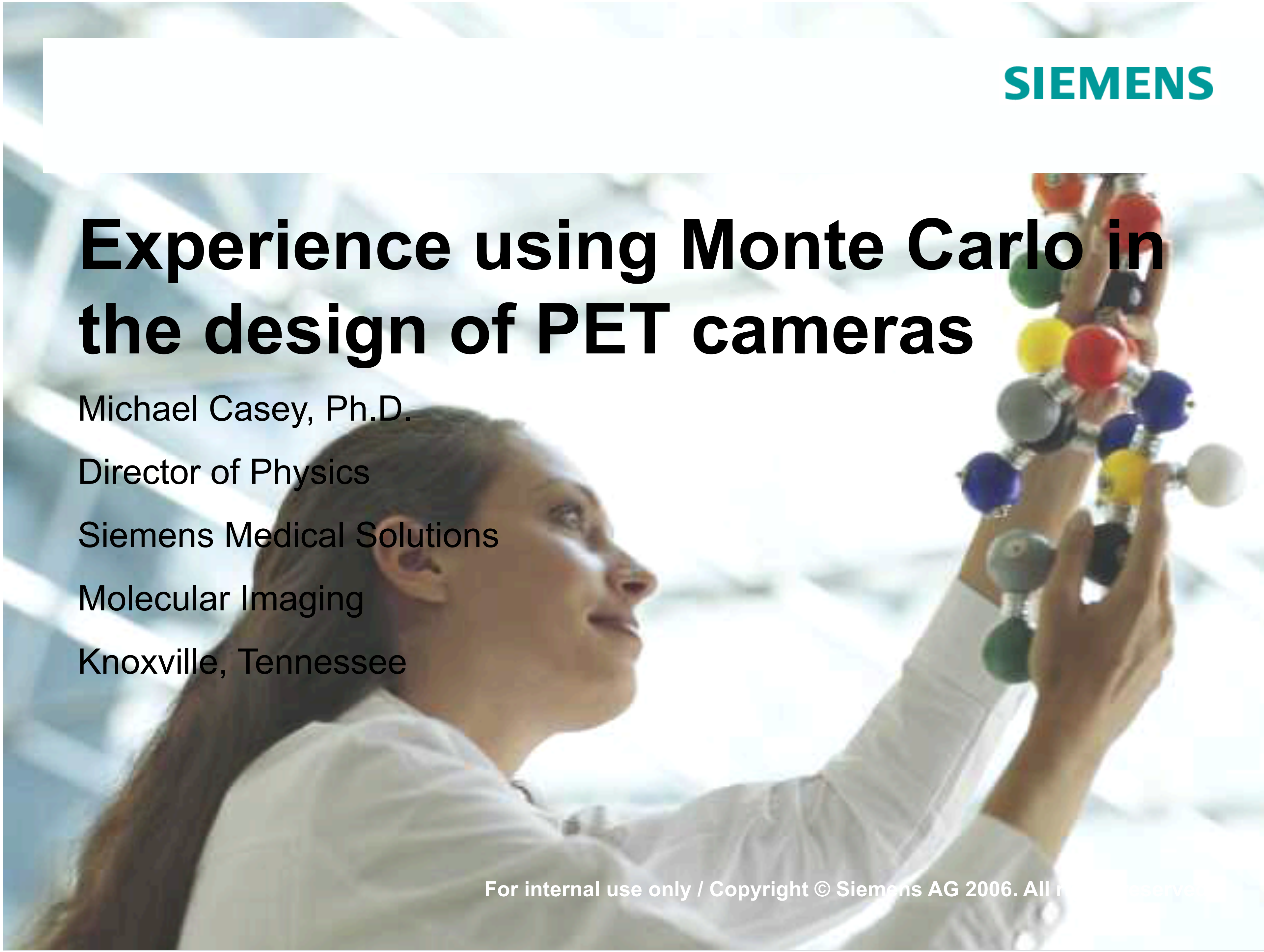


Experience using Monte Carlo in the design of PET cameras

Michael Casey, Ph.D.

Director of Physics

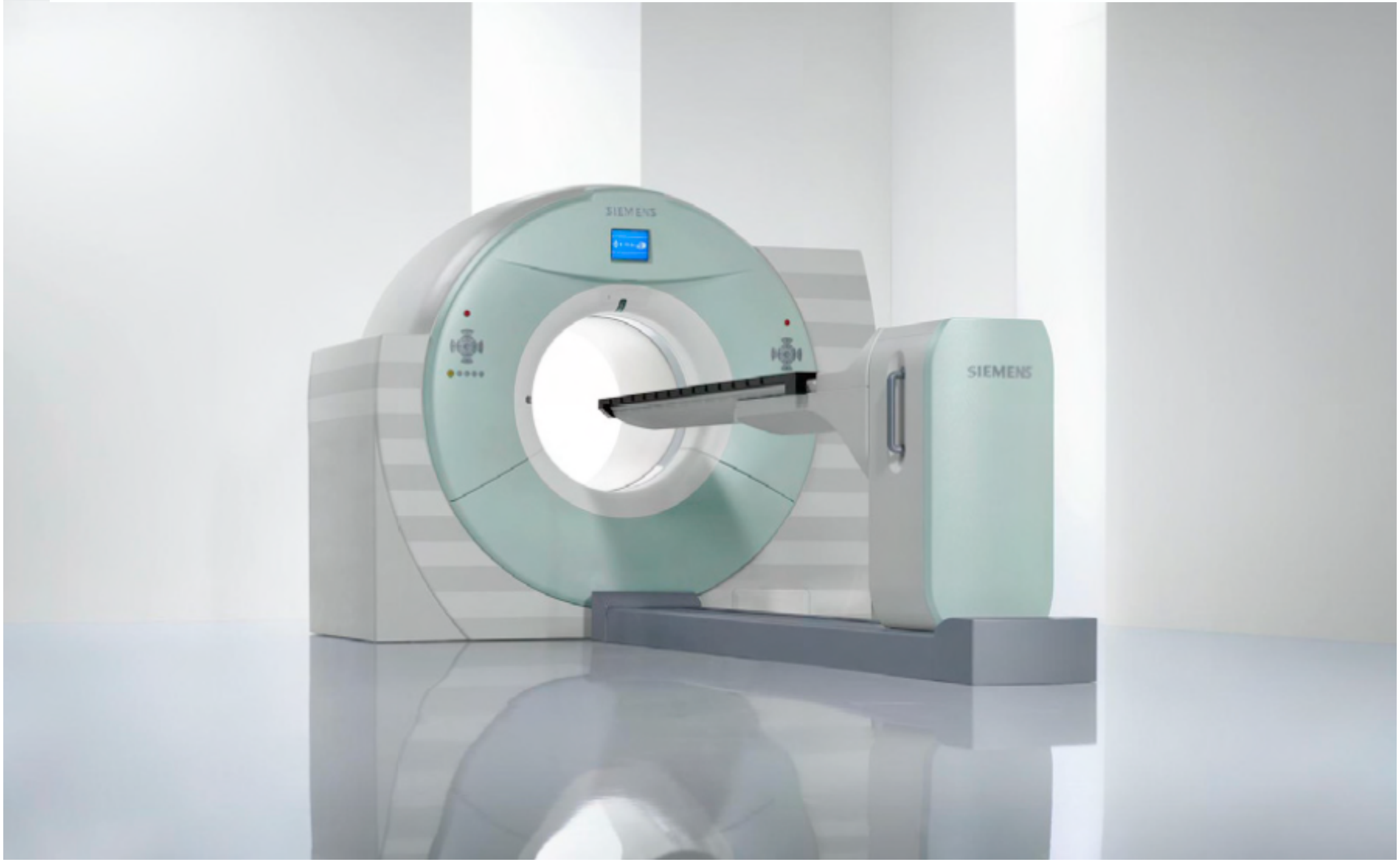
Siemens Medical Solutions

Molecular Imaging

Knoxville, Tennessee

SIEMENS

Biograph mCT new CT and time-of-flight PET



Outline

How is Monte Carlo used for PET camera design?

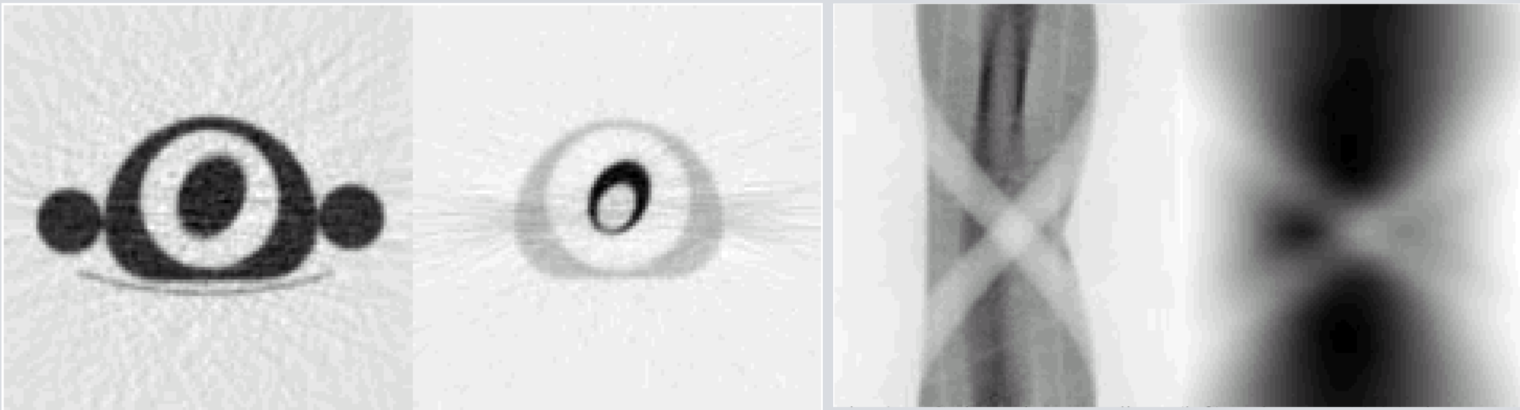
- Scatter Correction
- Time-of-flight Scatter Correction
- Prompt Gamma Correction
- Time Resolution
- PSF and Time-of-flight Reconstruction

Simulation based Scatter Correction

Rays are traced through the attenuating medium (transmission image) starting with the isotope distribution (emission image).

A scatter within each voxel of the image is predicted by the Klein-Nishina formula

The resulting scatter sinogram can be subtracted from the emission sinogram



μ Image | Emission Image | Emission Sinogram | Scatter Sinogram

J. M. Ollinger and G. C. Johns, “Model-Based Scatter Correction for Fully 3D PET,” *1993 IEEE Med. Img. Conf. Rec.*

C. C. Watson, et.al, “Evaluation of siniulation-based scatter correction for 3-D PET cardiac imaging,” IEEE Trans. Nuc. Sci., vol. 44, pp. 90-97, Feb.1997.

Simulation based Scatter Correction

In principal the process works except:

- The starting emission image contains both scattered and true coincidences.
- Larger patients have a larger ratio of scatter to trues than smaller patients.

Solution:

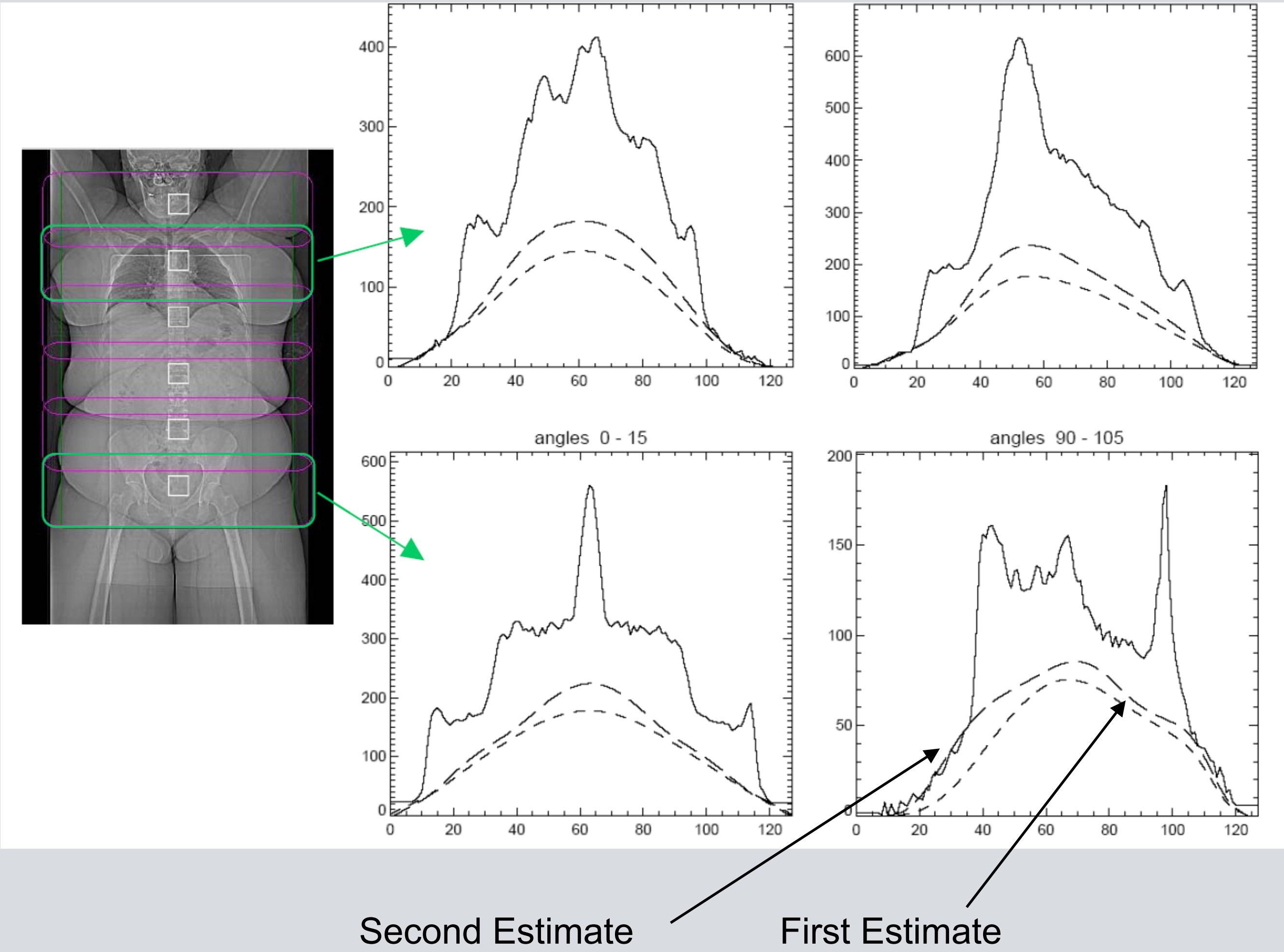
- The solution is an iterative approach where the second estimate of the emission image has been corrected by the scatter from the first estimate.

Simulation based Scatter Correction

135 kg female

Scatter fraction 52%-54%

Two iterations



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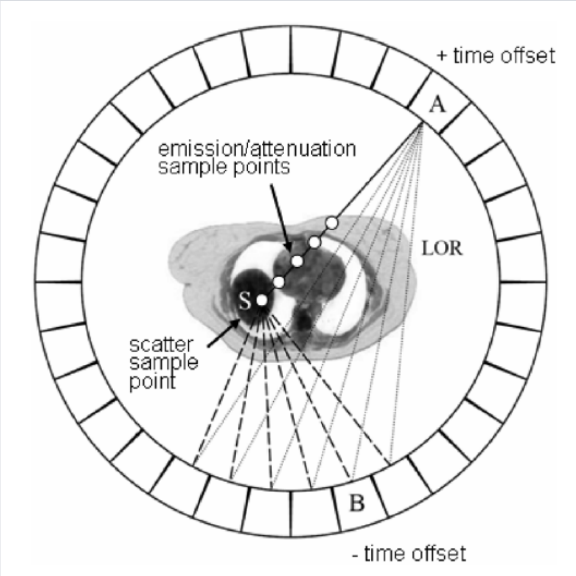
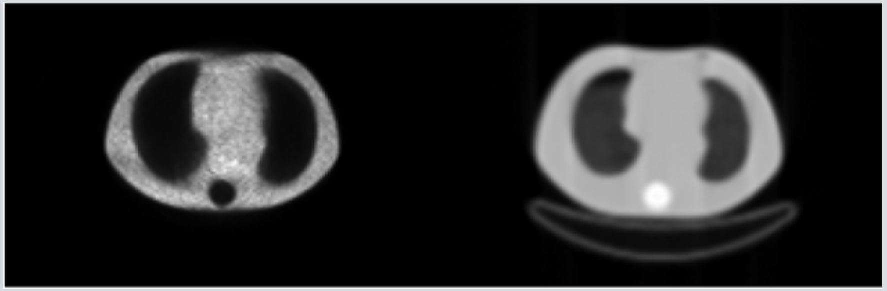
Time-of-flight Scatter Correction

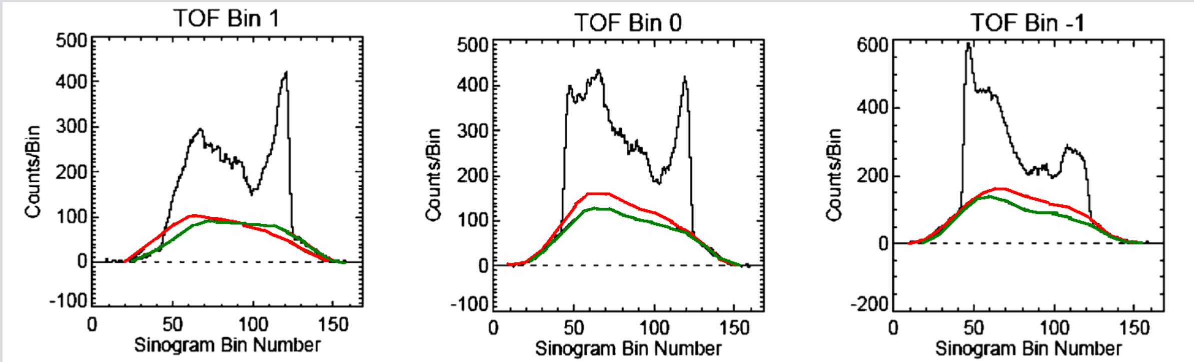


The time difference depends on both the path length and the angle of the scatter.



For a the torso phantom, the non-TOF scatter (RED) does not match the sinogram tails. The TOF scatter (Green) matches.

Time resolution was 550 ps or 8.25 cm



Watson, "Extension of Single Scatter Simulation to Scatter Correction of Time-of-Flight PET" *IEEE Trans Nucl Sci*, VOL. 54, NO. 5

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Prompt Gamma Correction

Several interesting isotopes have decay schemes that include prompt gammas as part of the decay scheme.

Isotope	Halflife	e^+ Abundance	γ Energy	Abundance	Use
^{82}Rb	76.4 sec	95.4%	776	13%	Mycardocal Perfusion
^{124}I	4.2 day	23.0%	603	63%	Iodine Chemistry (Amaloyd Imaging, Renal Imaging)
^{22}Na	2.6 year	90.5%	1274	99.9%	Phantoms

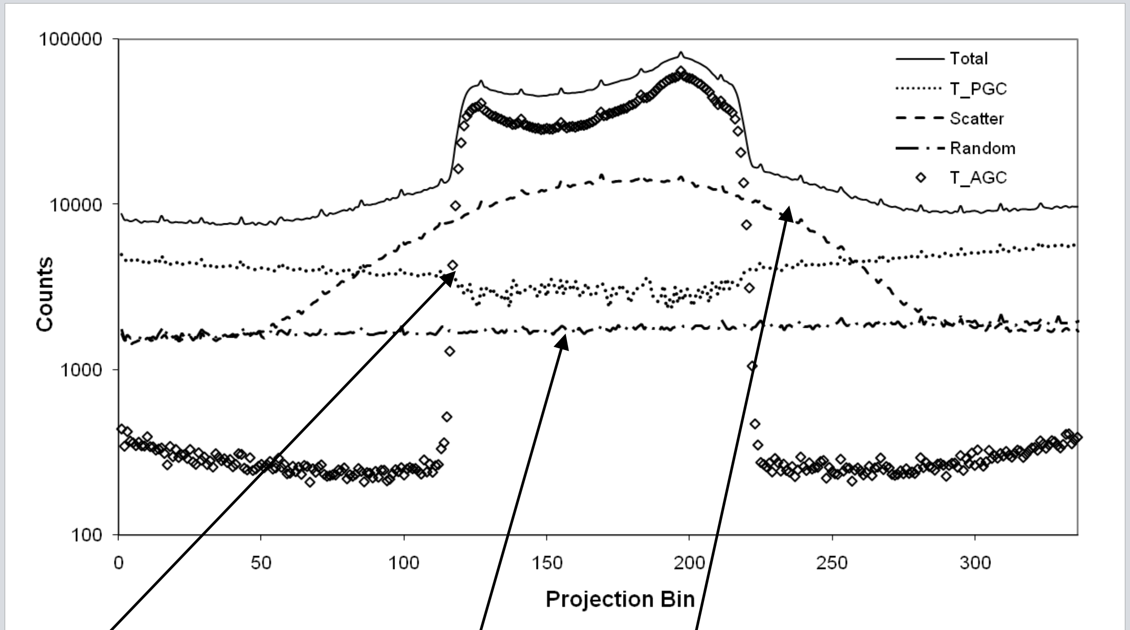
The prompt gamma cannot be discriminated based on time. Down scatter can cause the photon to appear within the energy window. The result is a low frequency bias much like randoms.

Siemens' correction follows the idea of Beattie that the prompt gamma background is similar to the randoms background. A simulation of the randoms background is used as a basis function for the prompt gamma background

Beattie, B. J., R. D. Finn, et al. (2003). Quantitative imaging of bromine-76 and yttrium-86 with PET: A method for the removal of spurious activity introduced by cascade gamma rays, AAPM. **30:** 2410-2423.

Simulation to validate correction

SIEMENS



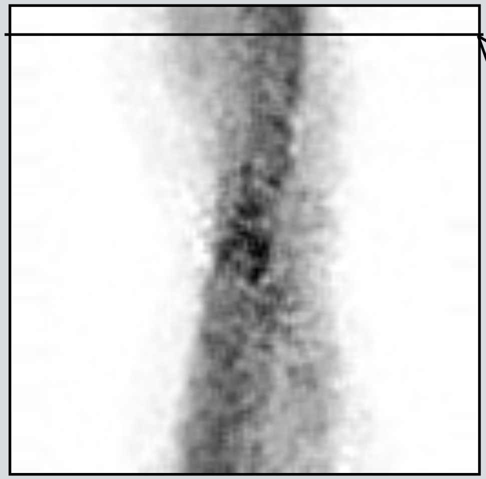
Simulation using GATE 3.0.0 modeled the contrast phantom shown. Isotope simulated was ^{124}I.

Prompt gamma background

Randoms background

Scatter

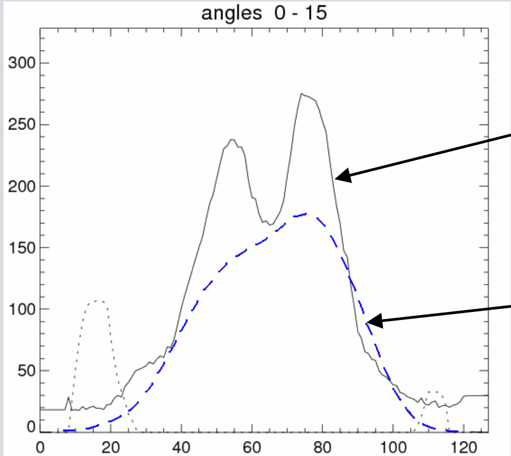
Prompt Gamma Correction ^{82}Rb



Sinogram

- The prompt gamma adds a background under the data.
- Without a correction, the scatter will be over scaled to compensate.
- Over scaling can increase contrast in some images and reduce quantitative accuracy.

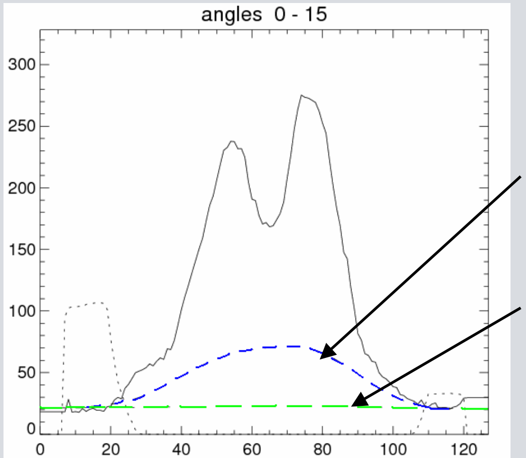
No prompt gamma correction



Solid line - profile through sinogram

Blue line – scaled simulated scatter

Prompt gamma correction



Blue line – scaled simulated scatter

Green dashes – simulated prompt gamma background

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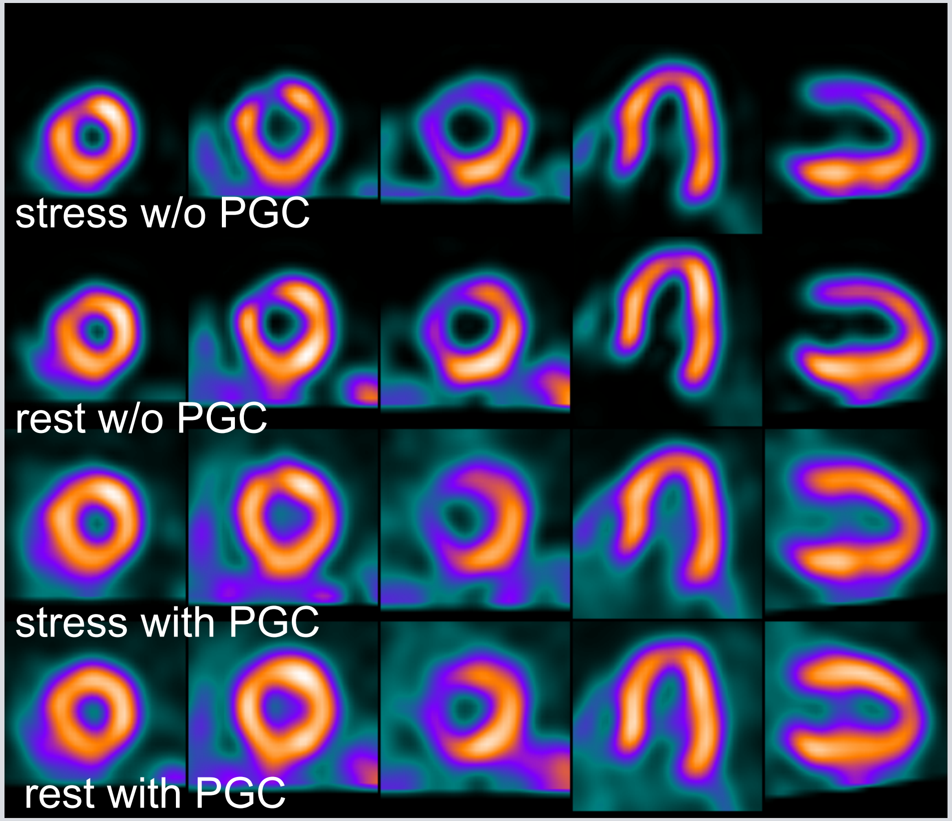
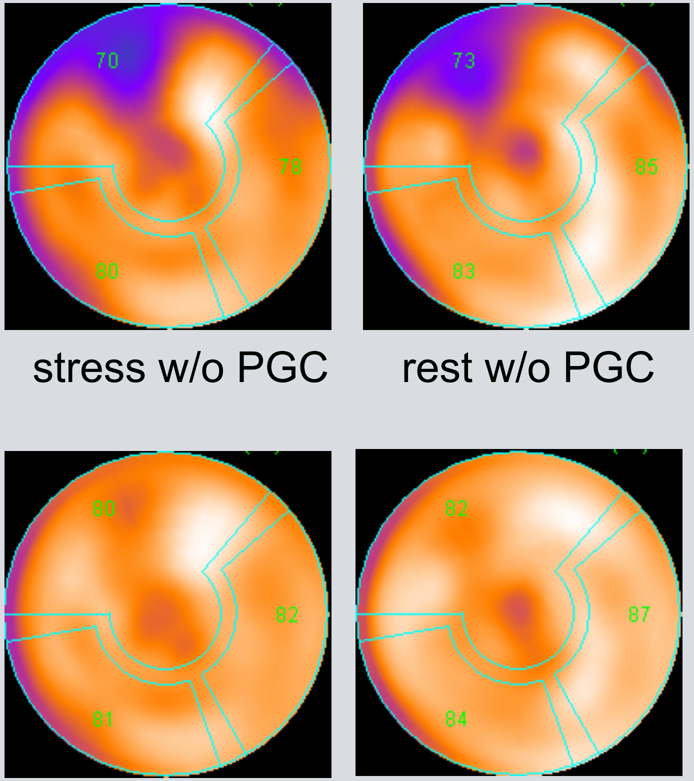
Prompt Gamma Correction ^{82}Rb



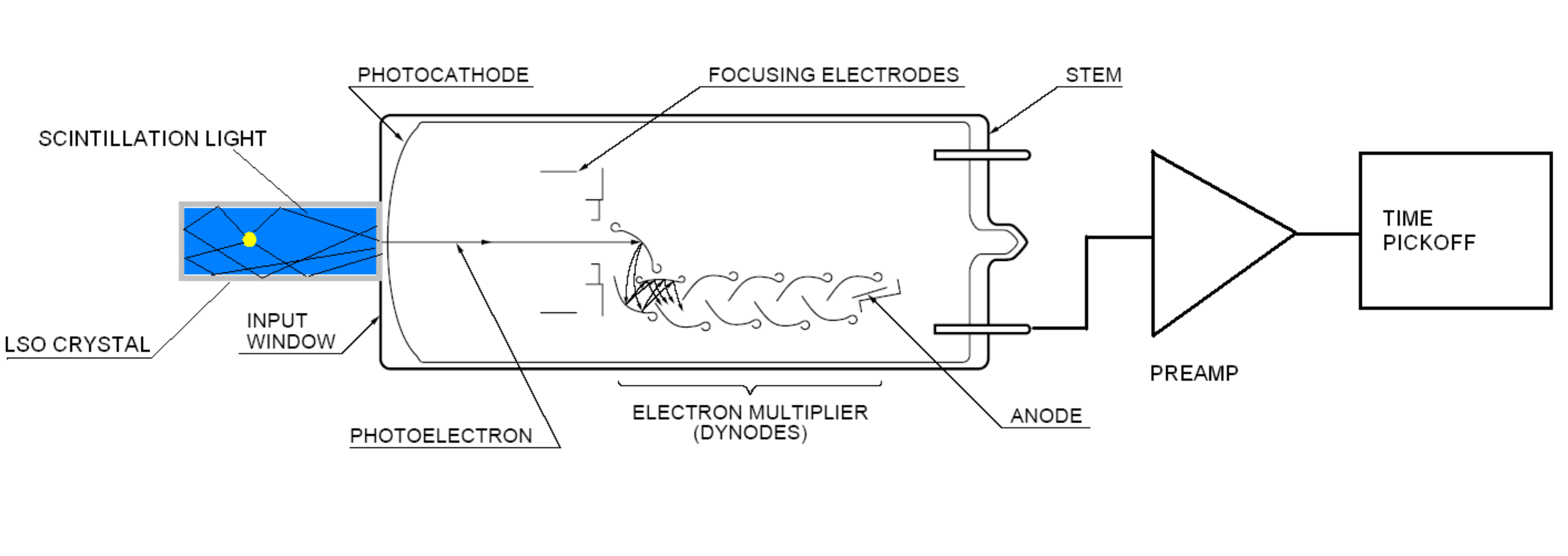


stress w/o PGC rest w/o PGC

stress with PGC rest with PGC

B . Hsu , M . Casey , T . Bateman , J . Case “Validation of prompt gamma correction for 3D Rb-82 myocardial perfusion PET/CT imaging” *Journal of Nuclear Cardiology* , Volume 15 , Issue 4 , Pages S4 - S4

Time Resolution Simulation
Data Flow of Signal



Time Resolution Simulation
Data Flow of Signal

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Scintillation Light: Random deviates from an exponential distribution

$$\lambda(t) = \lambda_0 e^{-\frac{t}{\tau_{LSO}}}$$

$$t(x+1) = t(x) + \frac{p(x)}{\lambda(t)}$$

→ **Light trapping crystal:** Random deviates from gamma distribution.

$$p(x) = \frac{x^{\alpha-1}}{\beta^{\alpha} \cdot \Gamma(\alpha)} \cdot e^{-\frac{x}{\beta}}$$

$$t(x) = t(x) + p(x)$$

→ **Single photoelectron transit time spread:** Random deviates from gamma distribution

$$p(x) = \frac{x^{\alpha-1}}{\beta^{\alpha} \cdot \Gamma(\alpha)} \cdot e^{-\frac{x}{\beta}}$$

$$t(x) = t(x) + p(x)$$

→ **Single photoelectron pulse height:** Modify intensity of photoelectrons with gaussian distribution

→ **Frequency response:** convolve impulse stream with impulse response

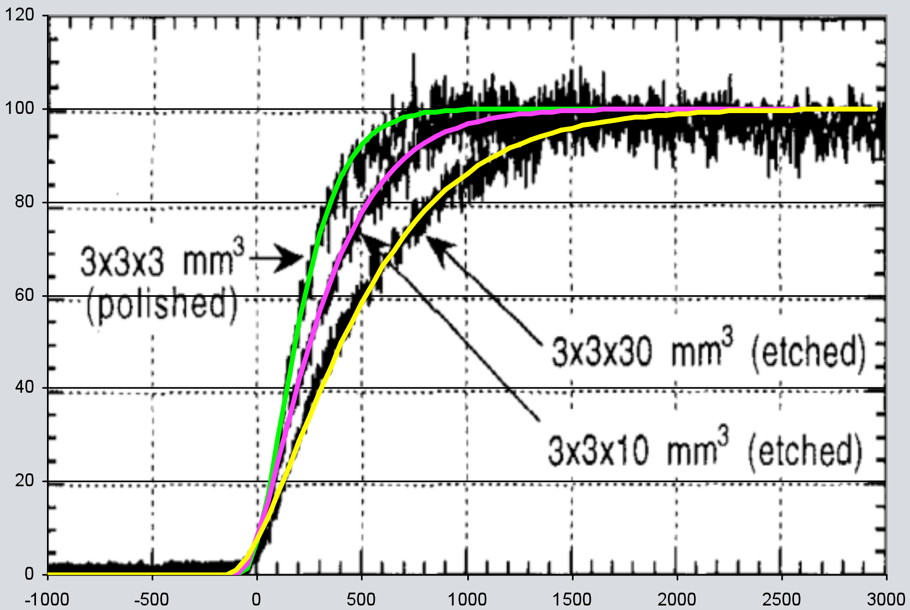
→ **Zero cross detect**

Light Trapping

- The gamma distribution represents the random delay of light from the crystal.
- Different size crystals can be simulated by using the parameters in the table.

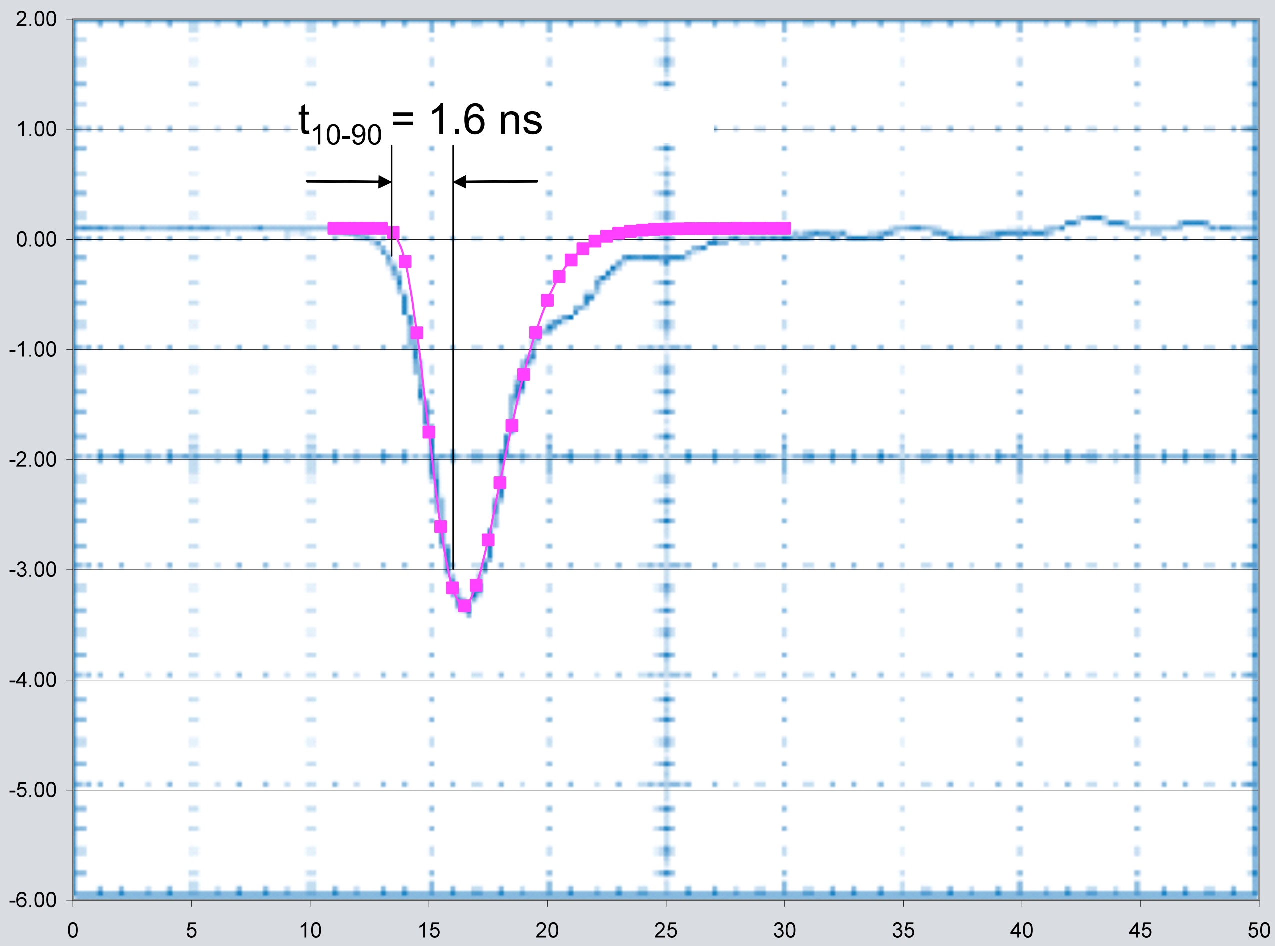
$$p(x) = \frac{x^{\alpha-1}}{\beta^{\alpha} \cdot \Gamma(\alpha)} \cdot e^{-\frac{x}{\beta}}$$

Crystal	α	β
3x3x3	**3**	**105**
3x3x10	2	210
3x3x30	2	330

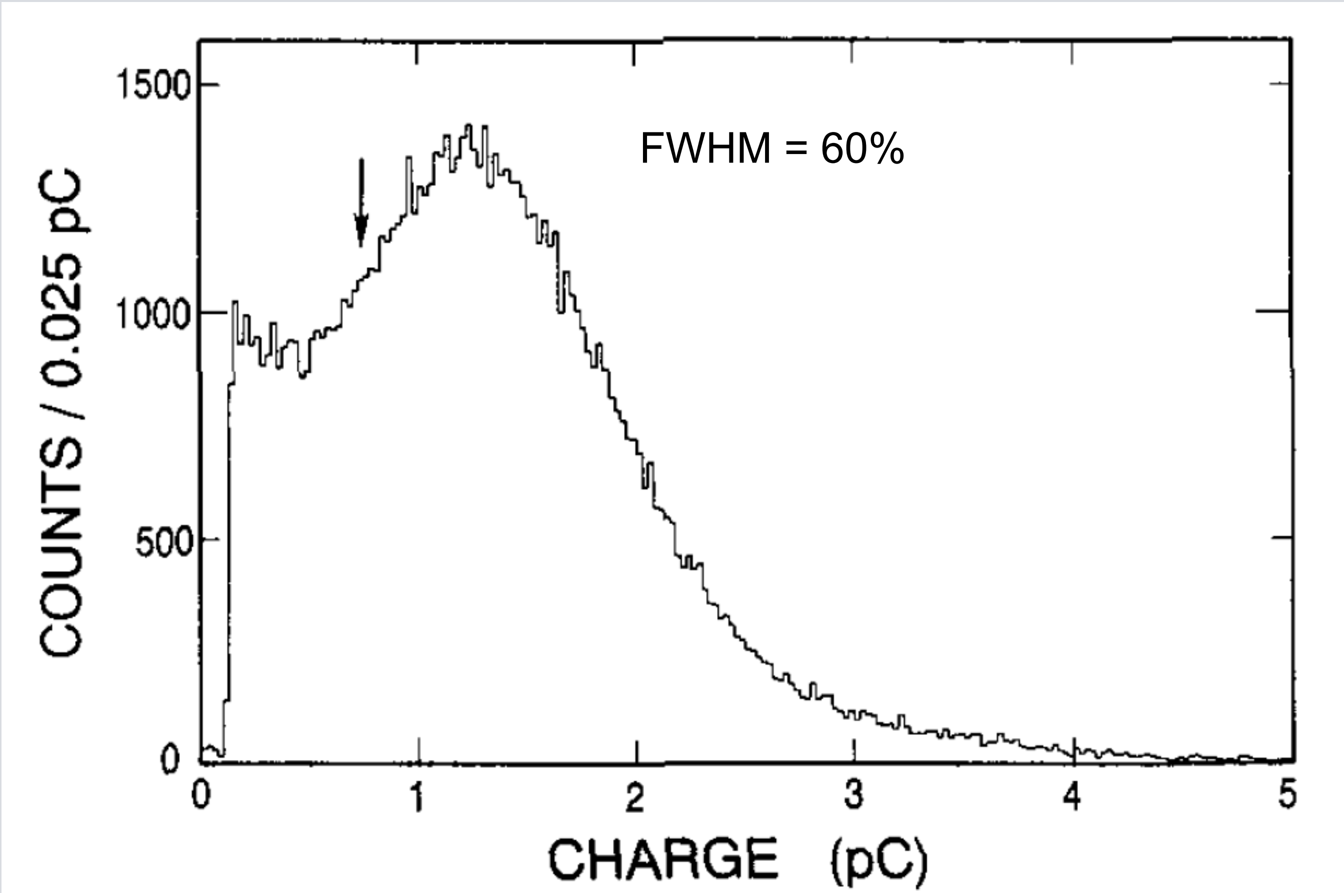


Moses, W., et. al. "Prospects for Time-of-Flight PET using LSO Scintillator" *IEEE Trans Nucl Sci*; VOL. 46, NO. 3, JUNE 1999

XP2020 PMT Response



XP2020 Single Photoelectron Pulse Height



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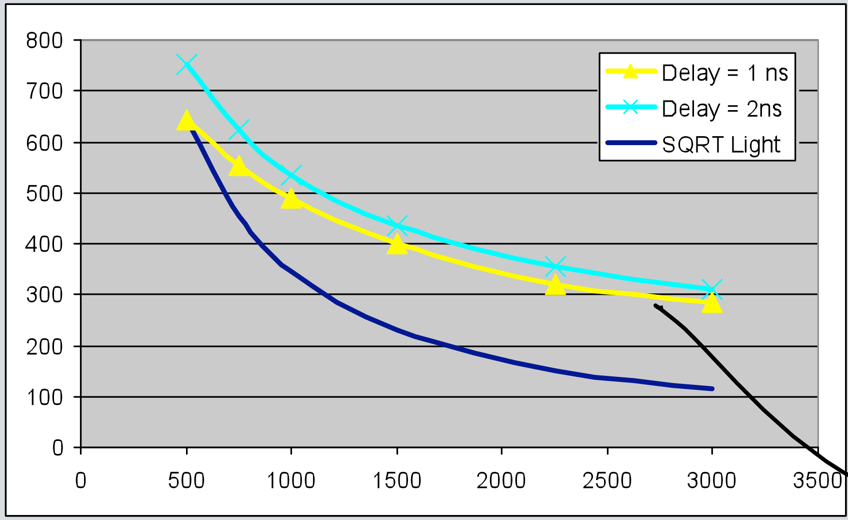
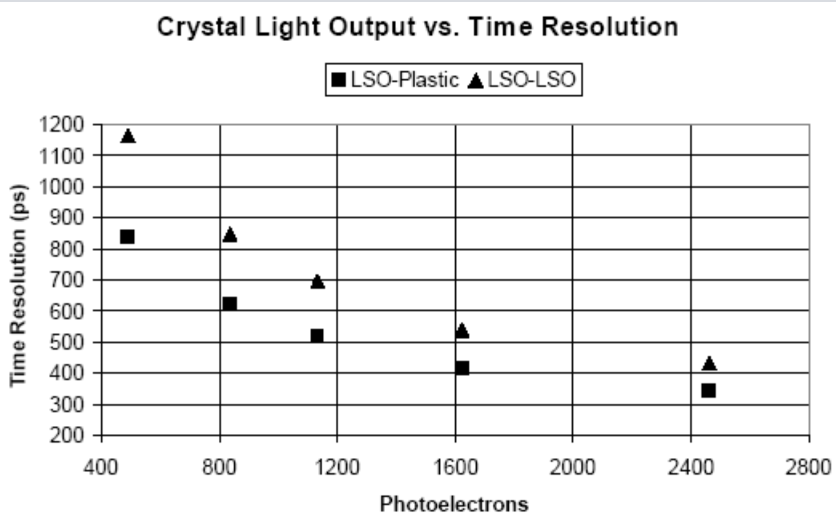
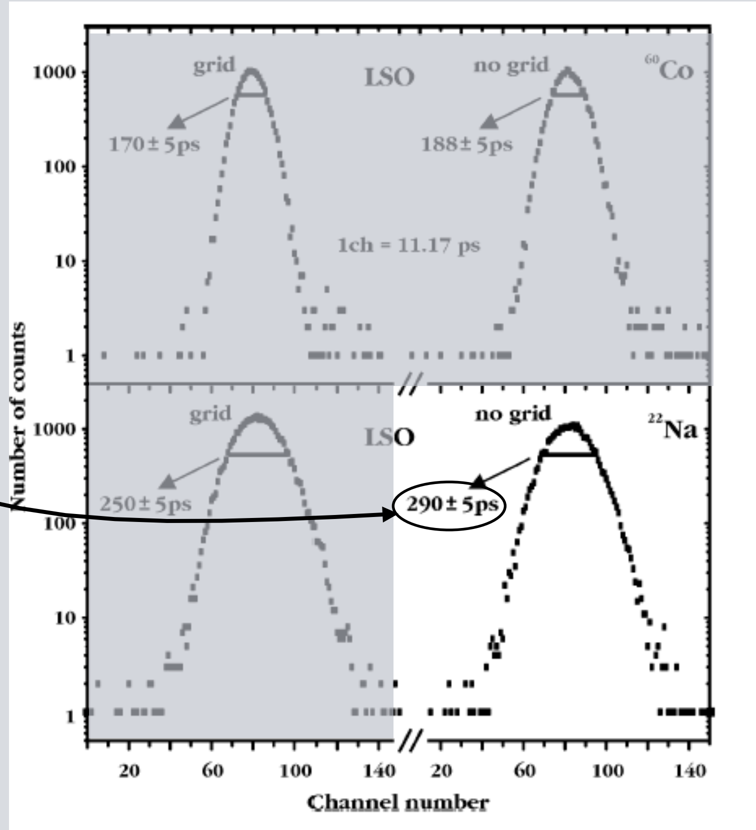
Simulated Time Resolution







Moszynski, M. et. al. "New Fast Photomultipliers With a Screening Grid at the Anode" *IEEE TRAN NUCL SCI*, VOL. 51, NO. 4, AUGUST 2004

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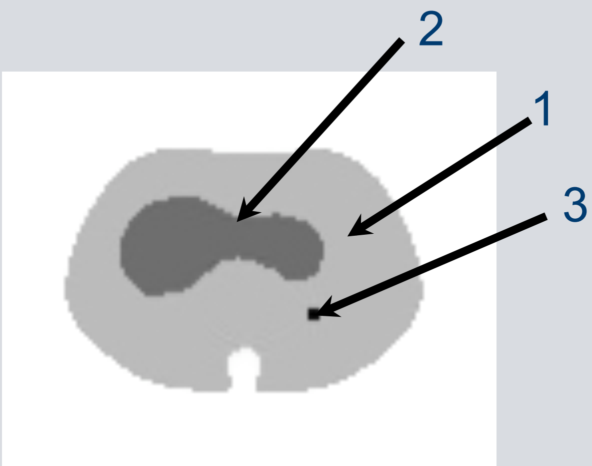
Early TOF Reconstruction Simulations

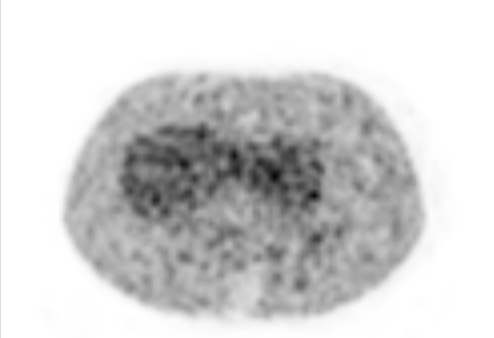


- Torso Phantom simulated with a background of 1, Liver 2:1 and Lesion 3:1
- Sinogram is made by forward projecting image
- Data were simulated with attenuation, randoms and scatter

$$sinogram = F(image) + randoms + scatter + Poisson\ noise$$

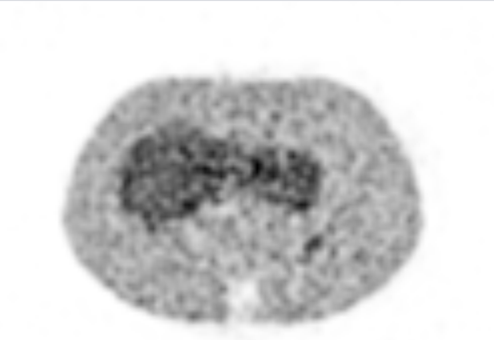
Early TOF Simulations

- Trues – 2e6
- Randoms – 1e6
- 2 iterations
- 14 subsets

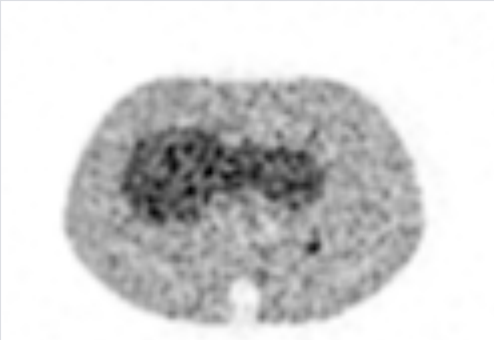


Conventional

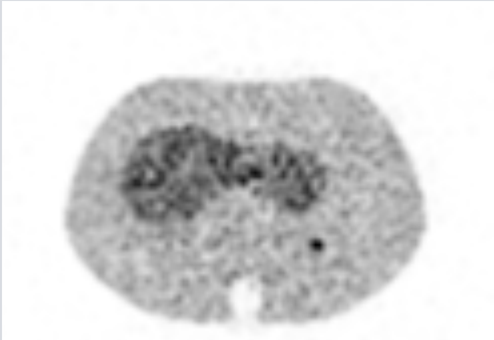
1.2 ns



700 ps



500 ps

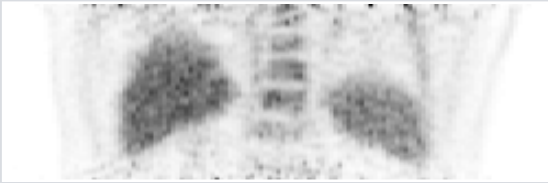


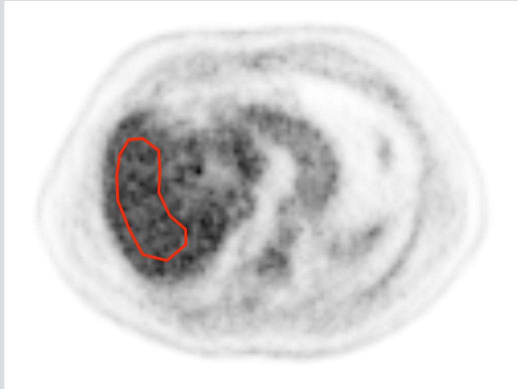
300 ps

Reconstruction algorithm was OP-OSEM with TOF extensions.

Introduction

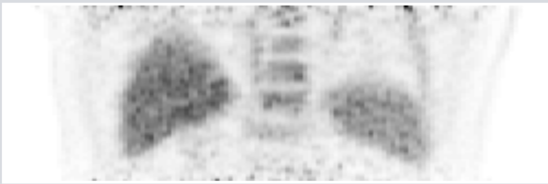
TOF

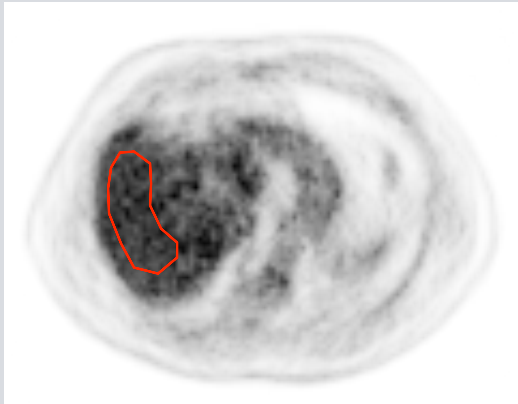




%RMS Noise = 8.5%

Conventional





%RMS Noise = 8.3%

256 lbs male arms up, 10.8 mCi injected, 130 min uptake, 20 min acquisition.

OP-OSEM reconstruction, 4 iterations, 14 subsets. 5 mm post smooth

1.2 ns time resolution

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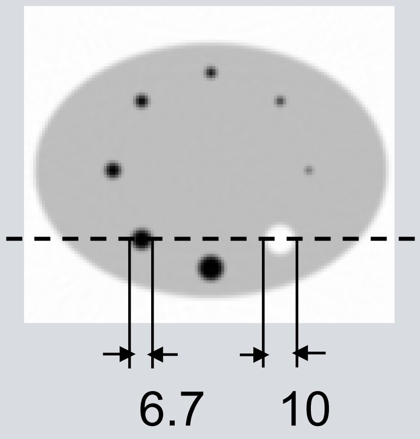
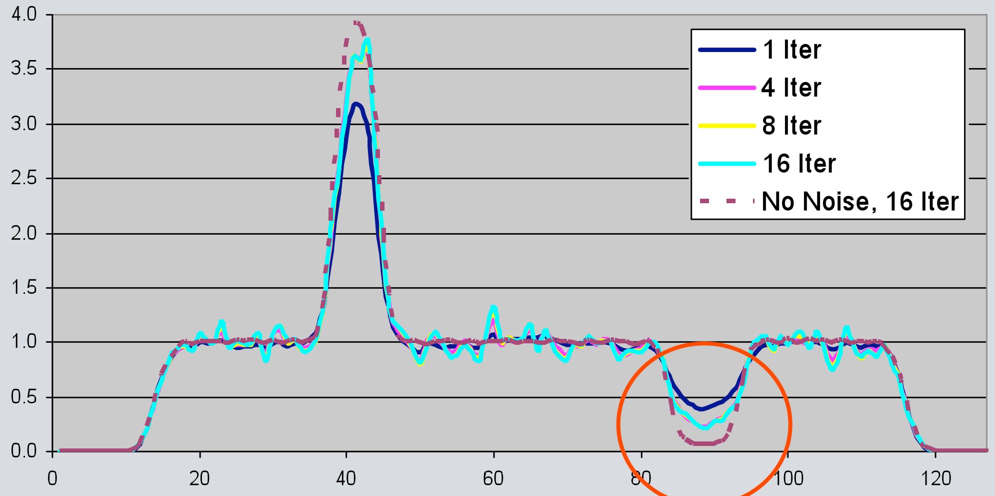
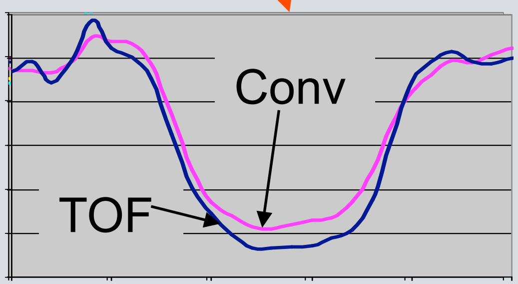
More Reconstruction simulations







TOF gain = 3.4

- Experiment used 500 noise realizations. Mean and variance images were saved
- OP-OSEM reconstruction, 16 subsets
- Iterations 4, 8, and 16 show no change in mean, yet image variance continues to increase.
- Noiseless image is closer to ideal than noisy image

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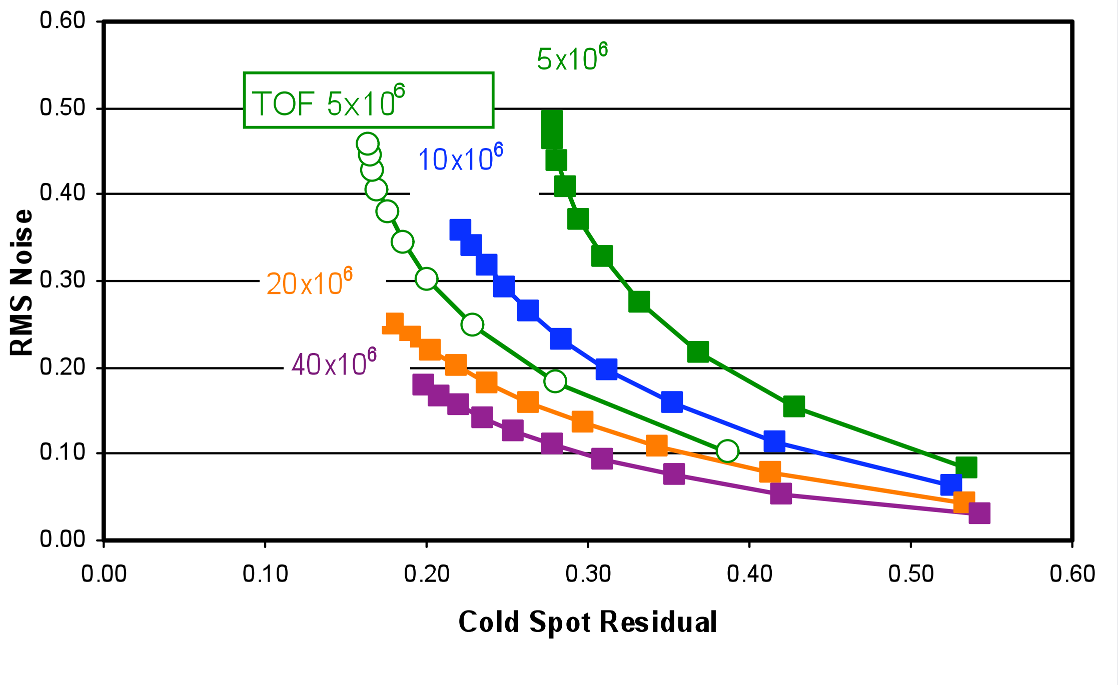
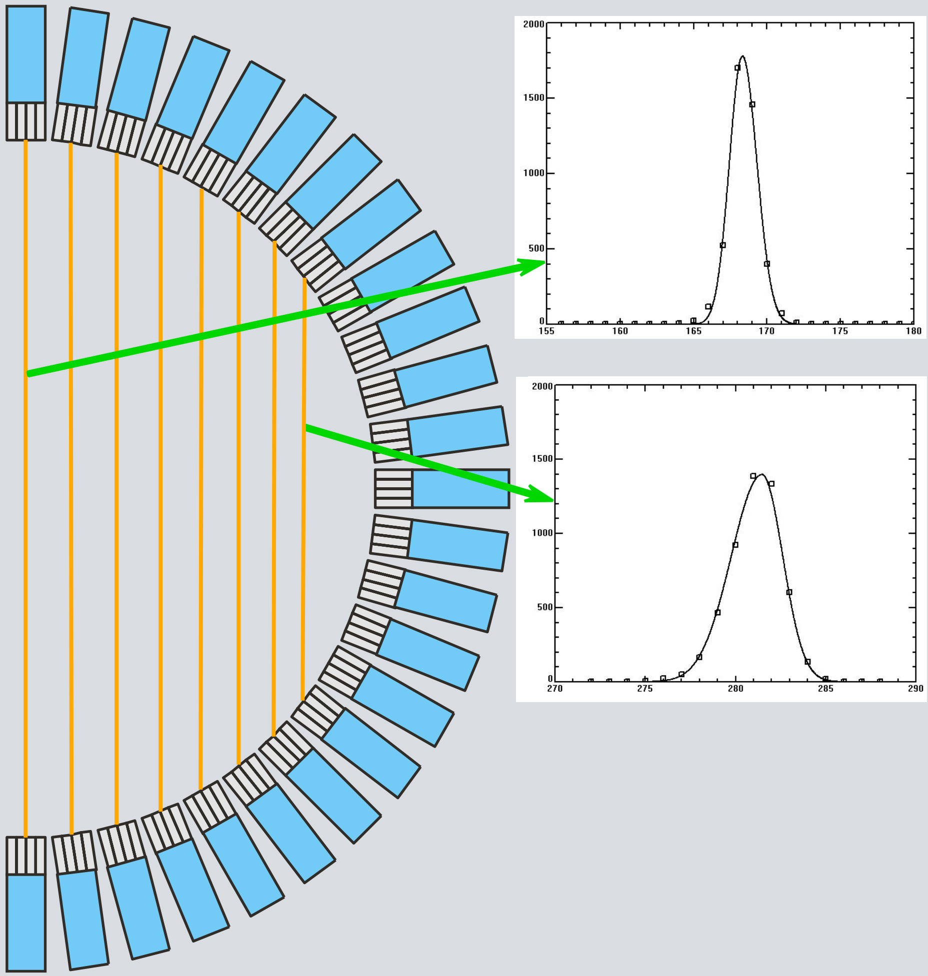
Effect of adding TOF information

- At each iteration, the residual decreases while the noise increases.
- More counts in the image allow more iterations before saturation occur.
- The TOF gain should be 3.4.



Maurizio Conti, Bernard Bendriem, Mike Casey, Mu Chen, Frank Kehren, Christian Michel and Vladimir Panin"First experimental results of time-of-flight reconstruction on an LSO PET scanner" *Phys. Med. Biol.* 50 No 19 (7 October 2005) 4507-4526

Distortion caused by circular tomograph



Lines-of-response become more closely spaced distance from center increases.

Point spread function for the central lines-of-response are symmetrical.

Point spread function for lines-of-response at large radius are asymmetrical and broader due to tilt of the crystal and depth of interaction.

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Measuring PSF

While others had relied on simulation to get the system matrix, we were skeptical of the accuracy.

We built a an X,Y,Z robot that allowed us to place a point source at precise positions within the tomograph.

Measurements were made a several radial positions. We fit the measurements using a piecewise gaussian. The widths were parameterized with respect to radial distance, allowing us to interpolate between the measurements.

To make a simulation using GEANT match our measurements, we found we had to add 0.5 mm of blurring to the simulation to compensate for the “Block Effect”

Ordinary Poisson (OP-OSEM)
Reconstruction is a now a simulation

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$$i_j^{k+1} = i_j^k \cdot \frac{1}{\sum_i \left(p_{i,j} \cdot \frac{1}{n_i \cdot a_i} \right)} \cdot \sum_i \left[p_{i,j} \cdot \frac{s_i}{\left(\sum_j p_{i,j} \cdot i_j^k \right) + a_i \cdot \left(r_i \cdot n_i + c_i \right)} \right]$$

Prompts (as collected)

Normalization

Attenuation

Contains geometric distortion

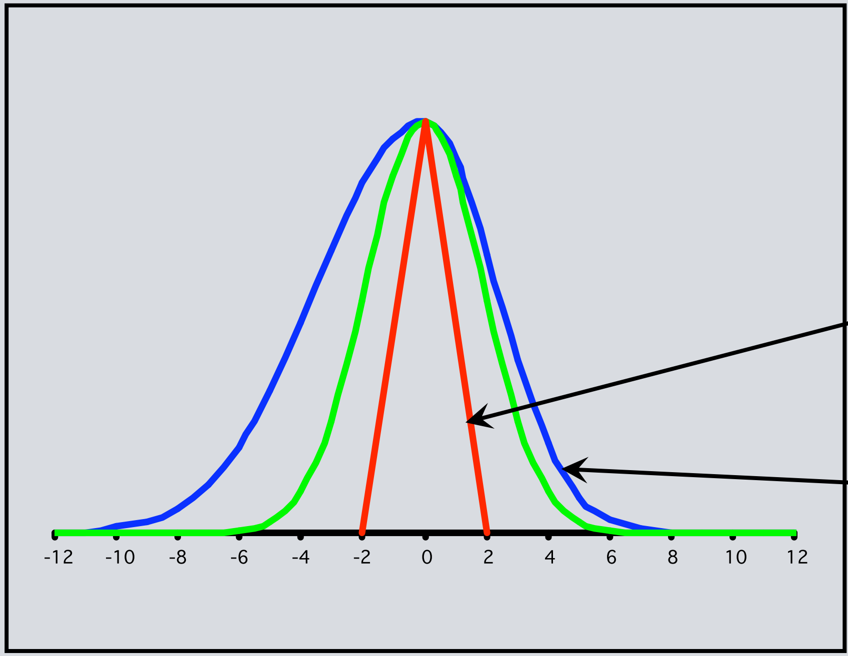
Forward projection

Simulation of data

Scatter

Randoms (Smoothed)

Inclusion of point spread functions



Recall that the $p_{i,j}$ matrix contains the probability that an event emitted from image voxel j is detected in sinogram bin i.

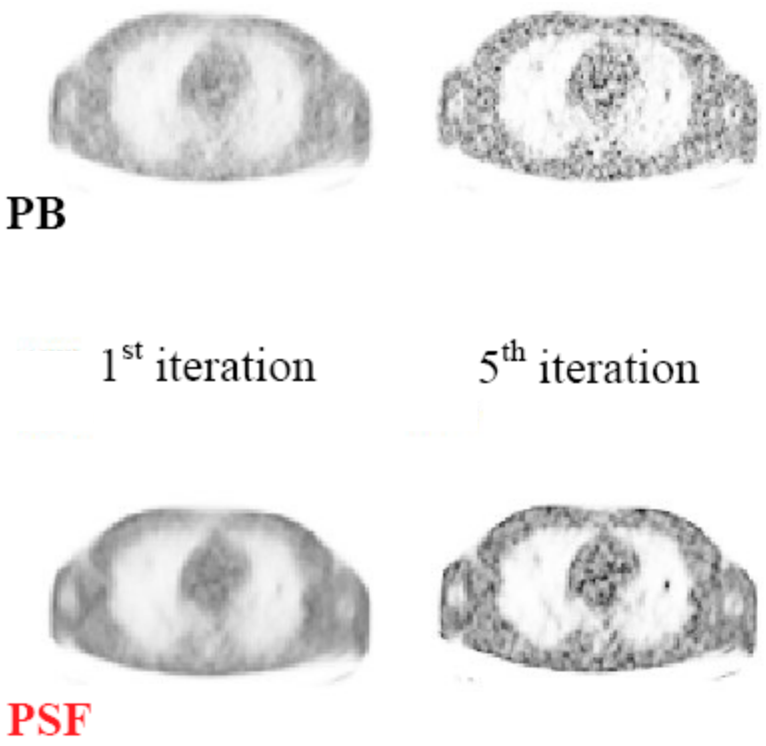
AW-OSEM interpolates between sinogram bins with linear interpolation

PSF interpolates between sinogram bins with spatially variant point spread functions

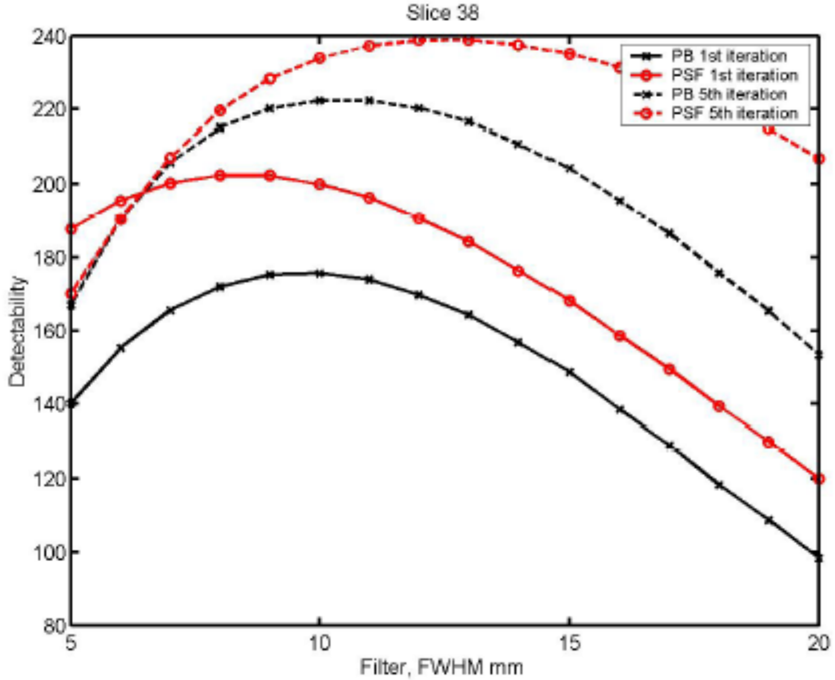
The inclusion of point spread function allows the simulation to better match the data with the effect of improving the resolution and lowering the noise.

PSF Simulations (TrueX)

One noise realization (signal present) reconstruction



Detectability in 2:1 hot spot central plane as function of postsmoothing



Panin, V. et. al. “Numerical observer evaluation of 3D statistical PET reconstruction method with system response modeling”, *2005 IEEE Nuclear Science Symposium Conference Record*

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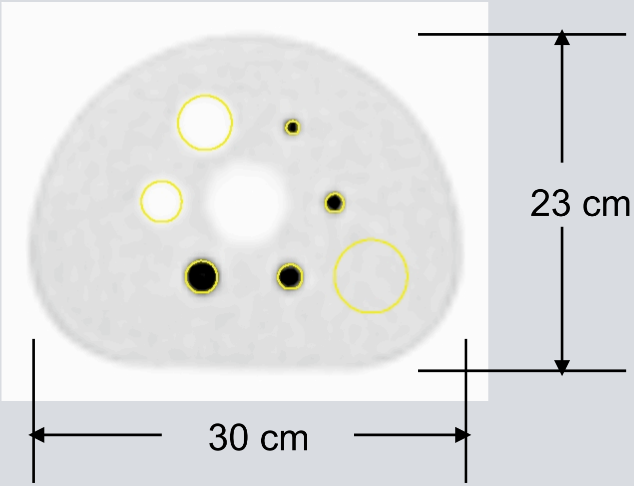
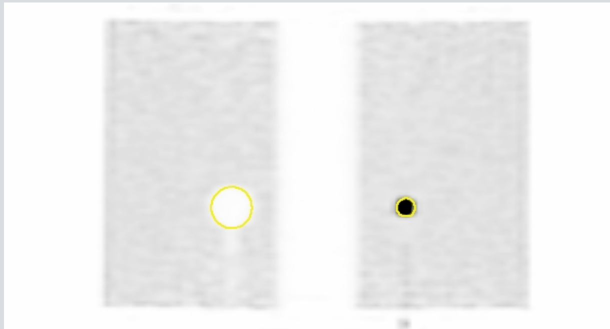
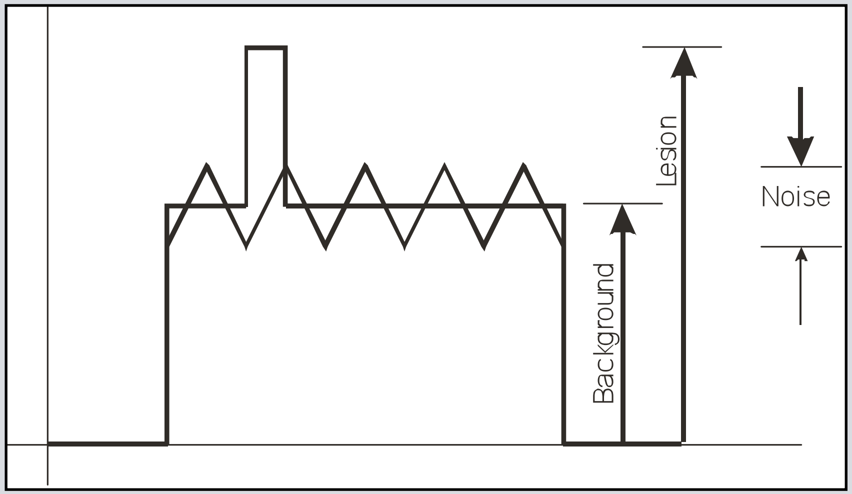
Detectibility







$$Detectibility = \frac{Lesion - Background}{Noise}$$

Detectibility is essentially a signal to noise ratio.

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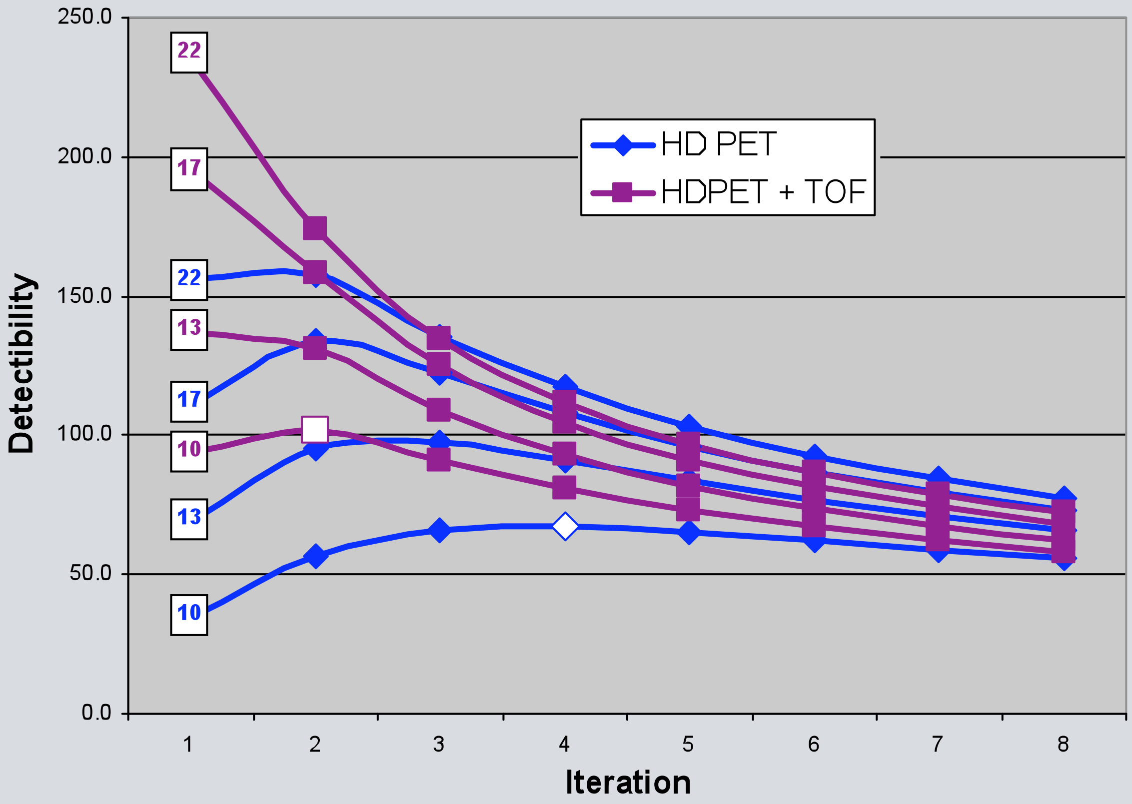
Detectibility



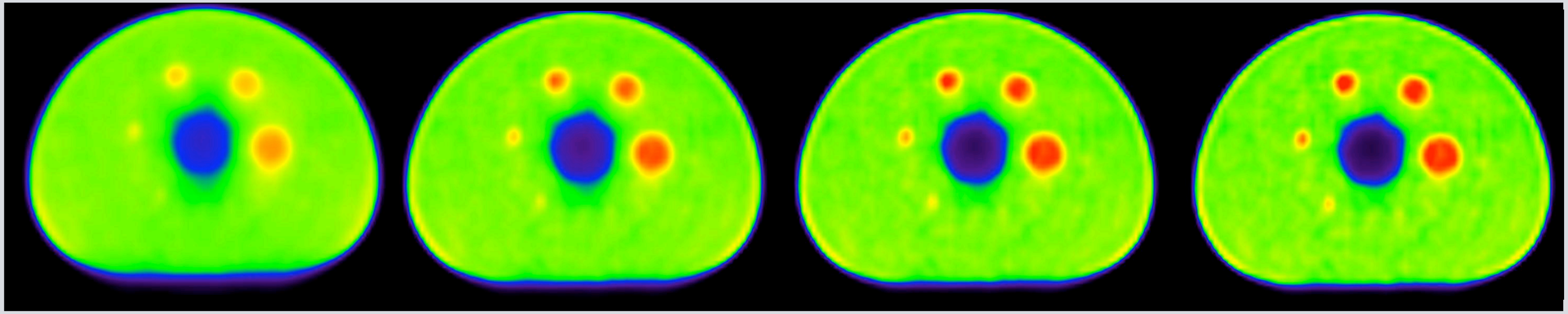
	Algorithm	
Iteration	HDPET	HDPET + TOF
1	2.5%	2.1%
2	3.4%	3.3%
3	4.2%	4.4%
4	5.0%	5.4%
5	5.8%	6.3%
6	6.5%	7.1%
7	7.2%	7.8%
8	7.9%	8.6%

- The information added by TOF allows faster convergence of TOF reconstruction. Detectibility peaks at an earlier iteration.
- For this phantom, with a time resolution of 550 ps, the TOF gain should be 3.2. The gain was $(5.0/3.3)^2 = 2.3$

Image quality phantom, **2:1 ratio** (!!)

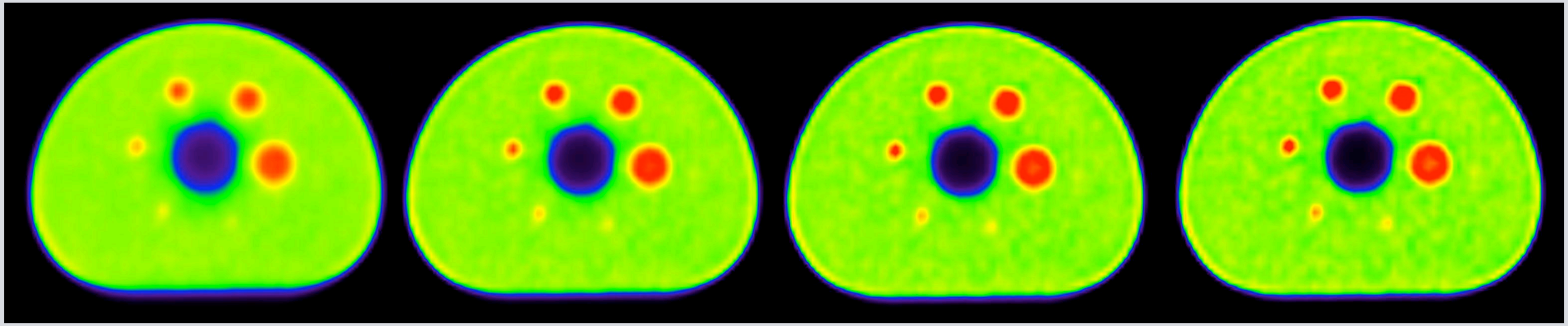
1-hour scan, about 1 mCi activity, 168x168 pixels (4mm pixel size)

non-TOF

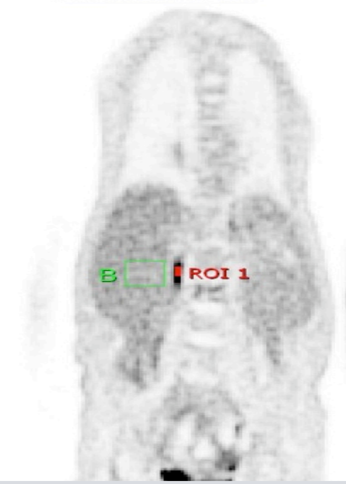


Iter 1 | Iter 2 | Iter 3 | Iter 4

TOF

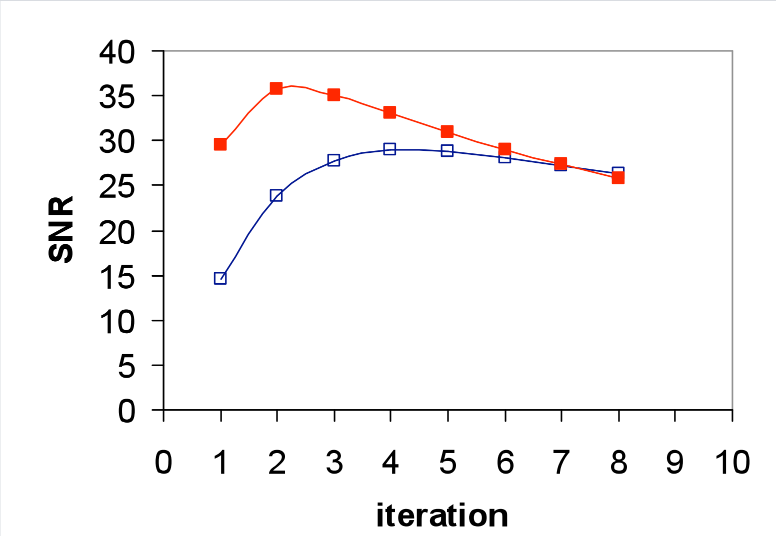


Patient studies: Signal to Noise and Contrast



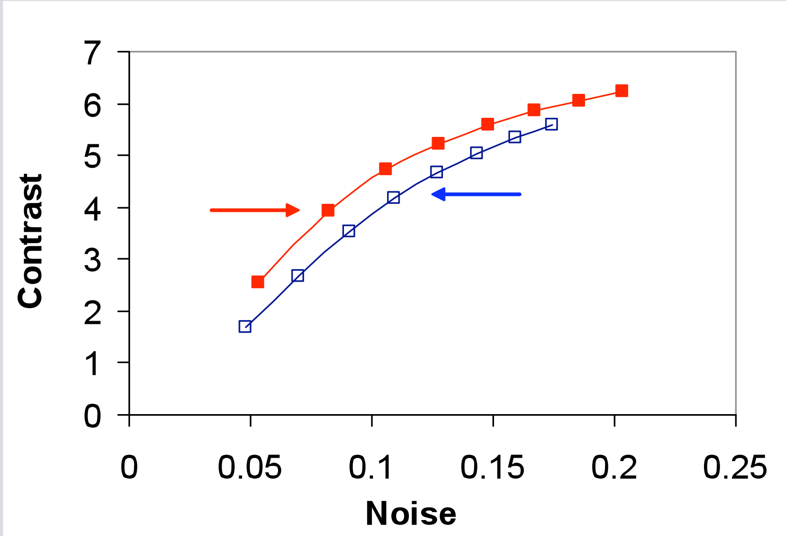
- small lesions: <2cm
- signal *S*: mean value in lesion (ROI < 1cm)
- background *B*: mean value in liver (large ROI)
- noise: standard deviation in background

$$SNR = \frac{S - B}{\sigma_B} \qquad Contrast = \frac{S}{B} \qquad Noise = \frac{\sigma_B}{B}$$



TOF

non-TOF



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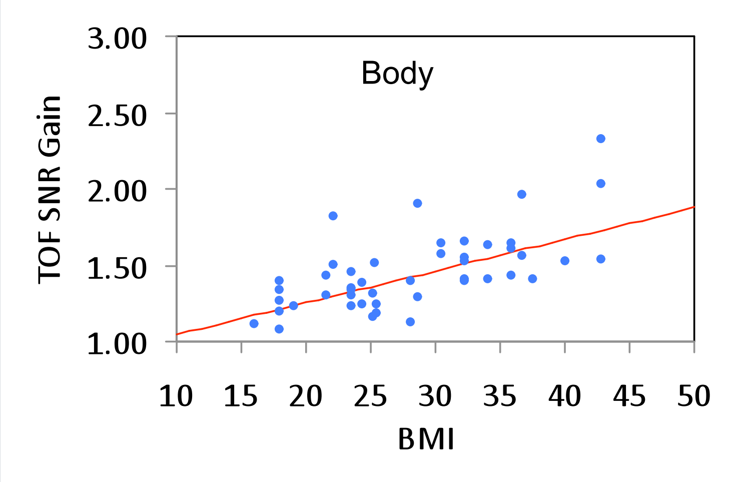
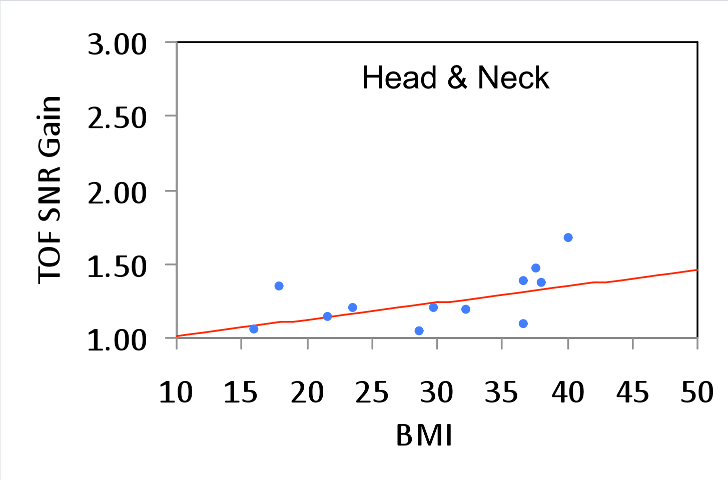
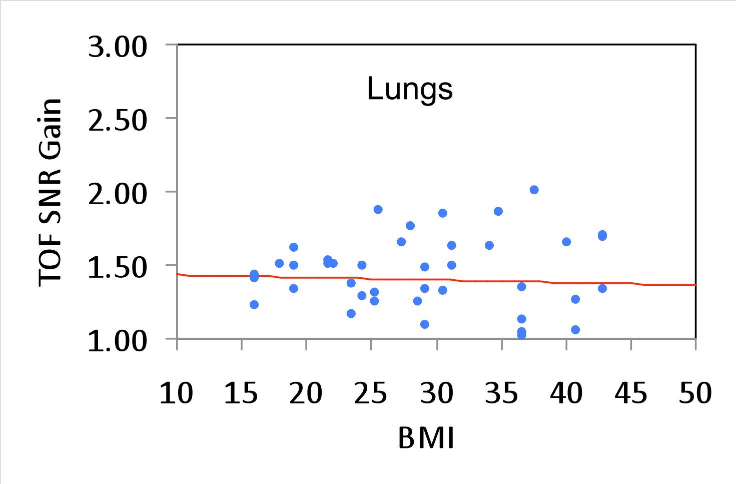
Patient studies: TOF Gain and patient size



- Multiple lesions (<2cm) analyzed in 30 patients
- Body Mass Index used a measure of the size

$$G_{SNR} = \frac{SNR_{TOF}}{SNR_{nonTOF}} \propto \sqrt{\frac{D}{\Delta t}}$$





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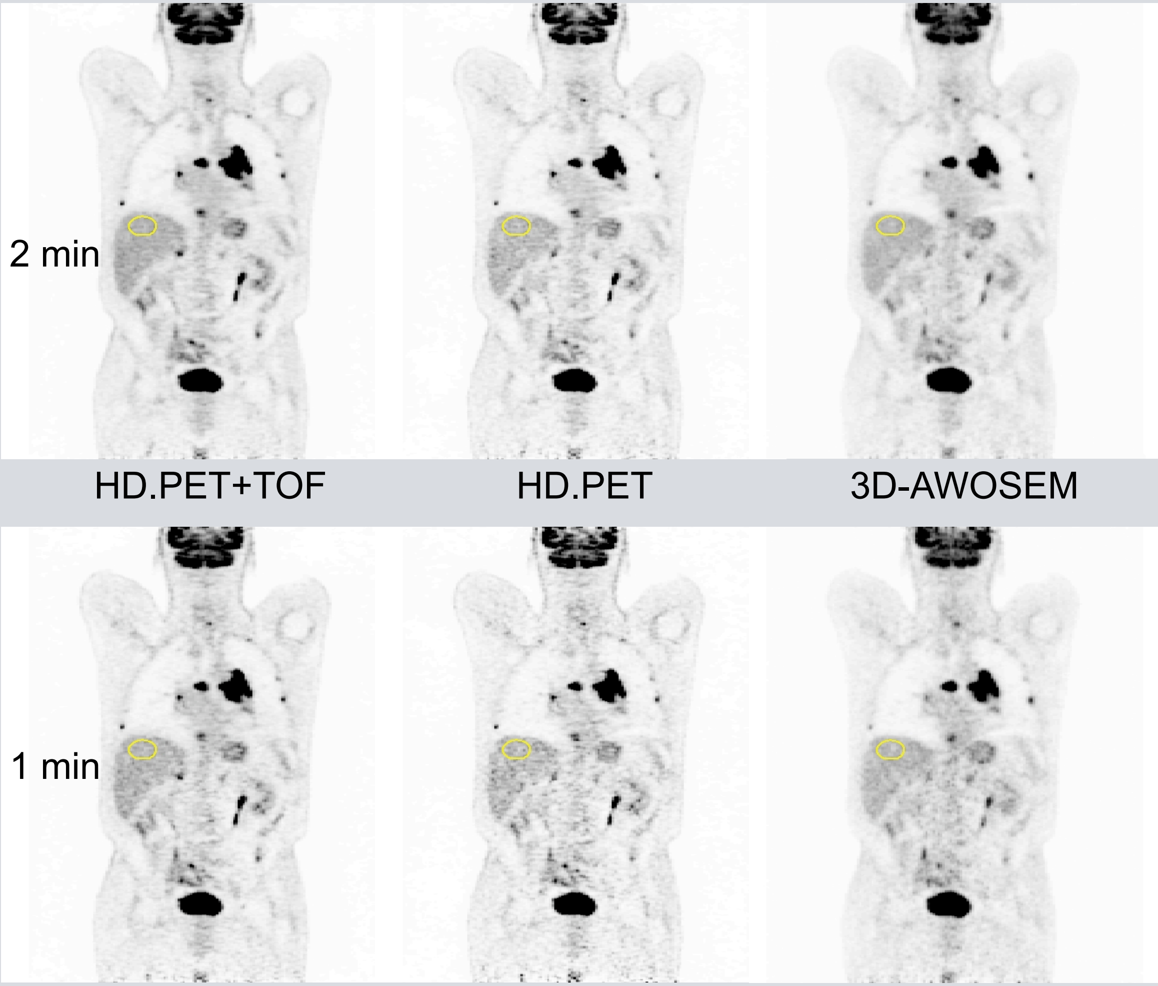
Simulation of shorter scan

One minute scan was simulated by throwing out every other event in a two minute list mode scan.

Contrast and noise in 2 min HD.PET is approximately the same as HD.PET+TOF for one minute

	Iterations	Subsets	Filter
3D-AWOSEM	3	8	5mm
HD.PET	4	14	none
HD.PET+TOF	2	14	none

		Contrast	Noise
2 Minute	3D-AWOSEM	3.24	10.4%
	HD.PET	5.38	16.3%
	HD.PET+TOF	5.46	11.6%
1 Minute	3D-AWOSEM	2.84	13.5%
	HD.PET	5.26	22.6%
	HD.PET+TOF	5.26	16.0%



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