration has to be measured all along the possible path of annihilation photons

= tissue content in terms of electronic concentration has to be determined



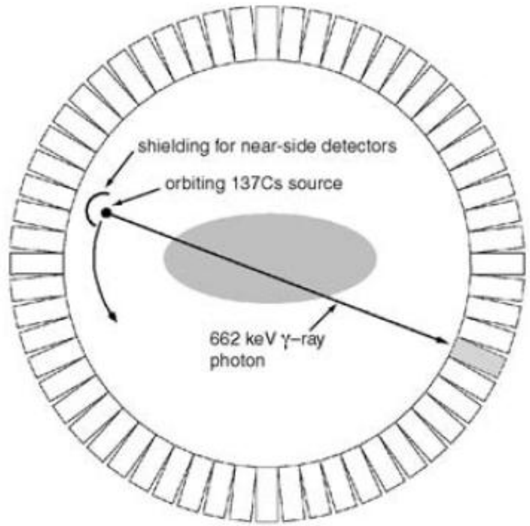
Coincident photon
Ge-68/Ga-68 (511 keV)

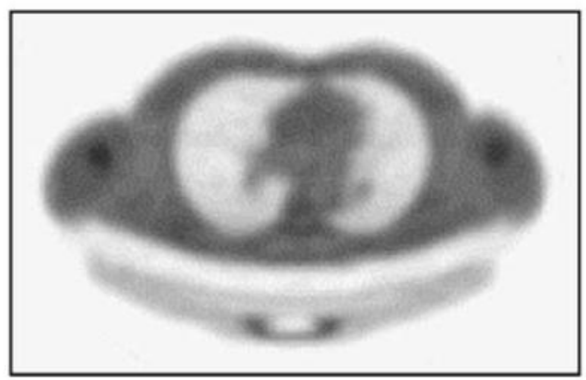




high noise
~12-45 min scan time
no bias
lowest contrast

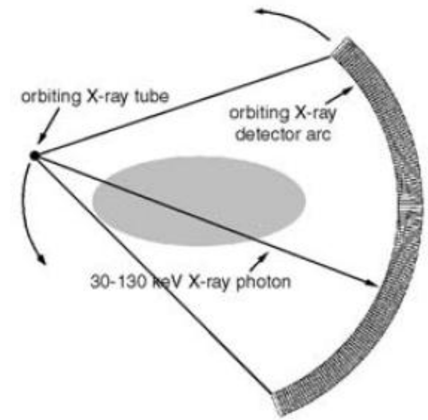
Single photon
Cs-137 (662 keV)

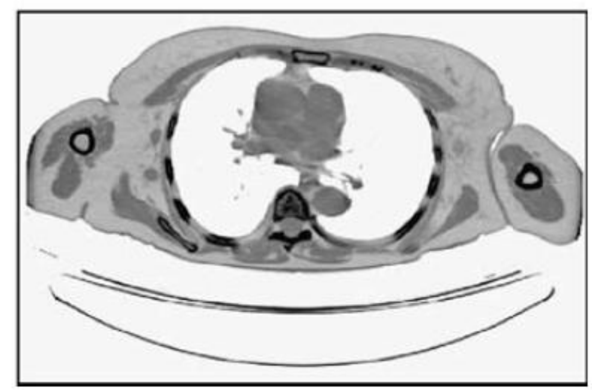




lower noise
~4-20 min scan time
some bias
lower contrast

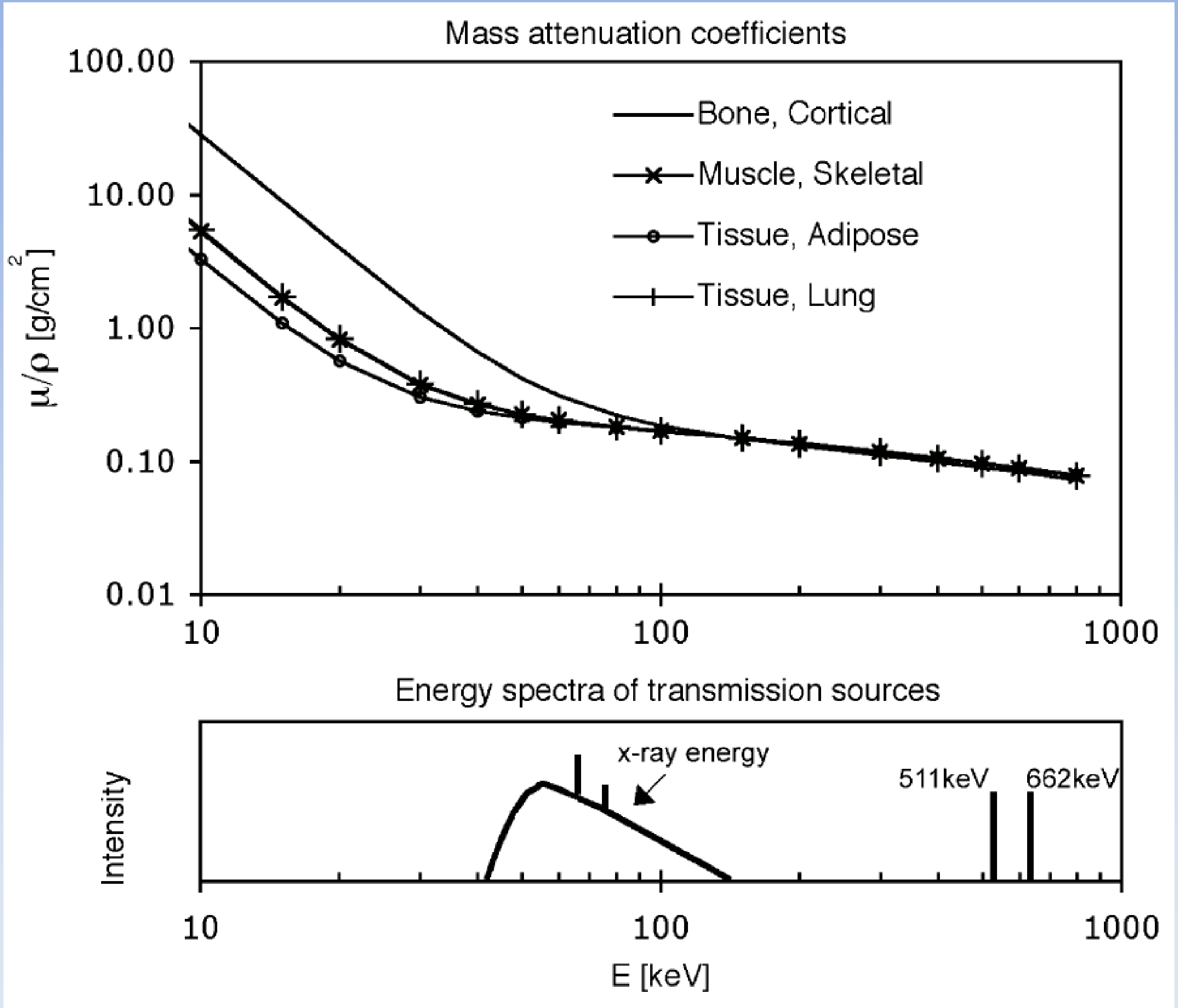
X-ray
(~30-140kVp)





no noise
~20 sec scan time
potential for significant bias
high contrast

Photons interactions probabilities depend both on the target and on the photon energy



Haunsfield Unit scale

The Hounsfield Unit (HU) scale is a linear transformation of the linear attenuation coefficient measurement into one in which the radiodensity of distilled water (at standard pressure and temperature) is defined as zero HU, while the radiodensity of air at STP is defined as -1000 HU. For a material X with linear attenuation coefficient μ_X, the HU value is therefore given by:

$$HU = \frac{\mu_X - \mu_{water}}{\mu_{water} - \mu_{air}} \times 1000$$

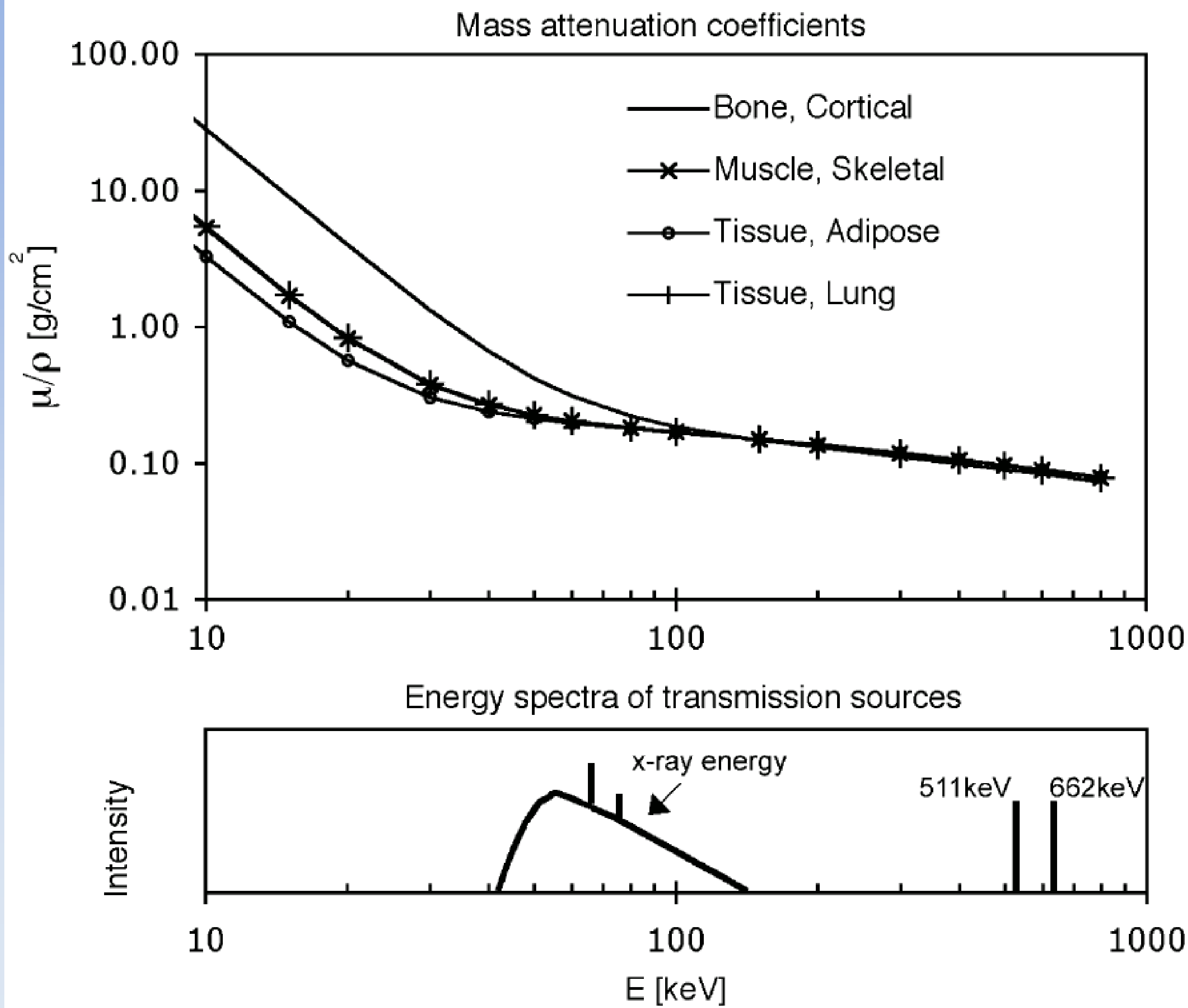
Computed Medical Imaging
Nobel Lecture, 8 December 1979
Godfrey N. Hounsfield
J Radiol. 1980 Jun-Jul;61(6-7):459-68

where μ_{water} and μ_{air} are the linear attenuation coefficients of water and air, respectively.

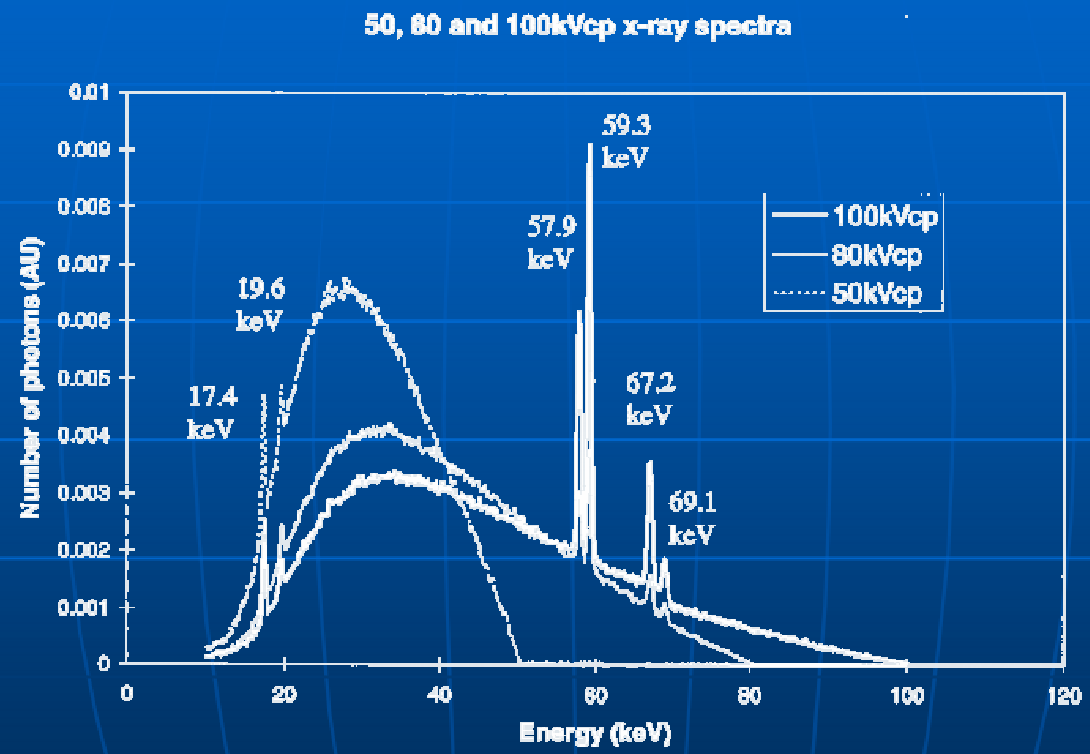
HU scale (indicative values)

material	HU value	notes
air	- 1000	and vacuum as well, for all practical effects
fat	- 100	
water	0	distilled, at STP
muscle	+ 40	
Blood	+40	
bone	~ > 400	spans over a large range, i.e. up to >1200
aluminum	2640	(at 60 KeV)

extrapolate μ's measured at the CT energy to μ's @ 511 keV



… unfortunately, real X-ray beams are NOT AT ALL monochromatic !



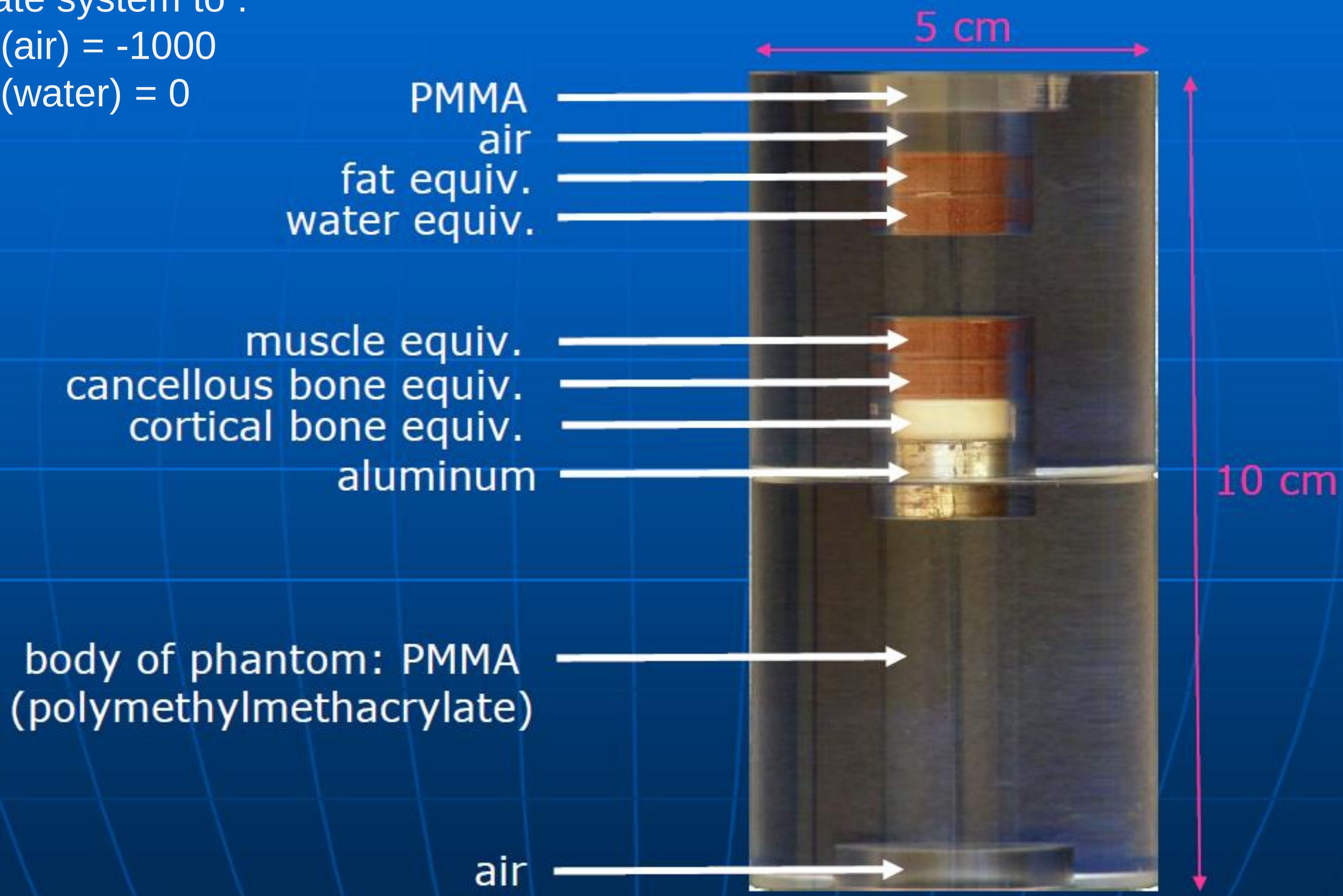
Diagnostic x-ray spectra: A comparison of spectra generated by different computational methods with a measured spectrum
M. Bhat, J. Pattison, G. Bibbo, and M. Caon Med. Phys. (1998) 25(1):114-120

Specific CT energy spectrum

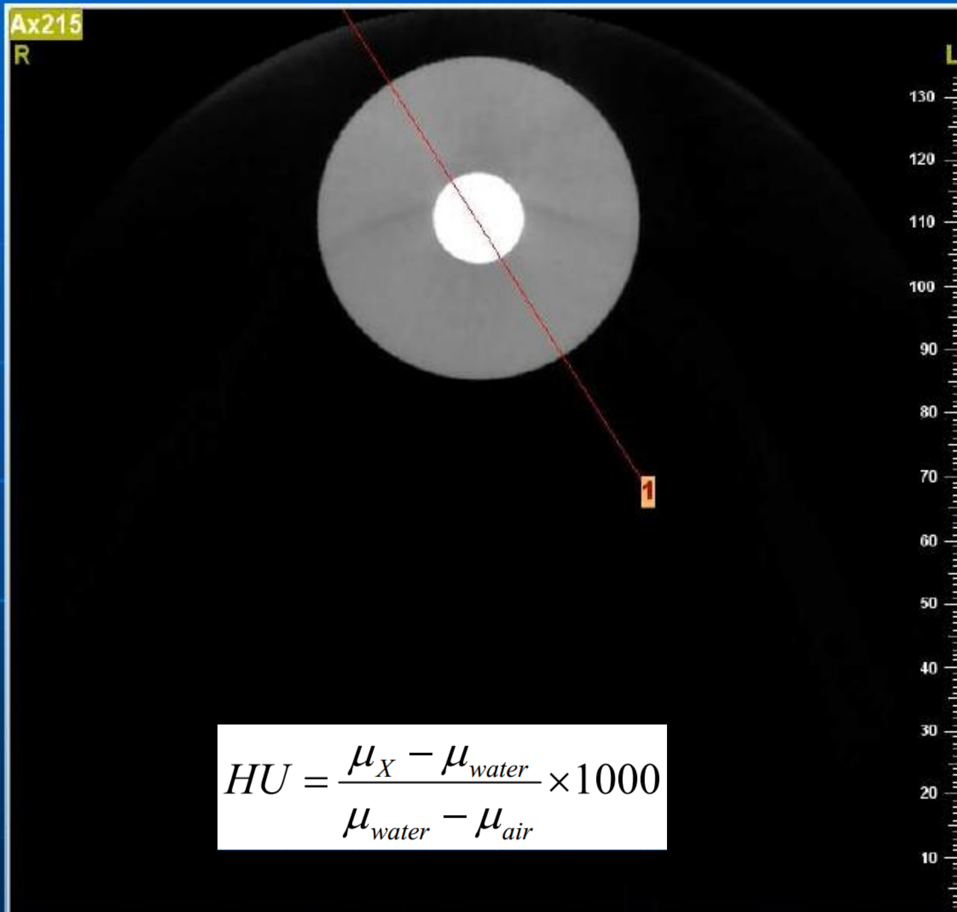
Calibrate system to :

- HU(air) = -1000
- HU(water) = 0

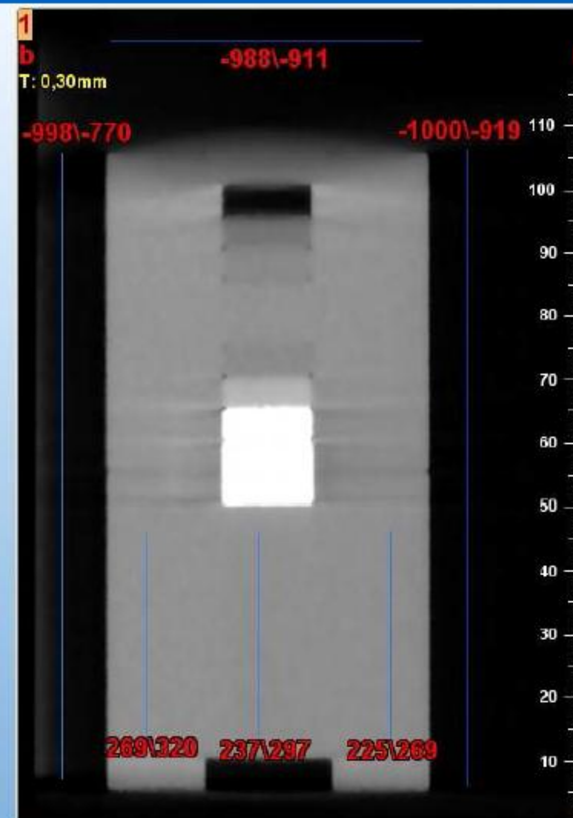
test phantom

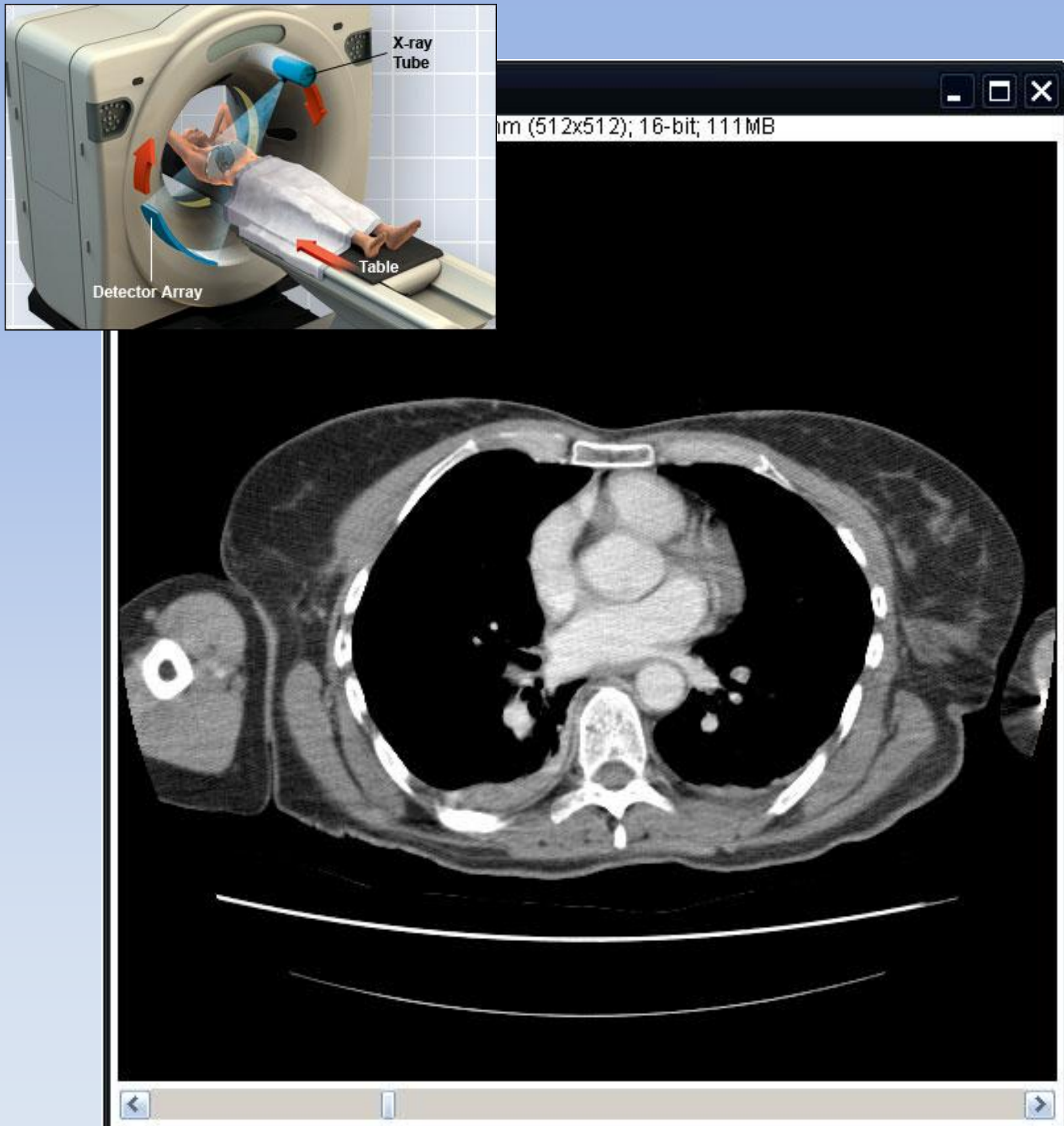


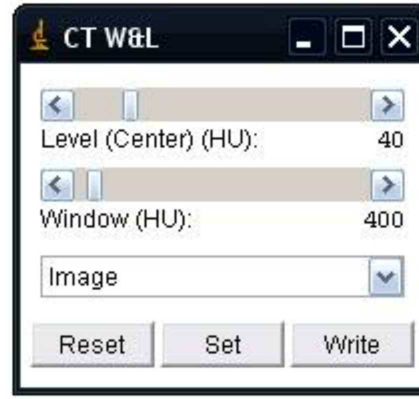
measured CT Numbers

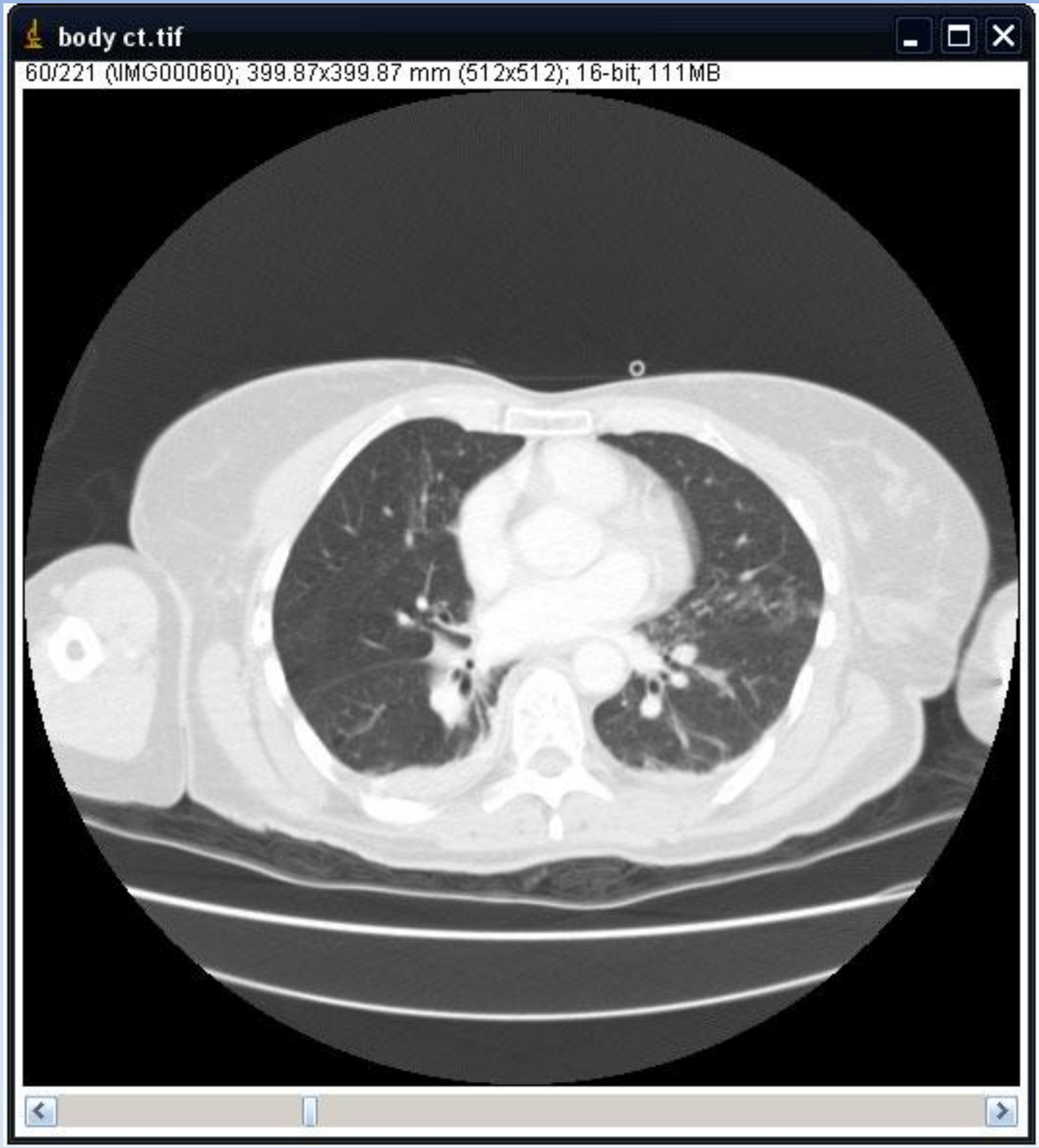


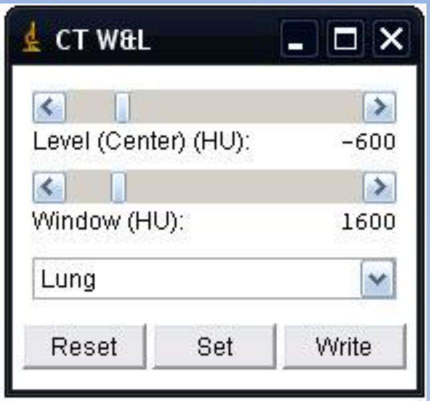
$$HU = \frac{\mu_X - \mu_{water}}{\mu_{water} - \mu_{air}} \times 1000$$




X-ray
Tube
Table
Detector Array
m (512x512); 16-bit; 111MB

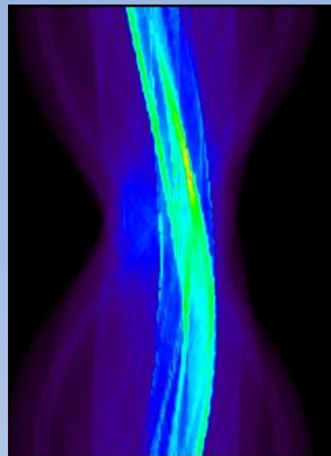

CT W&L
Level (Center) (HU): 40
Window (HU): 400
Image
Reset
Set
Write


body ct.tif
60/221 (IMG00060); 399.87x399.87 mm (512x512); 16-bit; 111MB

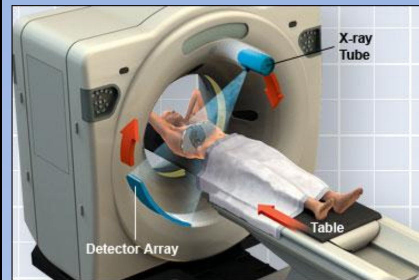

CT W&L
Level (Center) (HU): -600
Window (HU): 1600
Lung
Reset
Set
Write

PET calibration

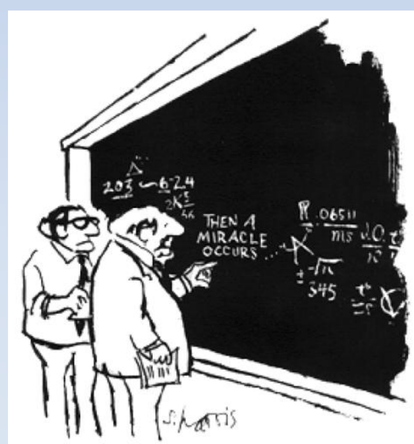
PET acquisition



CT acquisition

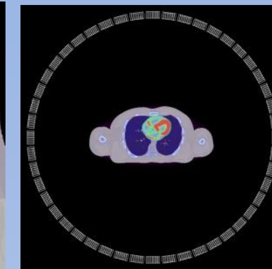


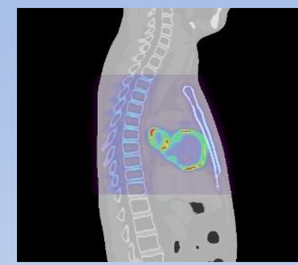
- HU to CT μ’s
- CT μ’s to μ’s @ 511 keV



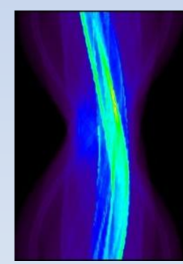
"I think you should be more explicit here in step two."

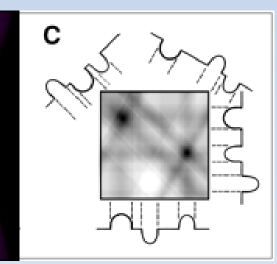
EMrecon: An Expectation Maximization Based Image Reconstruction Framework for Emission Tomography Data





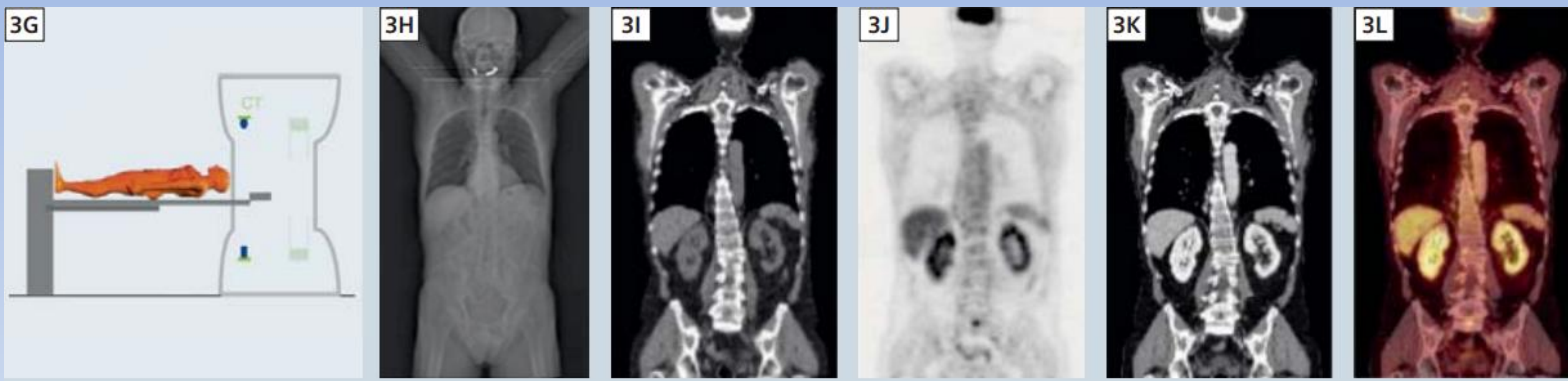
Dose quantification





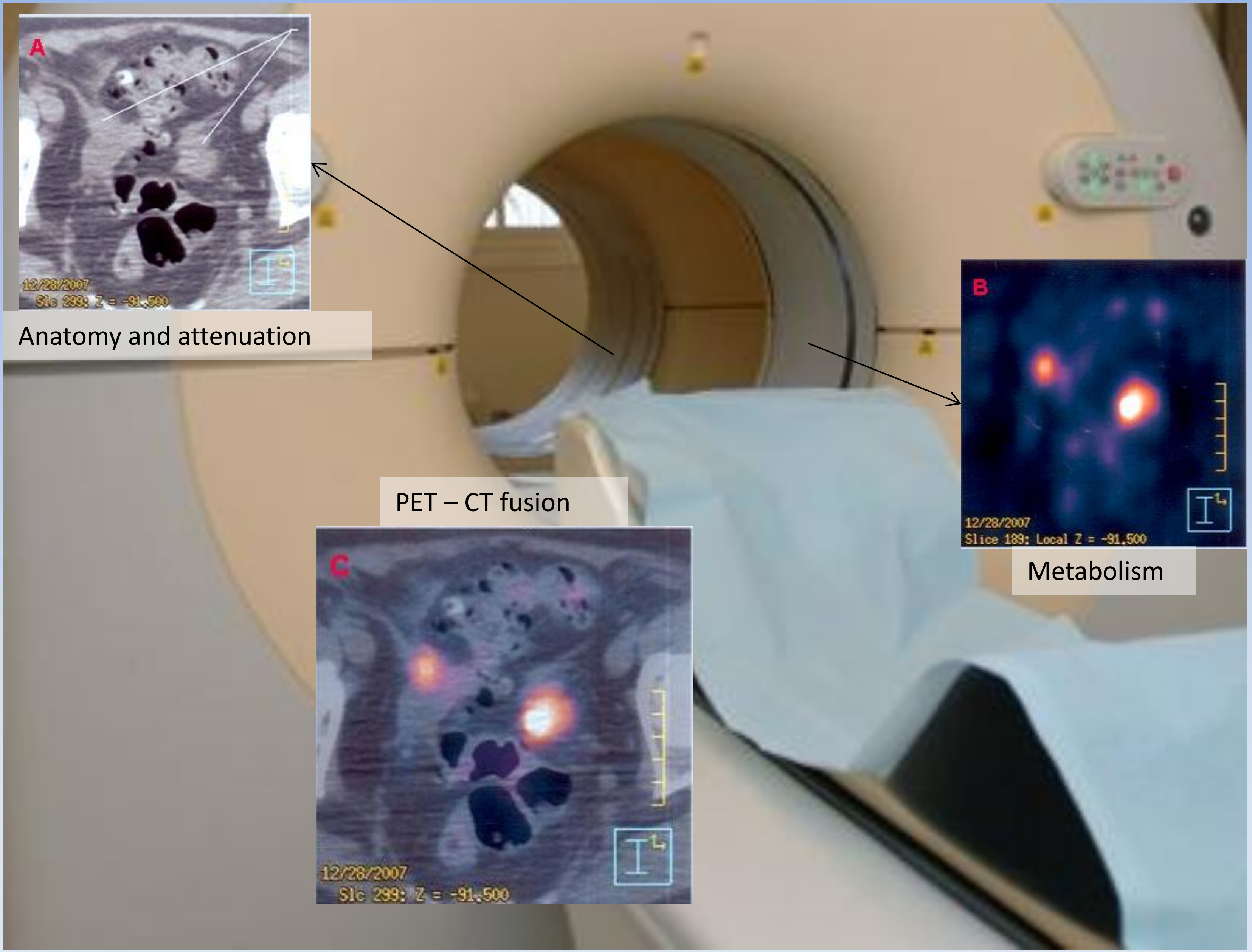
Back propagation algorithm

including attenuation map



3G–L Schematic illustration of a standard PET/CT investigation: tracer injection, uptake time and patient positioning **(G)**, topogram and scan range definition **(H)**, CT acquisition for attenuation correction **(I)**, emission scan **(J)**, contrast-enhanced CT scan **(K)** and image reconstruction/fusion **(L)**.

PET functional & CT anatomy imaging



PET cameras, a status report

- Positron Emission Tomography Products
- Acquisition
- Quantification

Based on material from :

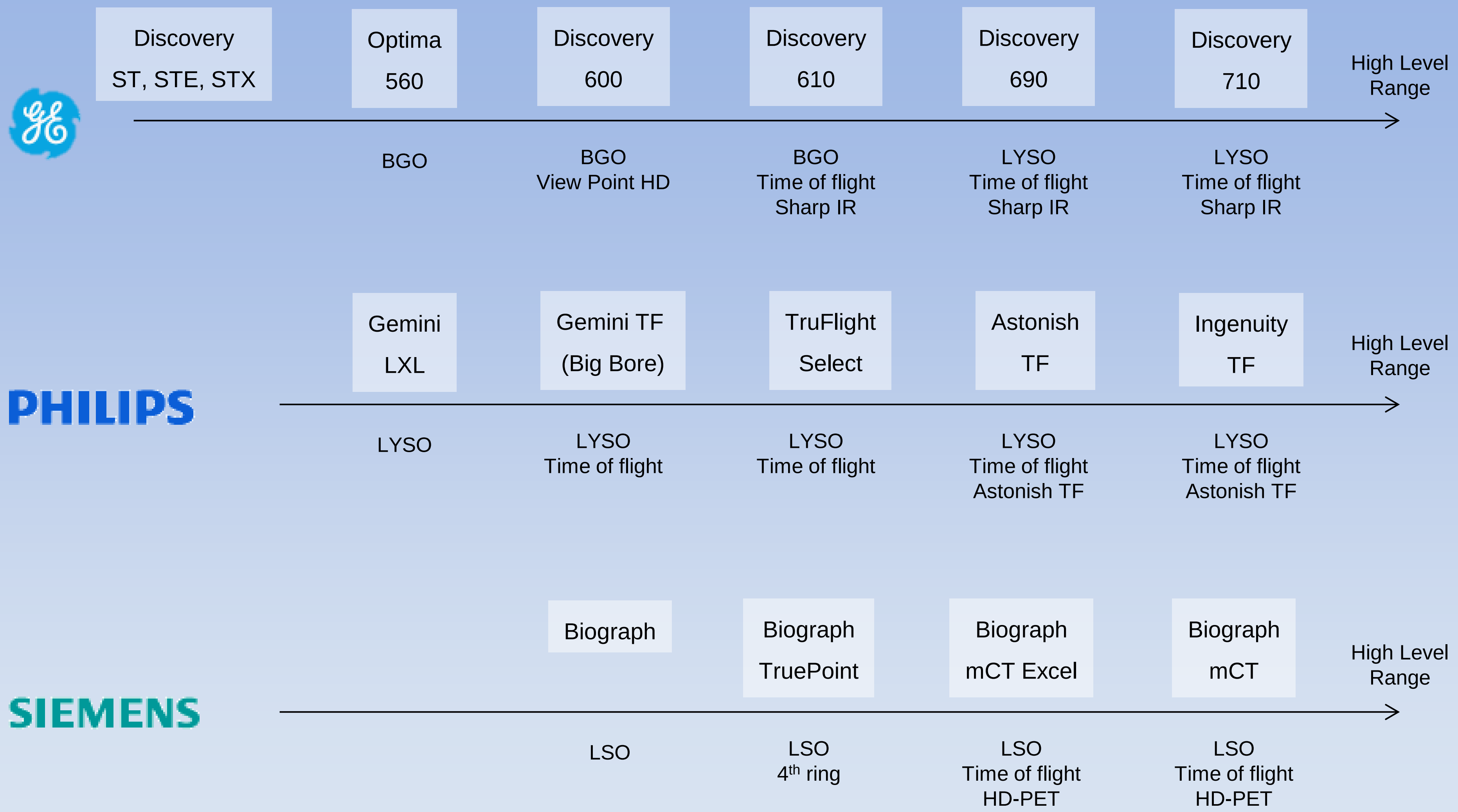
- Etat de l'art sur les caméras TEP, Pierre Malik Koulibaly, Medical Physicist, Centre Antoine Lacassagne, Nice, France
- Development of Hybrid Imaging, Uwe Pietrzyk, research center Jülich and University of Wuppertal, Germany

PET/CT Products

Three main suppliers :

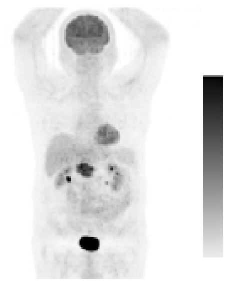
And also :

3A–F Current commercial PET/CT systems: **(A)** Discovery series from General Electrics Healthcare, **(B)** Gemini series from Philips Healthcare, **(C)** Biograph mCT from Siemens Healthcare, **(D)** Acquiduo series from Toshiba Medical Systems, **(E)** Sceptre series from Hitachi Medical Systems, **(F)** Anyscan from Mediso (this device can be combined with a SPECT to form a triple-modality imaging system). Variations apply to the individual performance parameters of PET and CT.


Discovery ST, STE, STX
Optima 560
BGO
Discovery 600
BGO
View Point HD
Discovery 610
BGO
Time of flight
Sharp IR
Discovery 690
LYSO
Time of flight
Sharp IR
Discovery 710
LYSO
Time of flight
Sharp IR
High Level Range
PHILIPS
Gemini LXL
LYSO
Gemini TF (Big Bore)
LYSO
Time of flight
TruFlight Select
LYSO
Time of flight
Astonish TF
LYSO
Time of flight
Astonish TF
Ingenuity TF
LYSO
Time of flight
Astonish TF
High Level Range
SIEMENS
Biograph
LSO
Biograph TruePoint
LSO
4th ring
Biograph mCT Excel
LSO
Time of flight
HD-PET
Biograph mCT
LSO
Time of flight
HD-PET
High Level Range

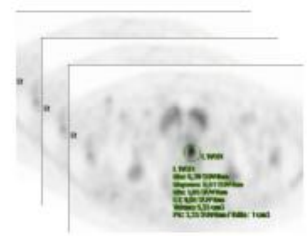
Available acquisition modes

Static



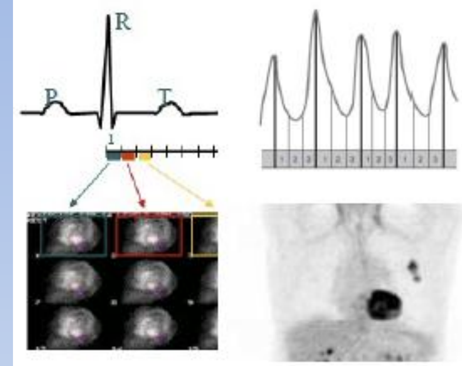
18F-FDG

Dynamic

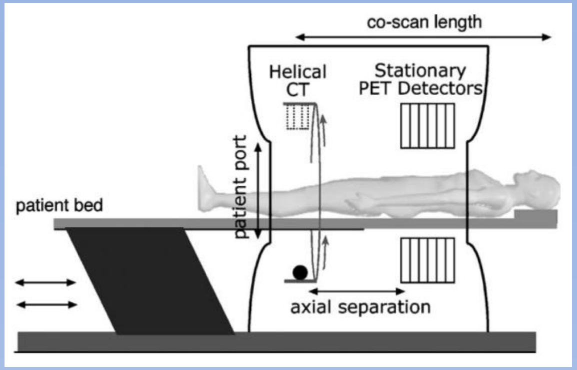


18F-Choline

Gated



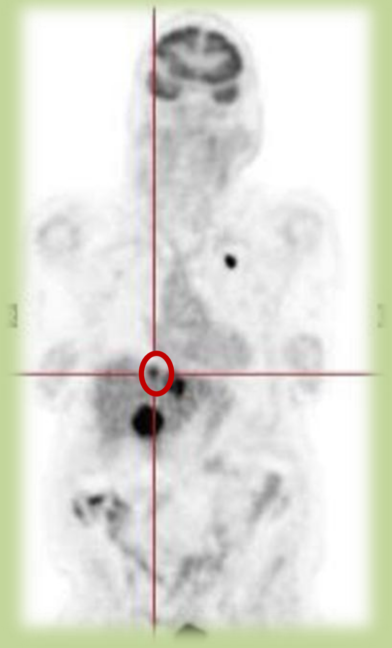
	Static	Dynamic	Gated
GE	☑	☑	☑
PHILIPS	☑	☑	☑
SIEMENS	☑	☑	☑



TEP module	Discovery 710	Ingenuity TF	Biograph mCT
	GE	PHILIPS	SIEMENS
Crystal type	LYSO	LYSO	LSO
Crystal size	4.7 x 6.3 x 30 mm^3	4 x 4 x 22 mm^3	4 x 4 x 20 mm^3
Number of crystals	13,824	28,336	32,448
Patient scan range	188 cm	194 cm	198 cm
Patient port	70 cm	70 cm	78 cm
Axial field of view	15.7 cm	18 cm	21.6 cm
time measurement resolution	550 ps	495 ps	555 ps
Coincidence window to accept events	4.9 ns	3.8 ns	4.1 ns
Energy window to accept events	425-650 keV	440-665 keV	435-650 keV
System sensitivity	7 cps per kBq	7.4 cps per kBq	9.5 cps per kBq
Transverse spatial resolution @ 1 cm	4.9 mm	4.7 mm	4.4 mm
Transverse spatial resolution @ 10 cm	6.3 mm	5.2 mm	4.9 mm
Axial spatial resolution @ 1 cm	5.6 mm	4.7 mm	4.5 mm
Axial spatial resolution @ 10 cm	6.3 mm	5.2 mm	5.9 mm

Quantification : extracting a physical value from an image

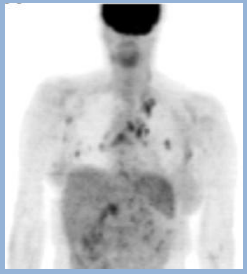
In PET, it could be the uptake value of FDG in a given volume.



Measurement of the concentration of a radioelement within a ROI (MBq/kg)

Prognostic : provides information on tumor grade (survival probability)

Therapeutic actions : visible slow down in the glucose uptake is a good indicator of an efficient treatment.



PET images have to be corrected from all distortions

Quantification in PET : Standard Uptake Value (SUV)

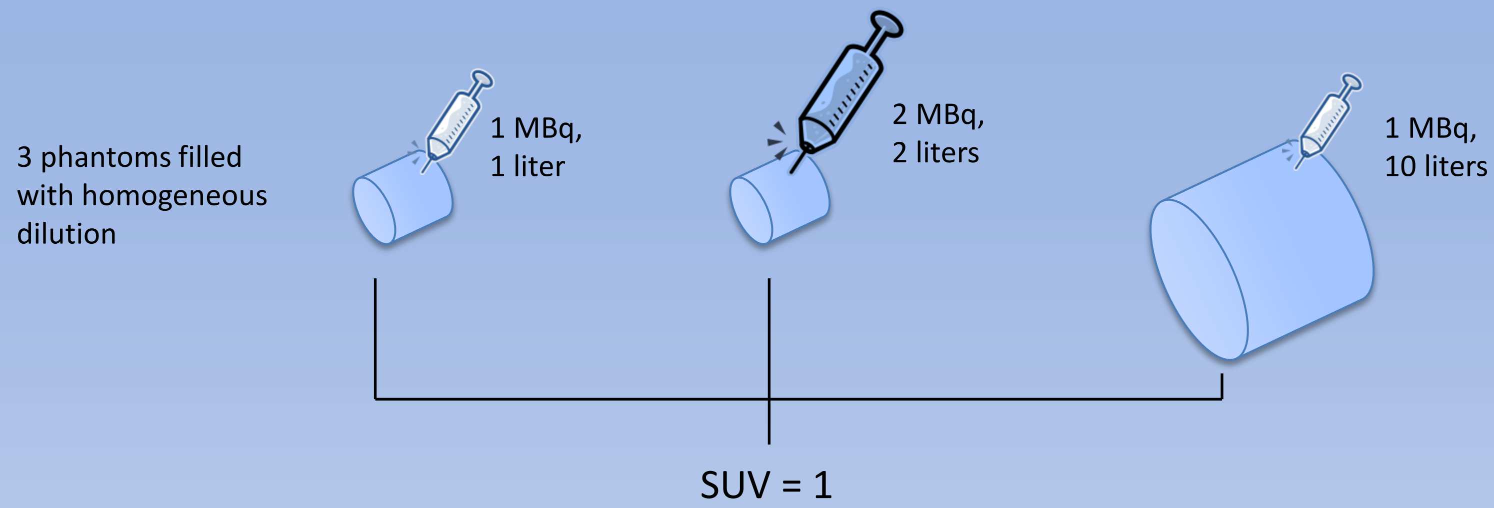
SUV is calculated over a region of interest

It is defined as follows :

$$\text{SUV} = \frac{\text{concentration of radiotracer (kBq/ml)}}{\text{Injected dose (kBq)/patient weight (g)}}$$

SUV is a precious common tool (quoted in all studies), even if not very accurate

- Injected dose is estimated with an accuracy of 5 -10 %
- The patient weight is known to a few percent
- The radiotracer fixation is estimated after many corrections



SUV = 1 indicates an homogenous spread of the radiotracer within the body. Any deviation from this value is a sign of hyper fixation.

Why is quantification so difficult to estimate ?

Patient side

- Spatial displacements (patient is moving, breathing, heart beating)
- Compton scattering
- Photon attenuation
- Random coincidences

PET side

- Detectors response non uniform
- Spatial resolution not constant
- Dead time
- Partial volume effect

Isotope decays

- Decreasing activity

Reconstruction algorithm

- Bias and nonlinearities

Corrections to reach correct quantification

Patient side

• Breathing :	triggered acquisition (gating)
• Heart beat :	gating
• Attenuation :	HU from CT-map to µ
• Compton scattering :	estimated then subtracted
• Random coincidences :	estimated then subtracted

PET side

• Detector uniformity :	normalization
• Arc geometry :	transverse and longitudinal normalization
• Dead time :	estimate real counting rate
• Spatial resolution not constant :	measure the Point Spread Function
• Spatial resolution not constant :	measure the Depth Of Interaction
• Partial volume effect :	correction function

Isotope decays

• Decreasing activity :	compensate for exponential decrease

Reconstruction algorithm

• Bias and nonlinearities :	algorithm correction

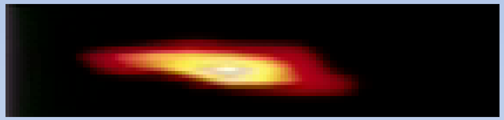
Corrections to reach correct quantification : breathing

Breathing induces 15-20 times a minute shifts with an amplitude up to 5 cm.

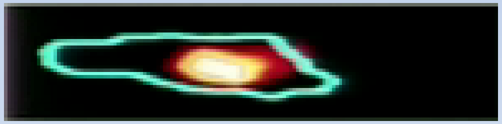
Shifts due to breathing affects many organs, in particular lung and liver

As a consequence, the volume and the intensity of the fixating target will be distorted.

Lung PET image



Tumor displacement



Fixed tumor

Apparent tumor volume increased by 10 to 30 %

Intensity of absorbed activity drops by 5% to 100%

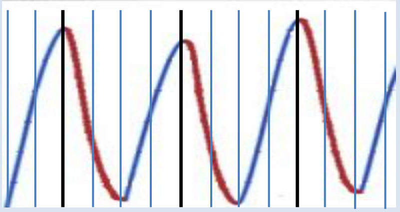
Solution :

Gating technics are adapted to respiratory motion

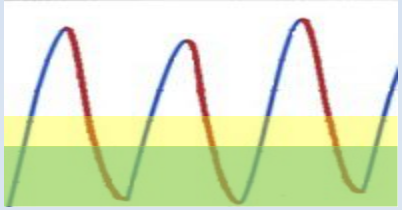
Chest amplitude checked :

- a belt around the patient chest measures the abdominal volume
- the motion of multiple fiducial markers are tracked simultaneously in two-dimensions (in the vertical plane) using a standard video camera.

Time splitting : The motion is split into different phases and images are reconstructed for each phase of the cycle.



Amplitude splitting : images are recorded if the motion lies between adjustable lower and upper limits



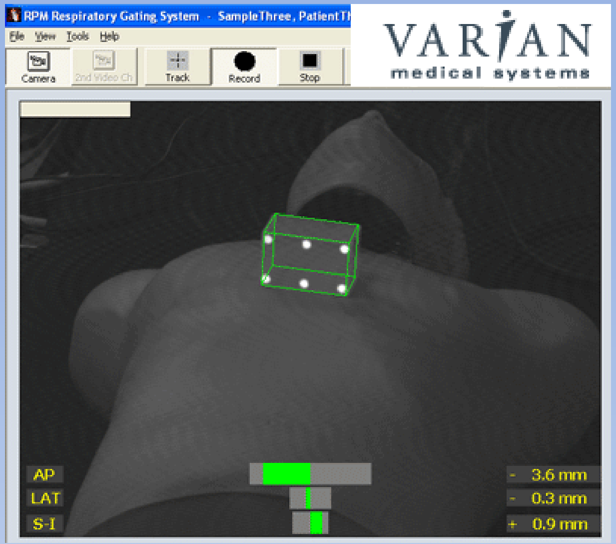
Belts for respiratory motion monitoring

PHILIPS

Deformable belt that when placed across the patient's chest/abdomen measures abdomen circumference.

It generates a breathing signal corresponding to the expansion and contraction of the abdomen. The breathing signal is fed to the PET/CT system.

SIEMENS ANZAI system



Real-time Position Management (RPM) System :

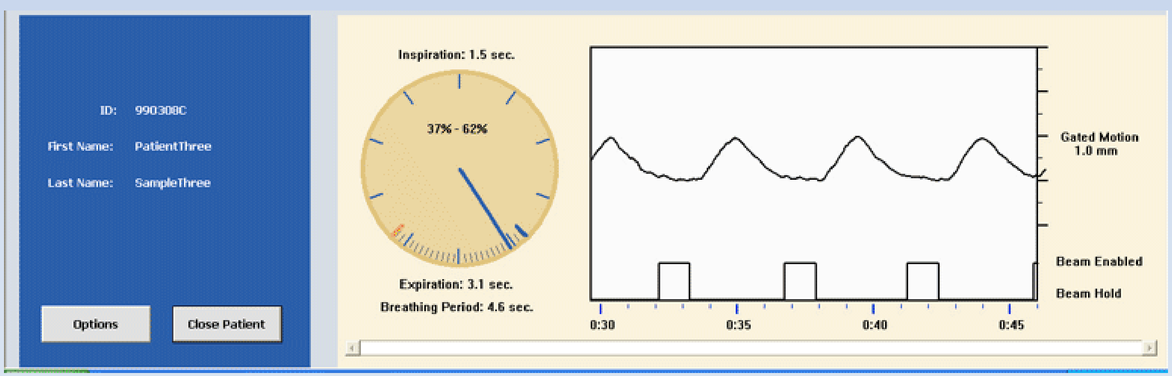
(used in CT / simulator / clinac)

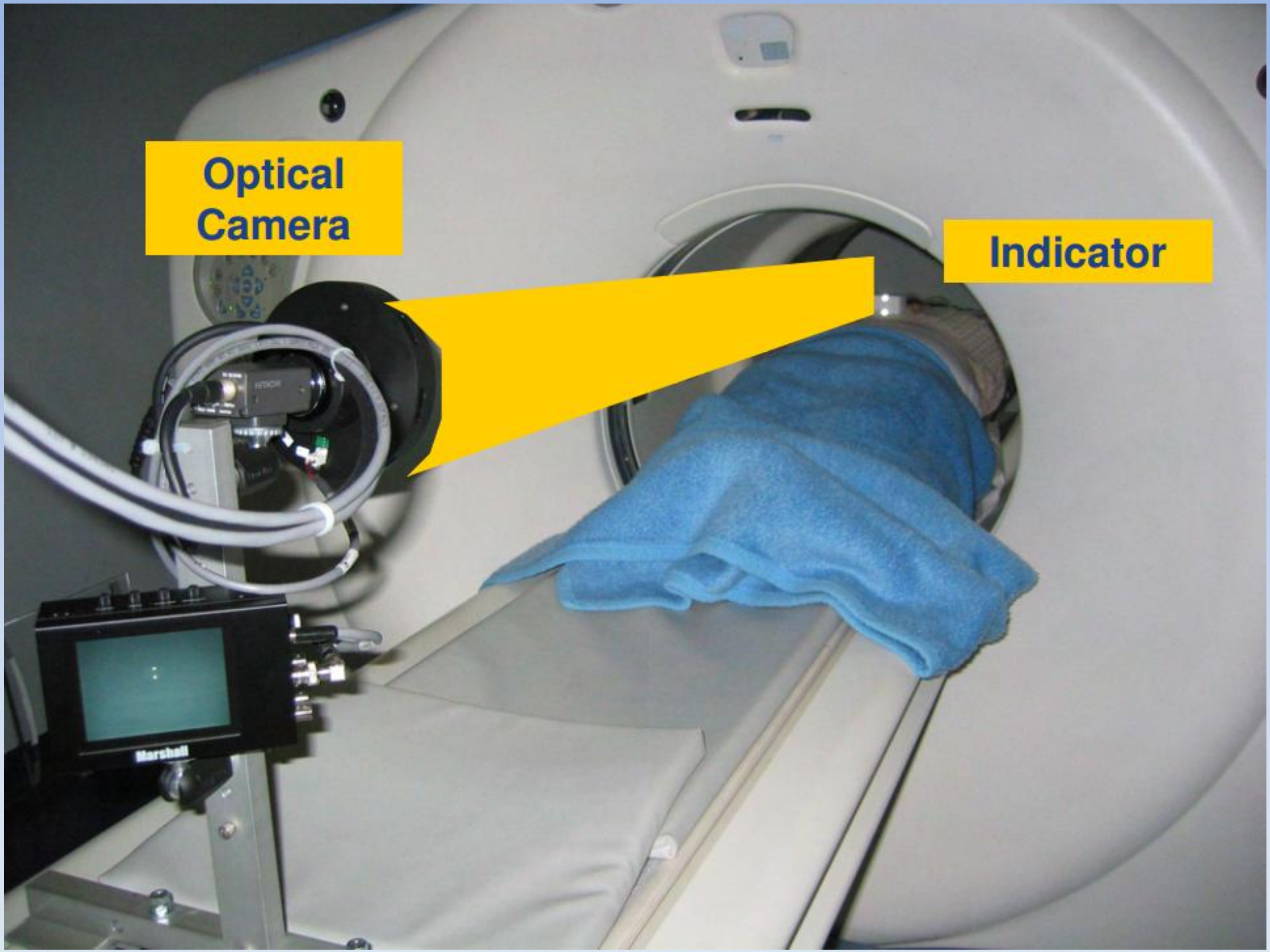
Reflective external marker placed on the abdomen of the chest

Infrared illuminator coupled with a CCD camera

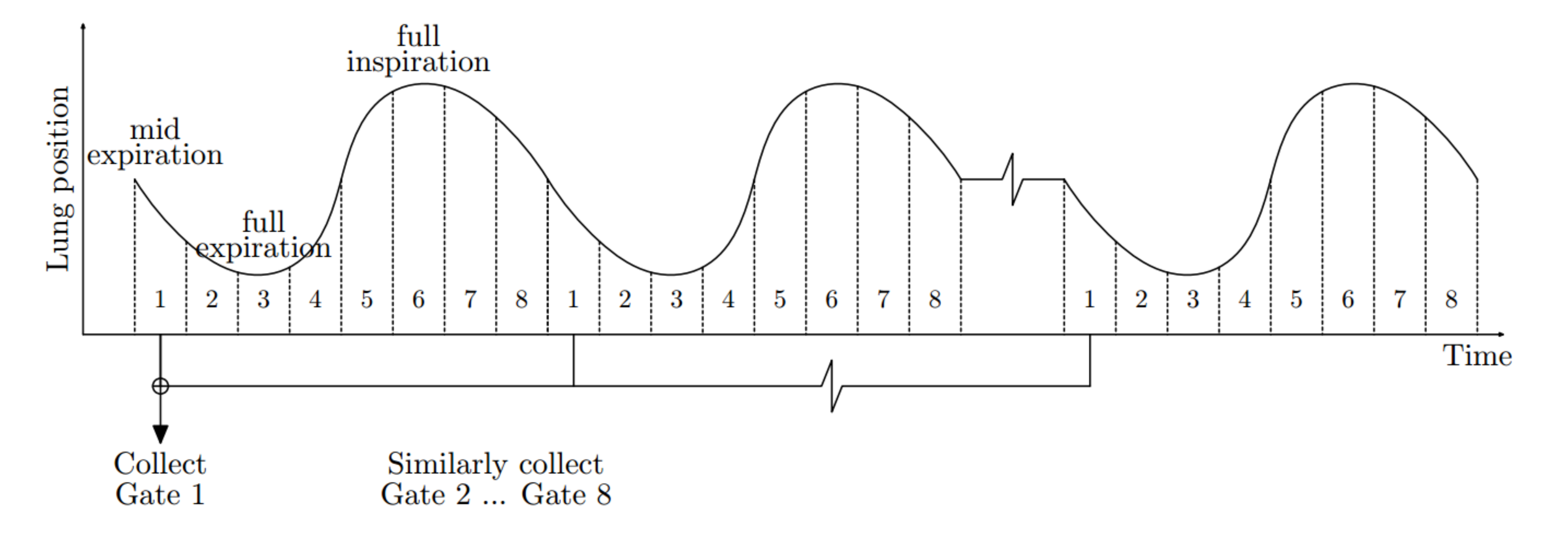
Workstation to process signals and generate triggers for PET CT acquisition.

GE PHILIPS SIEMENS





Optical Camera
Indicator



Mid expiration triggers a sequence of 8 sub samples, or gates.

Gates created assuming a perfectly periodic breathing motion.

PET data are divided into a number of gates : “motion freeze”

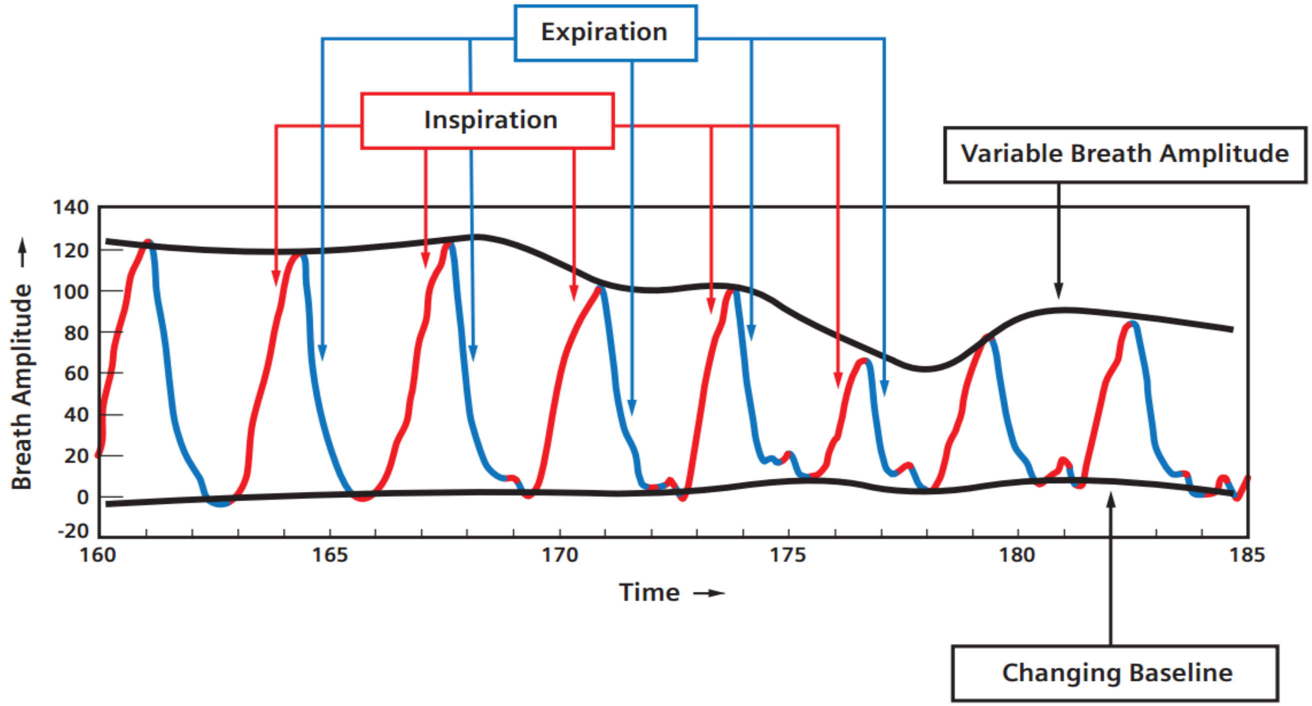
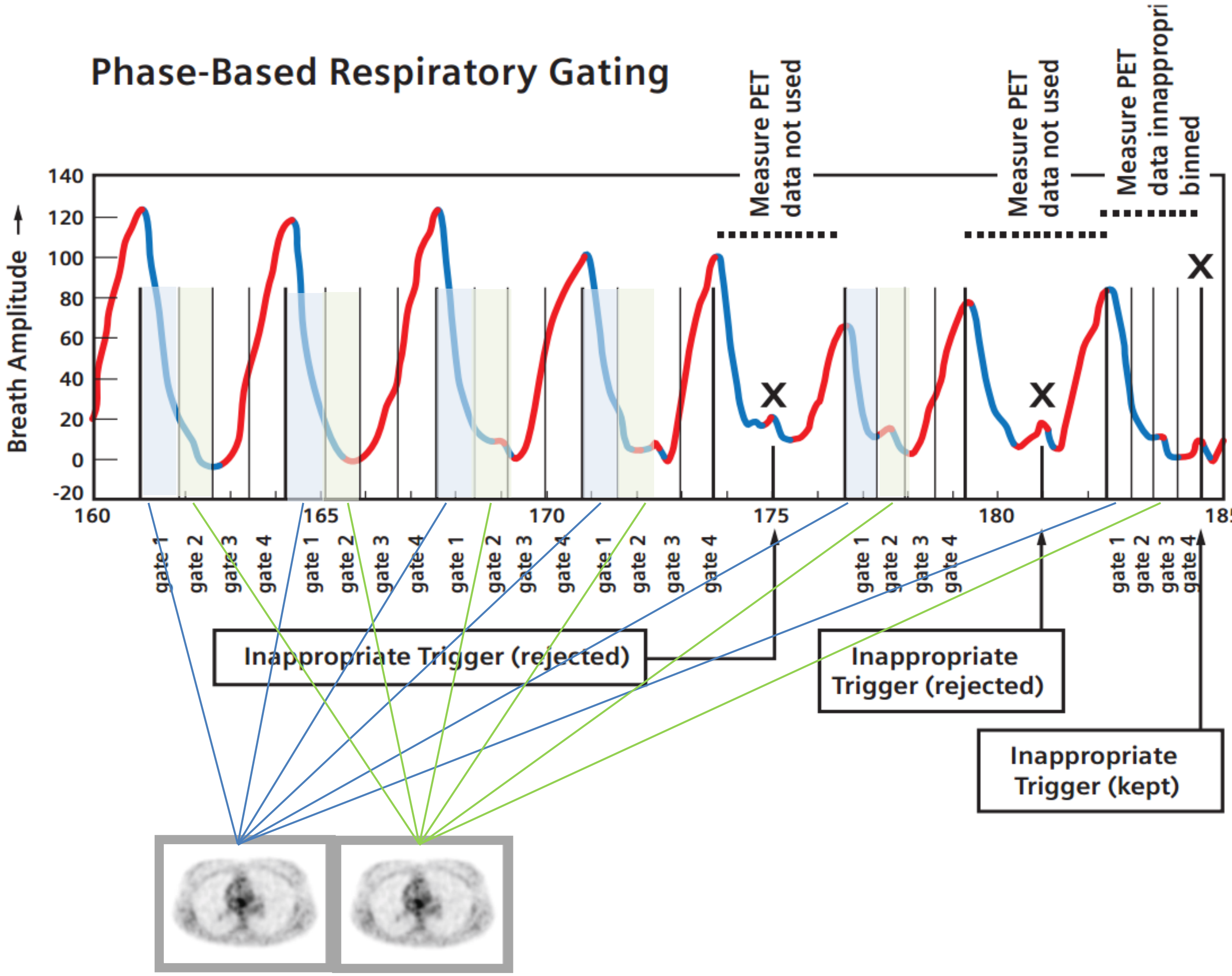


Fig. 1. Respiration signal vs. time in a 25-sec interval.

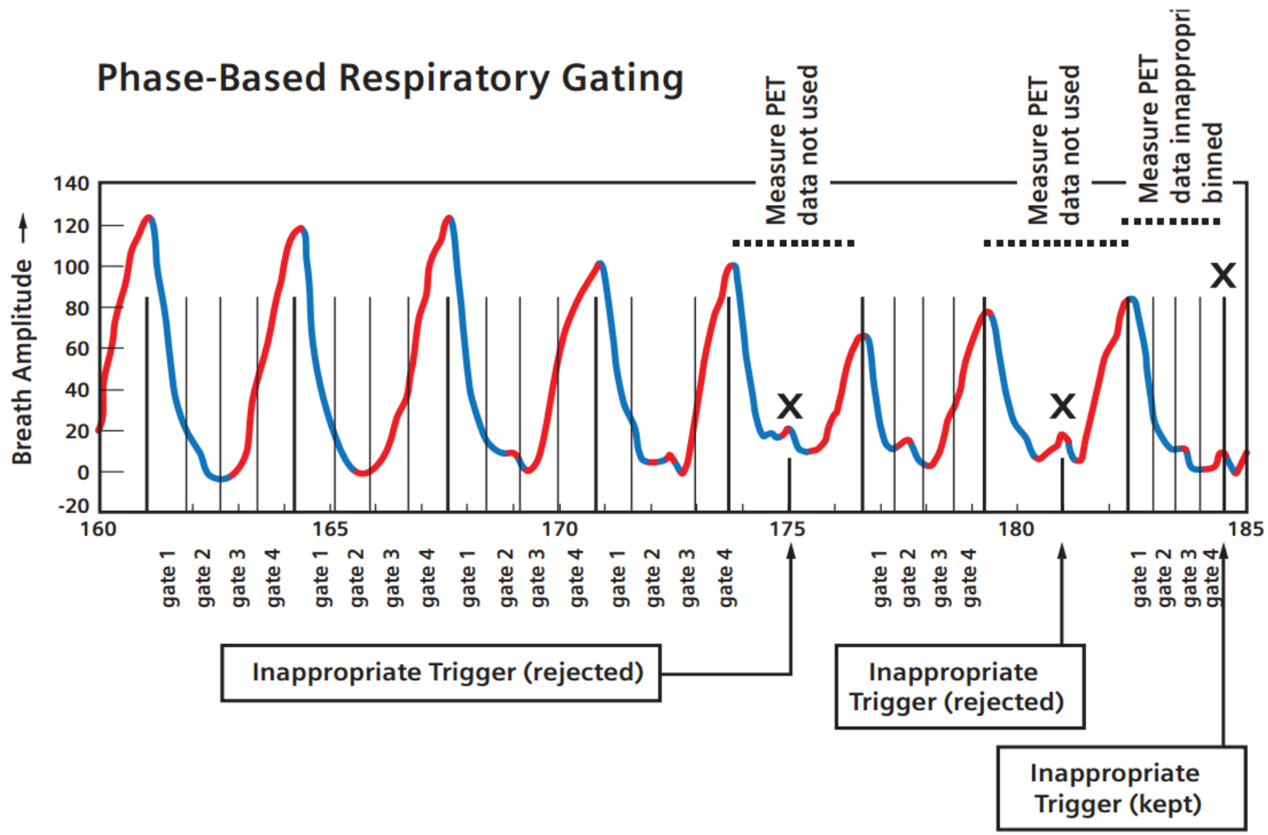
Example of time based (or phase-based) gating



Set parameters at the beginning of the PET session :

- Trigger defined at the max. of respiratory motion (end of inspiration).
- Four gates are defined after the trigger.

Decide online whether or not to keep the data.



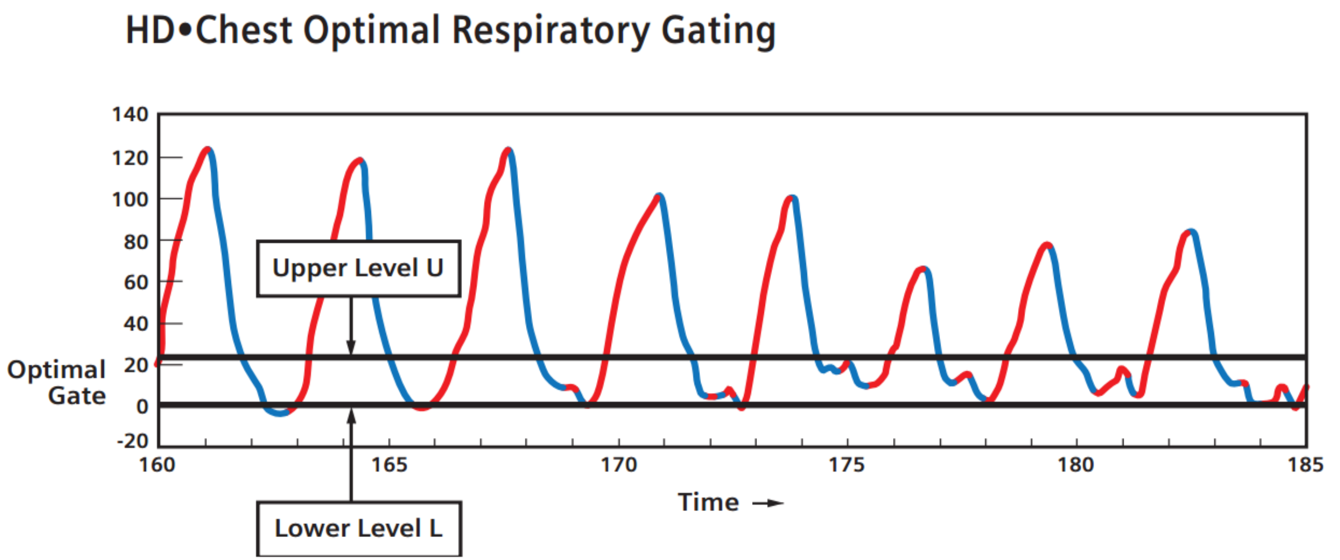
Time splitting :

Splitting into different phases reduces the amount of available data : images are noisy.

Bad triggers will reject additional data, making the images noise worse

Some inappropriate triggers are kept, mixing bad data with good data

Example of amplitude based gating



SIEMENS

Amplitude splitting :

SIEMENS HD Chest : acquisition is performed when the amplitude is above a lower level L but below an upper level U.

The user selects the percentage of data that will be kept, which set the value of U when L has been selected.

The data is analyzed at the end of the acquisition. After it has considers all possible L values, the computer determines which one gave the smallest range of respiratory motion. This determines the optimal setting for L and U,

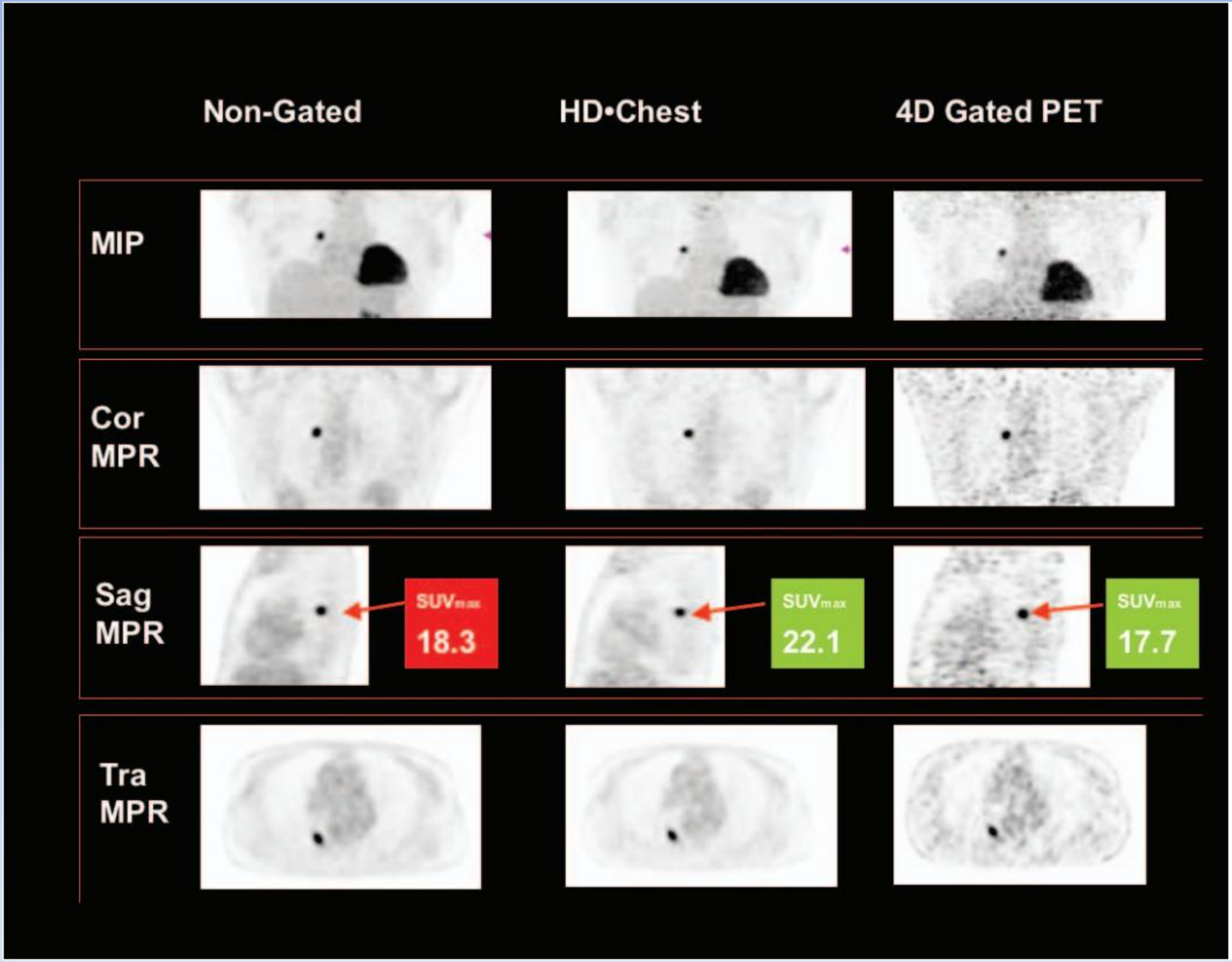


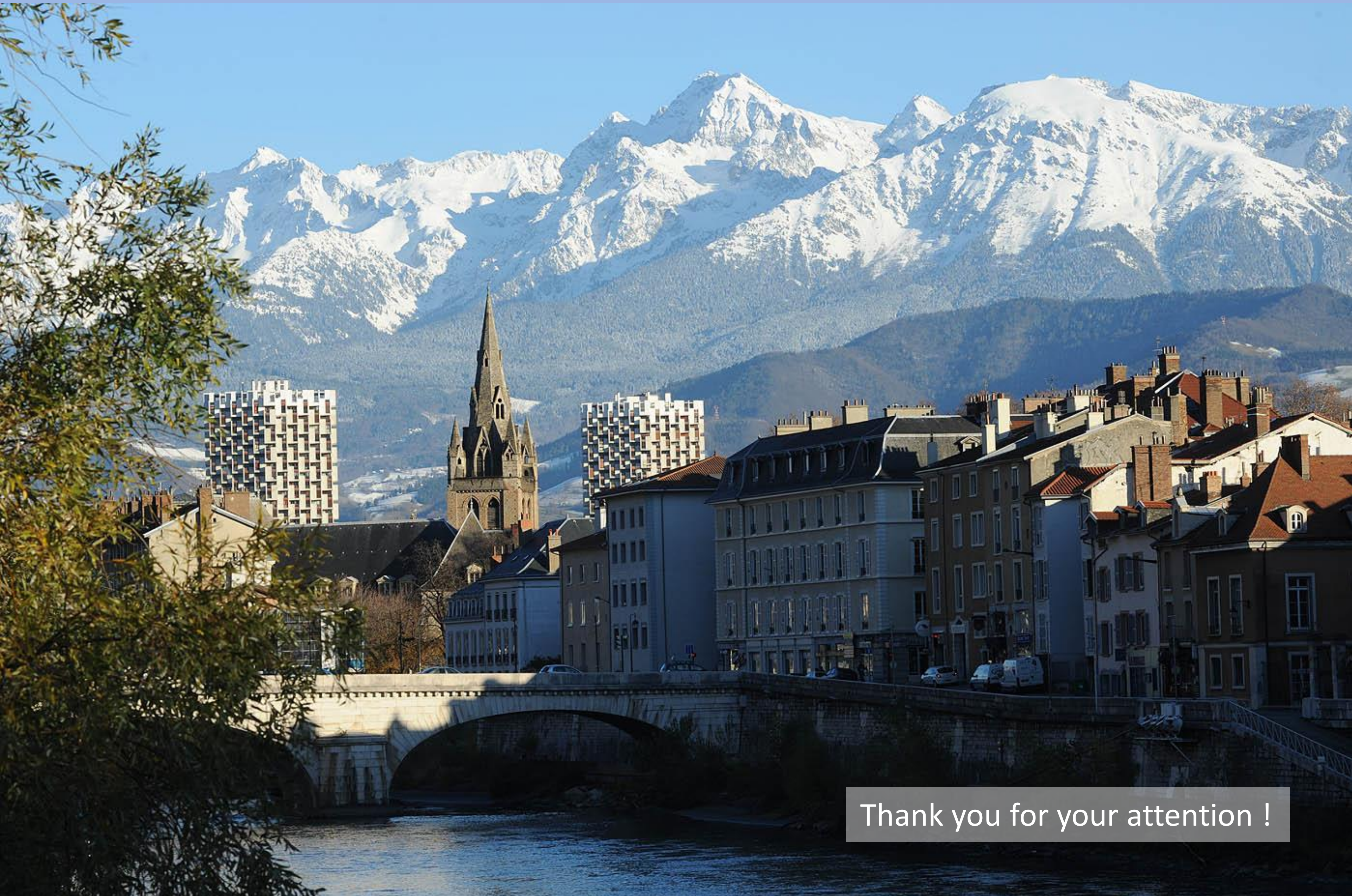
Figure 3. *A solitary pulmonary nodule in the posterior aspect of the right upper lobe, showing the improvement in SUV value with HD•Chest. Although the nodule is well delineated in the non-gated PET, respiratory motion related effects make the nodule appear slightly larger on the non-gated compared to the HD•Chest study. The HD•Chest study also shows higher SUV_{max} compared with non-gated PET. (Data courtesy of University of Michigan)*

Amplitude splitting : clear improvement in the SUV quality !

Apnea protocols with spirometer are being tested on selected patients (PET : 20 sec. hold x 3 times, CT : one 5s hold). Reduces acquisition time but gives results similar or worse as compared with gated acquisitions.

Conclusion

- PET is evolving rapidly in many aspects (quantification, image quality, PET/CT)
- Mastering its features is a least a ½ full time job for a medical physicist :
  - Commissioning (individual and geometrical efficiencies)
  - Correct from natural organ motions and systematic effects to achieve proper SUV quantification :
    - 4D-CT,
    - amplitude based PET acquisition,
    - partial volume correction
- PET, a precious tools in nuclear medicine



Thank you for your attention !