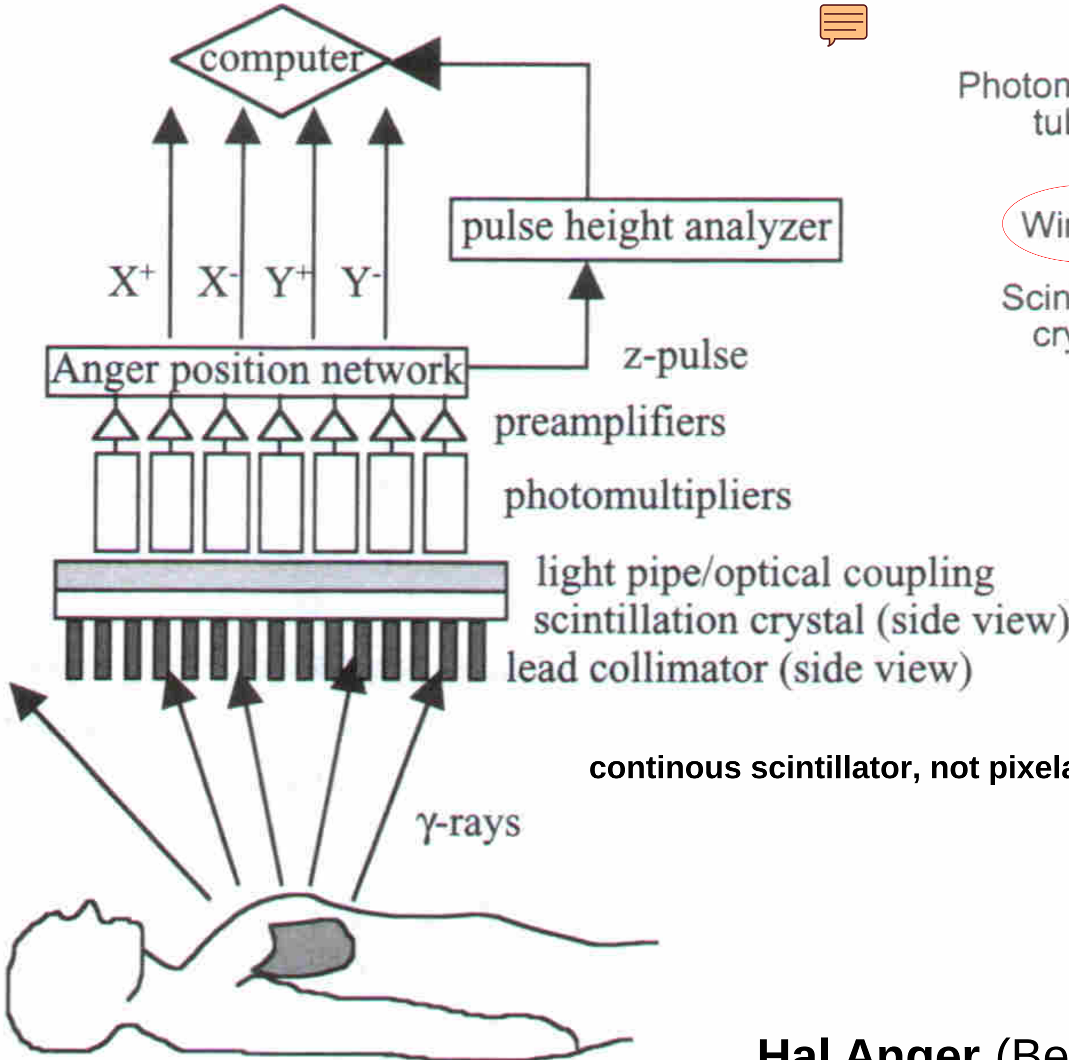
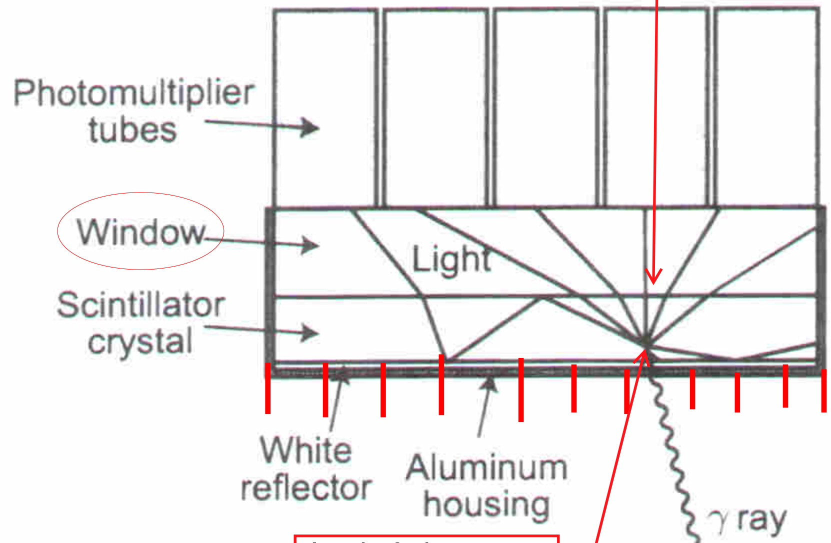
=> FOR GAMMA RAYS! *NOT FOR PET

*non sono PSPMT

# The Anger camera (1957)

divergence, light spread

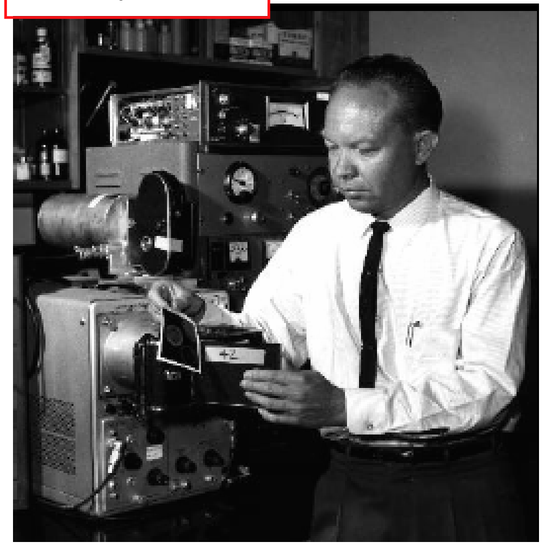



cloud of photons

**continous scintillator, not pixelated**

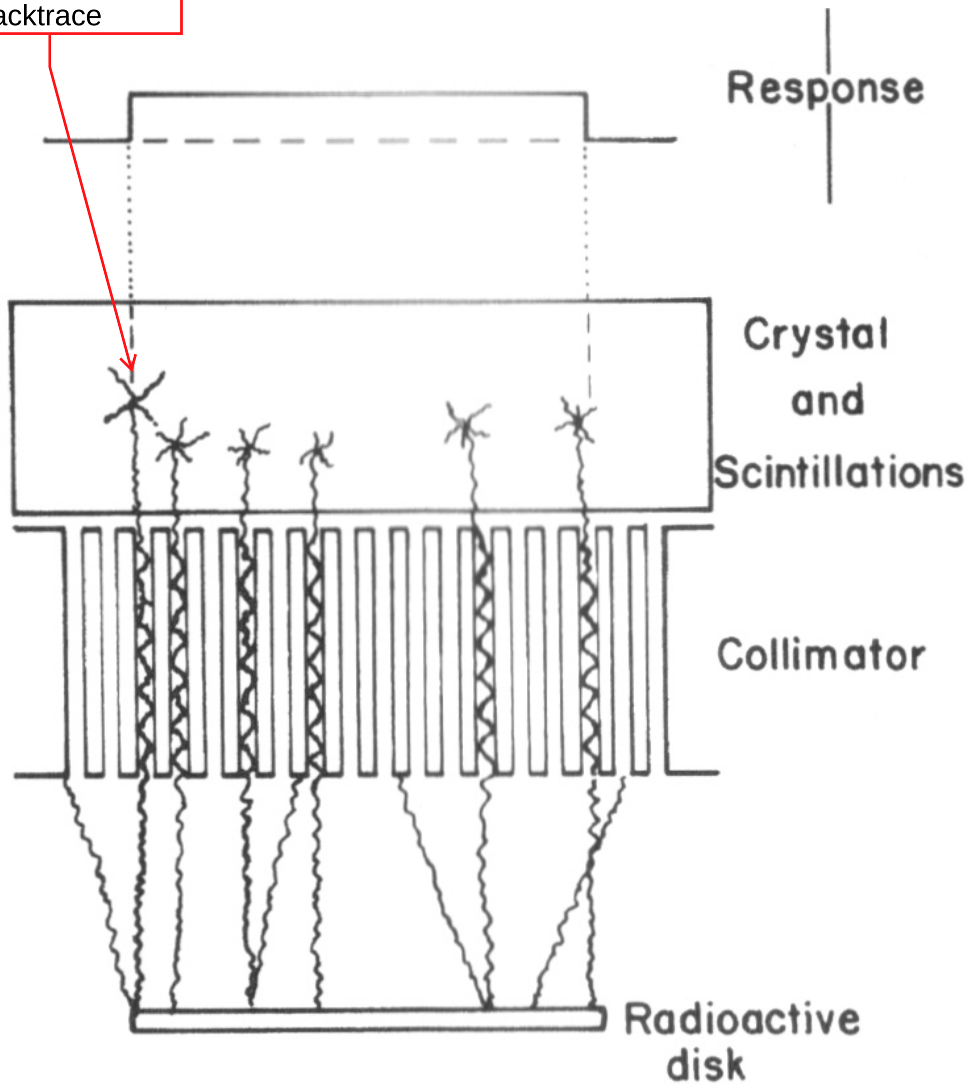
**Hal Anger** (Berkeley)

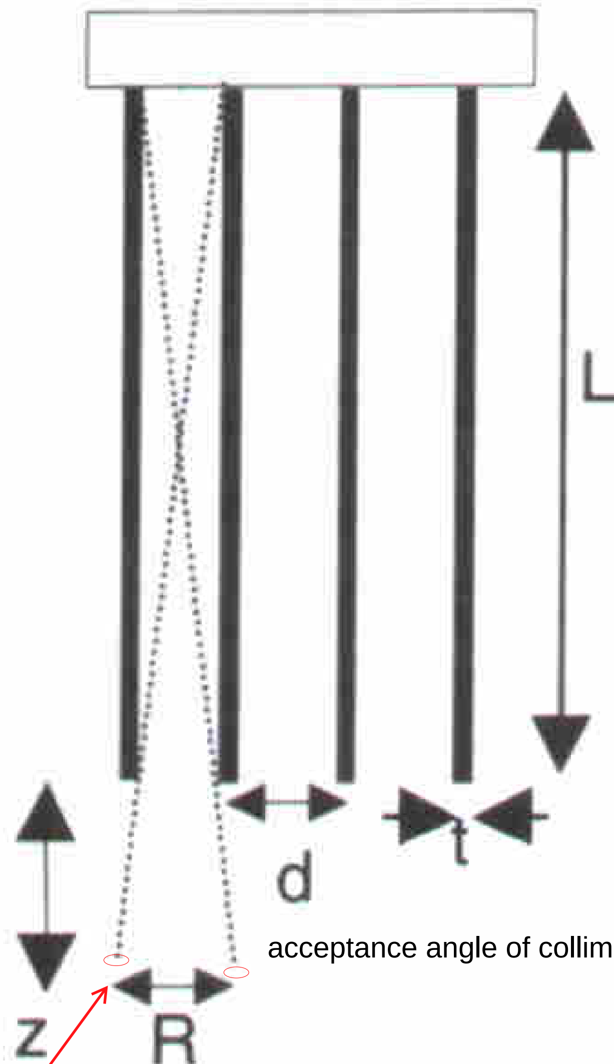


# The collimator

position of interaction calculated by the camera. Then we can backtrace

main parameter: lenghs of grids holes diamater, distance to patient




acceptance angle of collimator.

both can enter.. thus Resolution: capability of distinguish 2 sources

# Spatial resolution

Intrinsic of the camera: capability of the camera to detemrine the coordinate of interaction. Statistically independent to respect the collimator, depends on reconstruction capability [statistic of gen., **how scint spread over p.d. (the cloud of scint. photon is not deterministic**!)] Message: **whatever alg**., due to spread of signal, we have a spread of reconstructed coordinates. Also: no sense of using a collimator grid with R=1mm if the intrinsic is 10mm

2 contributions: quadratic sum (independent)

$$\boxed{R_{tot}^2 = R_G^2 + R_i^2}$$

$R_G$ : geometric resolution

limitation just due to collimator

$R_i$ : intrinsic resolution of the position detector

"the lower R the better resolution"

smaller the hole.. the better the resolution

longer z, the worse the res.

$$\boxed{R_G = \frac{d\,(L+z)}{L}}$$

longer L.. smaller is the acceptance angle

Not only res! also statistics plays a big role! Dictated by g-rays that surives entering the holes. Many are killed. We can calculate sensitivity efficiency! IT DOESNT DEPEND ON Z! the distance change the res. but not the senst. [If Z is high, you start to have some g-rays entering in the neighbour.. ofc with a worse res :D. So it can be demonstrated that changing Z we have same number of photons entering in the whole].

Can we have small R and large S? no, they depend reversly at same parameter. If we decrease d for example we reduce also S. Typical **tradeoff: RESOLUTION/SENSITIVITY.**

# Sensitivity (efficiency)

indeed depends on main factor

smaller the hole, less g-ray passes!

$$\boxed{S = k\left(\frac{d^2}{L\,(d+t)}\right)}$$

[larger the better]

longer.. smaller the acceptance angle

thickness of walls

k : costant dependent on the detector geometry

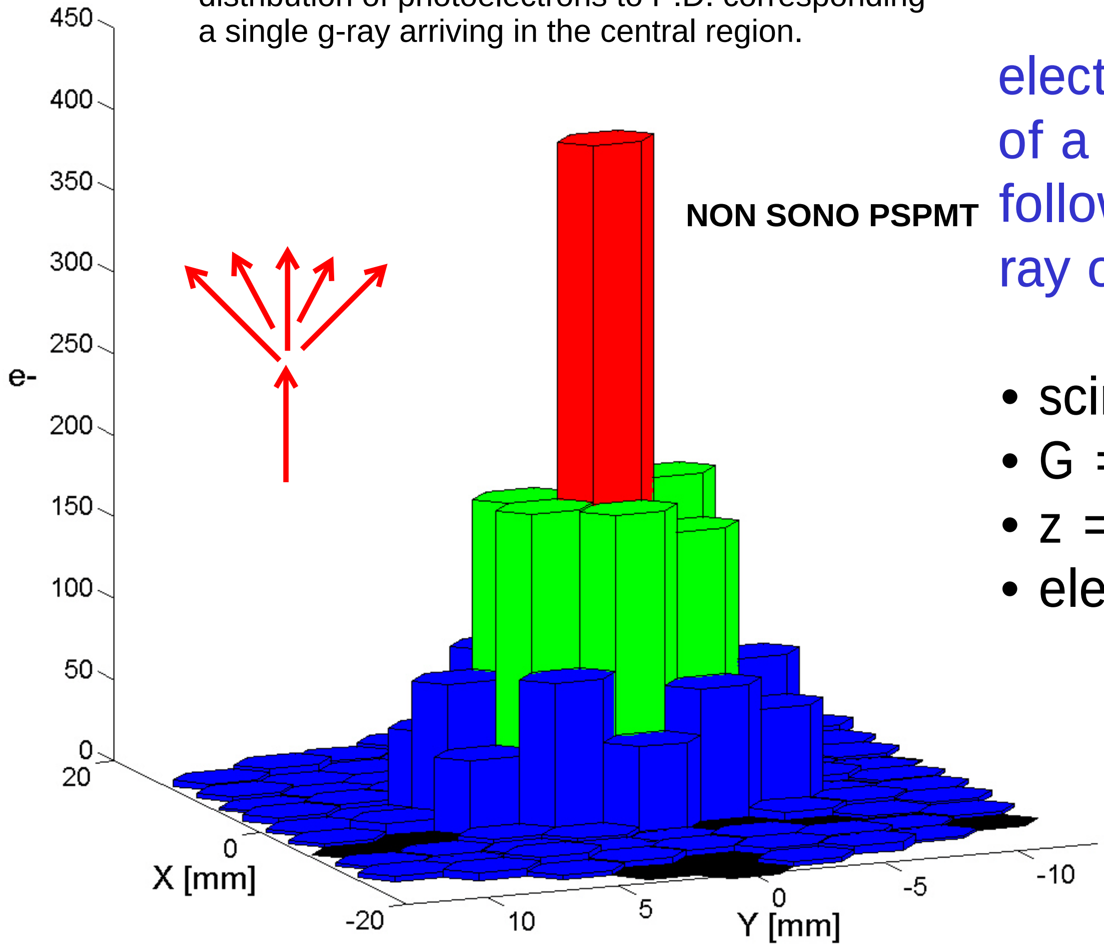
note: for $d \rightarrow 0$

$R_G \rightarrow 0$ but also $S \rightarrow 0$

$\Rightarrow$ resolution $\longleftrightarrow$ sensitivity technique to reconstruct position of interaction. First: problems. We have a g-camera with many P.D. for a single scint. (in this case, exagonal shape P.D., best way to sample a surface).

# Influence of the photodetector electronics noise to the spatial resolution of the Anger Camera

distribution of photoelectrons to P.D. corresponding a single g-ray arriving in the central region.



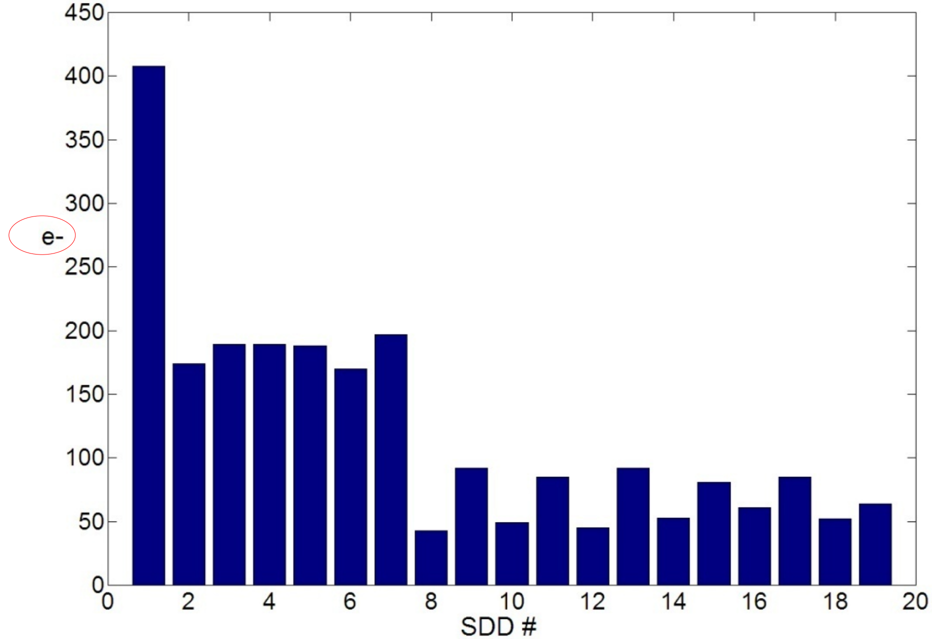

electrons collected by the units of a photodetector array following the absorption of a $\gamma$ ray of 140keV:

- scintillator: CsI(Tl)
- G = 15 e-/keV
- z = 5 mm
- electronics noise neglected

(Monte Carlo simulations)

We implement centroid algorithms: according to the distribution they reconstruct the x-y coordinates of the g-ray interaction. It's not determinisitic, b/c the **contribution of photoelectrons** to the signal **are affected by statistics**. This was a simulation created on purpose in a symmetric position.. but still contribs are not equal. That is because the **generation of light is a poisson process (Scint)**. conv->electrons also a statistical problem. And LASTLY: photons bouncing on the walls of scintillator. 1st conclusion: yuo have a statistical dependant, so with any reconstruction of x-y system we'll have an error. In addition: electronic noise distribution of signal among various det.

criterium to design for noise

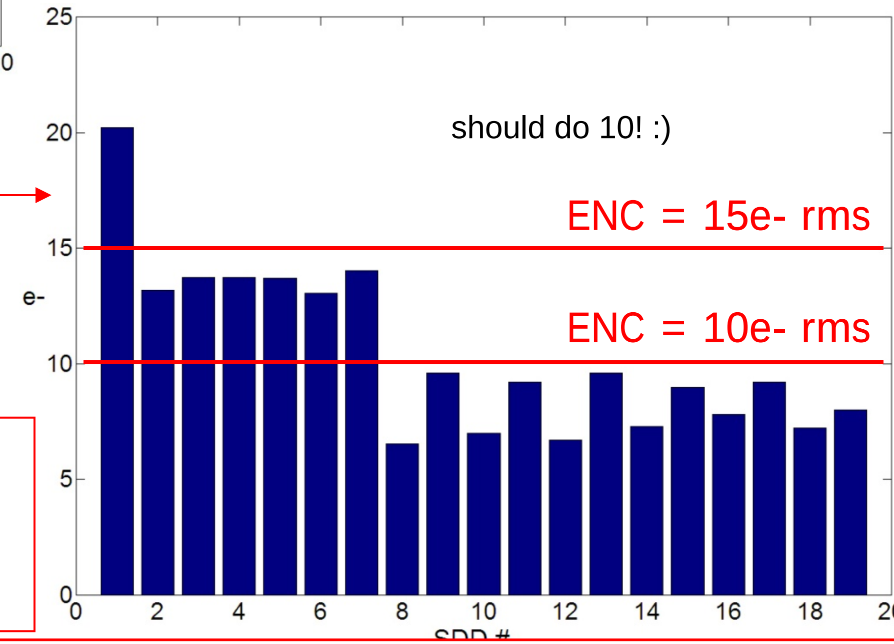



electrons collected by the first 19 photodetectors in decreasing order of signal

**(2) assuming poisson fluctuation, intrisic fluct. of each is the sqrt() of the signal... sigma [electrons]: this provides the reference! the electronic noise should be in the order of 10/15 electrons.**

**=sqrt(avg)**

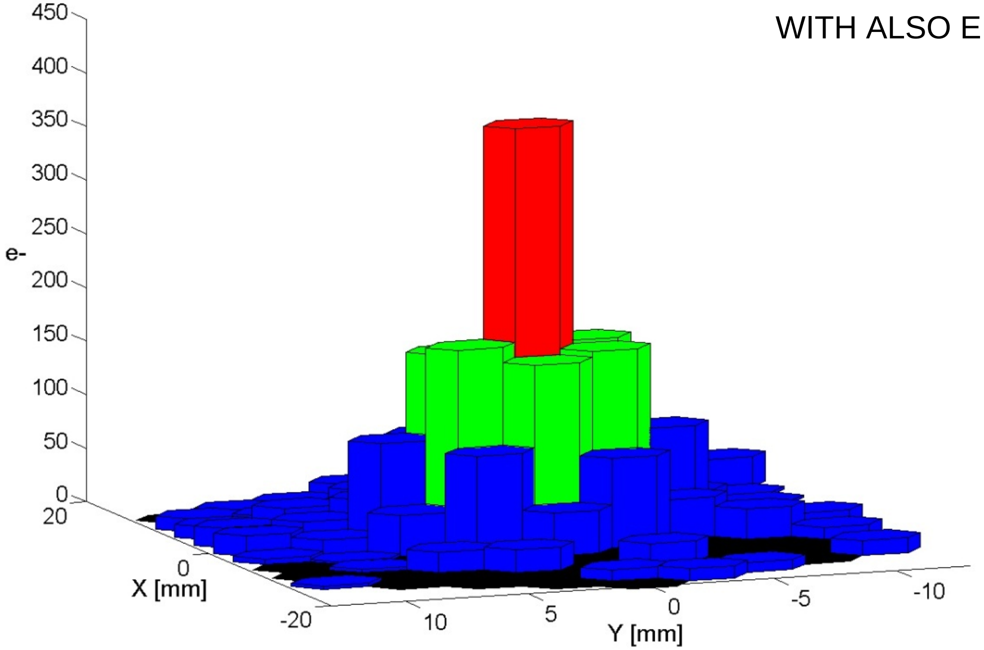
statistical fluctuation ($\sigma$ *rms*) of the charge collected by the photodetectors (Poissonian statistics)




$\Rightarrow$ to have the electronics noise smaller than the statistical fluctuation, ENC < 10-15 e- rms

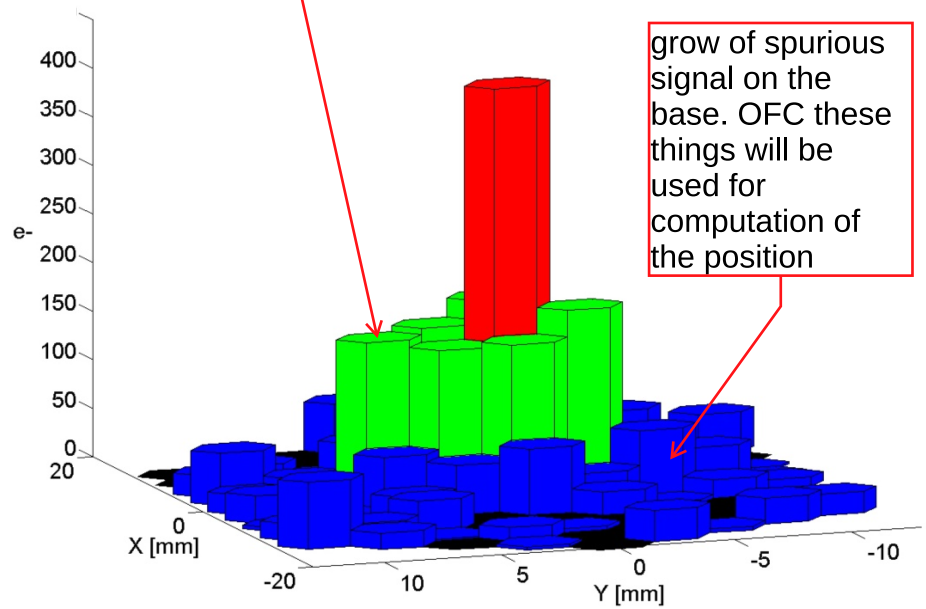
(1) How careful should be the design of electronics? we shouldnt design noise smaller than needed.
NOT ACCEPT AN ELECETRONIC NOISE WHICH IS OF THE SAME ORDER OF THE **INTRINSIC SPREAD OF THE SIGNAL**.
Indeed we have seen the signal has an intrinsic spread. with changes from g-ray to g-ray interactions. what's the intrinsic spread of signal? (3) if we do worse.. eg. 30: we get problems!

WITH ALSO ENC: additional spread.




ENC = 15 e- rms

this now is green!
fuck




grow of spurious signal on the base. OFC these things will be used for computation of the position

ENC = 30 e- rms

**1. spread intrinseco (gen. fotoni luce poissoniano)**
**+ spread dovuto al divergere**
**2. presenza di electronic noise**

*i grafici di prima sono in e-, ora vorremmo energia. Metodo easy: sommo ed uso un fattore di conversione G

# Energy resolution

conversion factor b/w signal in el- and g-ray energy

electronic!

$$E_g = G \times \sum_i (N_i + noise_i)$$

worse for P.D. where the signal is already low

$N_i$ : electrons collected by the unit *i* of the matrix

$noise_i$ : electrons associated to the electronics noise of the unit *i*

G : gain factor keV/e-

⇒ the sum of the electronics noise of all the units of the array (larger than a given threshold) contributes to the statistica fluctuation of the computed energy

Same for energy resolution. We need to reconstruct the en., because we may need to reject scattered g-rays. Easier: we just sum the signal. *scattered gRay avranno una energia TOTALE minore (totale eh! infatti somma tutti)

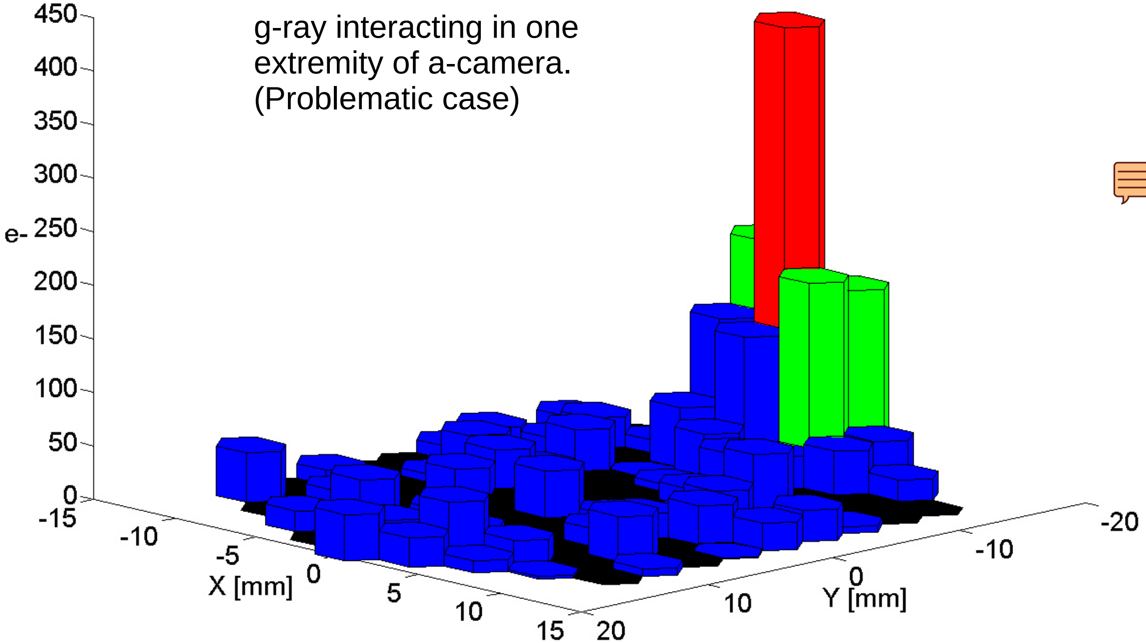
[in the anger-camera]

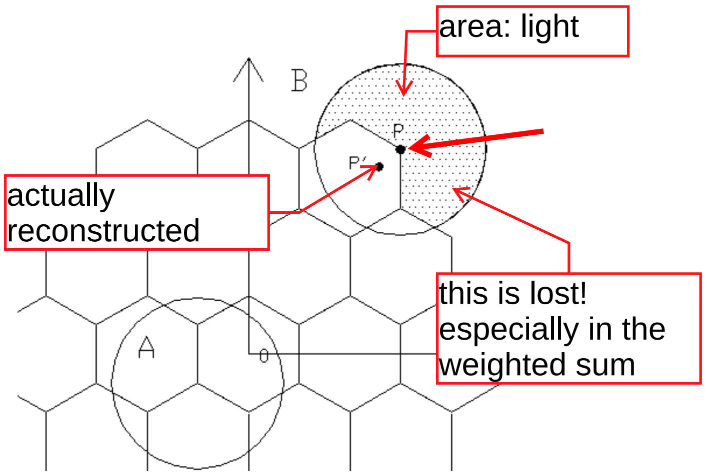
# Reconstruction of the position of interaction by using the centroid technique (center of mass)

WEIGHTED SUM

$$x_0 = \frac{\sum_i N_i \cdot x_i}{\sum_i N_i}$$

$$y_0 = \frac{\sum_i N_i \cdot y_i}{\sum_i N_i}$$

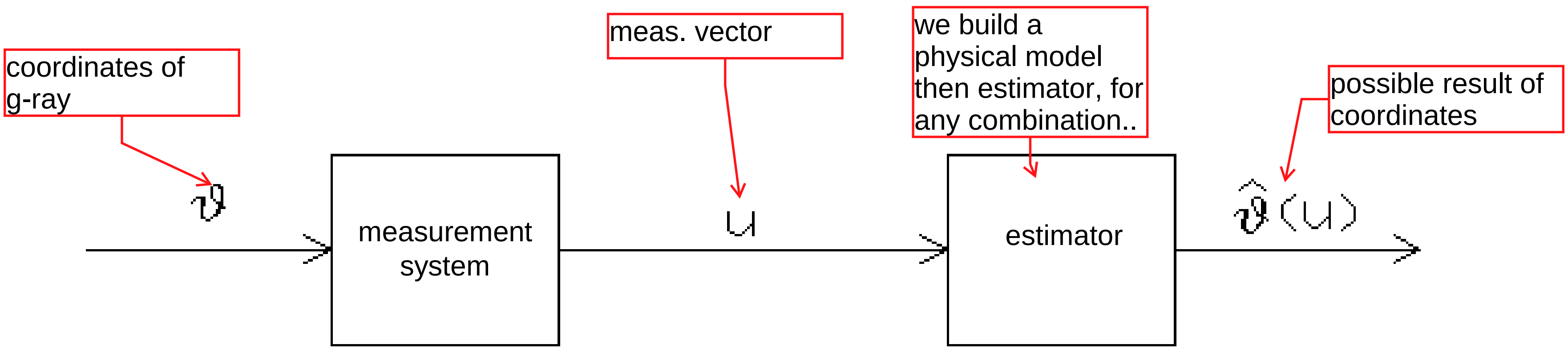


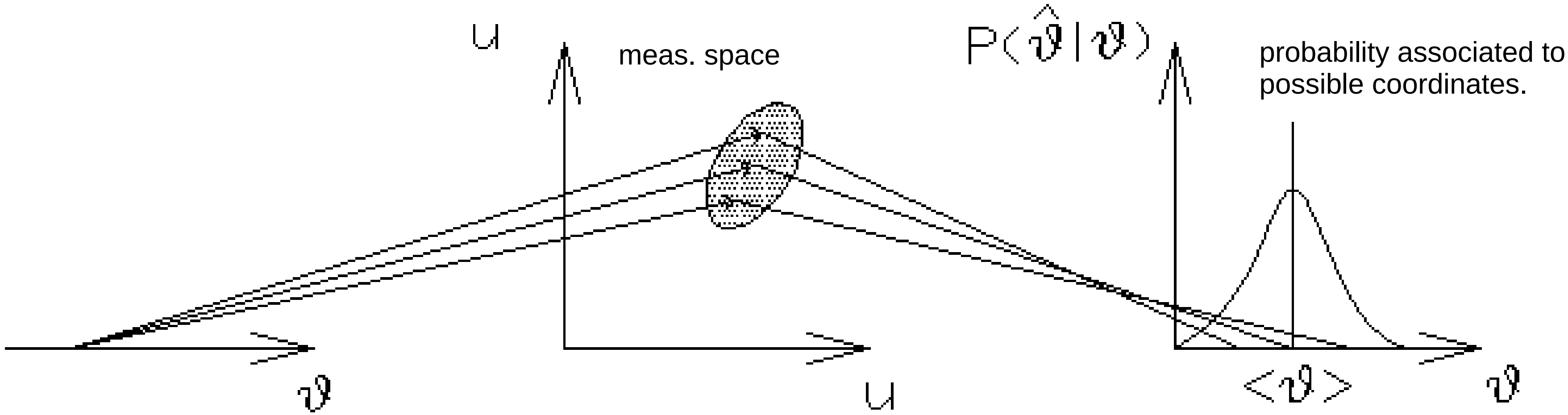




## the reconstruction may result distorded at the borders and the spatial resolution accordingly limited

We want to exploit that the distribution is spatially dependent on the interaction point.
1) Center of mass(or gravity): weighted sum of coordinates of each p.d. (center of each p.d.) then we multiply each coordinate by the number of photoelectron collected. It's easy and can be done by hardware. FAIL: 1st case: **borders.** 2nd case: **faulty** p.d.

# Reconstruction of the position of interaction by using the technique of the maximum likelyhood (ML)



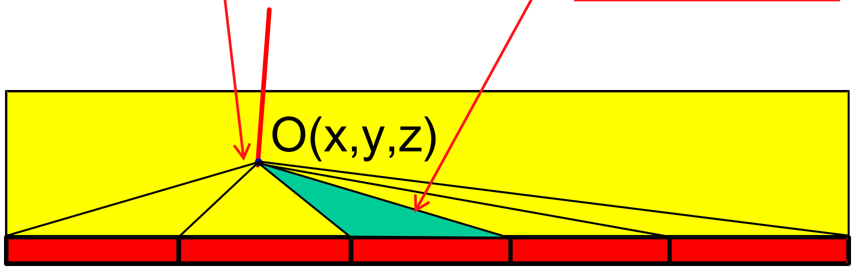




ML techniques: based on a prior physical model + statistical techninuques to reconstruct position. Most used: MAXIMUM LIKELYHOOD: statistical rec. method, not to reconst. all the image, just the position of interaction. **The resul will maximise a probability distribution**. We dont find coordinates thanks to analyitical calc, but a probability calc. Finds coordinates that better matches the meas.

any possible point of interaction

**1. MODEL => LOOKUP TABLE** (ni(x,y,z) stored on the pc)

solid angle




$n_i(x,y,z)$ number of e- collected by the unit *i*, calculated as function of the position O(x,y,z) **-> sarà considerato average probabilità**

ph. model determines the number of e- collected on each p.d. $n_i(x,y,z)$

1) we can find ni by the solid angle
or
2) optical tracing simulator

**2. WE COLLECT THE MEASURES**

$m_i$ number of e- measured by the unit *i*

**3. WE BUILD A PROBABILITY DIST.**

probability (Poisson) to obtain $m_i$ supposing the average number of electrons equal to $n_i(x,y,z)$:

$$P_i(m_i, n_i(x,y,z)) = \frac{n_i(x,y,z)^{*\,m_i} \exp(-n_i(x,y,z))}{m_i\,!}$$

I want to maximise this!

We assume ni average number of photoel. collected by each p.d. Given this, which is the probability to have mi on this detector? conditioned probability

**4. RESULT: global prob. maximized**

joint probability for all units:

(likehood estimator)

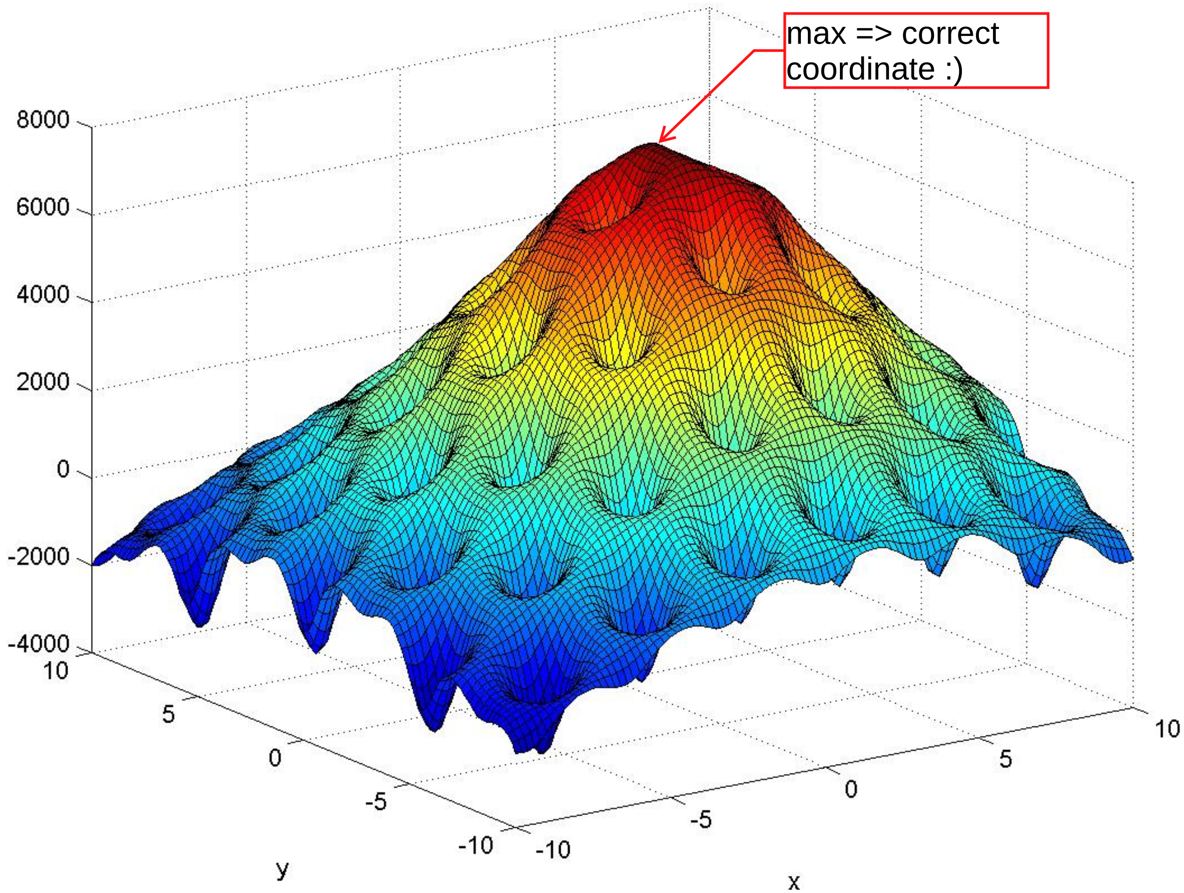
$$P_{tot} = \prod_{i=1}^{Ntot} P_i(m_i, n_i(x,y,z))$$

**SLOW TO RESPECT NN**

The best estimation of O(x,y,z) : $n_i(x,y,z)$ which maximizes $P_{tot}$

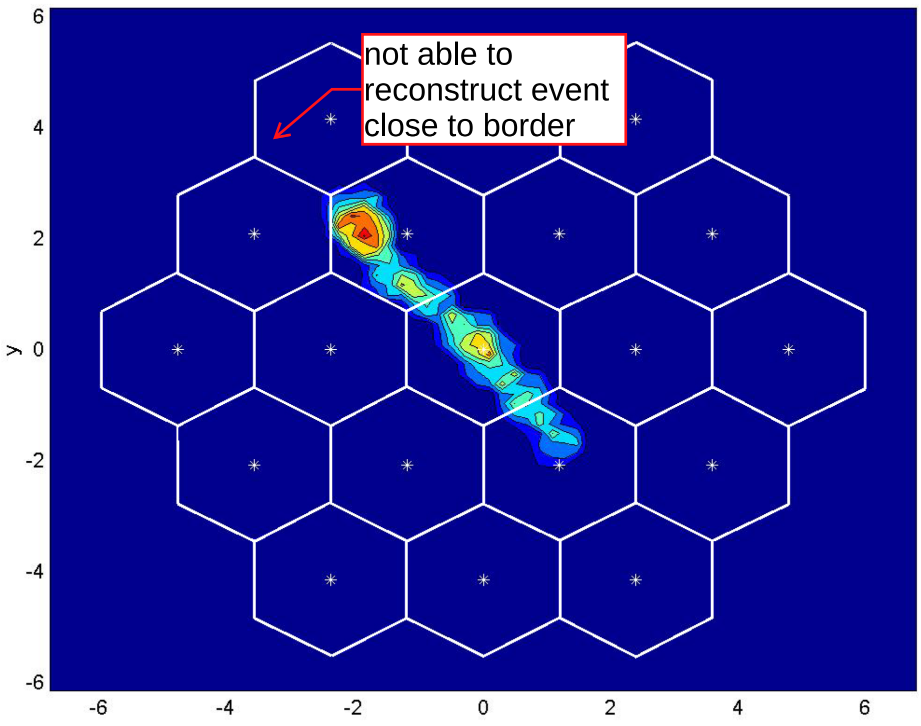
(1) We first have to build a model of the camera. For any point of interaction (x,y,z). Solid angle formula: first method. But we can do better: we also have contribution due to bouncing, spread around scint => optical tracing sim. We can also do real measurements. (...X) [audio] e.g. result: it's max when ni = mi, so we have mi = 137, we search for ni = 137 and we check for its coordinates. We do this for all the p.d., because one is affected by noise => joint probability. If I maximize this.. I found the coordinate which matches the probability better for all the p.d. **If one detector makes a mistake I compensate by another det. => I compensate for fluctuations.**

probability (likehood) corresponding to the interaction seen on slide 4



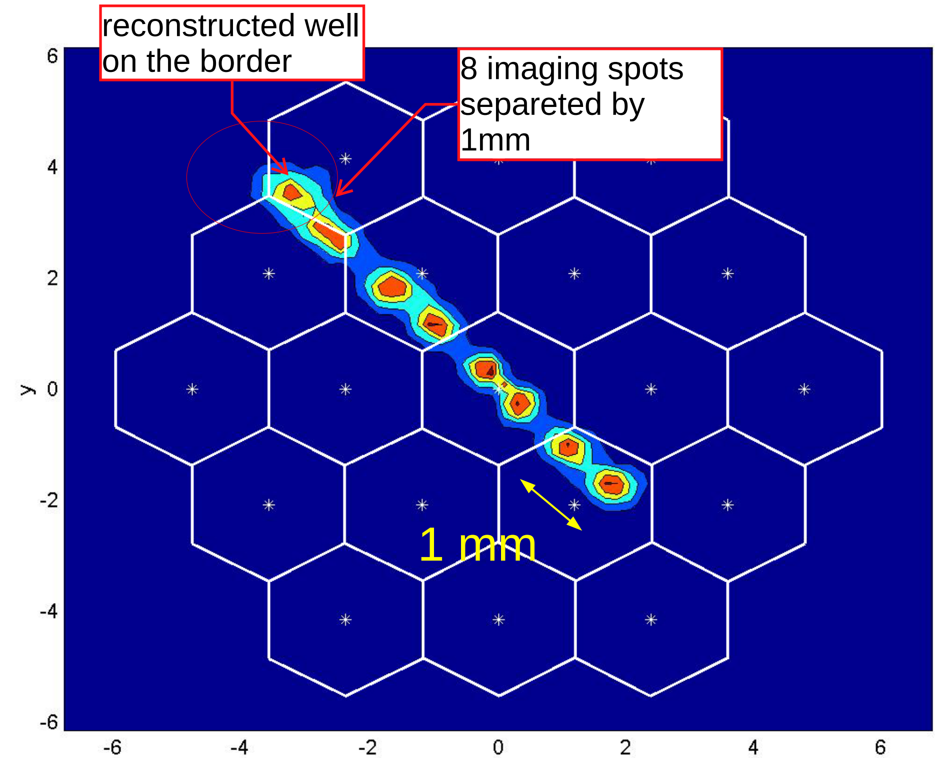

Function of the maximum likelyhood corresponding to a measured event (signals from a photodetector array)

# WHY IT'S BETTER




reconstruction with centroid

recostruction with maximum likelyhood



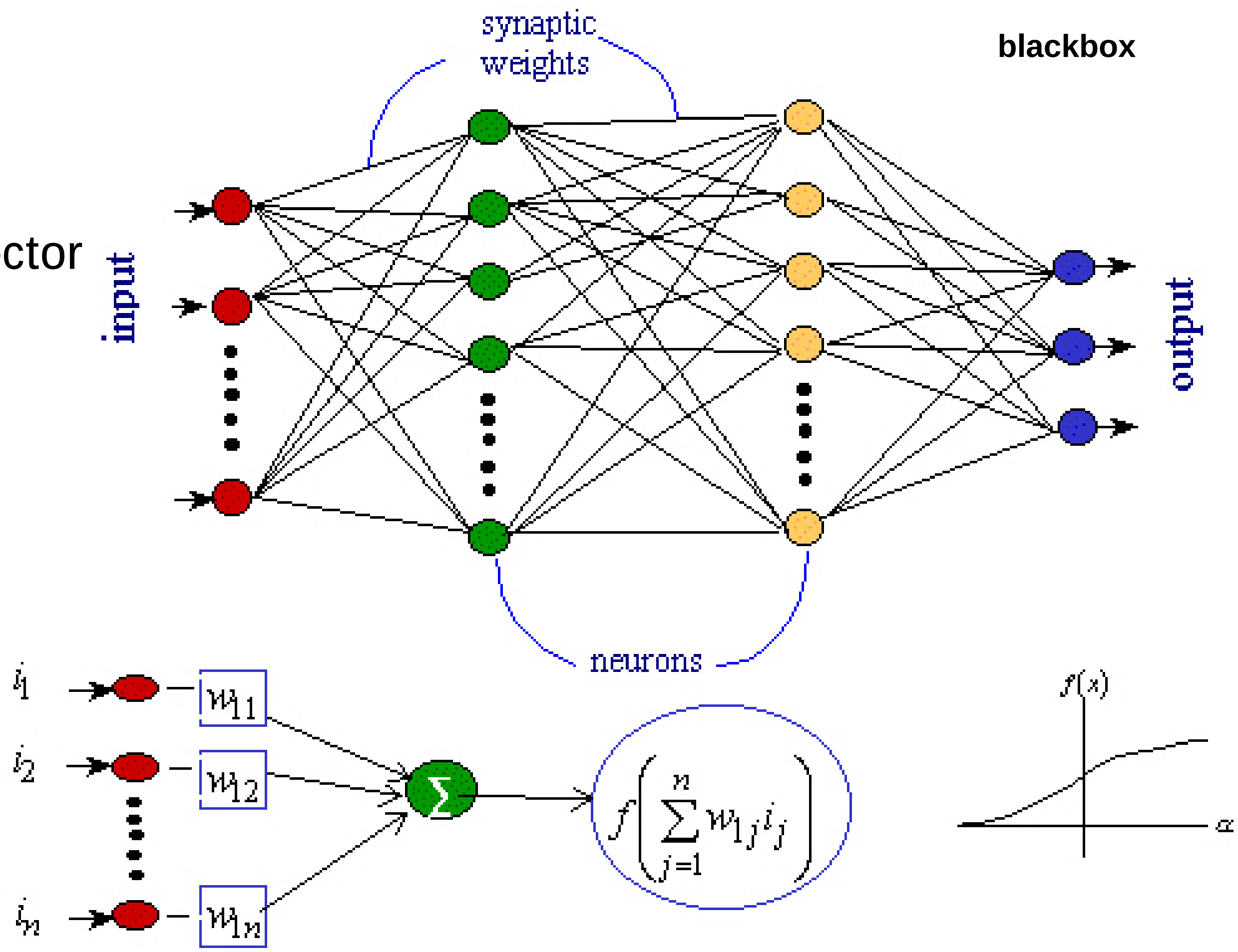

Also in max likelyhood we miss light at the borders, but it's based on previous knowledge of detector!! So the maximization performed takes this into account. And if one p.d. breaks it's a not a 0 in the sum.. I simply don't put it in the joint prob, (which is based only on the surrounding p.d. [?]). => you are not forced to use all the p.d., you can use a subset p.d., for e.g. the ones that have an high signal. [domanda: tutti contemporanei nell'esempio?]

# Recostruction of the position of interaction with the technique of the Neural Networks (NN)

fast

**blackbox**

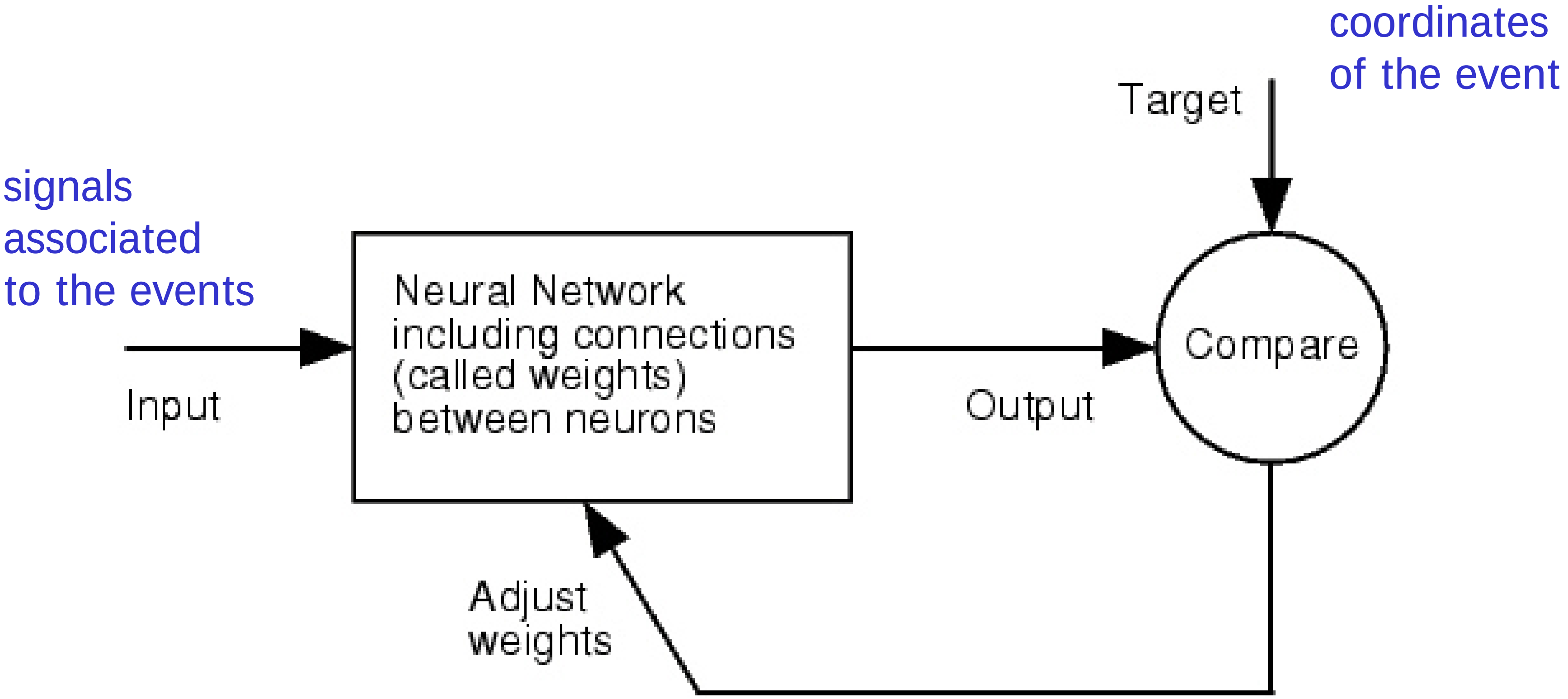
photodetector signals




coordinates of the interaction points

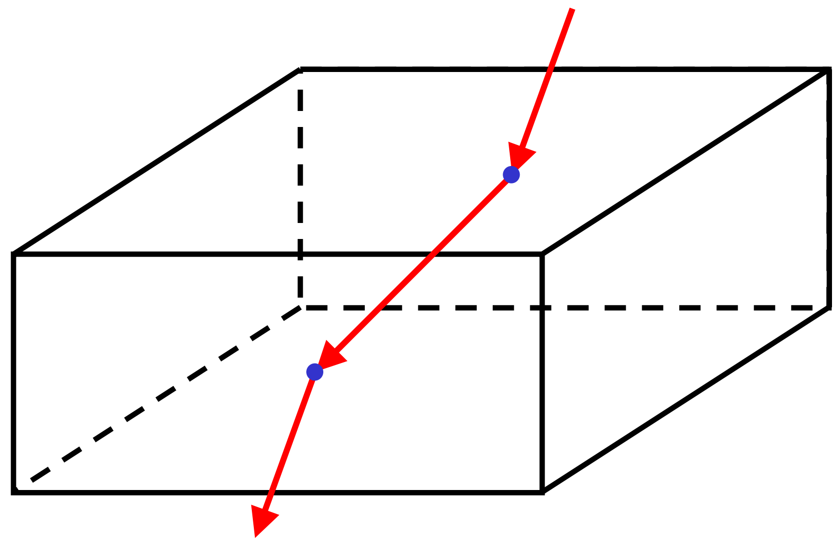
thanks gpu. same application but different method. stat => nice and conscious. nn. => not really, blackbox, and we need to train it, and at the beginning it's "empty".

# Learning of the Neural Network by means of “known” events

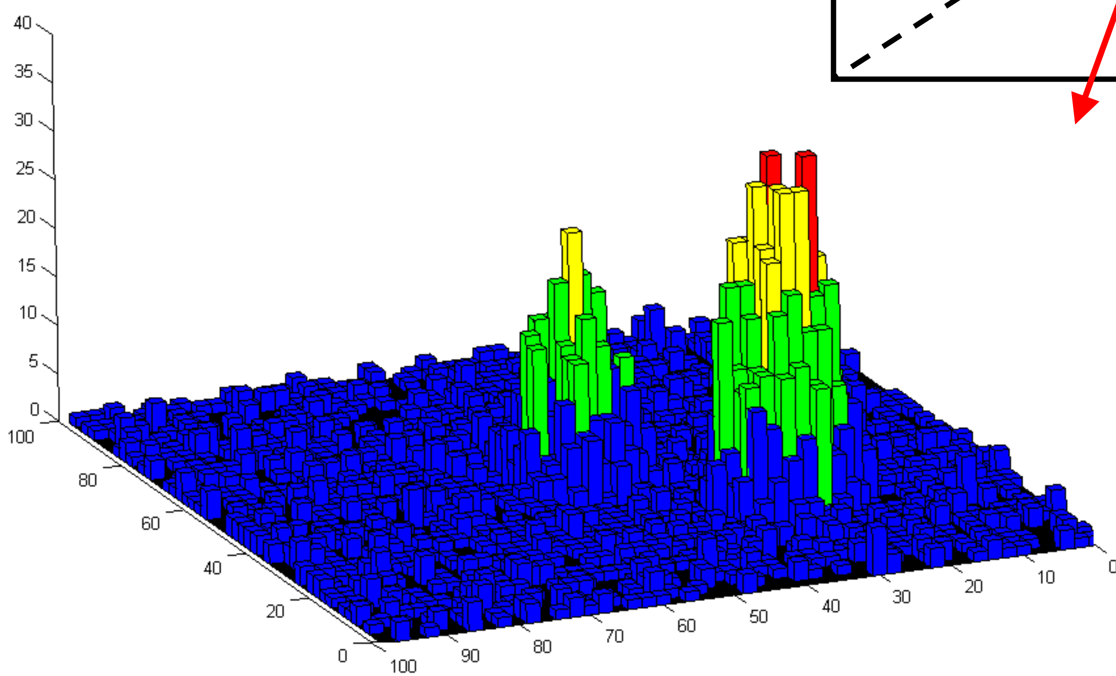



we apply at the model "test events",... training data.... target... error to adj. weights... trained... reliable output..
**Drawback of max likehol**: perform maximization of prob. for EACH gamma event! for each one we need to build p() and max it, computing costly :(. in **NN:** once it's trained, the network is fast and just need some "*" and sum, done quickly in fpga, in one clock you do like 500 (same for gpu)

# Possible application: reconstruction of Compton events

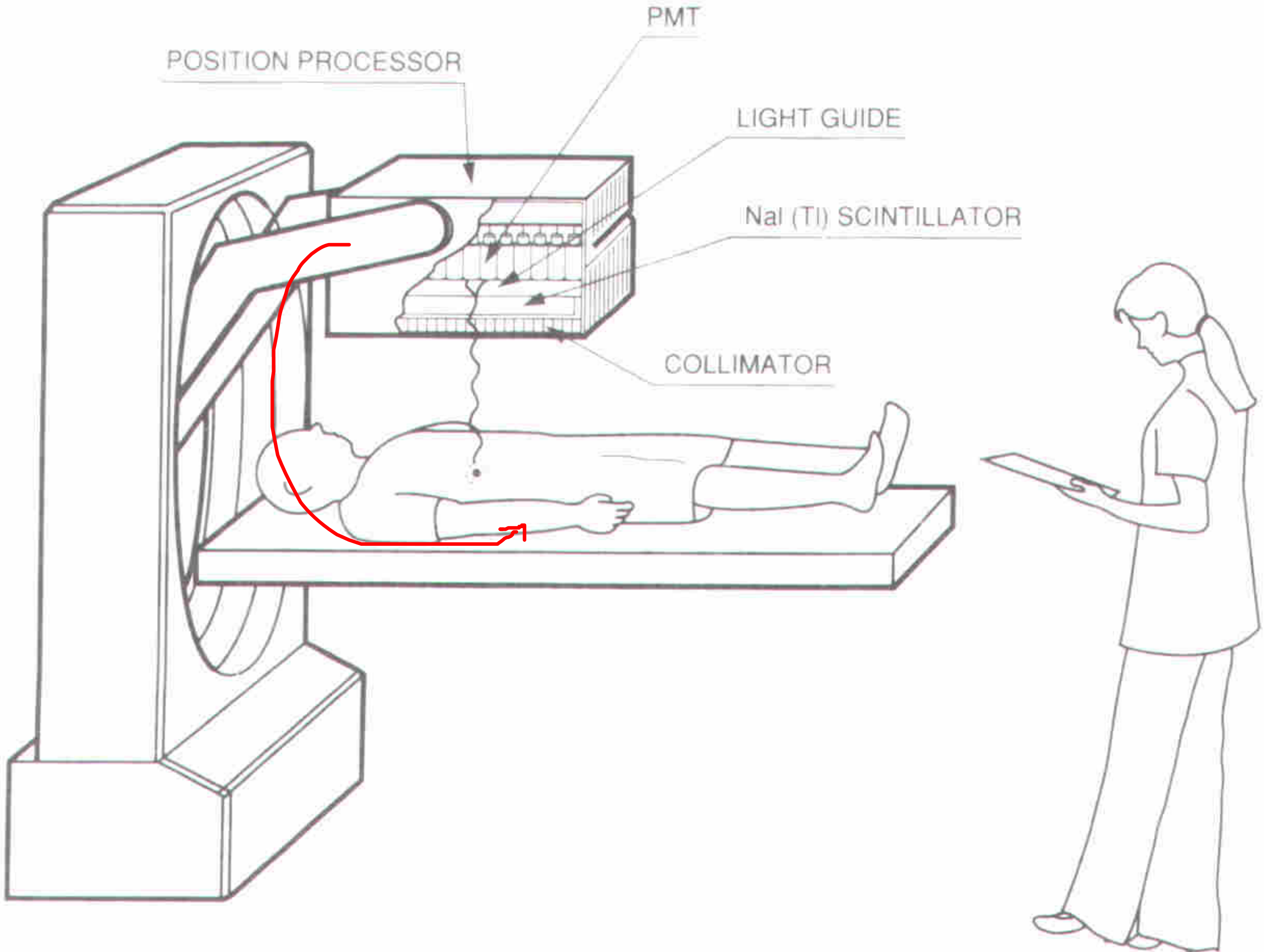


no



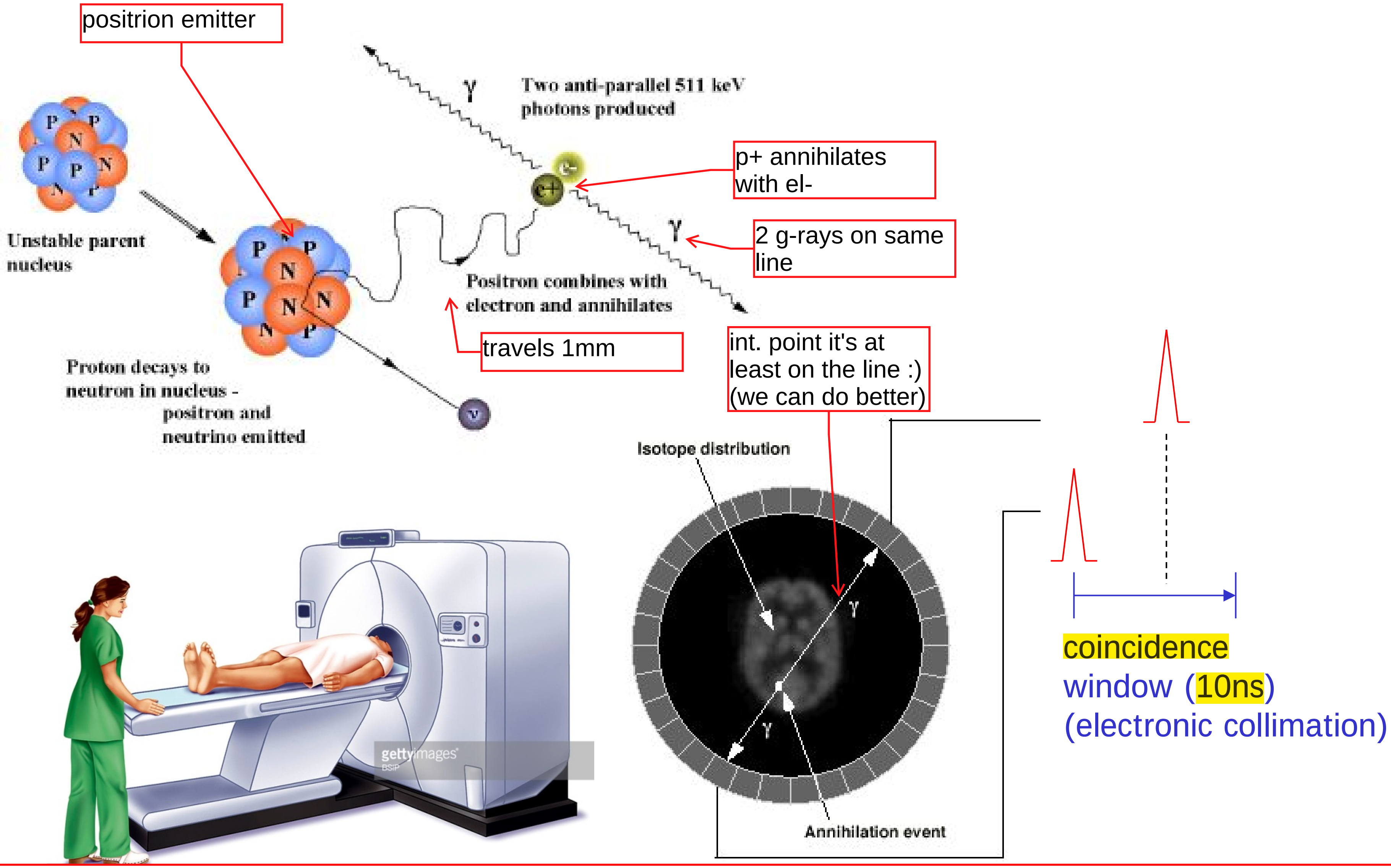
"... spect detector is anger camera"

# Anger Camera for SPECT




single cam => single proj.
if you turn it around you take many projection and can make tomography (spect)

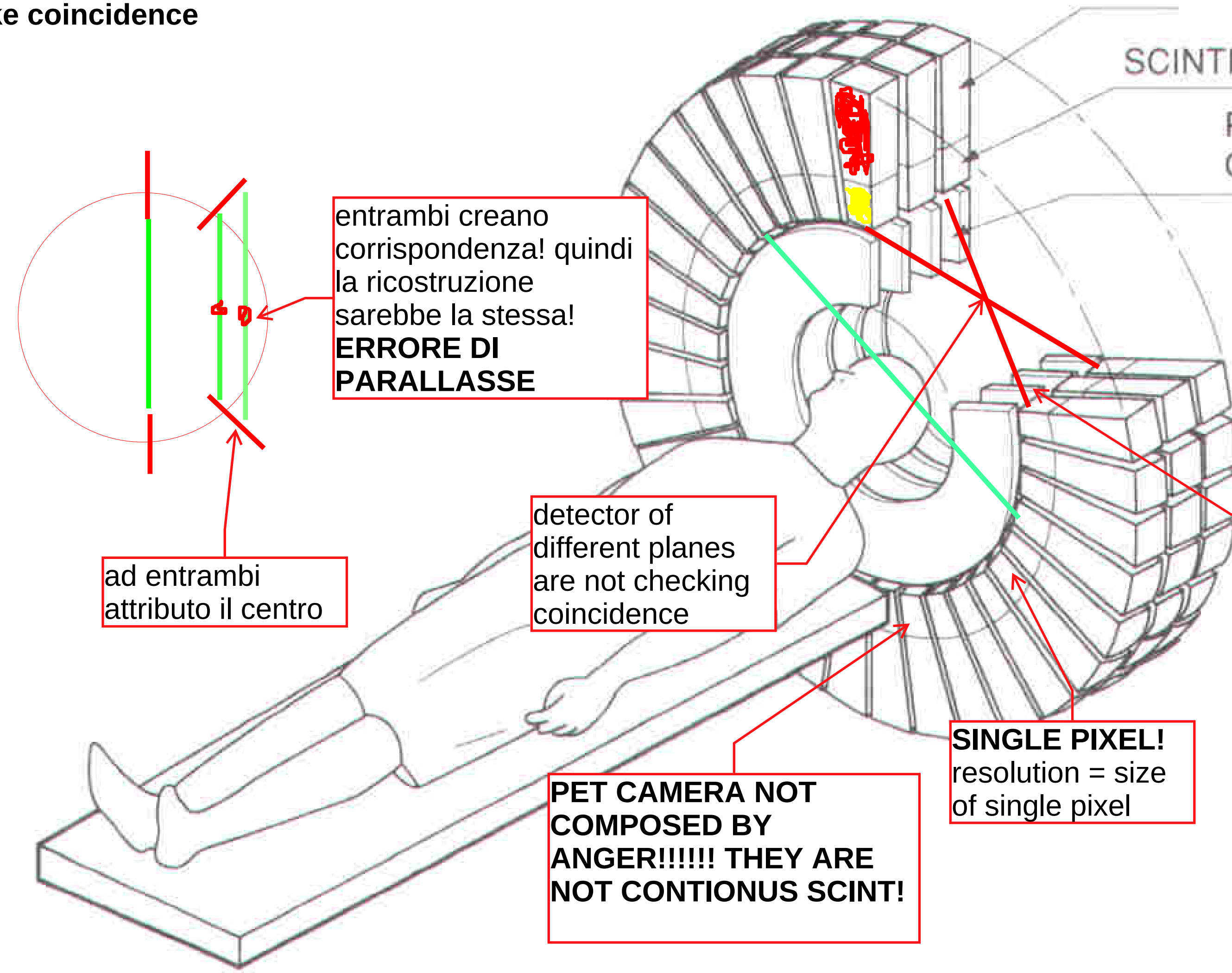
# Positron Emission Computed Tomography (PET)



# Gamma Camera for PET

detector configuration of a PET camera.
We dont do the rotation! We need to have **fully surround to take coincidence**

**<-- NON SONO PSPMT**




entrambi creano corrispondenza! quindi la ricostruzione sarebbe la stessa! **ERRORE DI PARALLASSE**

ad entrambi attributo il centro

detector of different planes are not checking coincidence

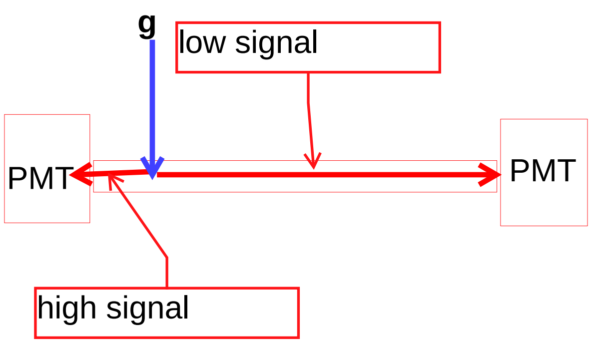
**PET CAMERA NOT COMPOSED BY ANGER!!!!!! THEY ARE NOT CONTIONUS SCINT!**

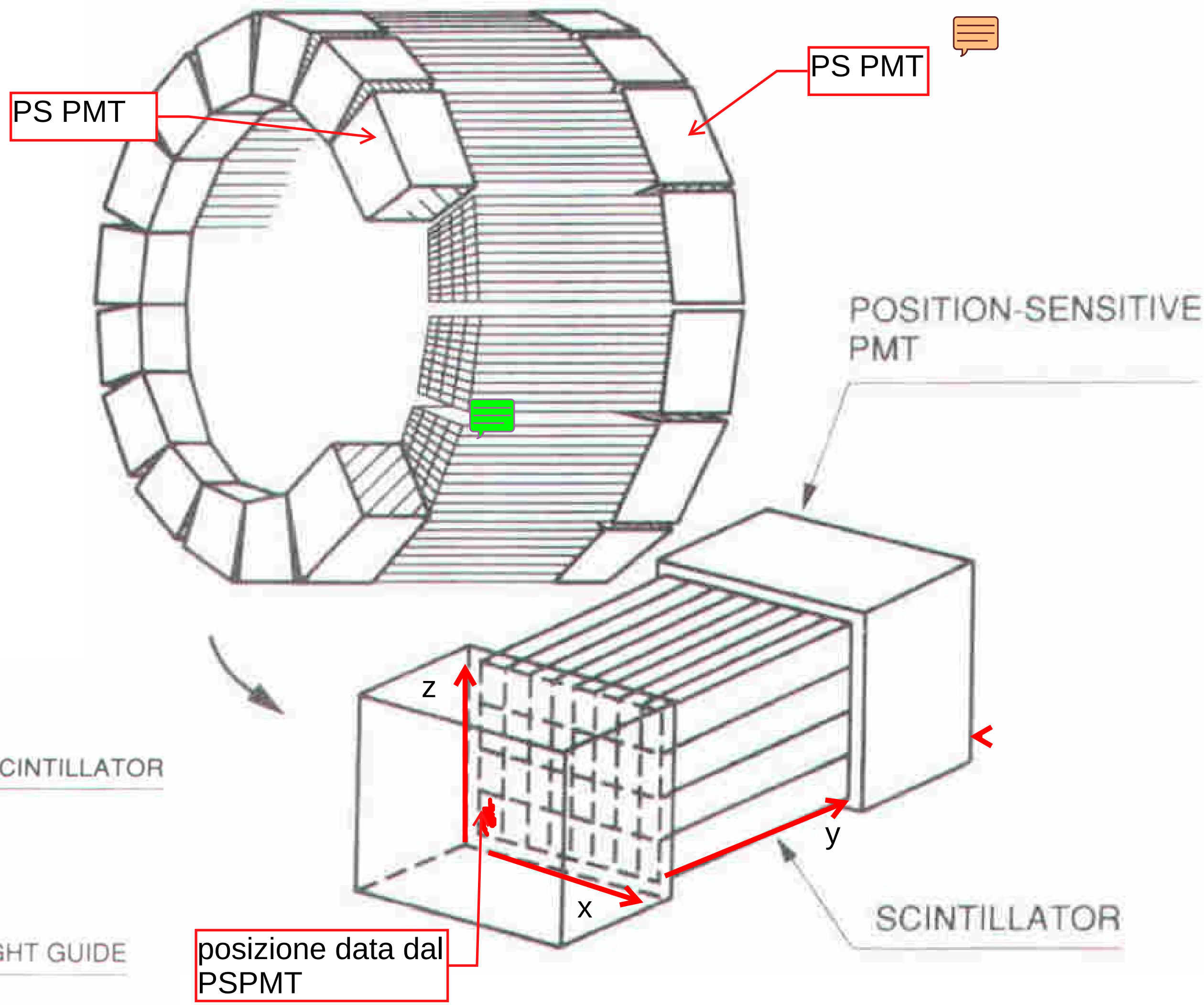
**SINGLE PIXEL!** resolution = size of single pixel

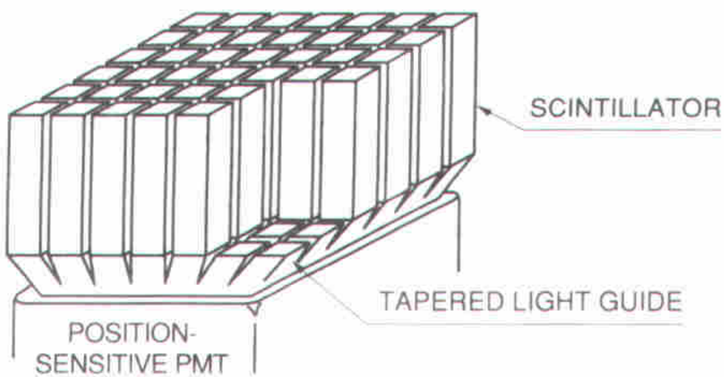
camera divided in **3 layers** of detector, each plane separed by a ring shaped collimator. This is a 3D problem! infatti avendo un paziente abbiamo una emissione in 3D! se abbiamo i rings (pet larga) potremmo avere coincidenze tra ogni p.d., ed è computazionalmente hard. Noi così introduciamo planar limitations, cioè le possibili coincidences solo per p.d. nello stesso slab. *prendiamo solo slice parallele.
**we reduce efficiency!**
adesso grazie alle gpu possiamo farlo senza i collimator e controlliamo ogni possibile coincidenza

Soluzione: DOI

# Alternative detection system for PET



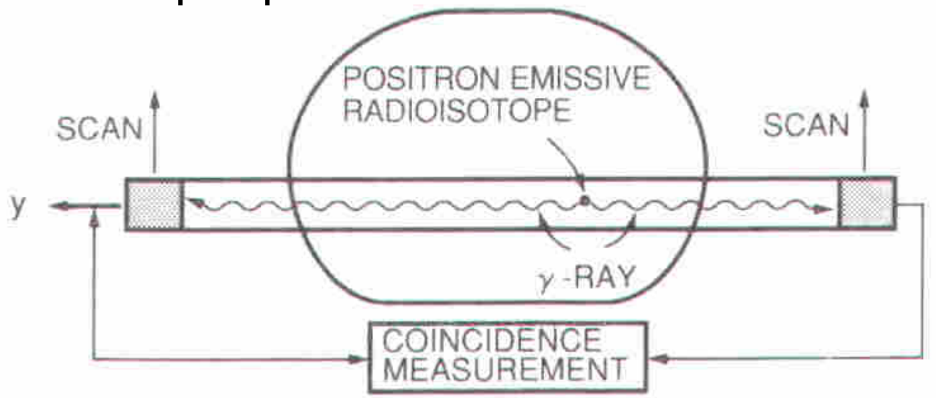

Change detector assembly. Scintillator are pixelated, but not in the RADIAL anymore!!! in the **LONGITUDINAL** direction with PSPMT to detect coordinates of the unit hit: in coordinata x and z. **La y la ricavo perché ho DUE!! PSPMT!!. Dato che lo scintillator è lungo, da un segnale ottico ai due PMT che è attenuato più è lunga la distanza (se è nel centro stesso segnale). Ratio signal => y position.**

STATE OF THE ART OF PET:

# PET-TOF (Time of Flight)

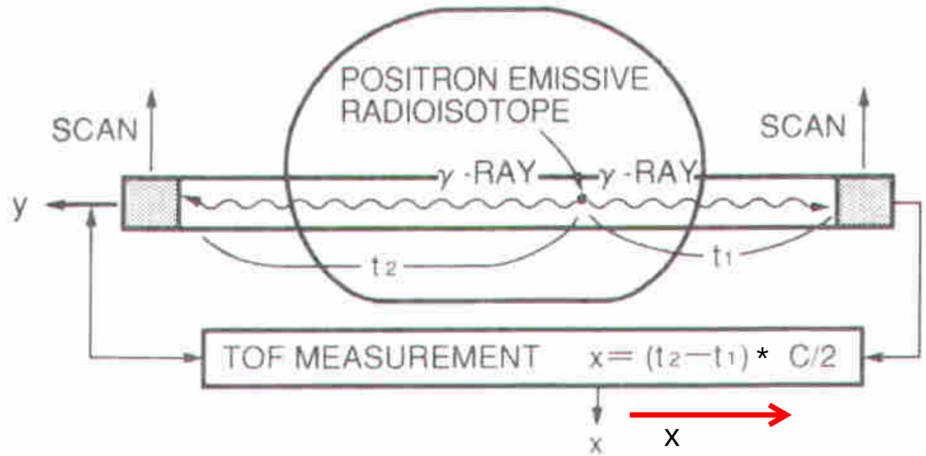
a) Positron

**Pet principle:**




(associate with tmax=10ns)

b) TOF: Time of Flight




a bit of PET history:

- PET: 1950
- PET-TOF: 1960
- PET-TOF first prototype: 1980
- PET-TOF clinical scanner: 2005

**Q: why 25 years to get a first useful instrument?** due to lack of suitable el systems

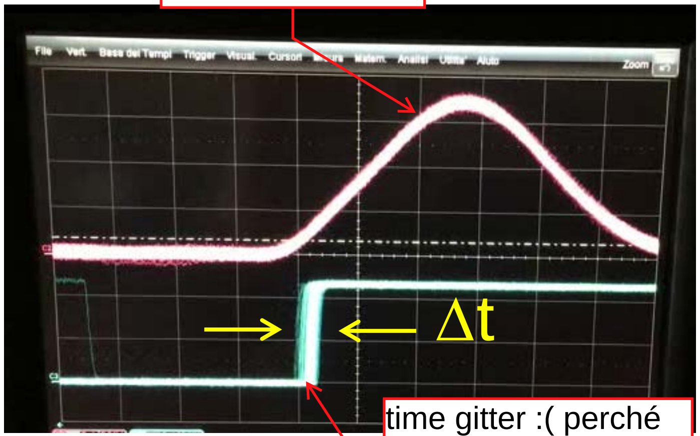
Prima: arrivano in questo slot, sono in questa linea. Dopo: possiamo **misurare la differenza di tempo** con la quale sono arrivati => identifichiamo la coordinata x dell'origine del g-ray lungo l'asse. [C/2, diviso 2 perché x0 è scelto al centro, infatti x0 appare se t1=t2]. The problem: enough timing resolution to measure this difference.

# Resolution of the measurement in PET-TOF

Risoluzione data dal time jitter... con la stessa formula:

$$\boxed{\Delta x = \Delta t \times c/2}$$

pulse to check where it passes a threshold




time gitter :( perché ris. temporale limitata [= time res.]

$$c = 30\text{cm/ns}$$

$$\text{if } \Delta t \sim 3\text{ns}$$

$$\Rightarrow \quad \Delta x \sim 45\text{cm}$$ lol, inutile

..the uncertainty in the position of annihilation is of the same order of the diameter of the body of the patient....!

... few years ago ... thanks to electronics development:

$$\boxed{\Delta x = \Delta t \times c/2}$$

$c = 30\text{cm/ns}$

"today"

if $\Delta t \sim$ 300ps (now state of the art)

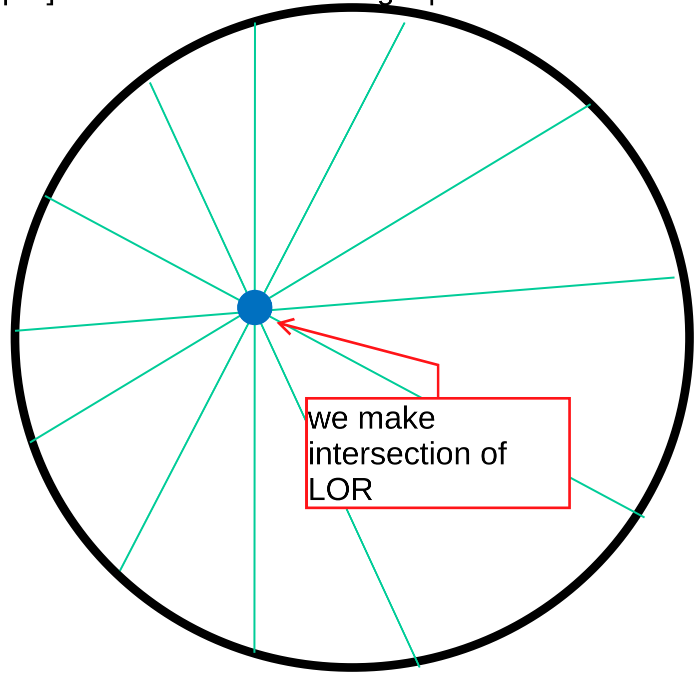
$\Rightarrow \quad \Delta x \sim$ 4.5cm Coordinate of interaction => spatial resolution of 4.5cm

- recent improvements in electronics timing resolution have pushed an 'old' idea of PET TOF to reality….**but**
- resolution still not adequate for the reconstruction of the annihilation point with a resolution required for PET (~mm)

Still low :( but we still use it because..

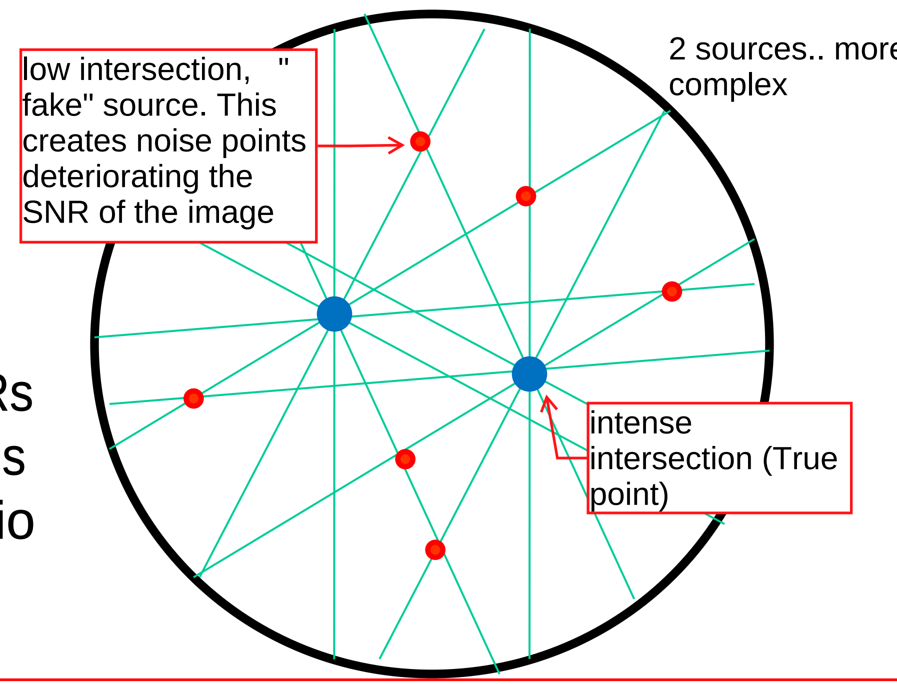
**we dont use 4.5cm to imrpove res. ... but to improve SNR in the image!!**

[Traditional pet] reconstruction of a single point source:




single source determined by the intersection of the LORs (lines of range)

• spurious intersections (no sources)

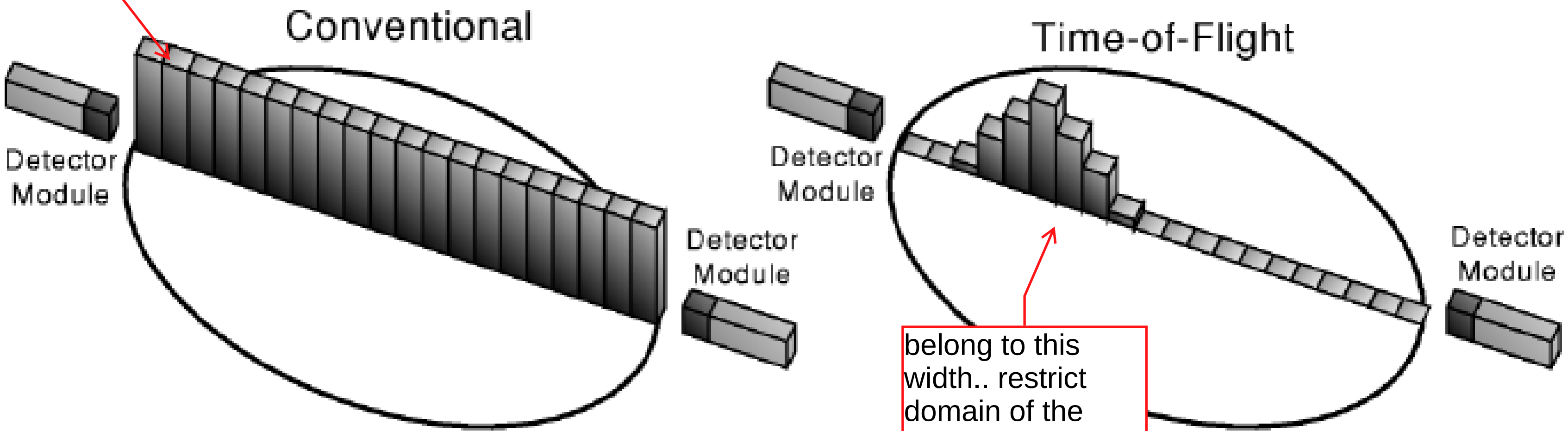



the image of more than one sources reconstructed with the criterion of LORs intersection only is affected by spurious intersections which reduce the S/N ratio of the reconstructed image

... quindi il TOF PET può aiutare.. see next slide

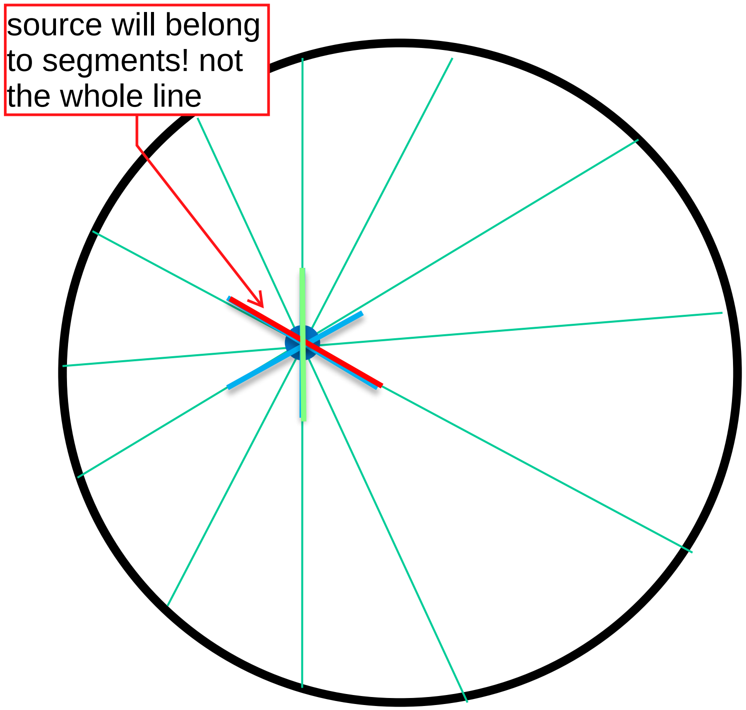
# Benefit to S/N provided by TOF

conv: source belong to any point.

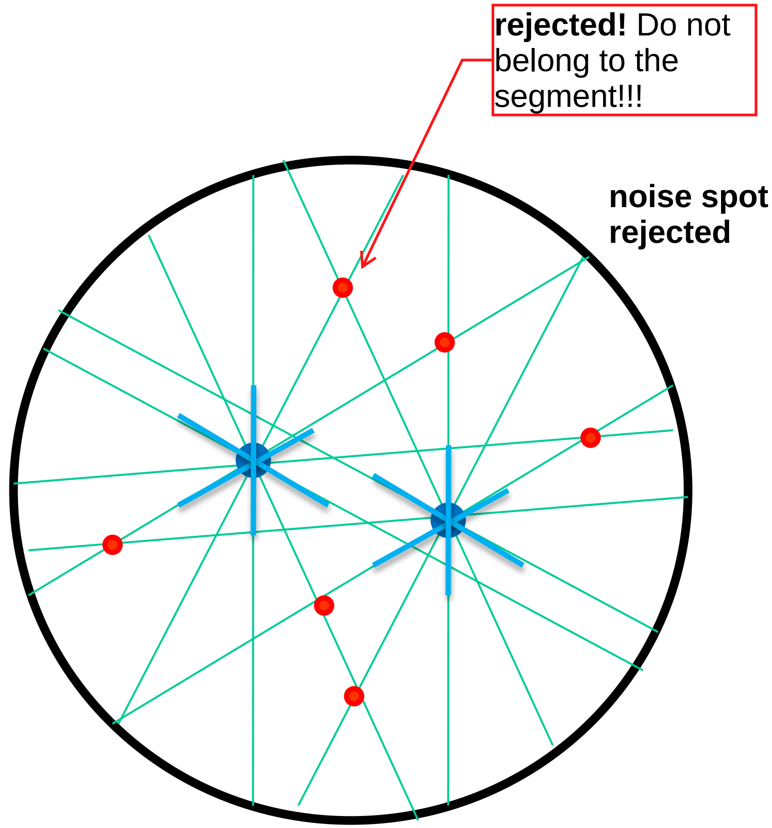



belong to this width.. restrict domain of the source to this

Fig. 2. TOF reconstruction. With conventional reconstruction (shown on the left), all pixels along the chord are incremented by the same amount. With TOF reconstruction (shown on the right), each pixel on the chord is incremented by the probability (as determined by the TOF measurement) that the source is located at that pixel.



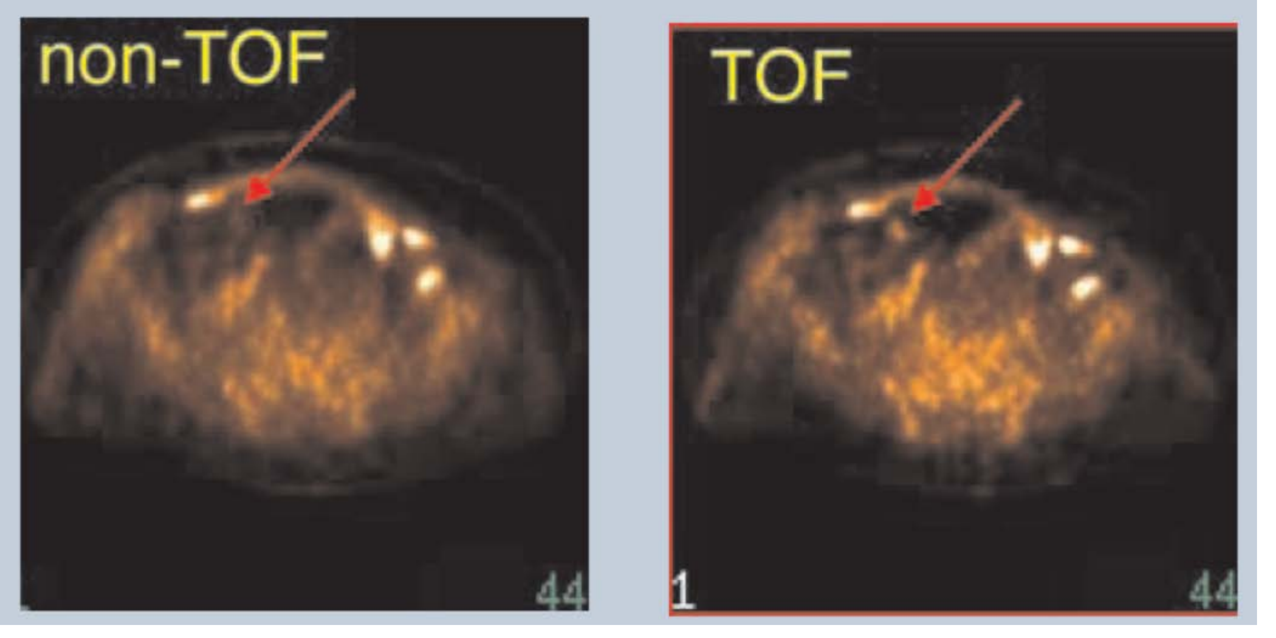

single source identified by intersection of segments with width determined by the spatial resolution of TOF




the spurious intersections are reduced to only intersections between segments (lower in number with respect to previous LORs intersections)
$\Rightarrow$ improvement on S/N
of the reconstructed image

Use of TOF infos. are not use to imrpove the location, but to CLEAN the image by crossing of "unfortunate" LOR audio

- **Variance Reduction Given by $D/\Delta x = 2D/c\Delta t$**
**(D: diameter of the object)**
- **500 ps Timing Resolution $\Rightarrow$ 8 cm localization**
**$\Rightarrow$ 5x Reduction in Variance!**




We have the imrpovement depending on dx. If the segment is smaller than patient diameter we improve! (by that factor). x5 improvement in SNR. We can **improve image quality**, or we can **LOWER THE DOSE** (by a factor of 5) to get the same image quality.