

Signal transport in Computed Tomography detectors

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Computed Tomography



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

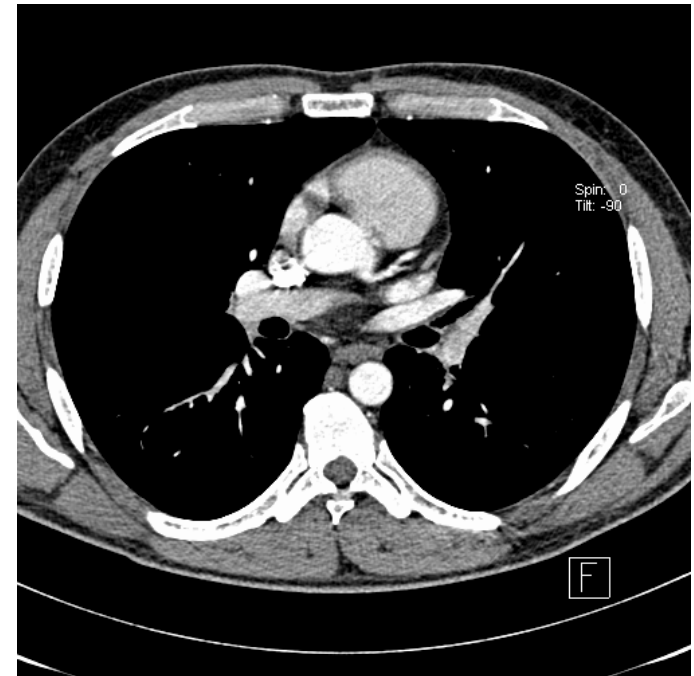
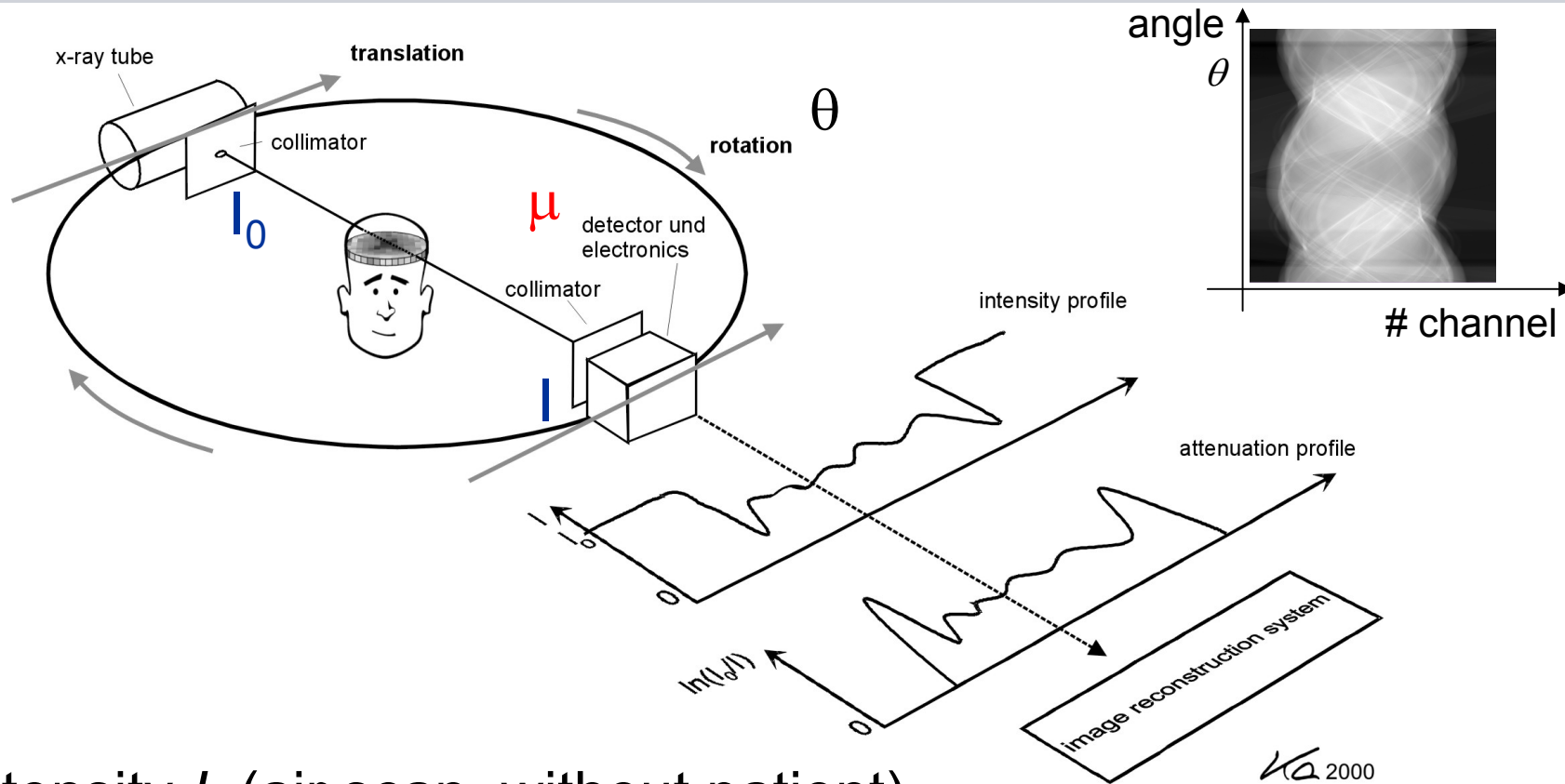


Image data
patient 3D morphology

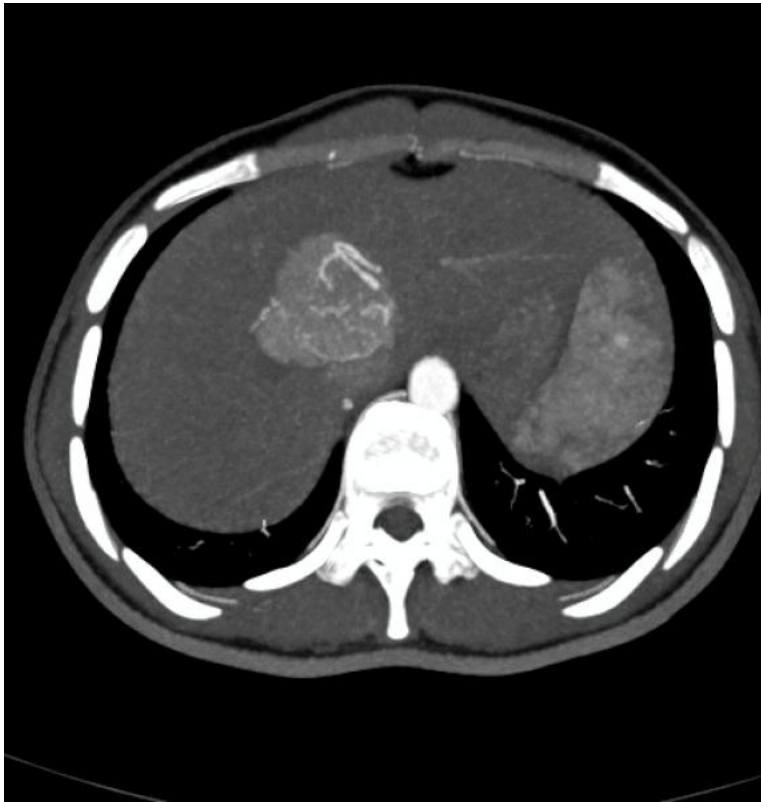
CT measurement



- 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

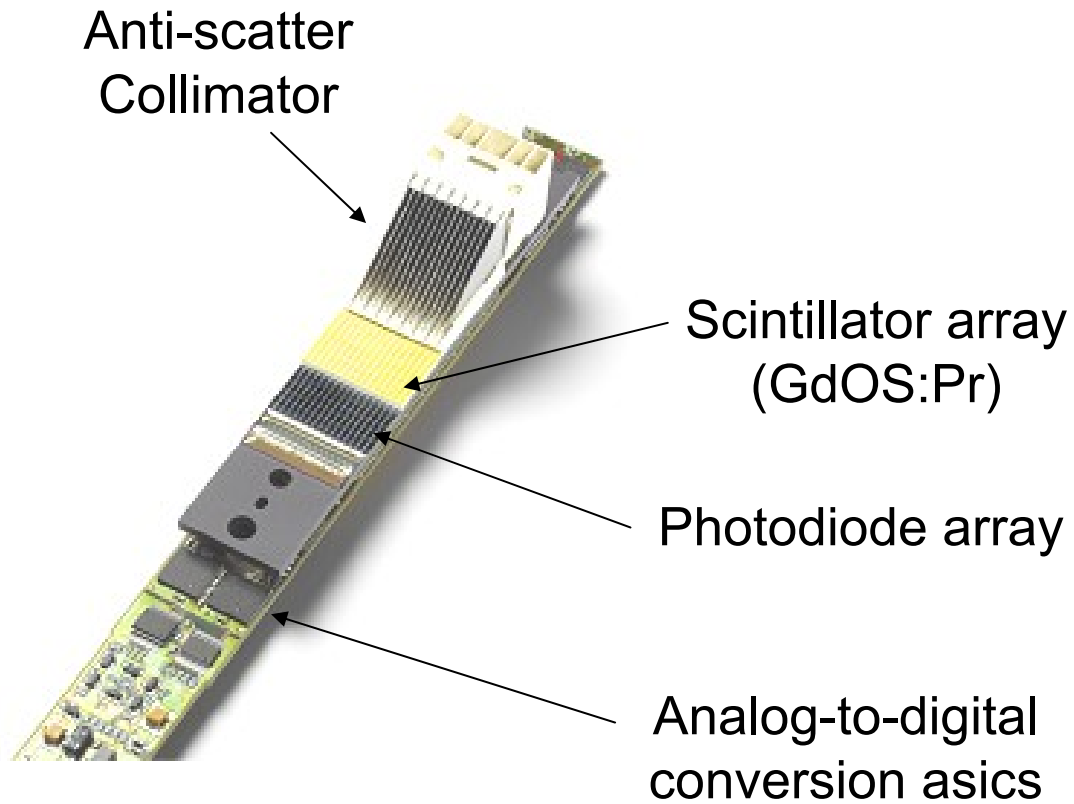
Liver tumor
contrast-to-noise



Carotid Aneurysma
resolution

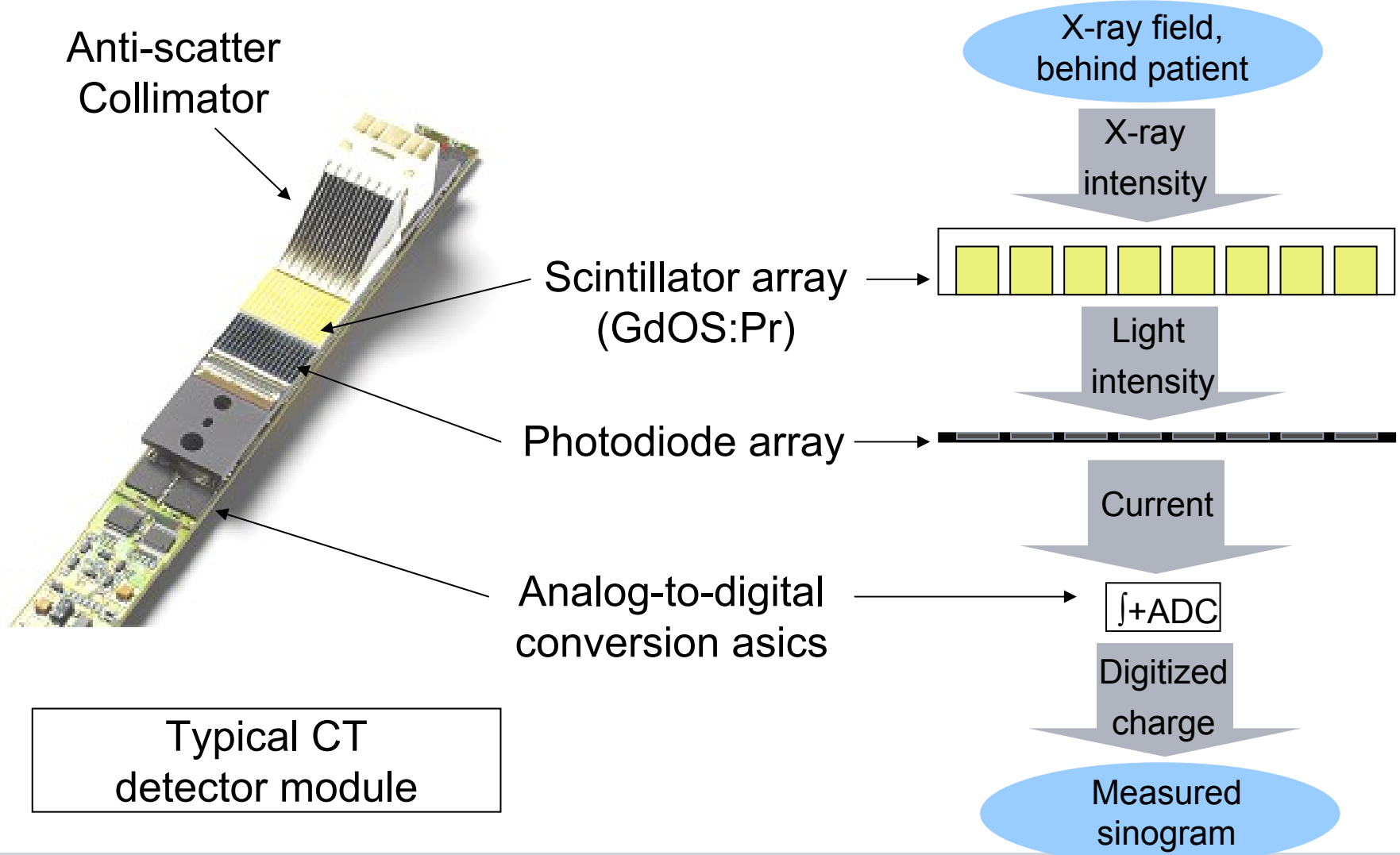


Scintillator detectors

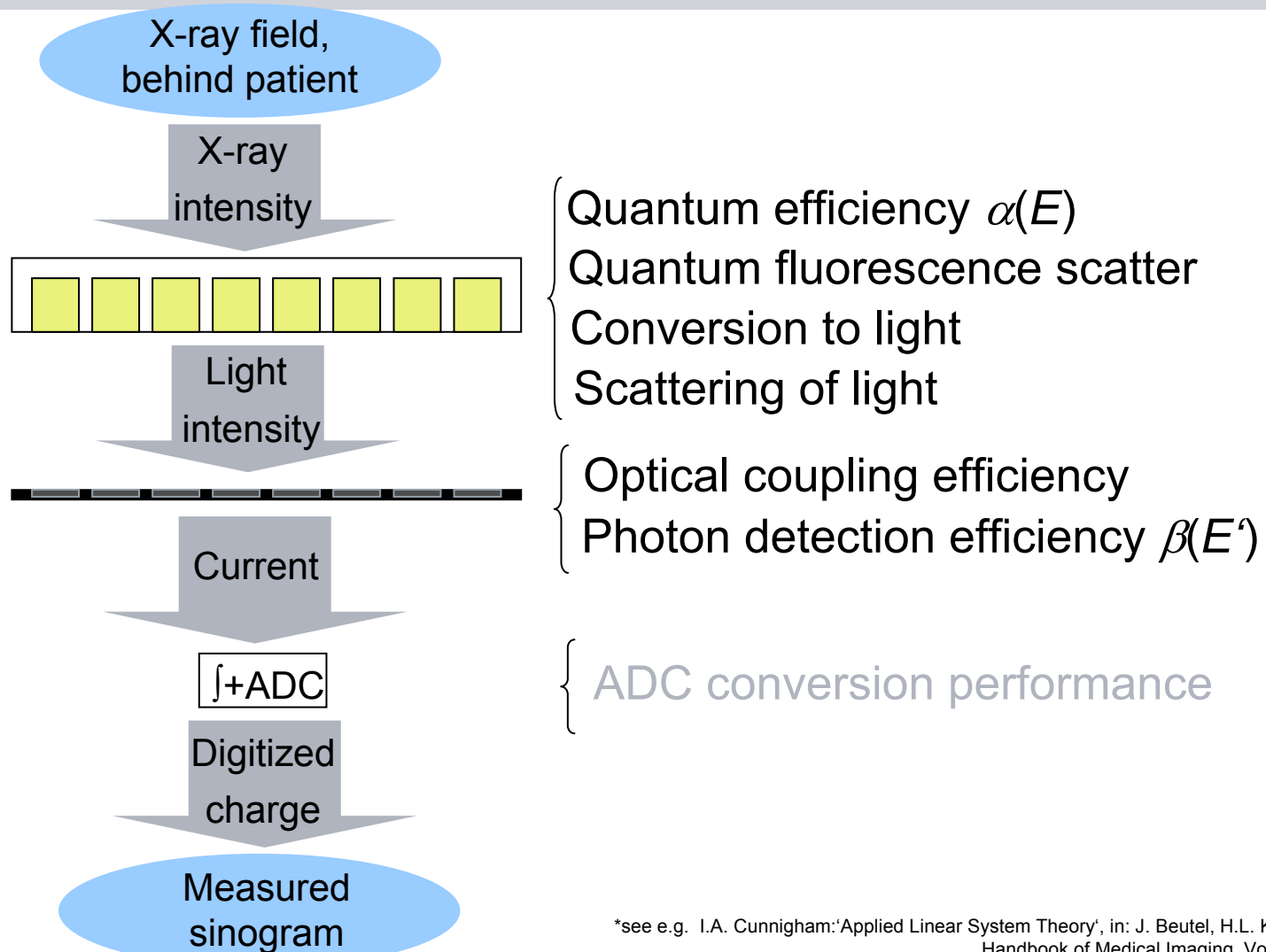


Typical CT detector module

Scintillator detectors



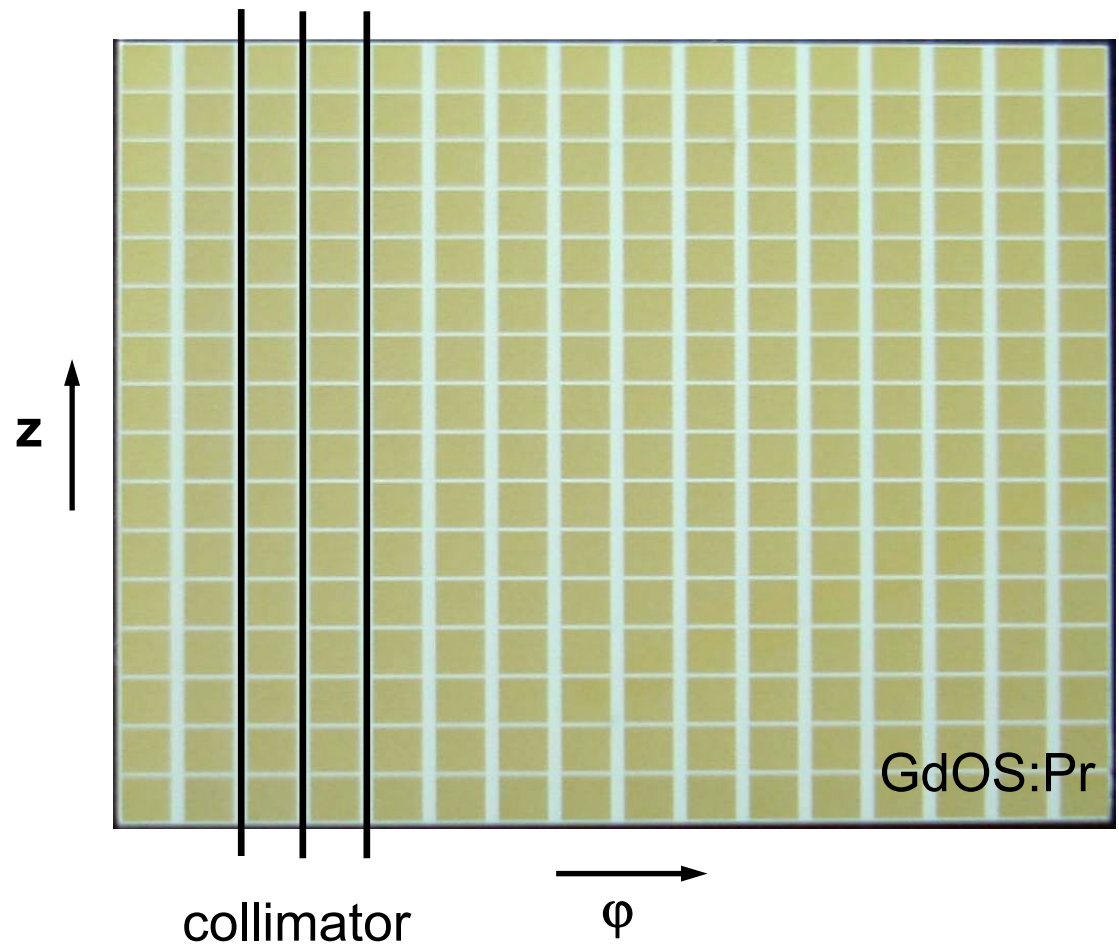
Cascaded Detection Model*



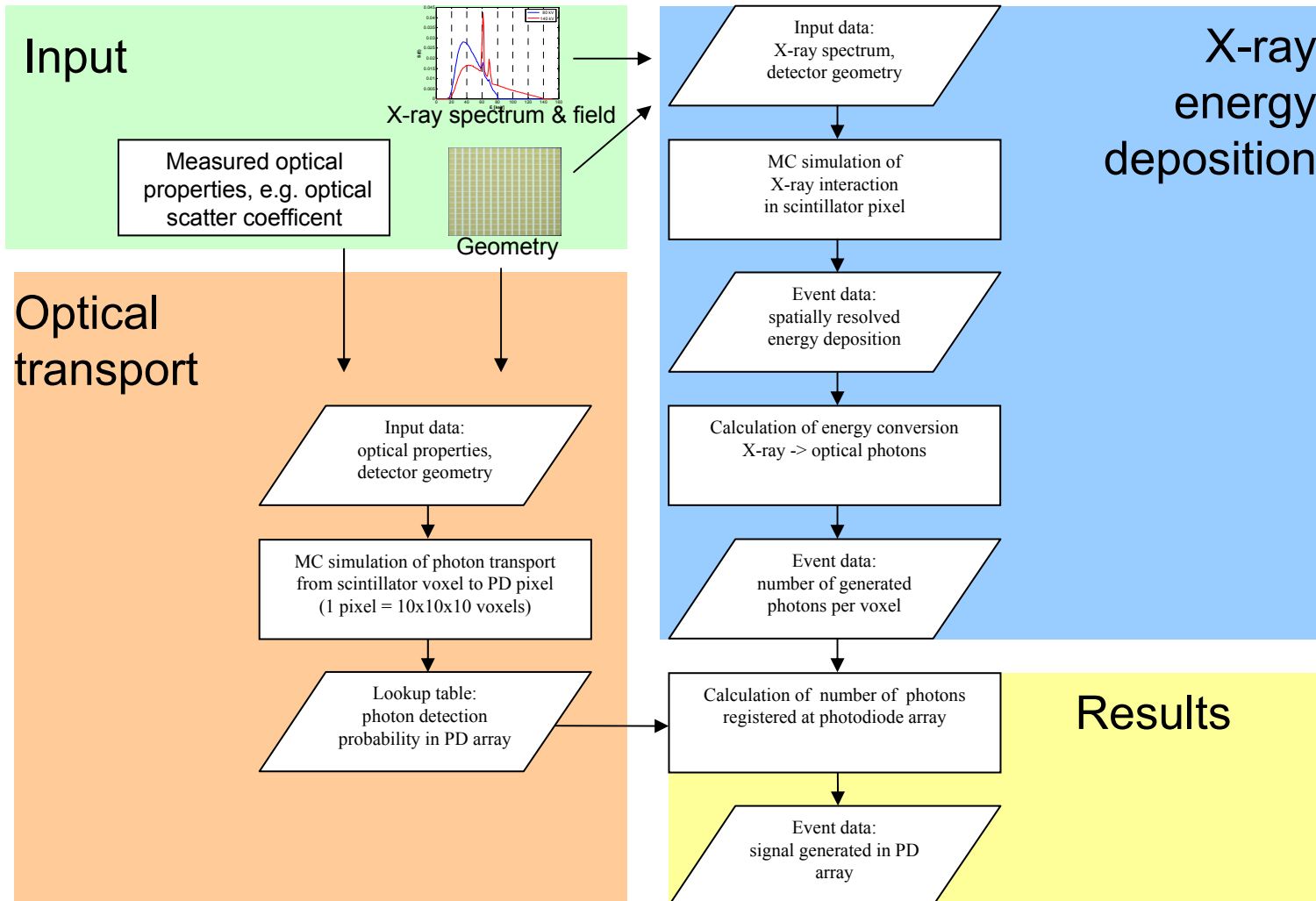
*see e.g. I.A. Cunningham: 'Applied Linear System Theory', in: J. Beutel, H.L. Kundel, R.L. van Metter, Handbook of Medical Imaging, Vol. 1, pp 79, SPIE (2000)

Reference scintillator geometry

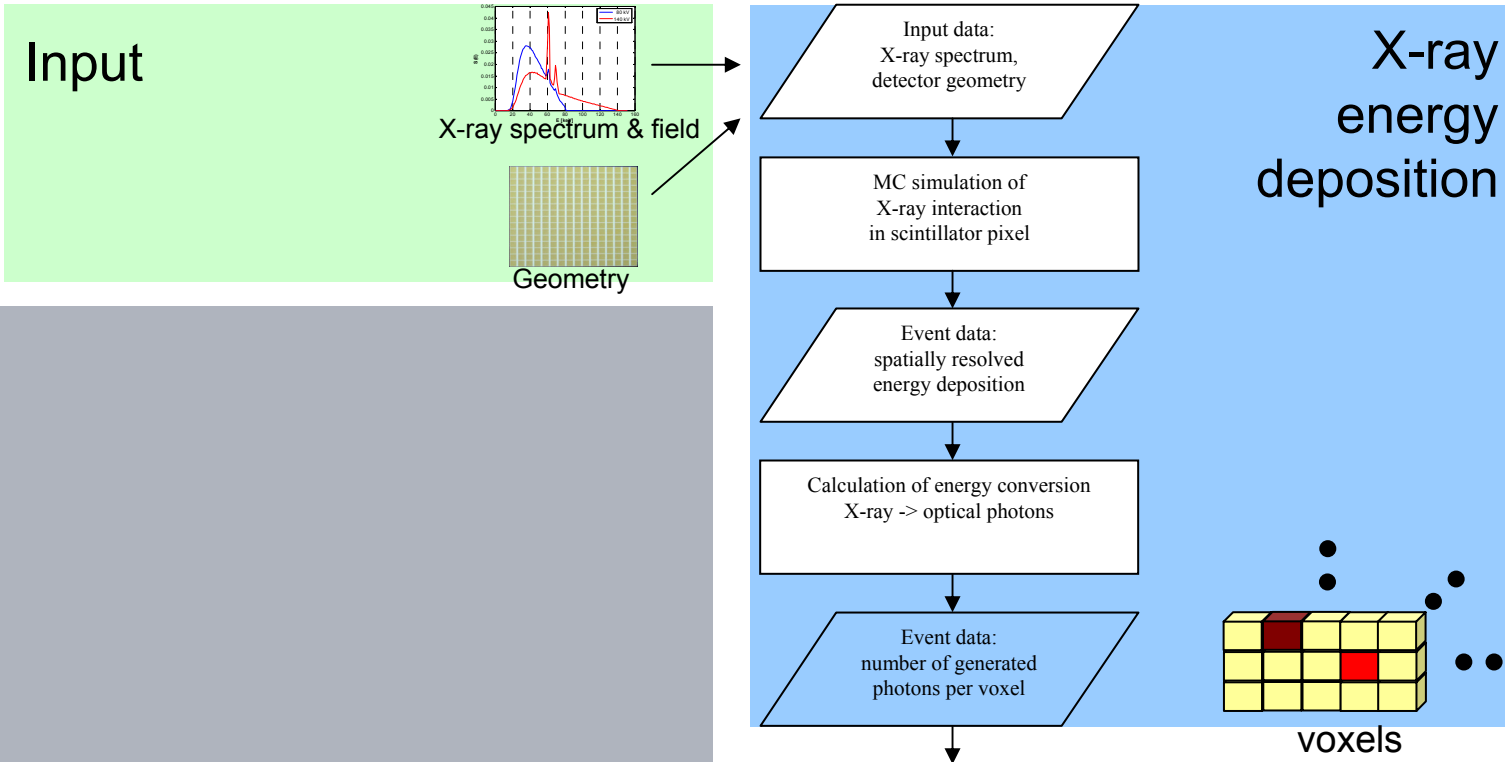
- **Pitch ϕ** ~ 1.2 mm
- **Pitch z** ~ 1 mm
- **Gap ϕ** ~ 250 μm
(incl. collimator dead-zone)
- **Gap z** ~ 80 μm



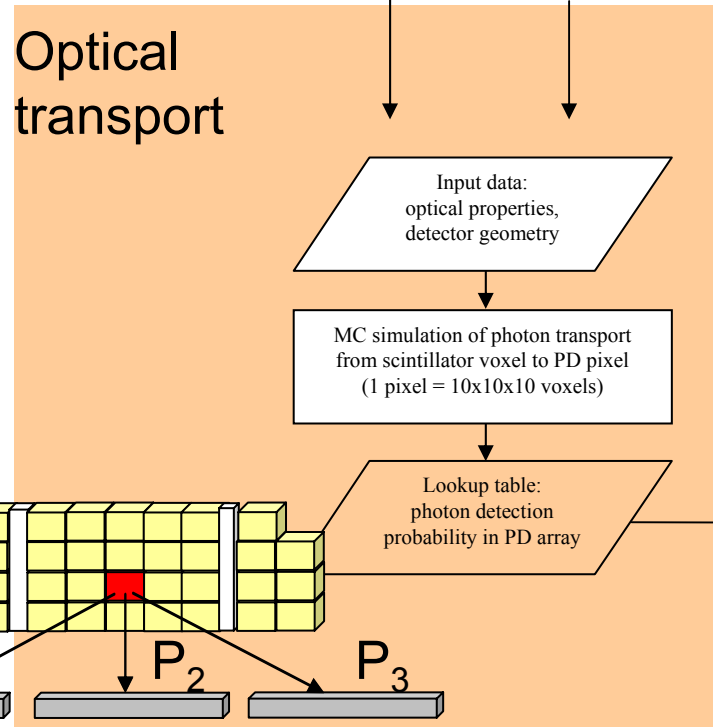
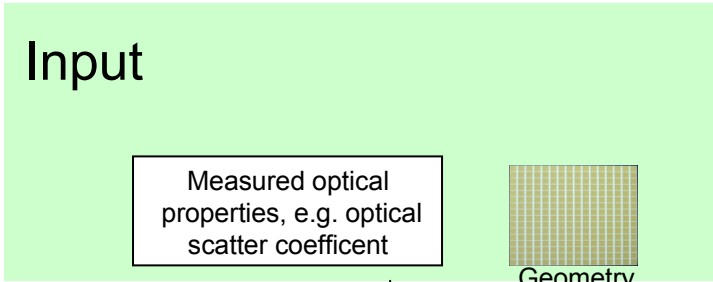
Simulation flow chart



Simulation flow chart (X-ray)

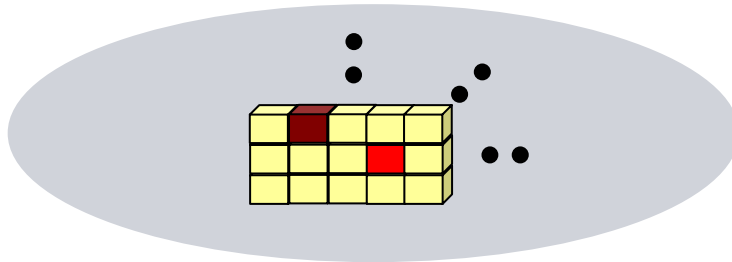


Simulation flow chart (optical)

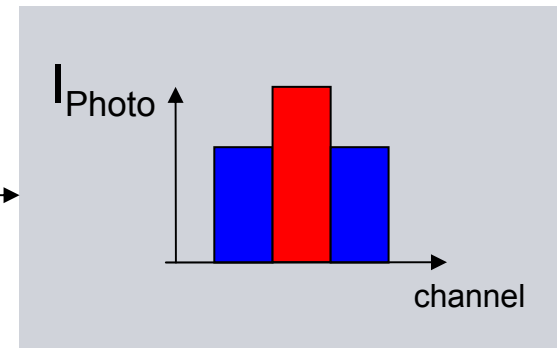
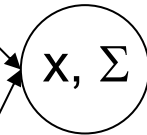
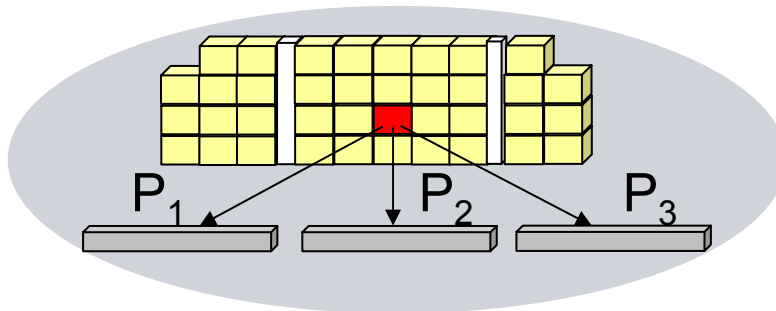


Simulation flow chart (results)

1. X-ray energy deposition:
Event list: $\#i, \langle E_i, x_i, y_i, z_i \rangle$



2. Photosensor signal:
detection probabilities

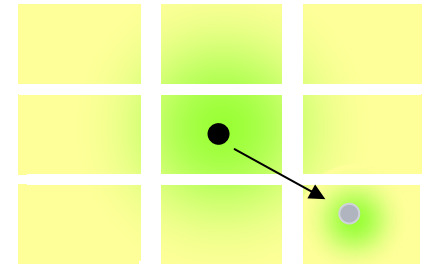


Result:
Detected number of light photons on sensor

Microscopic characteristics

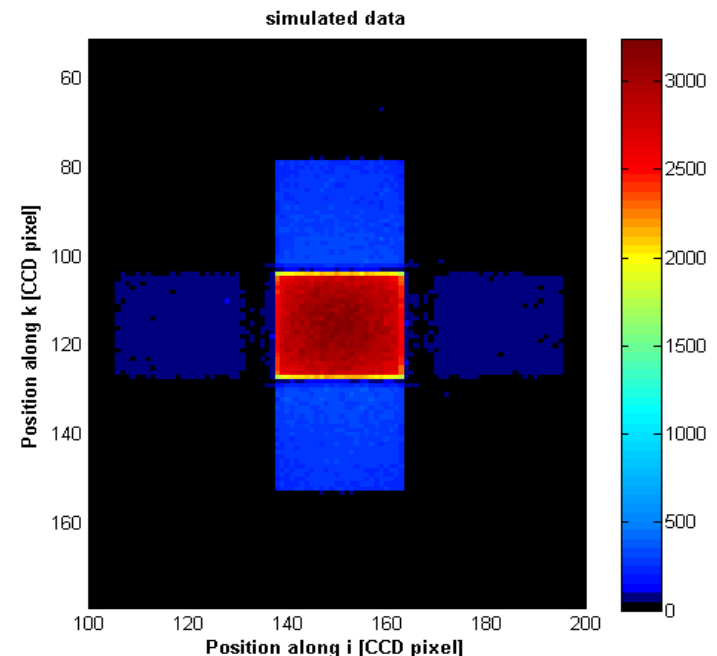
Individual quantum absorption events:

- 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

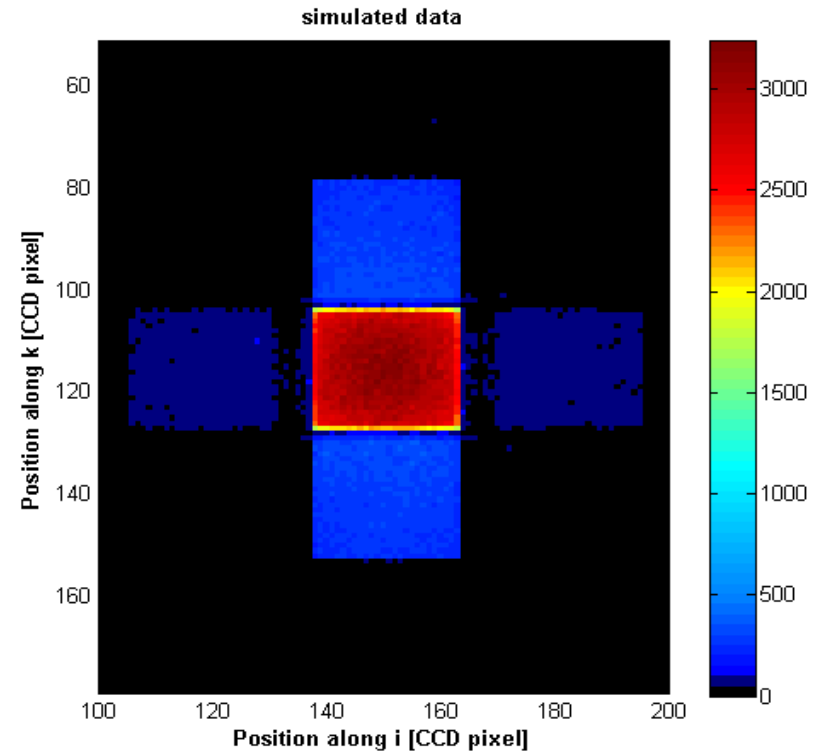
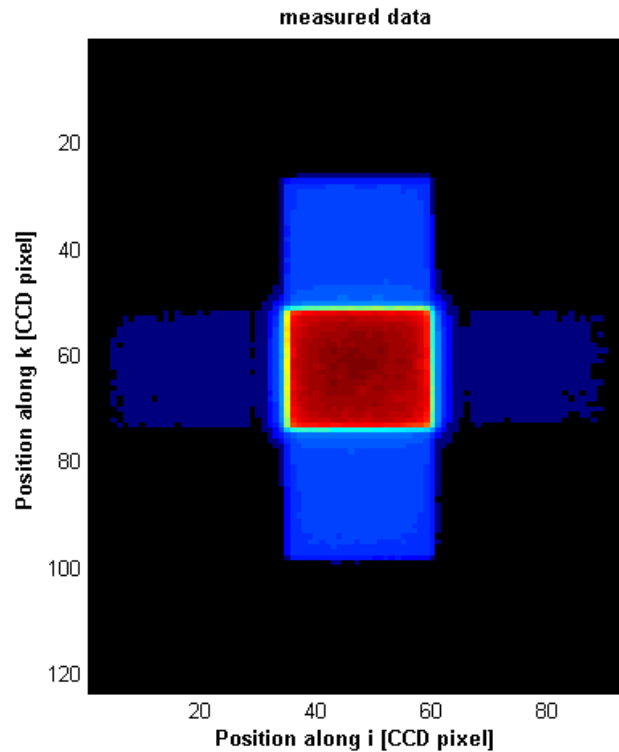


Averaged result of a 5 x 5 pixel array:

- 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



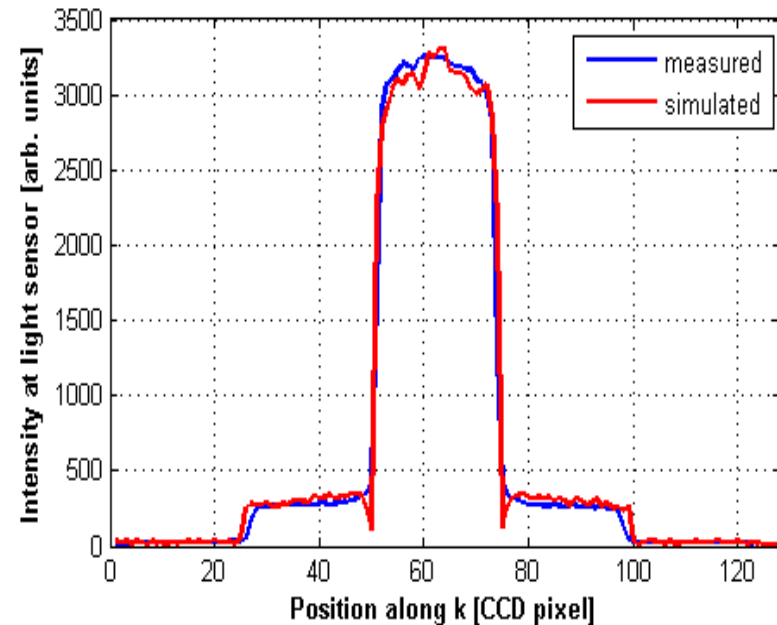
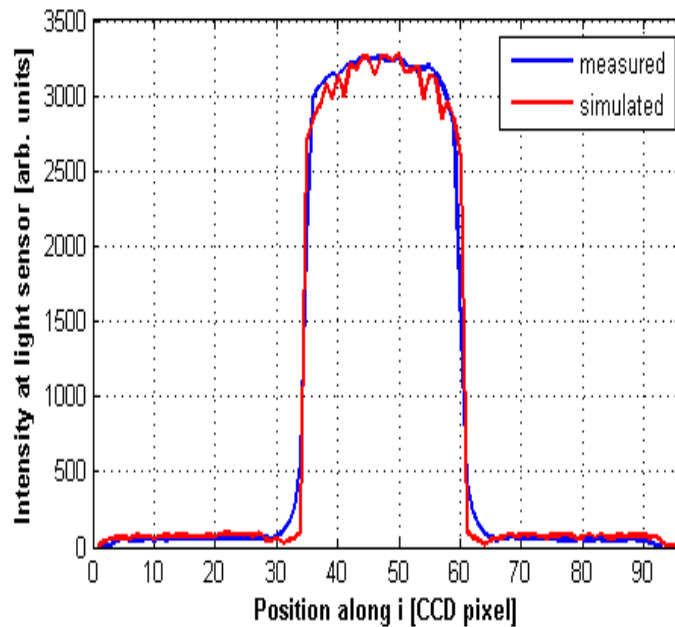
CCD measurement, fibre coupled

Simulation result

43 μm resolution

central X-ray needle beam excitation

Profiles

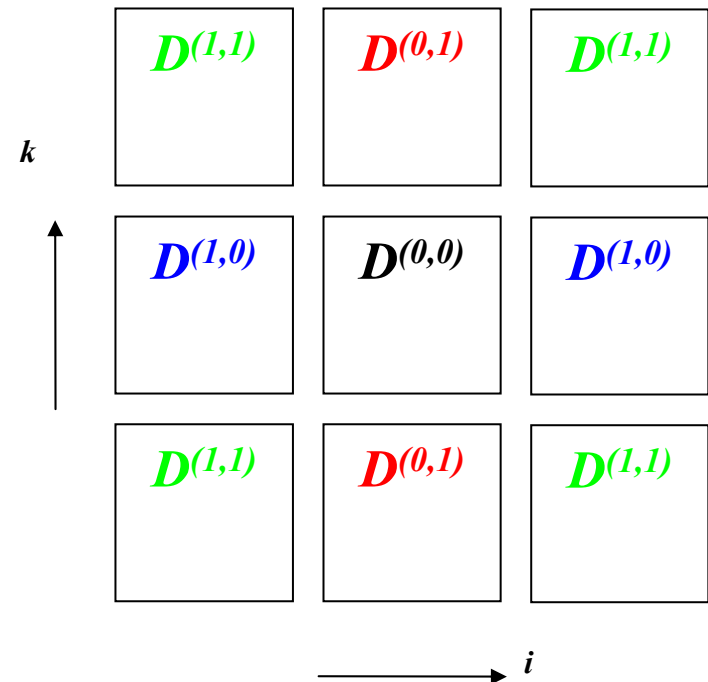


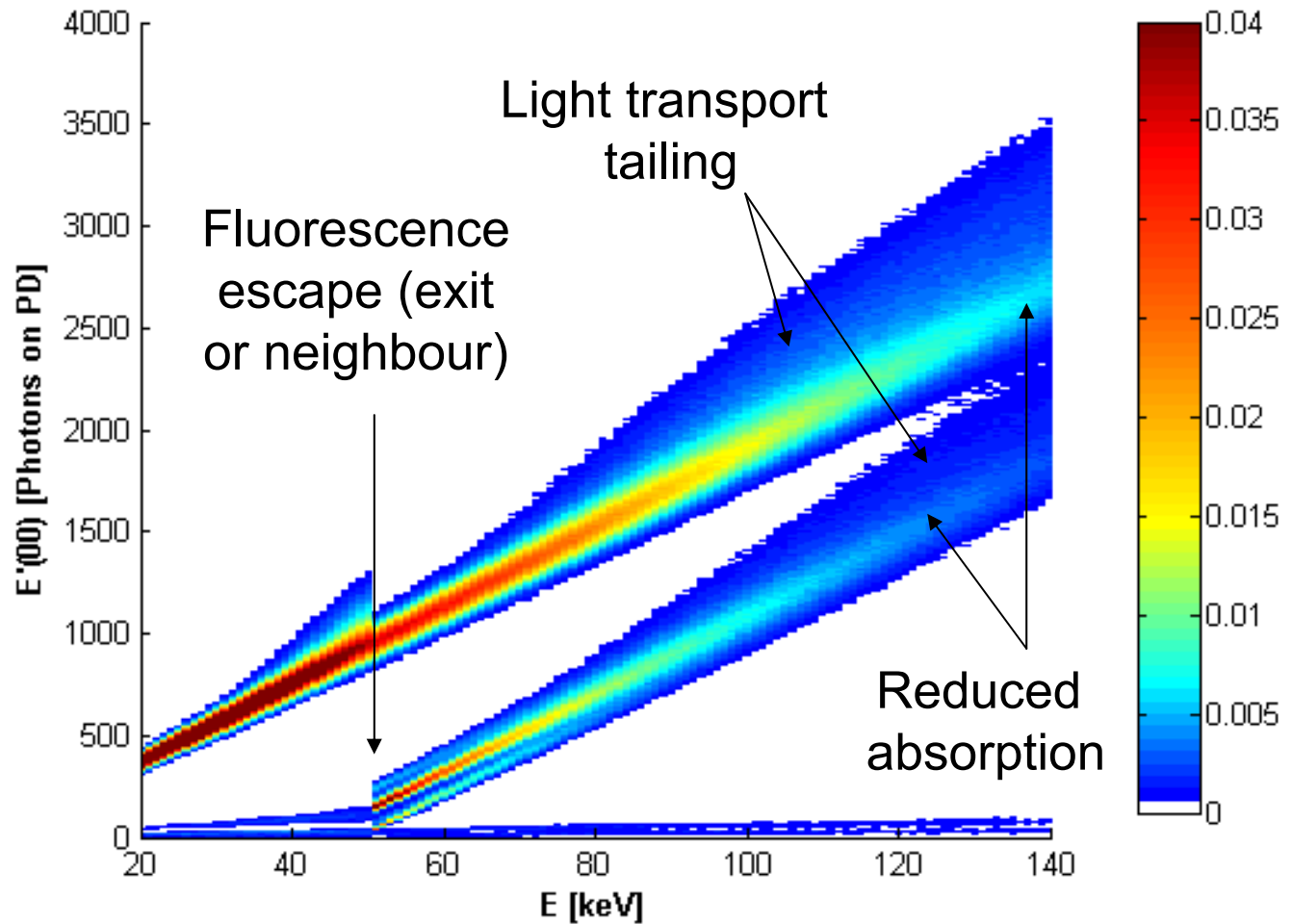
- good agreement, typical error $< \sim 5\%$
- smoother edges in measurement:
probably due to fibre coupling cross-talk not included in simulation

Macroscopic (average) figure

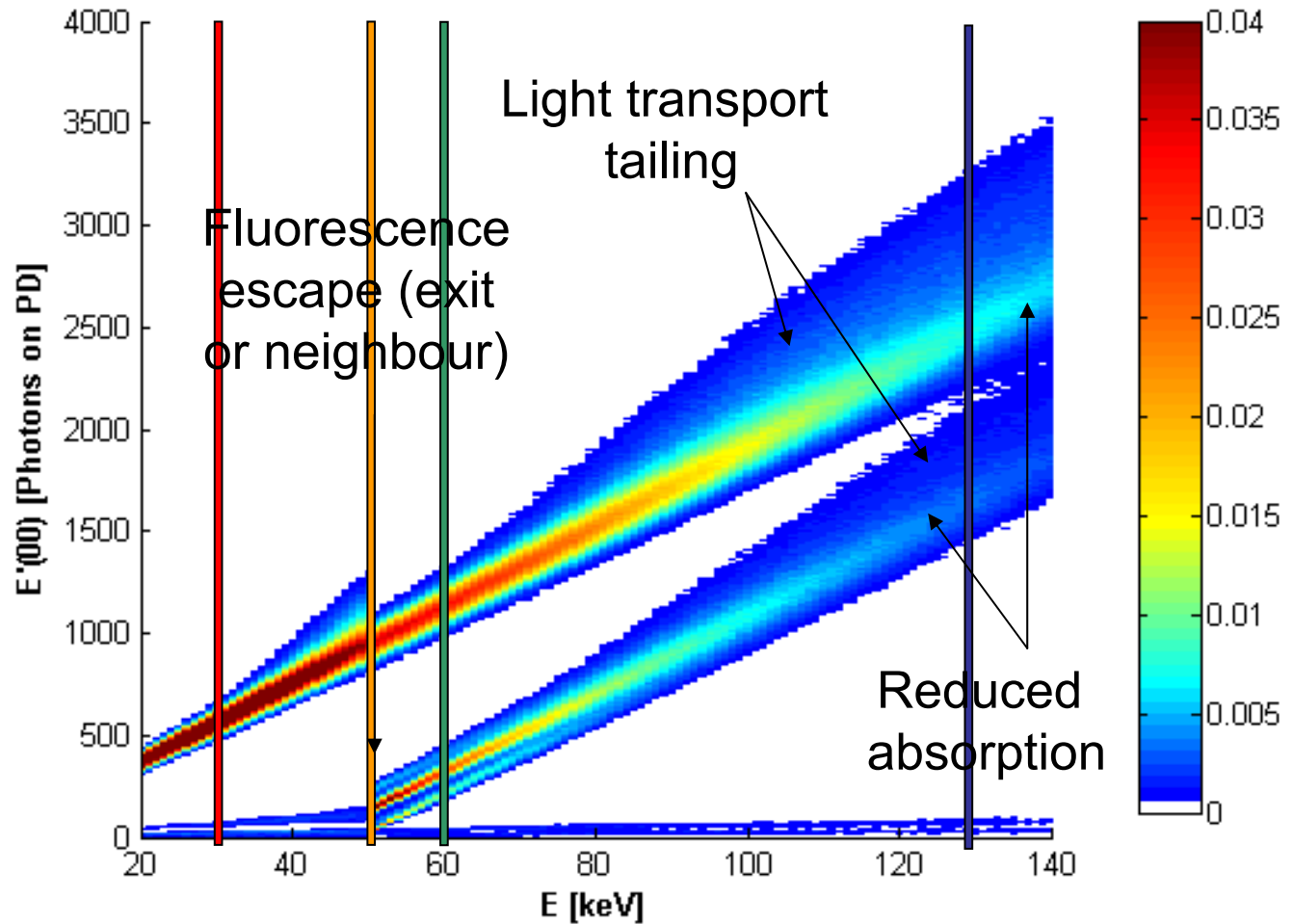
Detector response function $D^{(i,k)}(E, E')$

- Probability density function for measuring output energy E' at input energy E
- i, k denote pixel class (symmetry):
 - $i = 0, k = 0$: central pixel
 - $i = 1, k = 0, \dots$: horizontal next neighbours
 - $i = 0, k = 1, \dots$: vertical next neighbours
 - $i = 1, k = 1, \dots$: diagonal neighbours

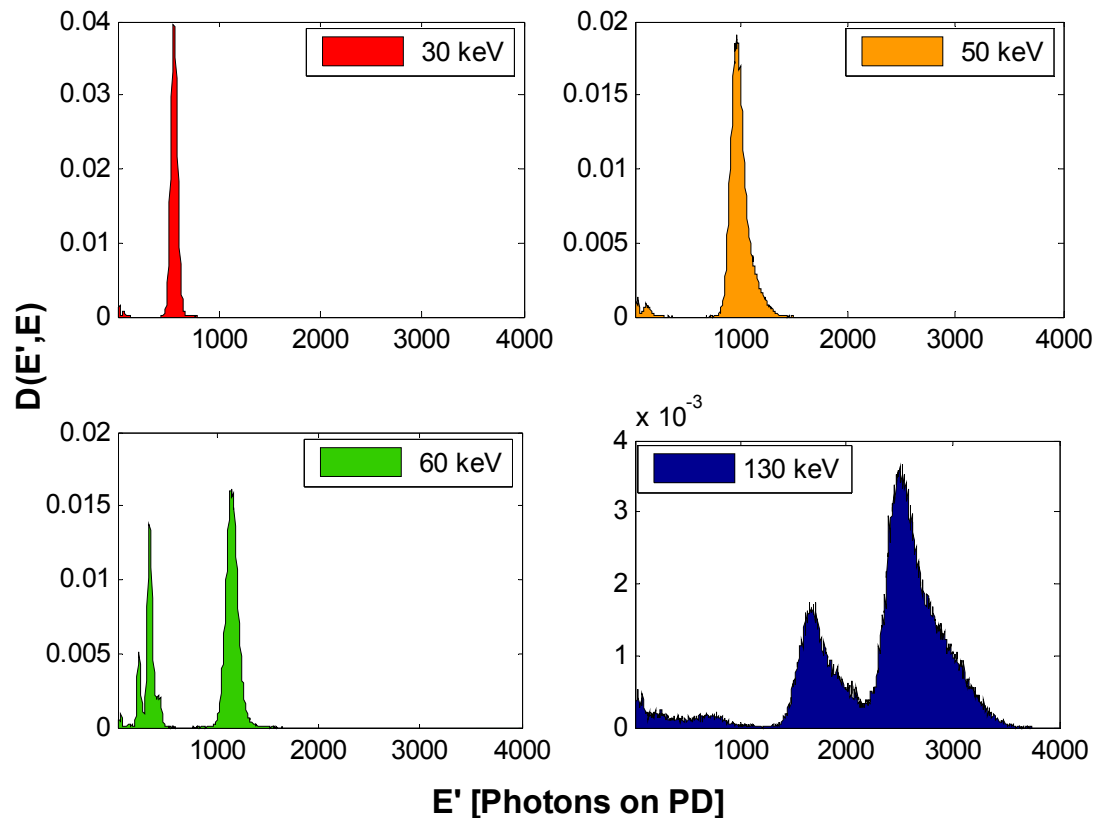


Central pixel: $D^{(0,0)}(E, E')$ 

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



Behaviour at constant input energies E



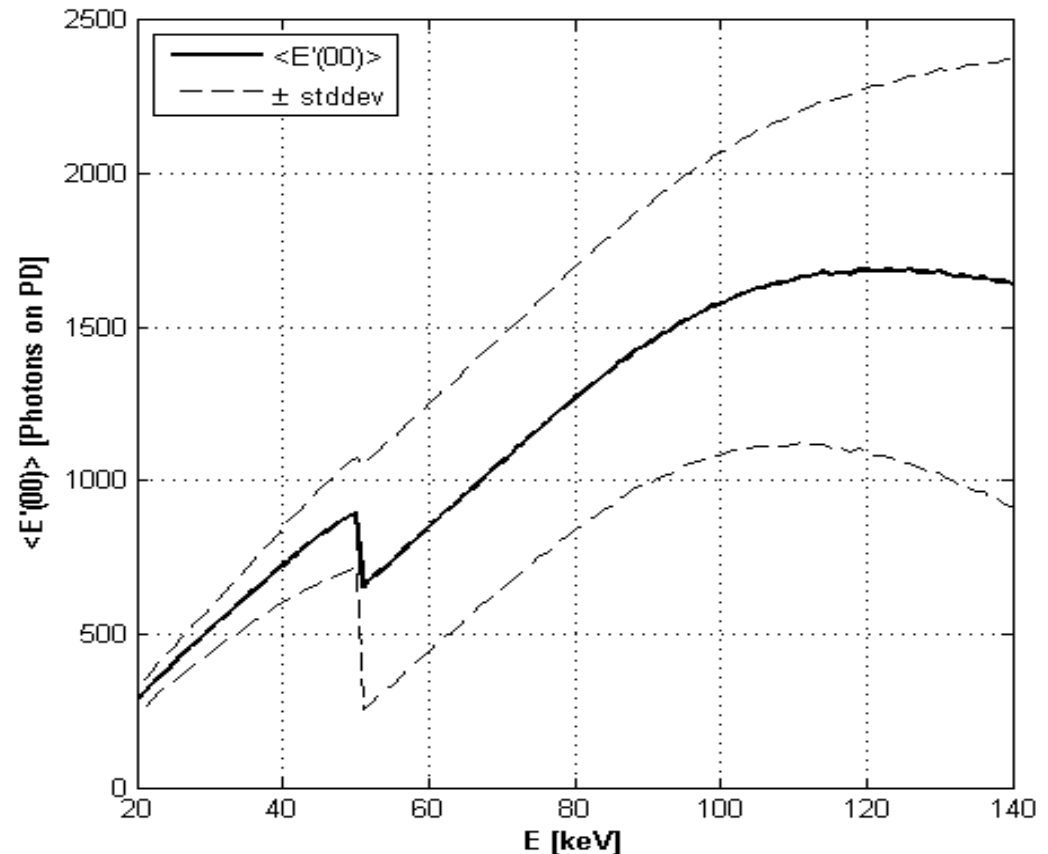
- Two peaks occur above the Gd K-edge
- Signal tailing increases with input energy due to longer light paths

Expected signal $\langle E'(E) \rangle$

$\langle E'(E) \rangle$ 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*):

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

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

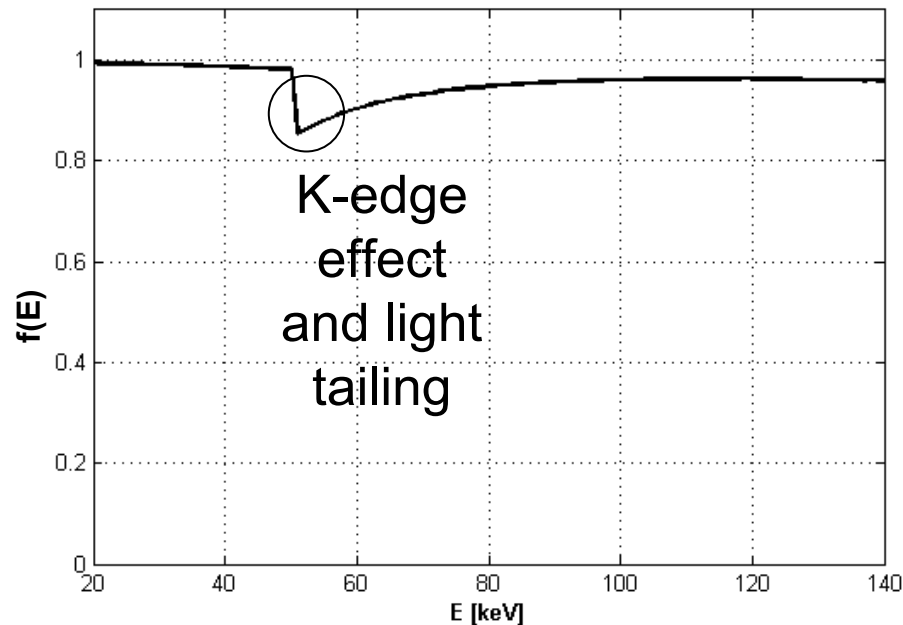
*) M. Rabbani, R. Shaw, and R. Van Metter, 'Detective quantum efficiency of imaging systems with amplifying and scattering mechanisms,' *J. Opt. Soc. Am. A*, vol. 4, pp. 895-901, 1987

Comparing SNRs

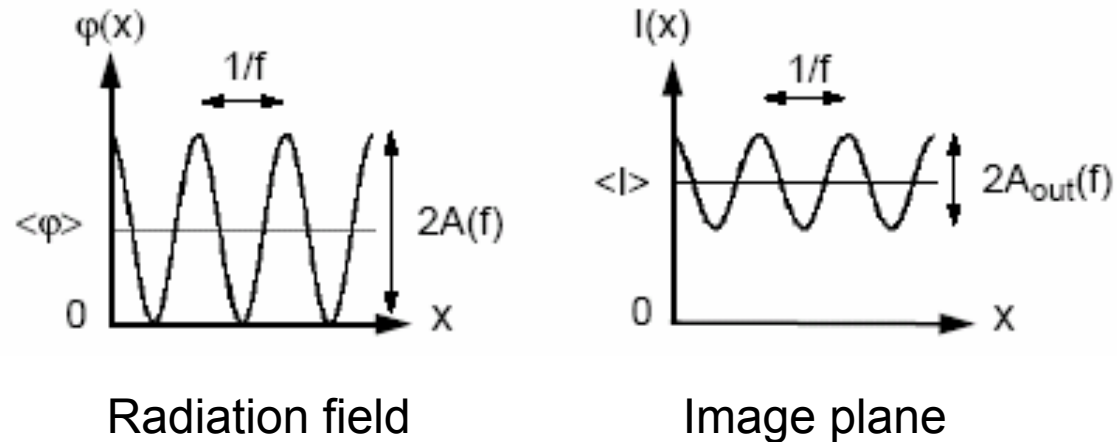
SNR decrease by signal transport variance:

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

with $\alpha(E)$:
detection efficiency



Modulation transfer function (MTF)



MTF definition:

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

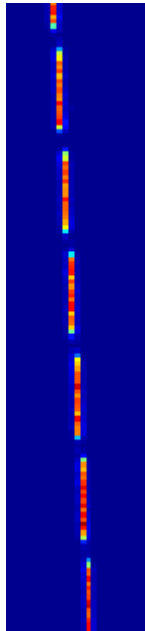
Prerequisites:

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

Both only approximately true for CT detectors

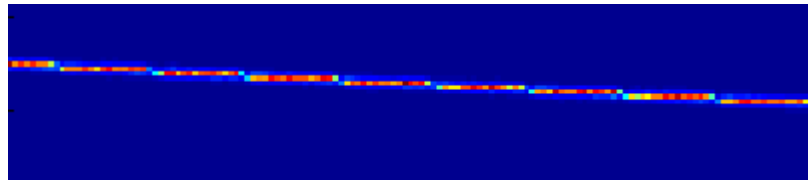
X-ray slit simulation

a)



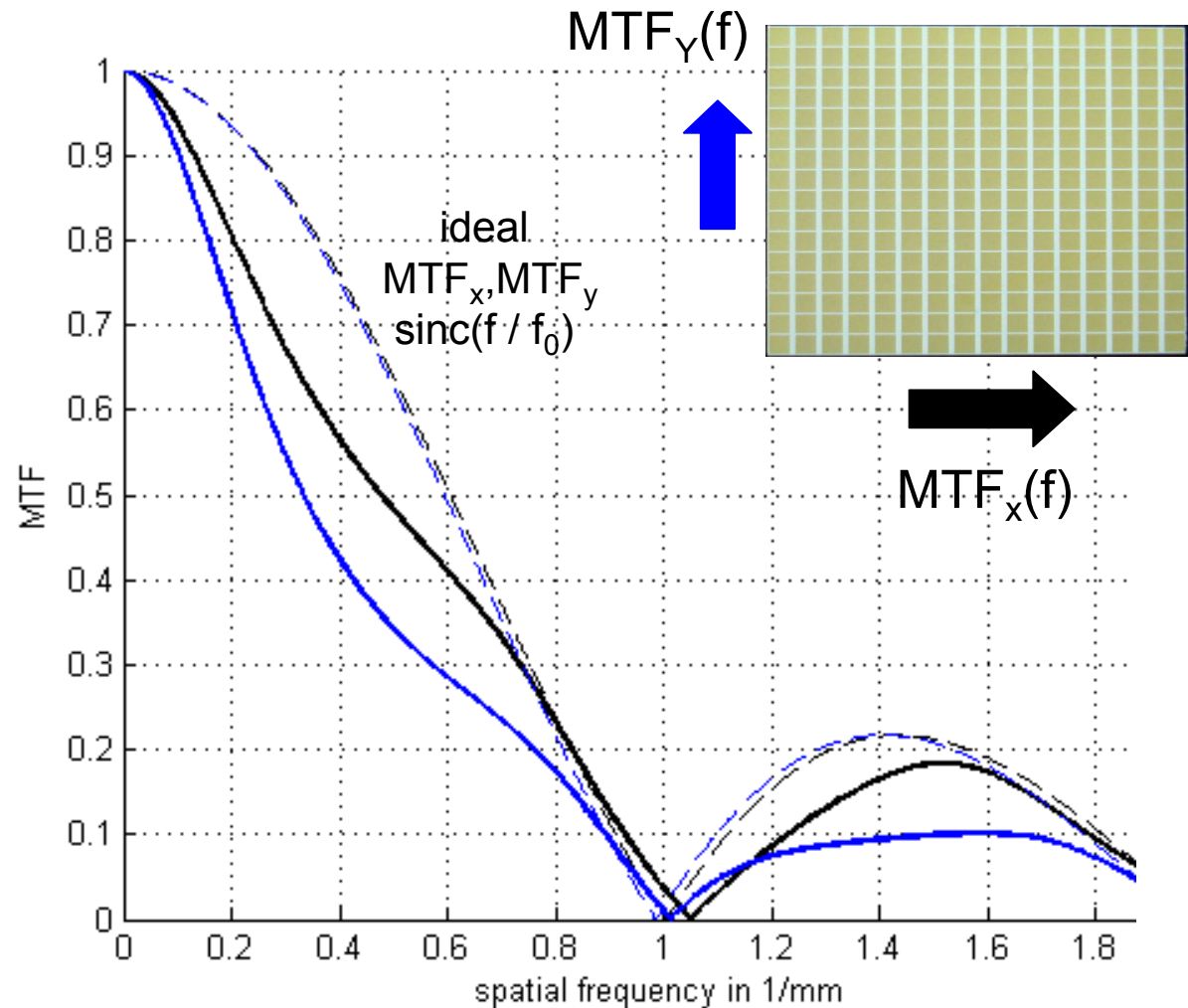
- Geometry: 100 μm slit in thick absorber
- X-ray irradiation: 120 kV spectrum, high flux case
- Simulate two images with slit tilted by 3° relative to
 - a) vertical and
 - b) horizontal axis
- Summarize over-sampled PSFs along tilted lines
- Calculate MTFs

b)



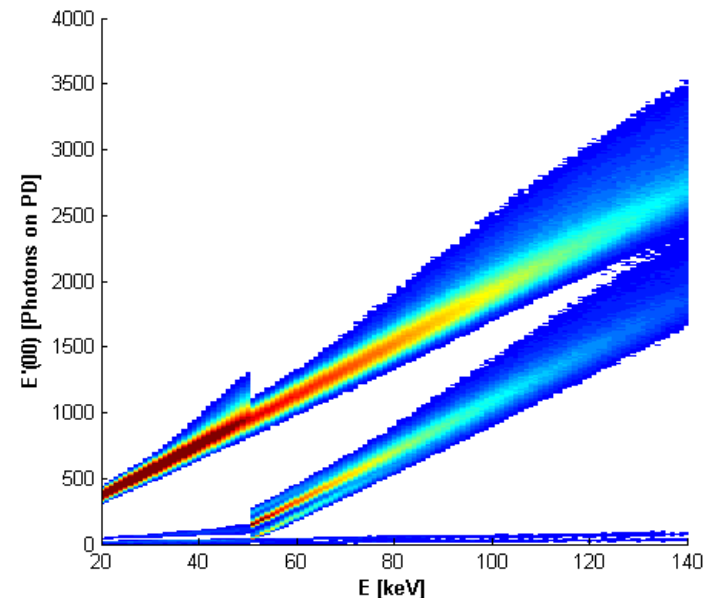
MTF results

- 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

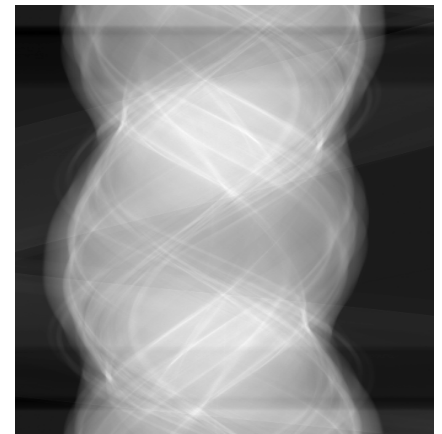
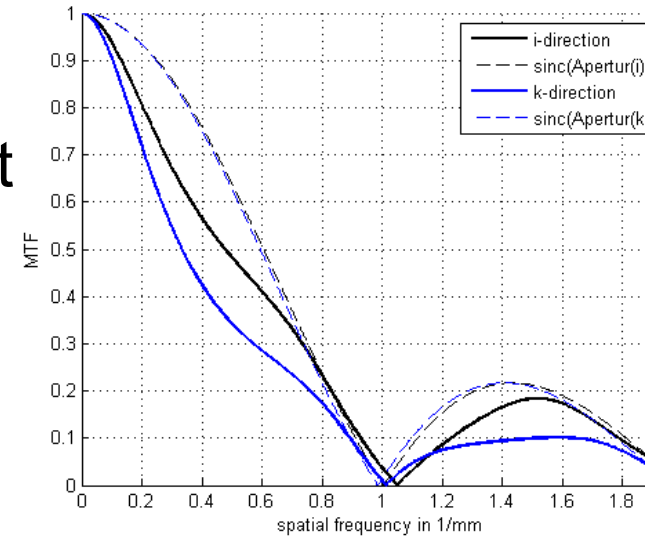
- Cascaded full 3D physical modelling of X-ray deposition and optical transport phenomena in scintillator arrays developed
- Light output measurements agree to simulation within ~5%
- Macroscopic spectral response $D^{(i,k)}(E, E')$ calculated
- $\langle E'(E) \rangle$ has significant deviations from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