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Signal transport in Computed Tomography detectors

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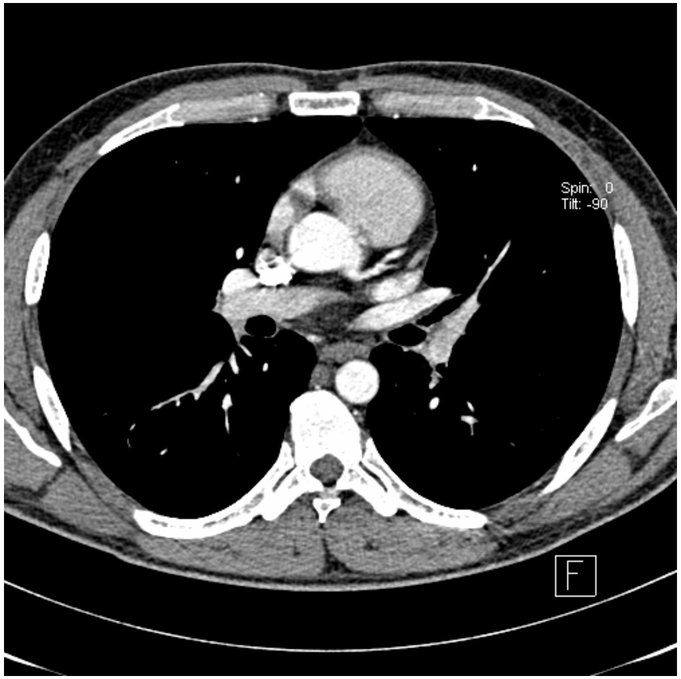
Computed Tomography



CT system
measures X-ray attenuation coefficient $\mu(\boldsymbol{r})$

Image data
patient 3D morphology

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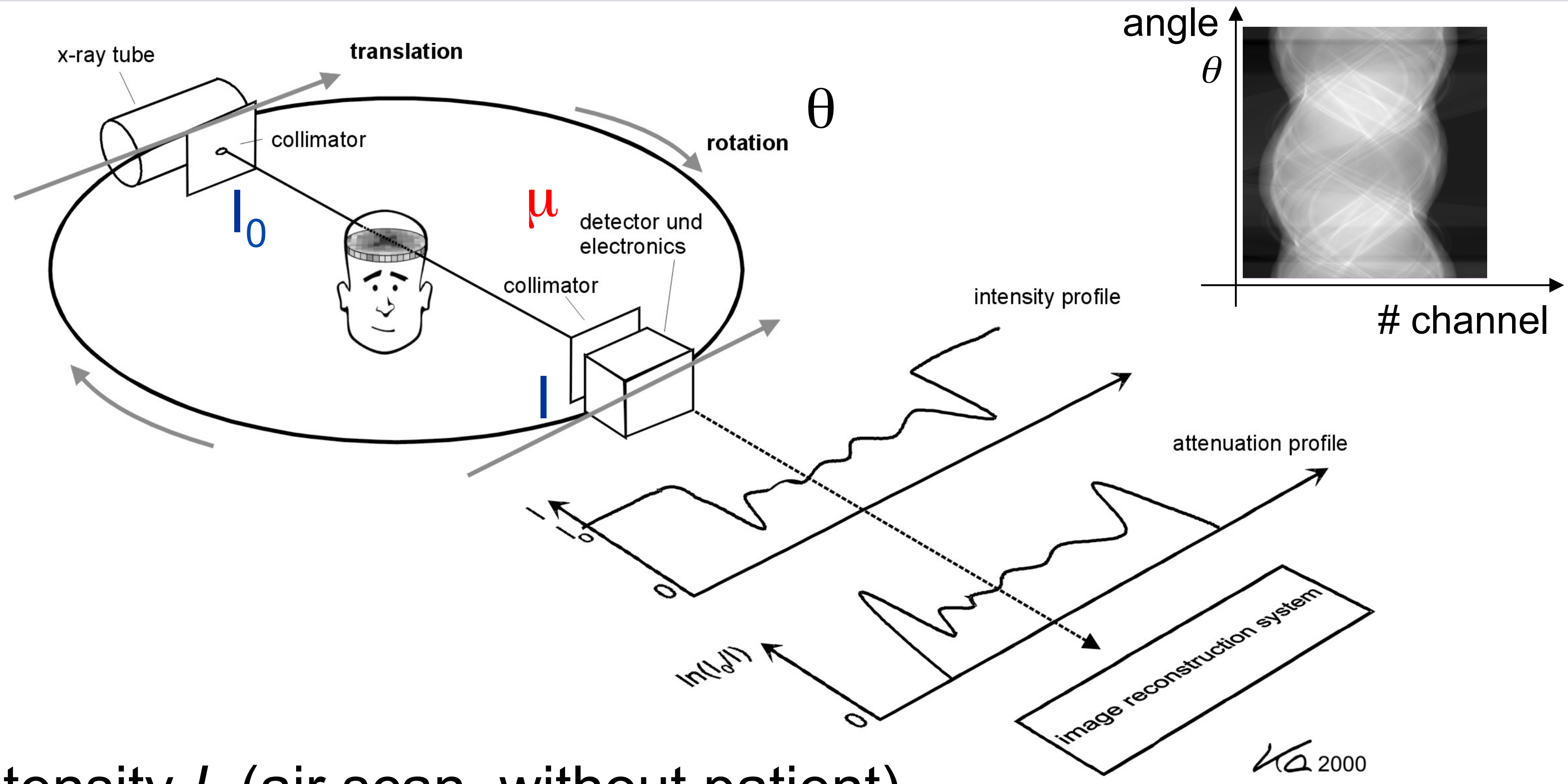
CT measurement



- Intensity I_0 (air scan, without patient)
- Intensity I (with patient)
- Measured: Sinogram $\ln(I_0 / I)$

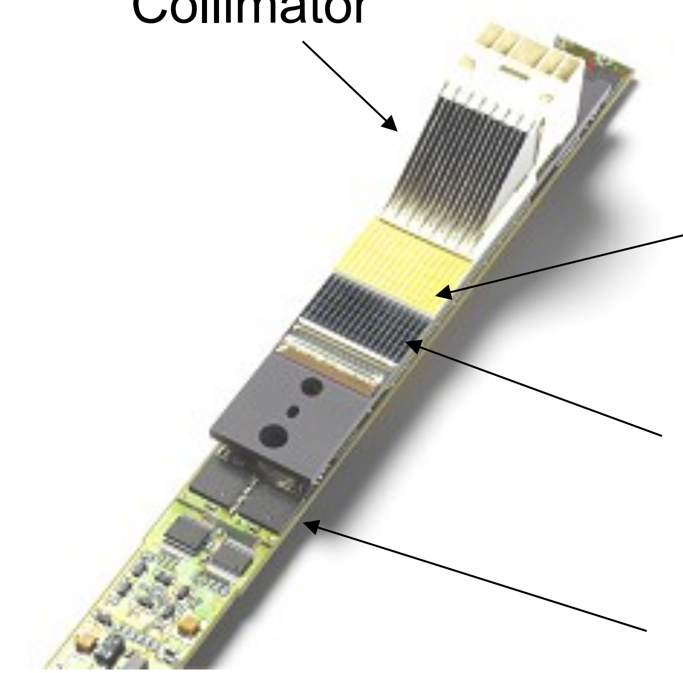
Examples: Noise vs. X-ray dose and resolution in CT

Liver tumor
contrast-to-noise

Carotid Aneurysma
resolution

Scintillator detectors

Anti-scatter Collimator



Scintillator array (GdOS:Pr)

Photodiode array

Analog-to-digital conversion asics

Typical CT detector module

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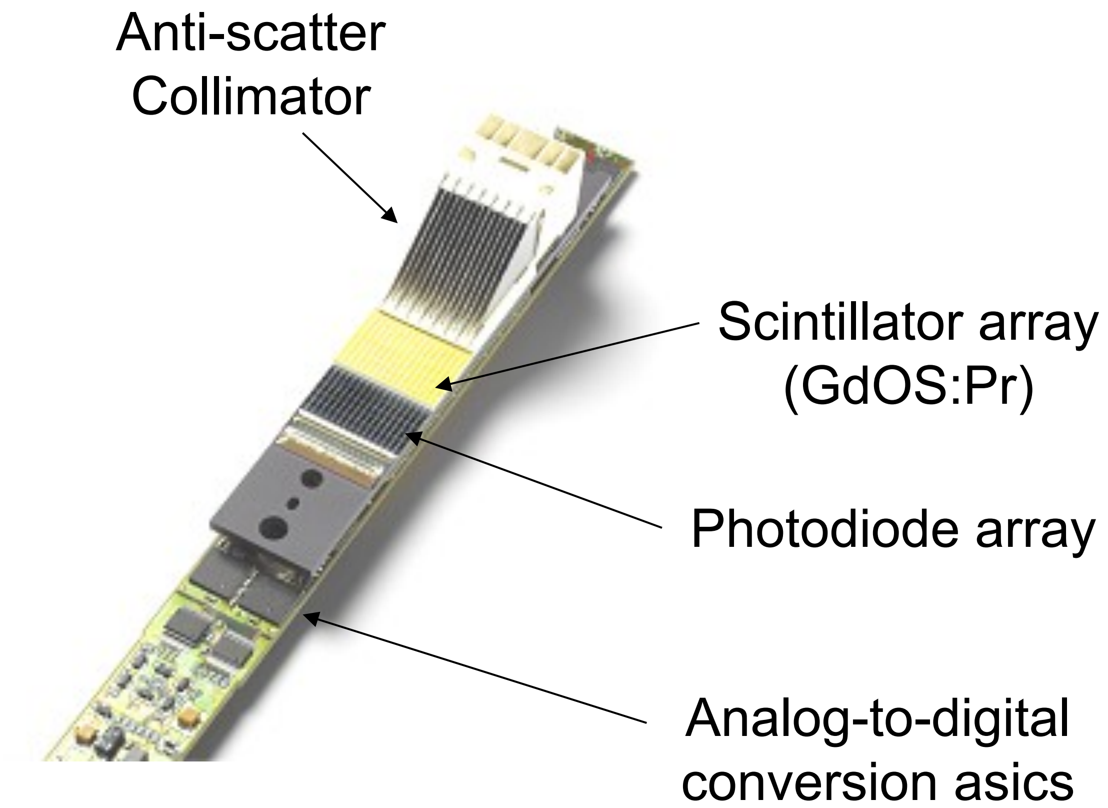
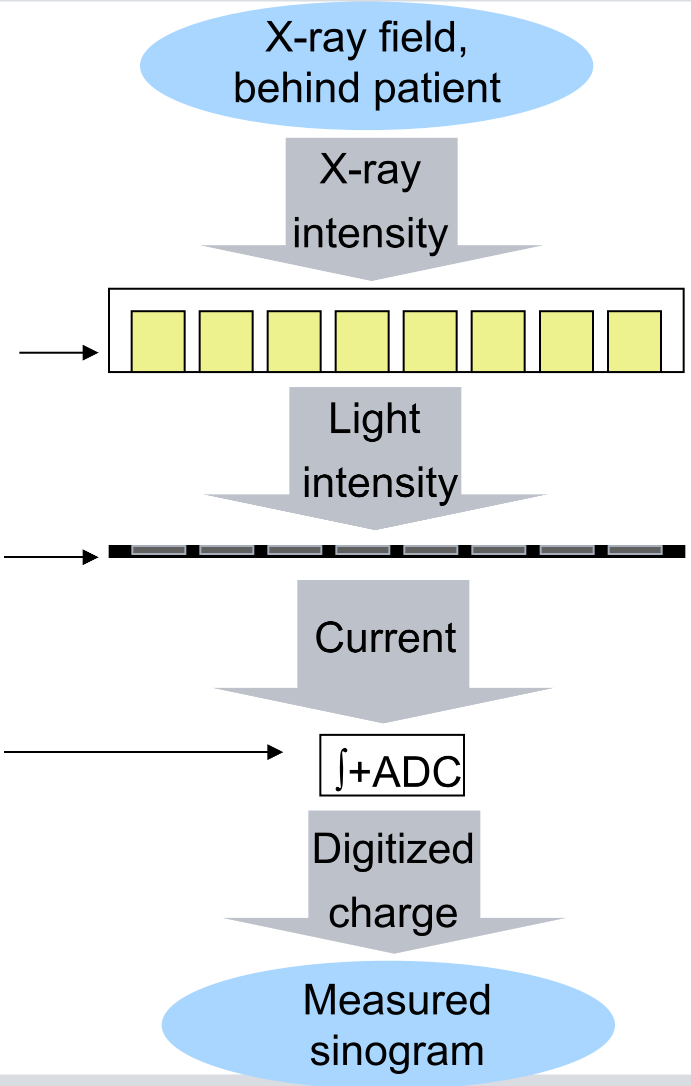
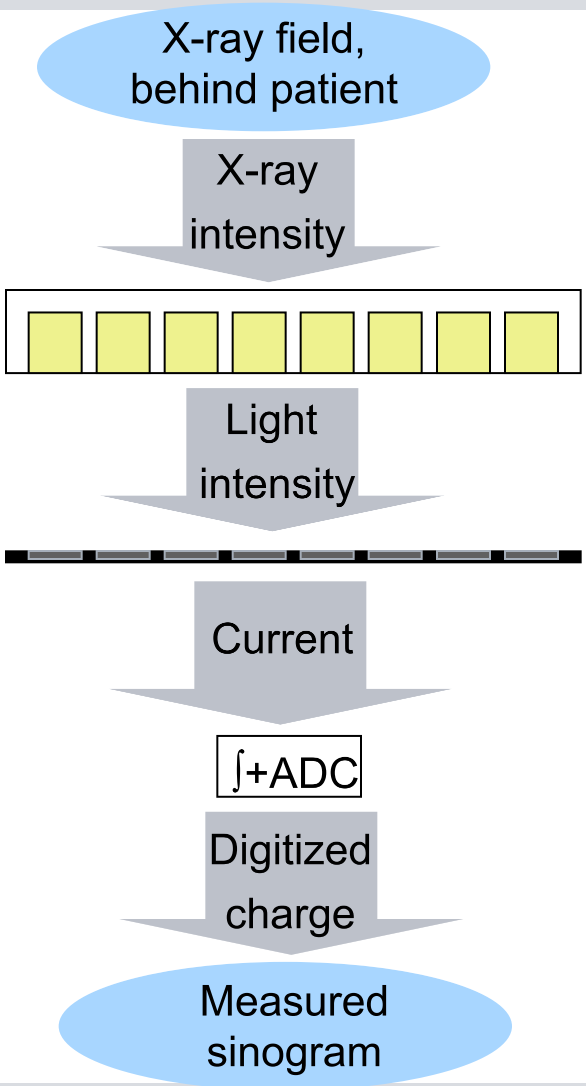
Scintillator detectors





Typical CT detector module

Cascaded Detection Model*



- Quantum efficiency $\alpha(E)$
- Quantum fluorescence scatter
- Conversion to light
- Scattering of light

- Optical coupling efficiency
- Photon detection efficiency $\beta(E')$

- ADC conversion performance

*see e.g. I.A. Cunnigham: 'Applied Linear System Theory ', in: J. Beutel, H.L. Kundel, R.L. van Metter, Handbook of Medical Imaging, Vol. 1, pp 79, SPIE (2000)

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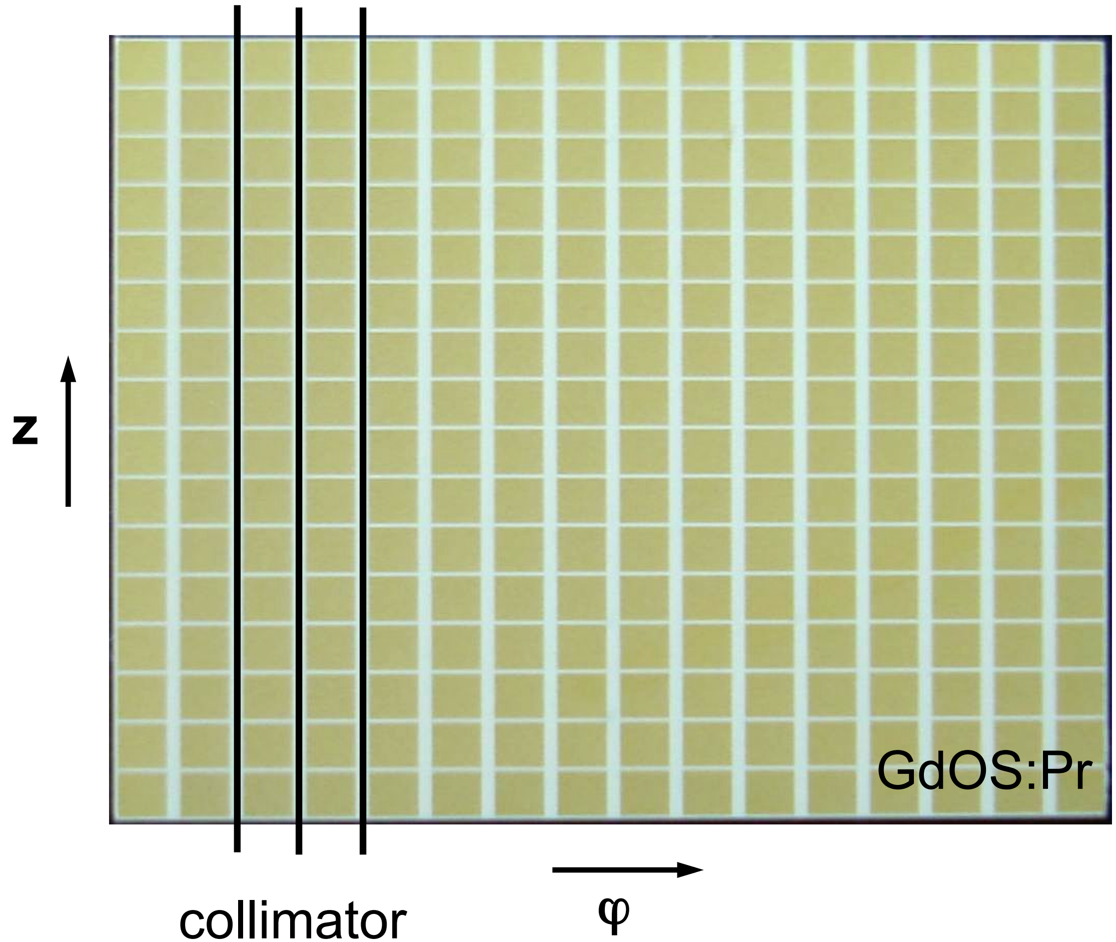
Reference scintillator geometry

- **Pitch φ** ~ 1.2 mm
- **Pitch z** ~ 1 mm
- **Gap φ** ~ 250 µm
  (incl. collimator dead-zone)
- **Gap z** ~ 80 µm



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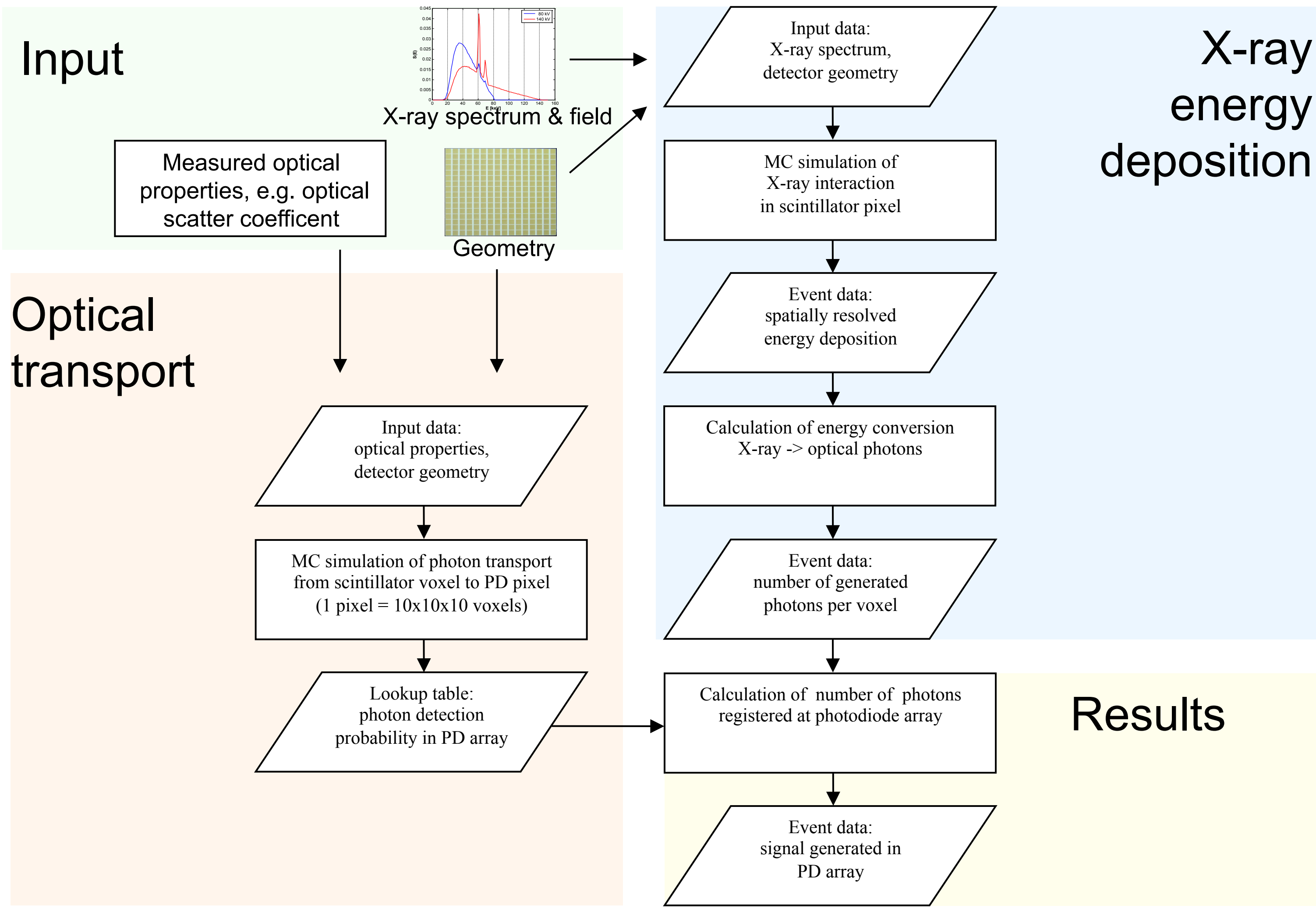
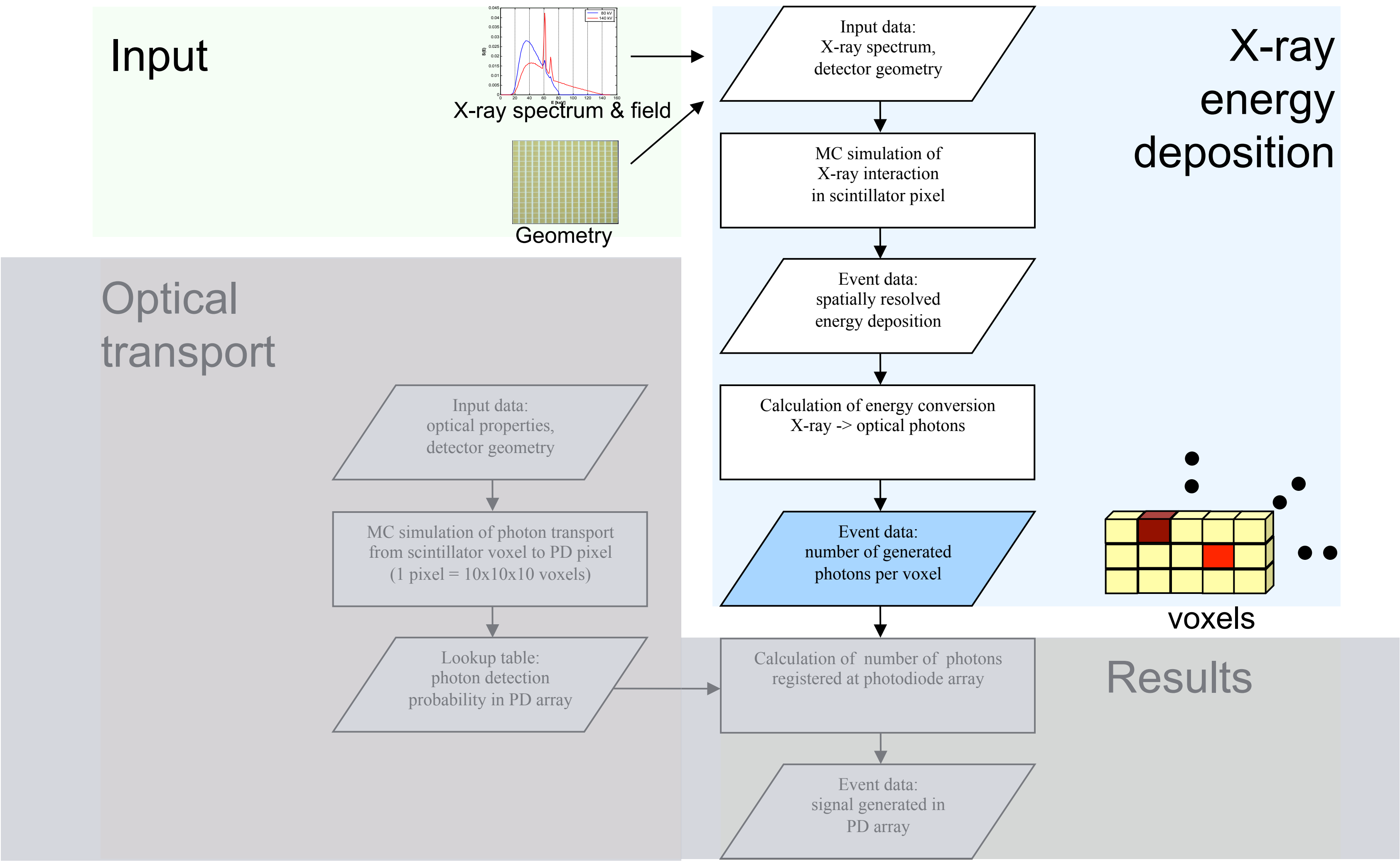
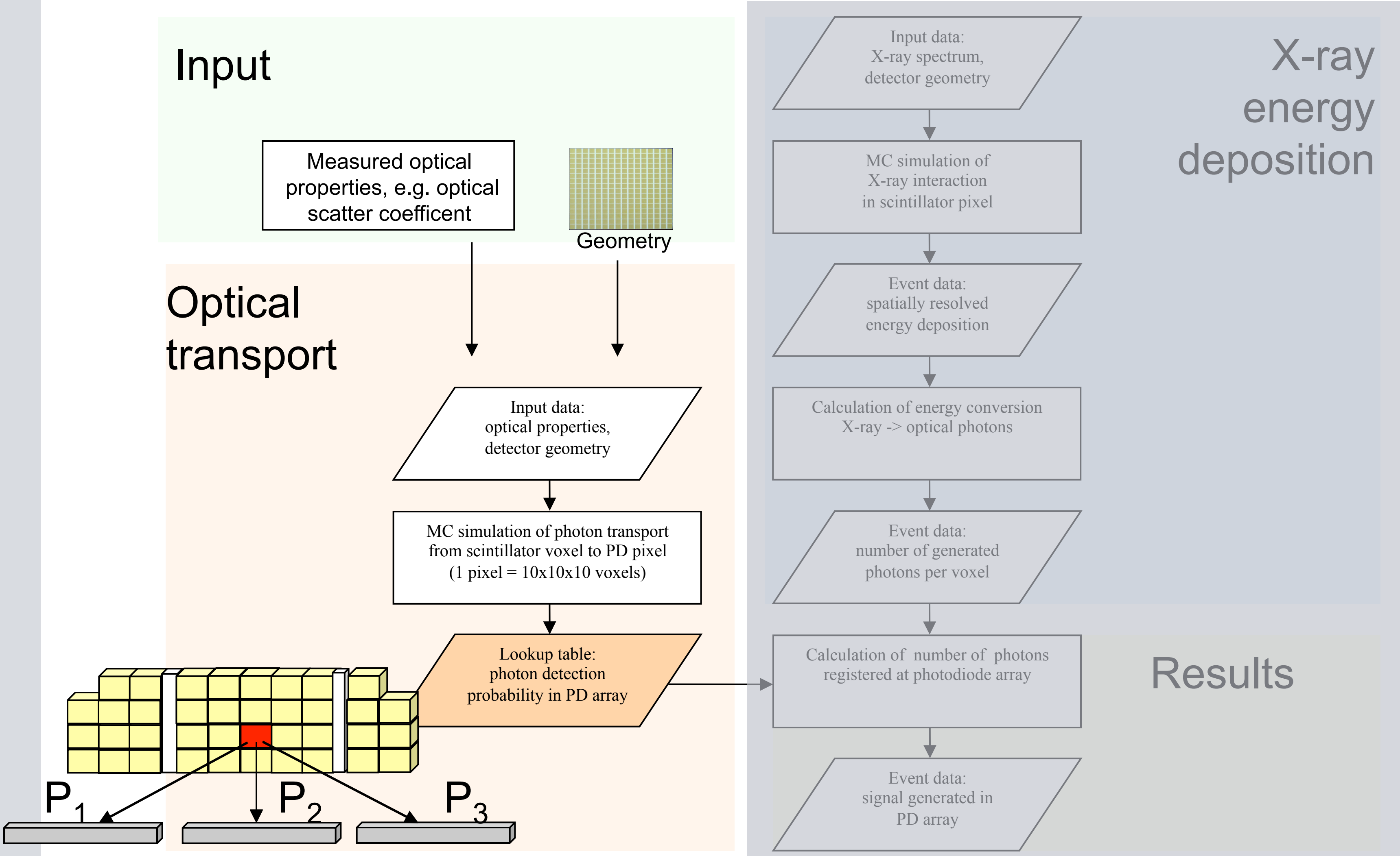
Simulation flow chart



Simulation flow chart (X-ray)

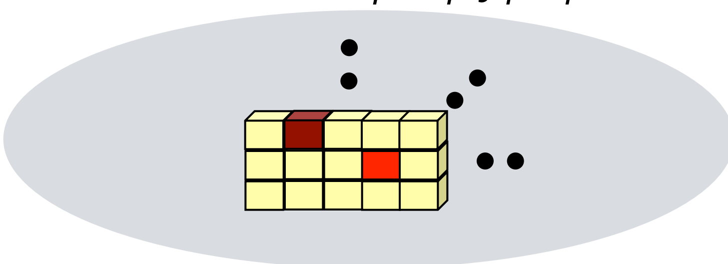


Simulation flow chart (optical)

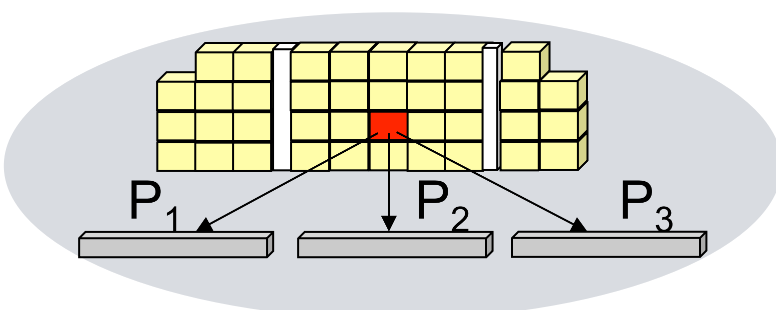


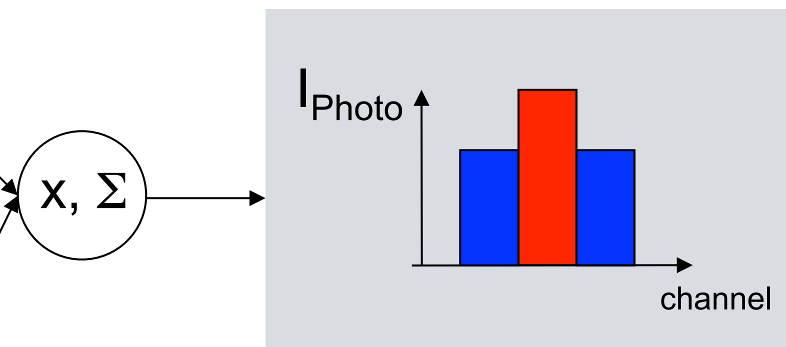
Simulation flow chart (results)

1. X-ray energy deposition:
Event list: #i, <E_i, x_i, y_i, z_i>



2. Photosensor signal:
detection probabilities



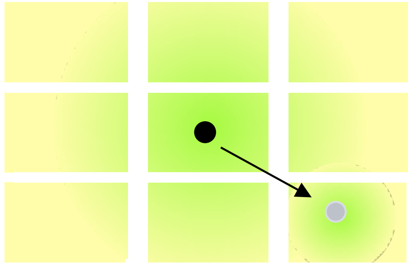


Result:
Detected number of light photons on sensor

Microscopic characteristics

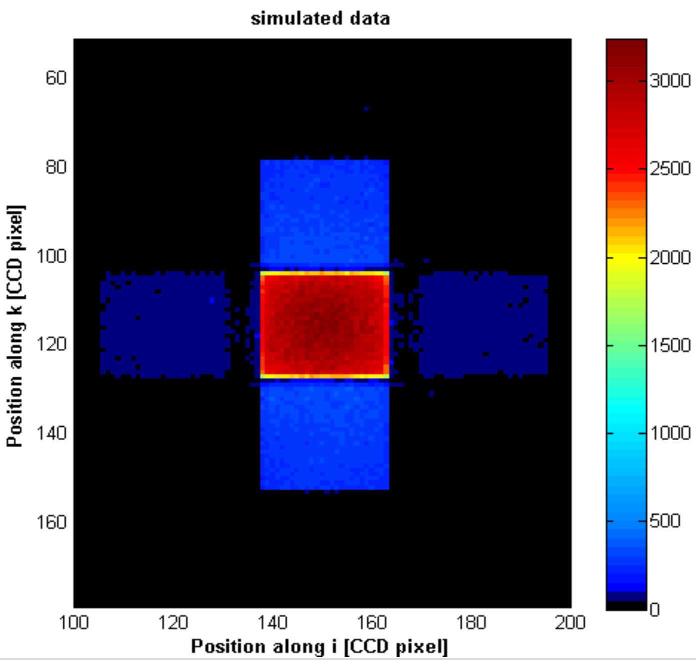
Individual quantum absorption events:

- X-ray cross-talk due to fluorescence (*E* > 50.2 keV), affects mostly next neighbours and only 2 pixels
- Optical cross-talk involves up to 5 x 5 pixels

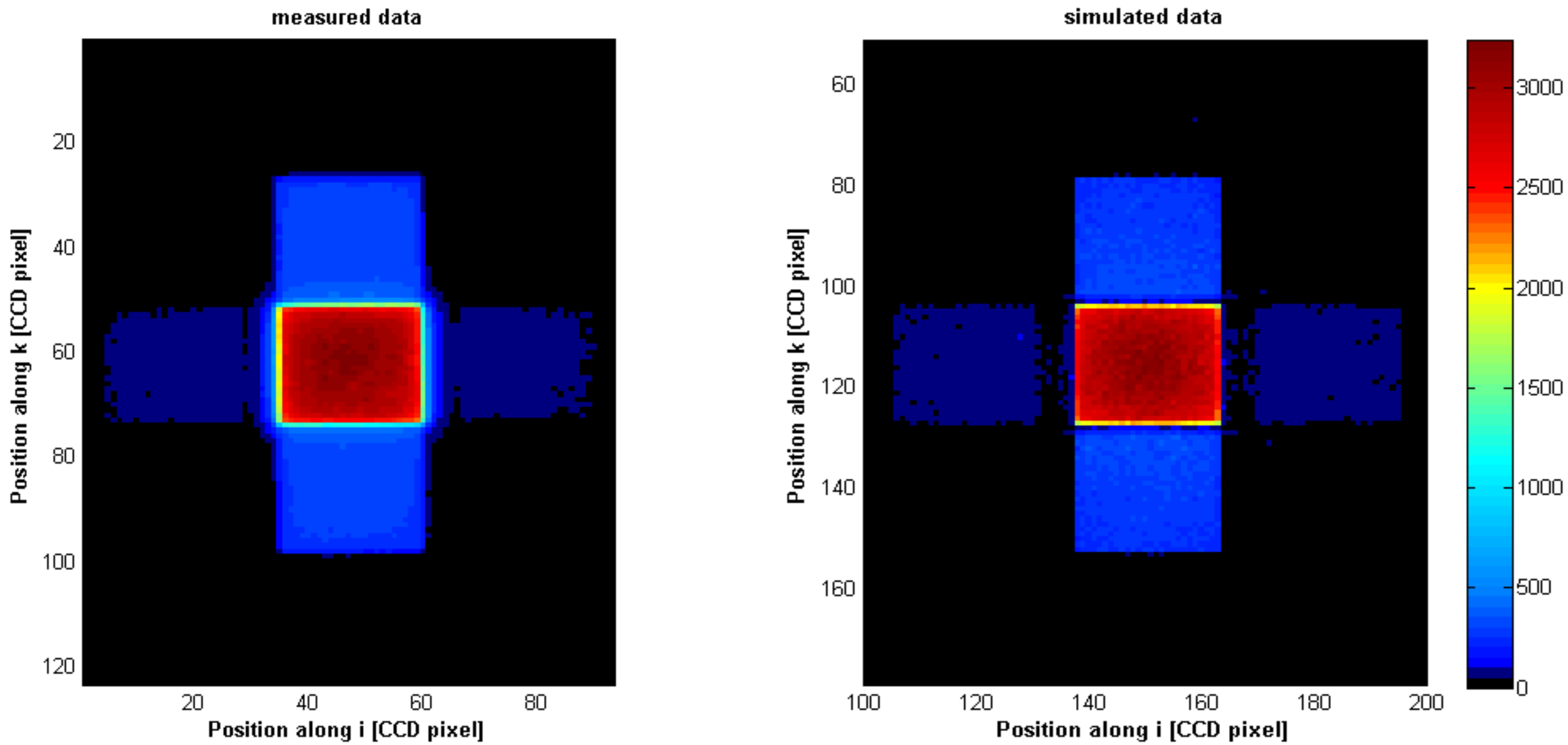


Averaged result of a 5 x 5 pixel array:

- 120 kV CT spectrum
- *n* = about one million quanta
- Needle beam excitation of central pixel
- Photosensor: CCD with sub-pixel resolution



CCD measurement vs. simulation



CCD measurement, fibre coupled

Simulation result

43 µm resolution

central X-ray needle beam excitation

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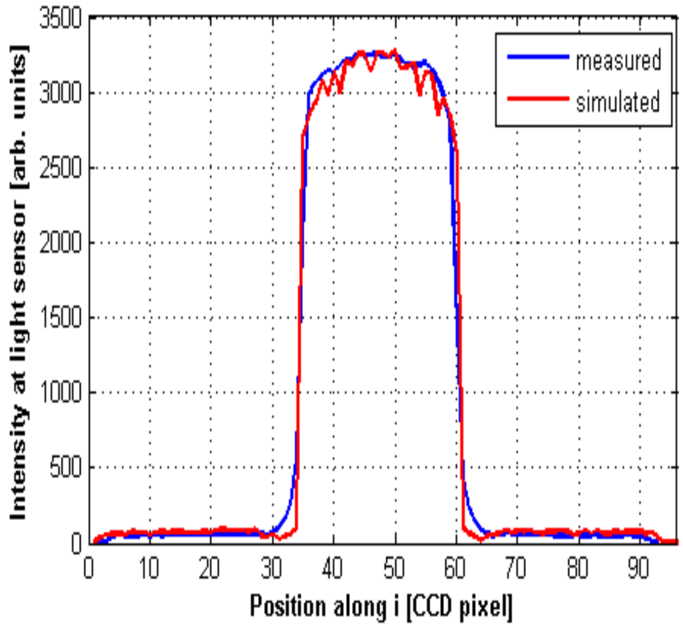
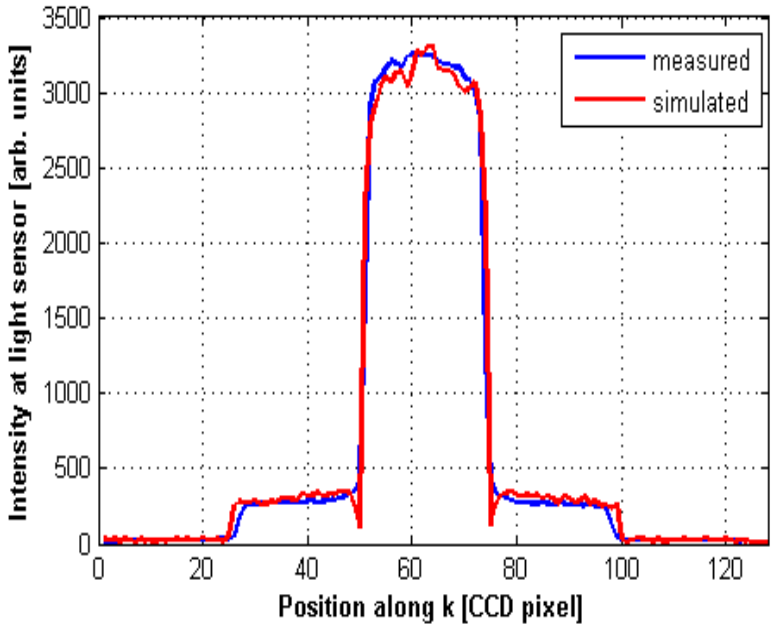
Profiles





- good agreement, typical error < ~ 5%
- smoother edges in measurement:

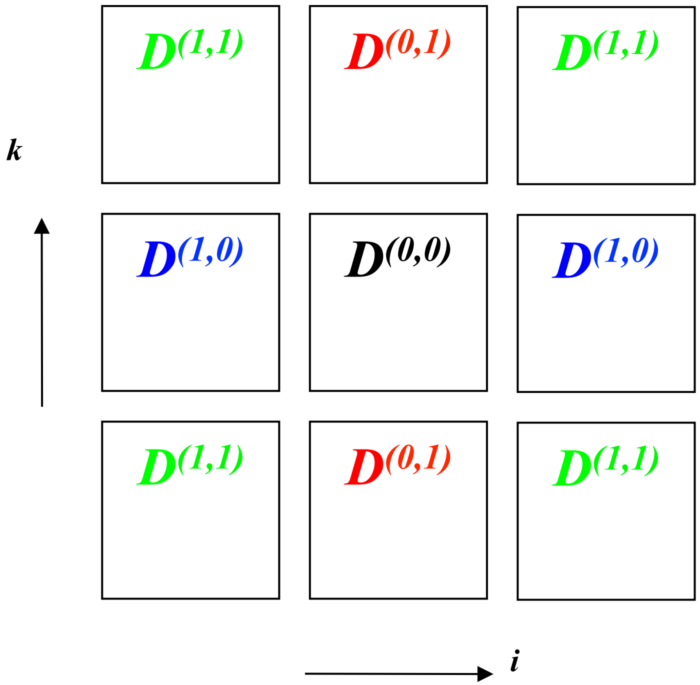
probably due to fibre coupling cross-talk not included in simulation

Macroscopic (average) figure

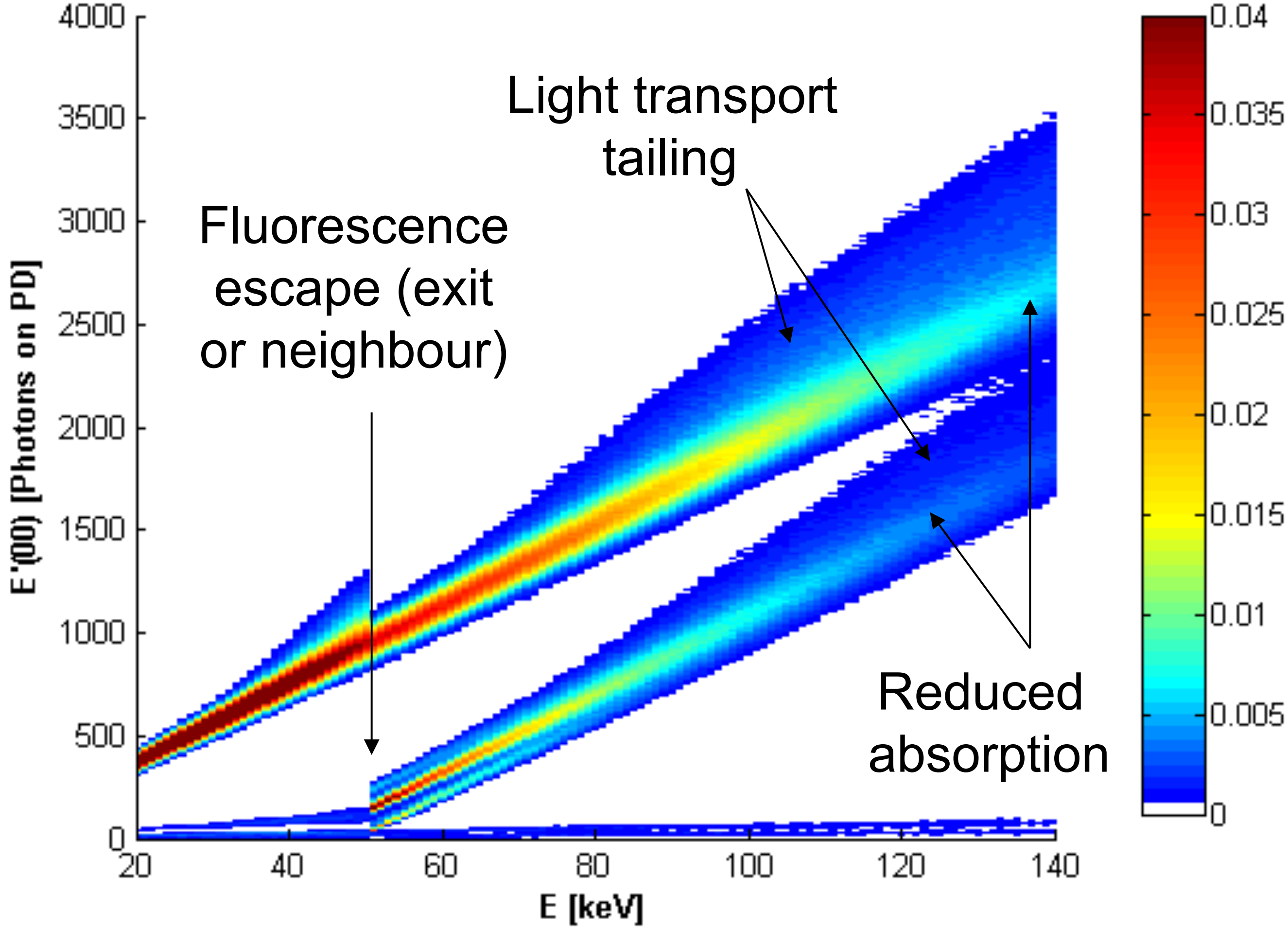
Detector response function $D^{(i,k)}(E, E')$

- Probability density function for measuring output energy E' at input energy E
- i, k denote pixel class (symmetry):

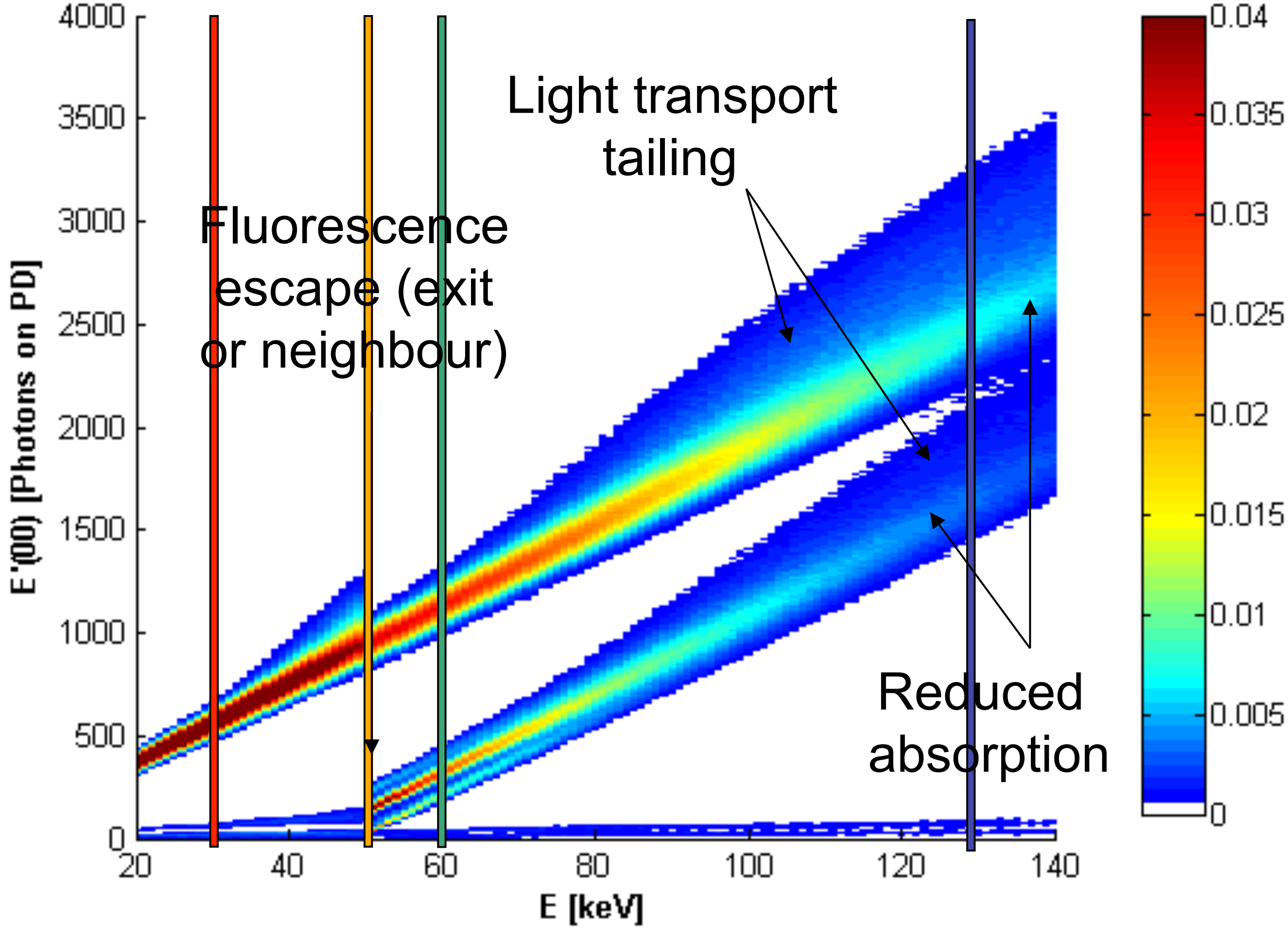
$i = 0, k = 0$: central pixel
$i = 1, k = 0, \ldots$: horizontal next neighbours
$i = 0, k = 1, \ldots$: vertical next neighbours
$i = 1, k = 1, \ldots$: diagonal neighbours



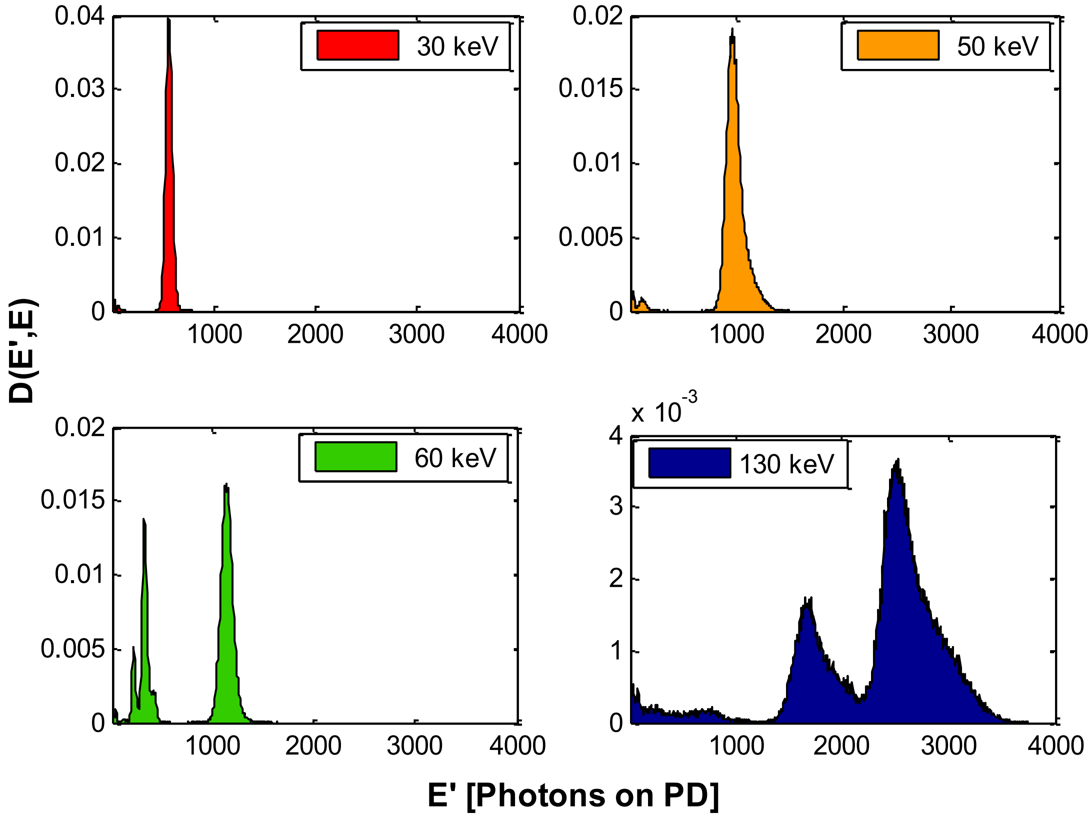
Central pixel: $D^{(0,0)}(E,E')$



Central pixel: $D^{(0,0)}(E,E')$



Behaviour at constant input energies *E*



- Two peaks occur above the Gd K-edge
- Signal tailing increases with input energy due to longer light paths

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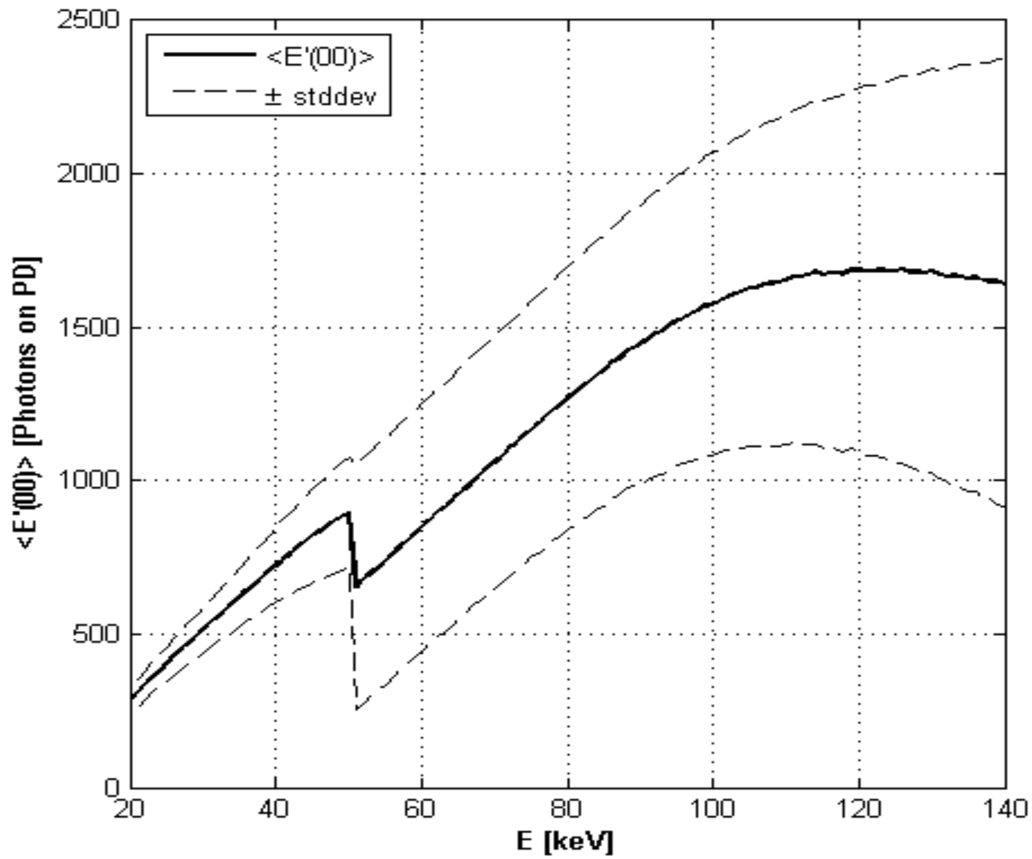
Expected signal <*E ‘*(*E*)>

<*E ‘(E)*> deviates from $\propto E$ due to

- Fluorescence escape
- Reduced absorption
- Light tailing
- Compton escapes

How does the signal noise affect the quantum statistics?



Statistics of scintillating detectors

Measured signal

$$Y = \sum_{i=1}^{M} E'_i$$

with M : number of detected X-ray quanta
and E'_i: number of detected light photons of event $\#i$

M and E'_i are random variables described by their means and standard deviations. We have*):

$$\langle Y \rangle = \langle E' \rangle \cdot \langle M \rangle$$

$$\sigma(Y) = \sqrt{\langle M \rangle}\sqrt{\left(\langle E' \rangle^2 + \sigma^2(E')\right)}$$

*) M. Rabbani, R. Shaw, and R. Van Metter, 'Detective quantum efficiency of imaging sytems with amplifying and scattering mechanisms,' *J. Opt. Soc. Am. A*, vol. 4, pp. 895-901, 1987

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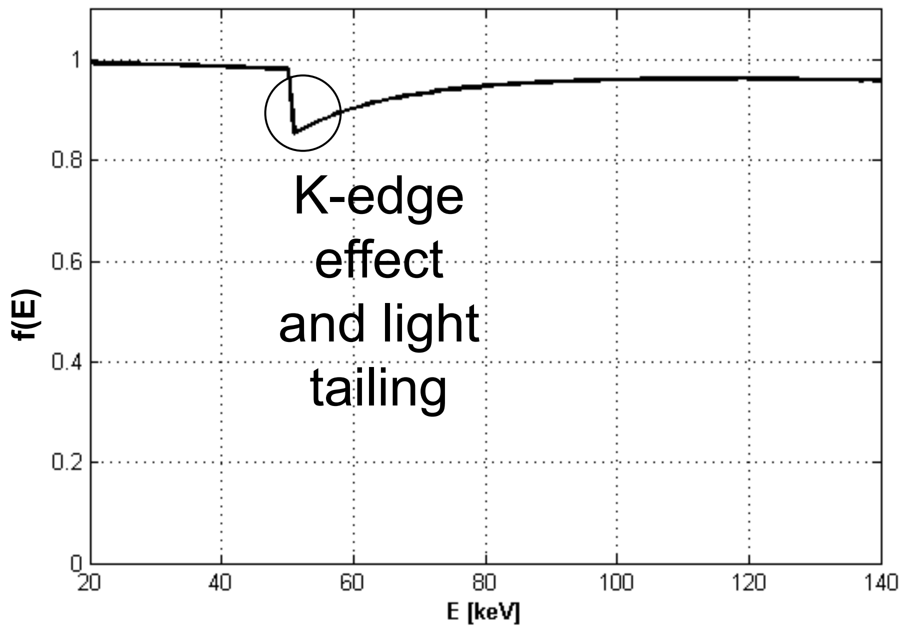
Comparing SNRs

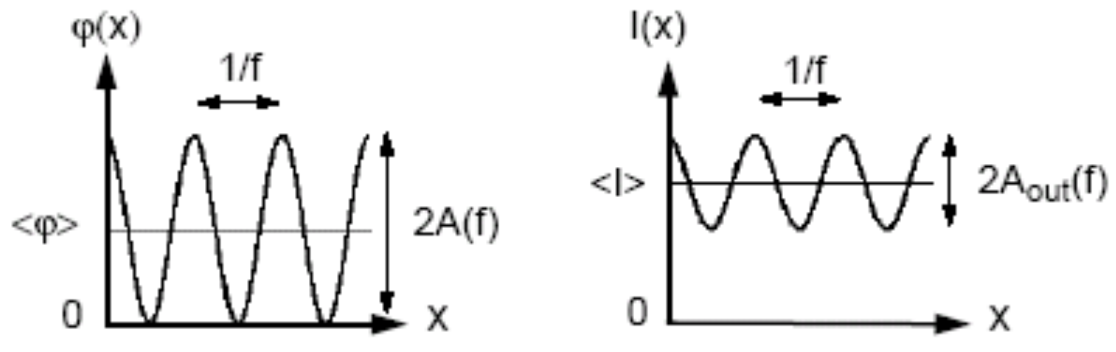
SNR decrease by signal transport variance:

$$f(E) = \frac{1}{\sqrt{\alpha(E)}} \frac{SNR_{out}}{SNR_{in}} = \frac{\langle E' \rangle}{\sqrt{\left(\langle E' \rangle^2 + \sigma^2 (E') \right)}}$$

with α(E):
detection efficiency



Modulation transfer function (MTF)



Radiation field   Image plane

MTF definition:

$$MTF(f) = \frac{\left| FT(PSF(x)) \right|}{FT(PSF(0))}$$

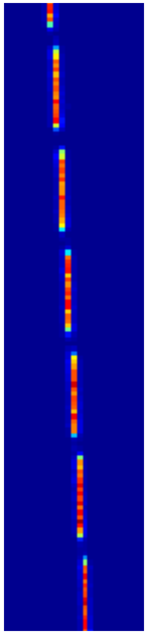
Prerequisites:

- Linear, shift-invariant (LSI) system
- Even Point Spread Function (PSF)

Both only approximately true for CT detectors

X-ray slit simulation

a)

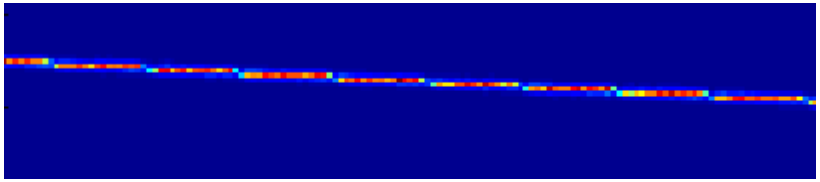


- Geometry: 100 µm slit in thick absorber
- X-ray irradition: 120 kV spectrum, high flux case

- Simulate two images with slit tilted by 3° relative to
  a) vertical and
  b) horizontal axis

- Summarize over-sampled PSFs along tilted lines
- Calculate MTFs

b)



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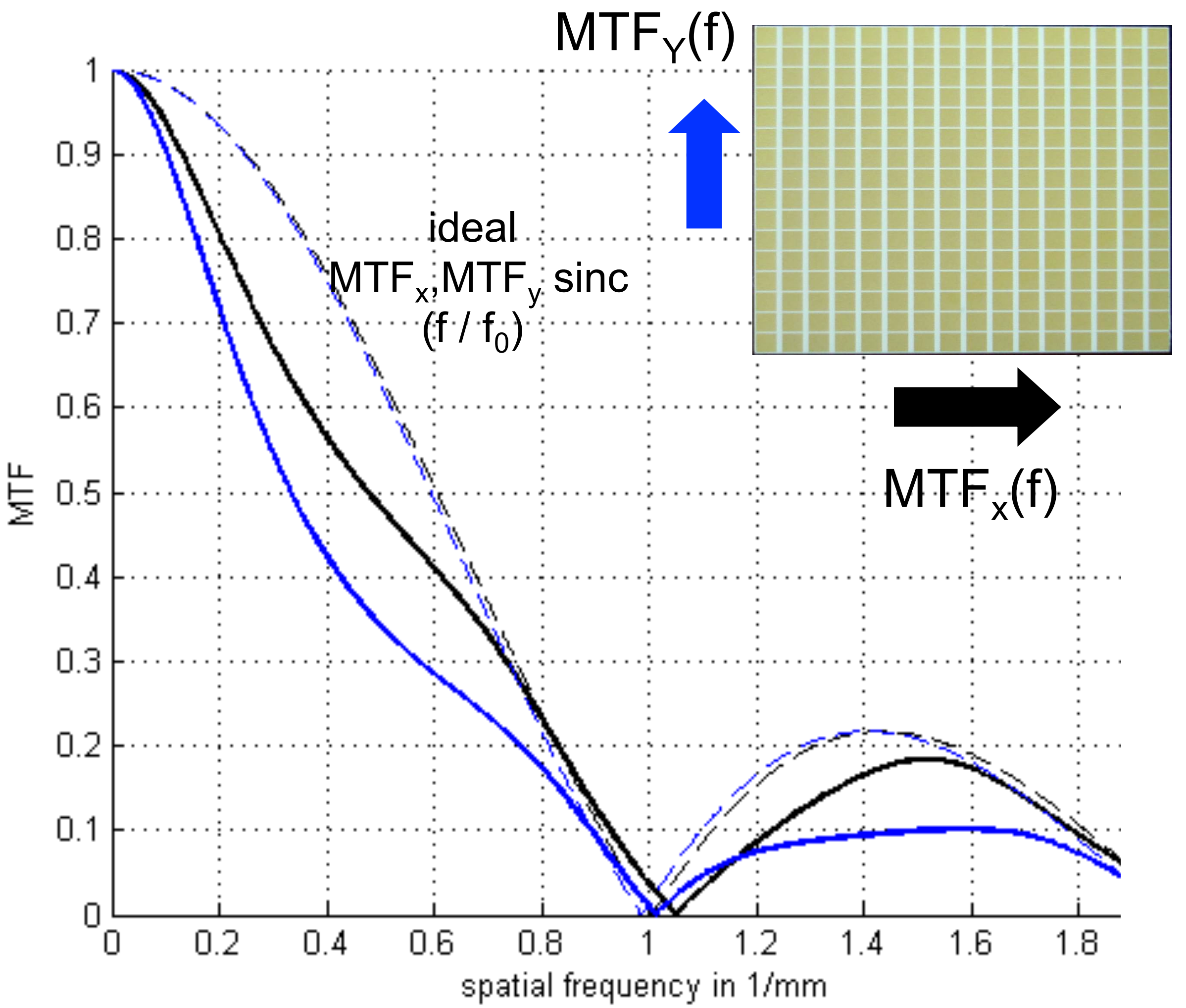
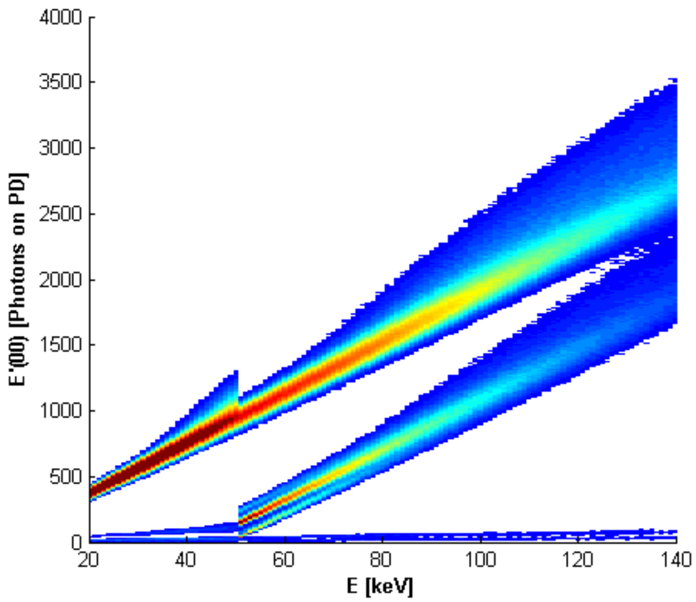
MTF results

- Detector aperture sets resolution limit
- Crosstalk reduces MTF for intermediate frequencies
- Stronger reduction in y (thinner septum)
- Compensated in CT by filtering kernels, lossless for Poisson noise characteristics



Summary I

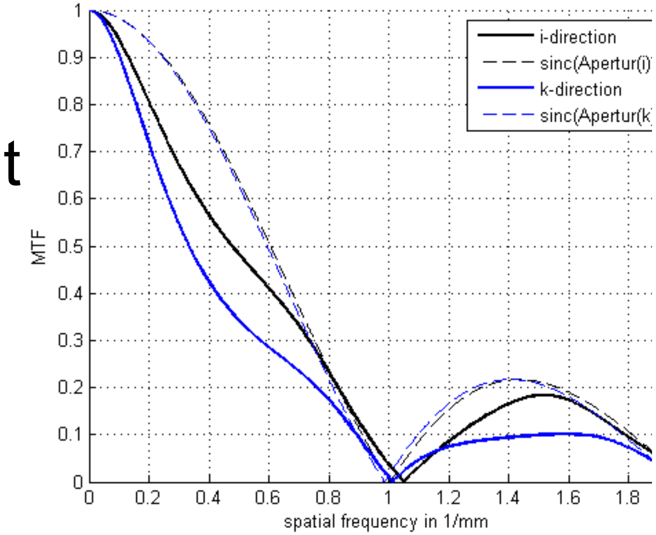
- Cascaded full 3D physical modelling of X-ray deposition and optical transport phenomena in scintillator arrays developed
- Light output measurements agree to simulation within ~5%
- Macroscopic spectral response $D^{(i,k)}(E,E')$ calculated
- $\langle E'(E)\rangle$ has significant deviations from $\propto E$
- SNR loss above K-edge

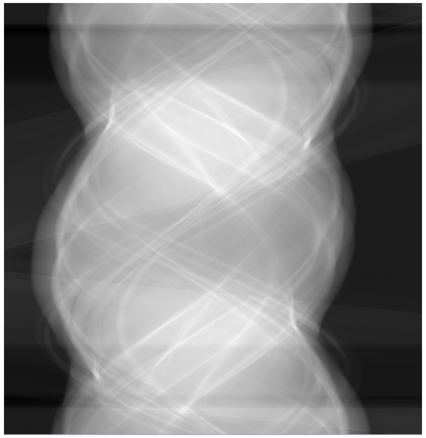


Summary II

Modulation transfer function:

- Detector aperture sets resolution limit
- MTF reduced for intermediate frequencies due to cross-talk, lossless recovery for Poisson noise characteristics
- Next steps:
  - Compare MTF results to measurements
  - Sinogram and image quality analysis with full 3D scintillator modelling





THANK YOU!