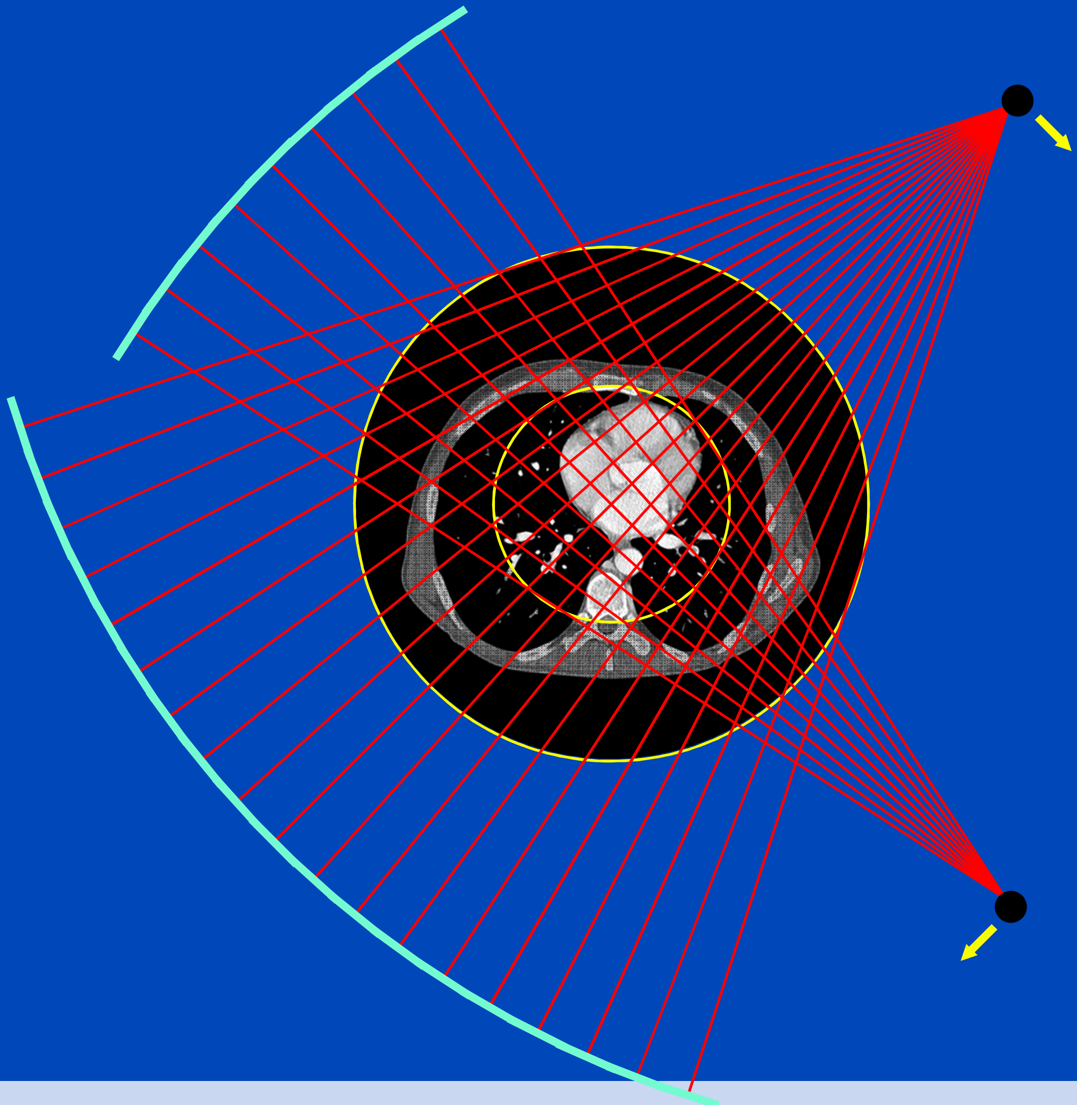
Photon Counting CT

Marc Kachelrieß

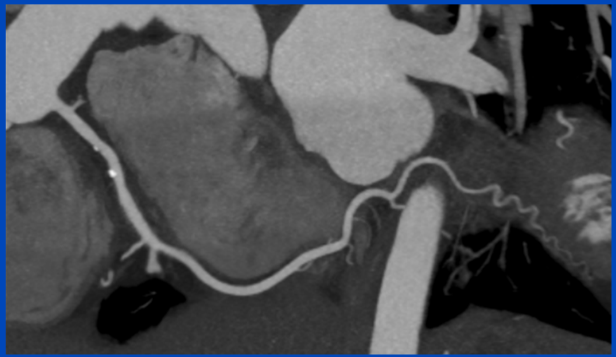
German Cancer Research Center (DKFZ),
Heidelberg, Germany

DEUTSCHES
KREBSFORSCHUNGSZENTRUM
IN DER HELMHOLTZ-GEMEINSCHAFT

Dual-Source-CT (since 2005)



Siemens SOMATOM Force
3^{rd} generation
dual source cone-beam spiral CT

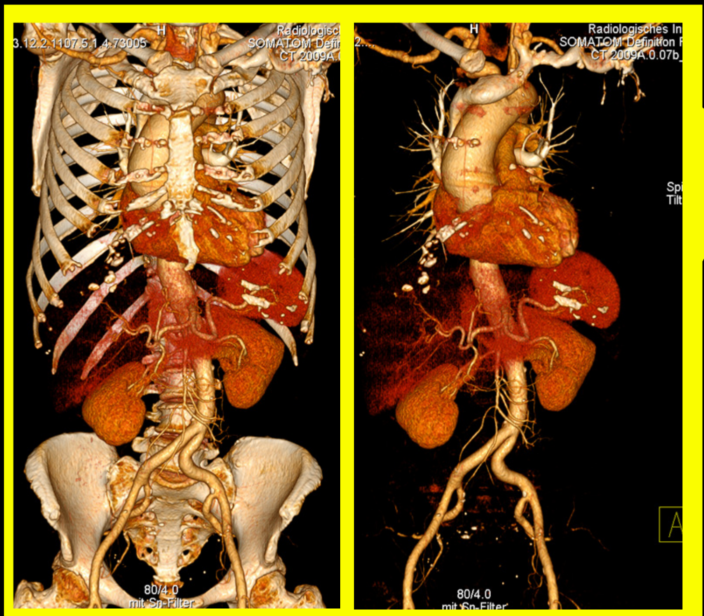


Turbo Flash, 70 kV, 0.55 mSv
63 ms temporal resolution
143 ms scan time

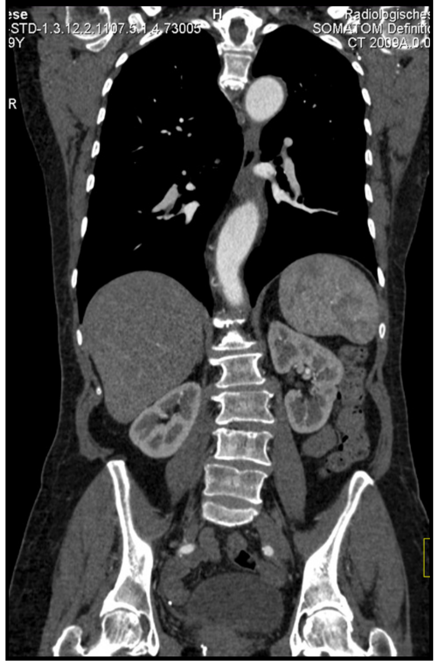
Examples

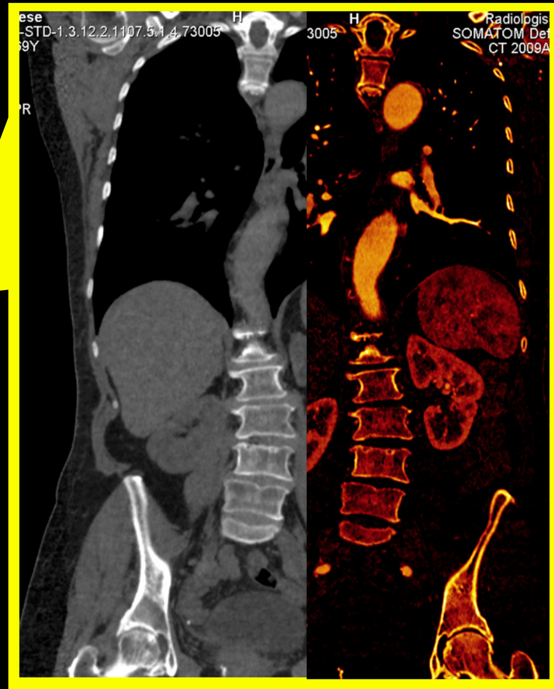
(Slide Courtesy of Siemens Healthcare)

DE bone removal



Single DECT Scan

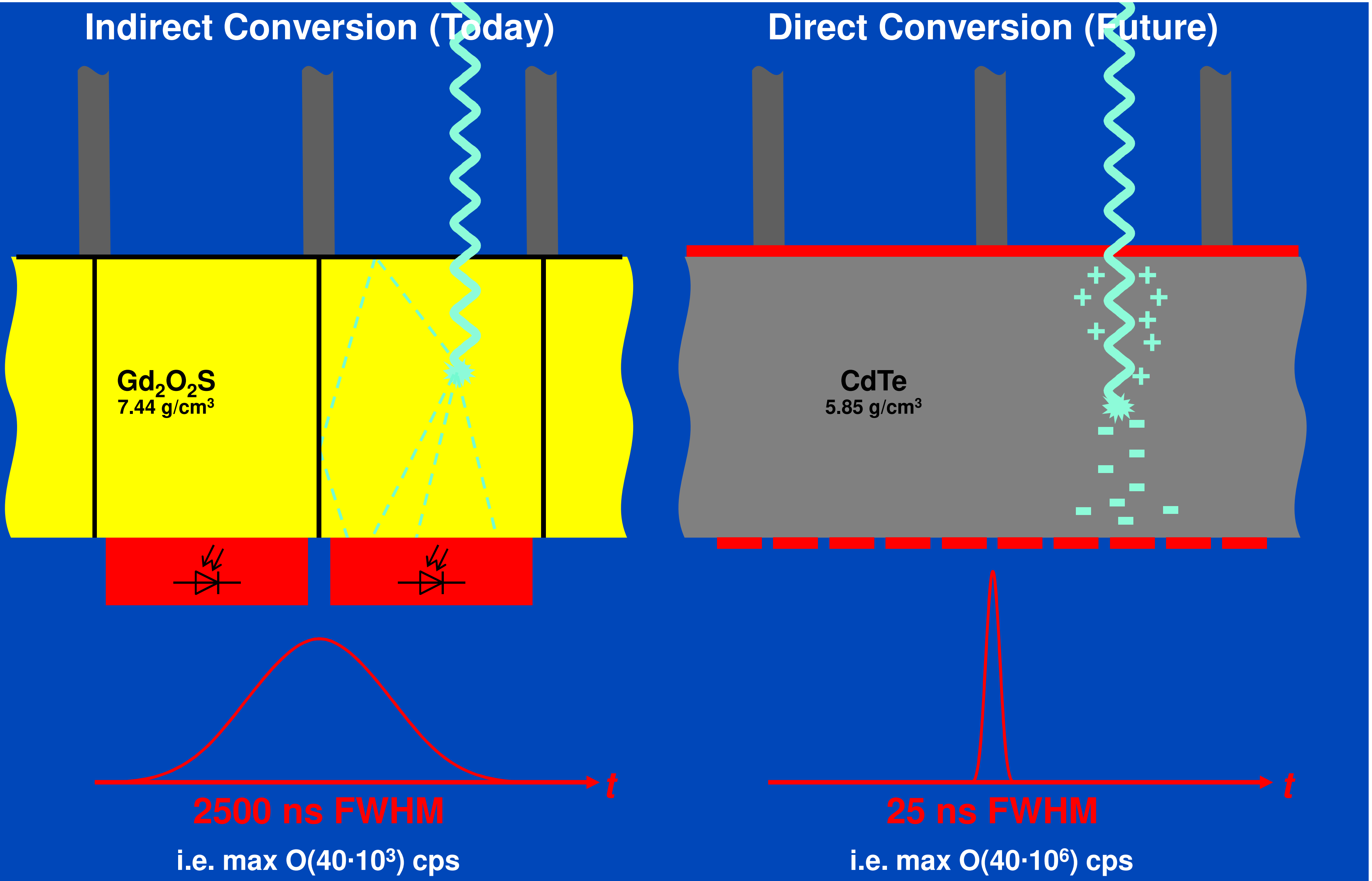




Virtual non-contrast and Iodine image

Dual Energy whole body CTA: 100/140 Sn kV @ 0.6mm

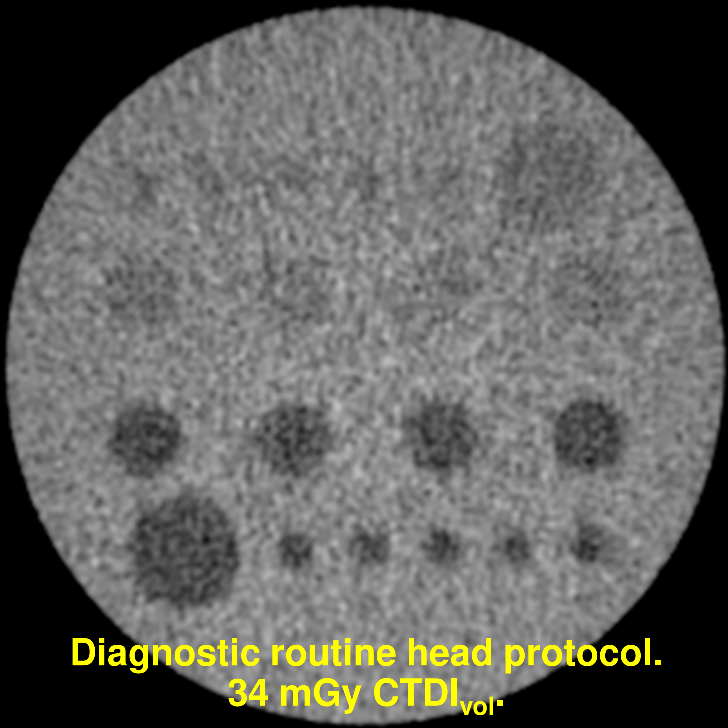
Courtesy of Friedrich-Alexander University Erlangen-Nürnberg



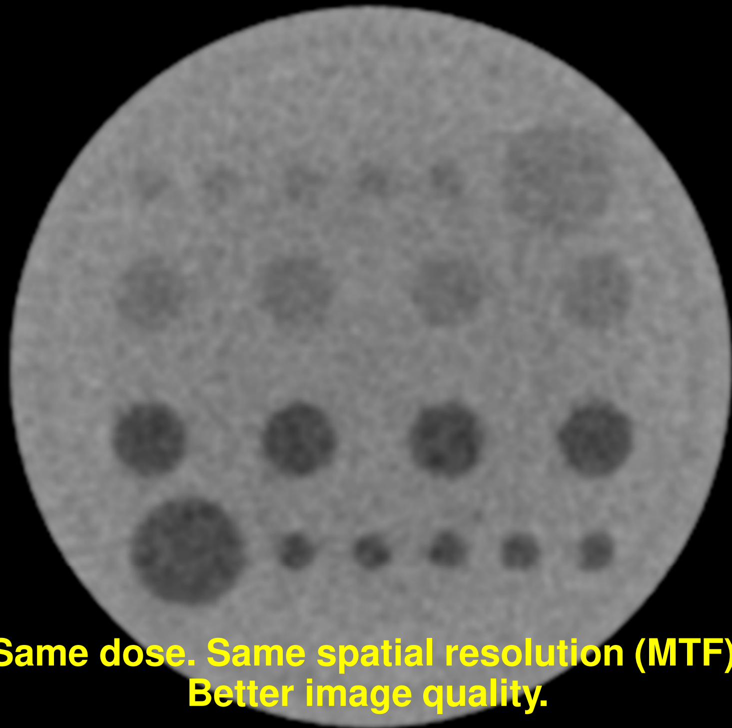
Requirements for CT: up to 10^9 x-ray photon counts per second per mm^2.
Hence, photon counting only achievable for direct converters.

Diagnostic CT (Conventional Detector) of a Low Contrast Phantom

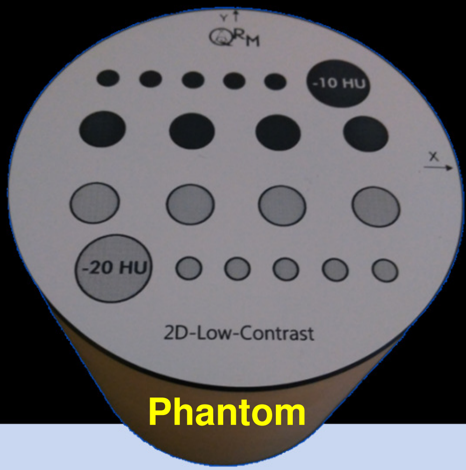
Photon Counting Detector CT of a Low Contrast Phantom

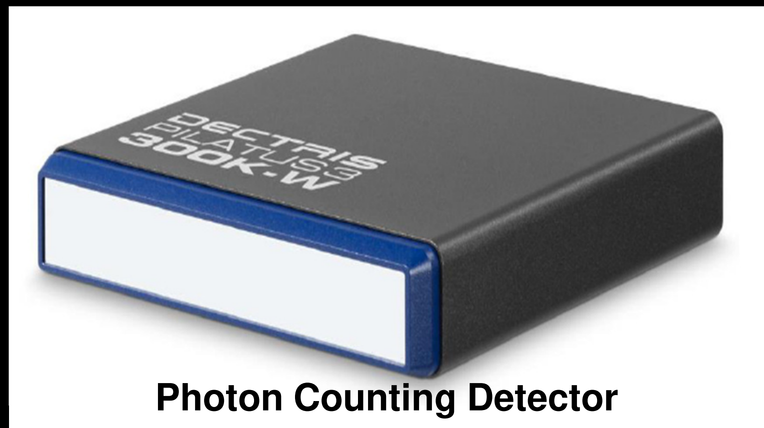


Diagnostic routine head protocol.
34 mGy $CTDI_{vol}$.



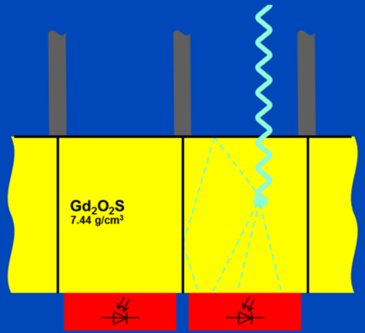
Same dose. Same spatial resolution (MTF).
Better image quality.



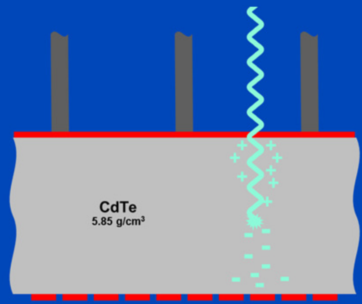


C = 0 HU, W = 80 HU

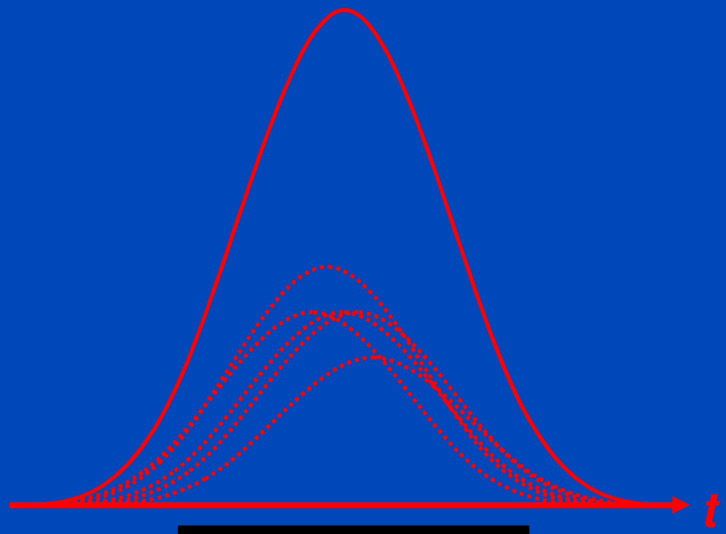
Energy Integrating (EI) vs. Photon Counting (PC) Detector Technology

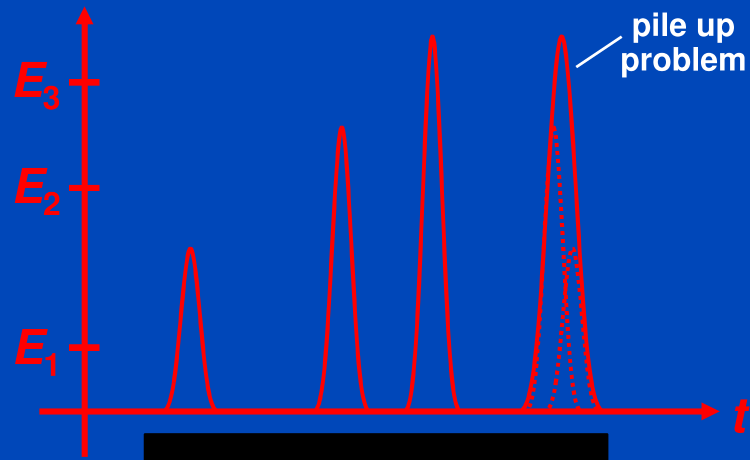


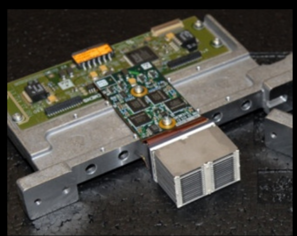
indirect conversion
= pile up only
= energy integrating



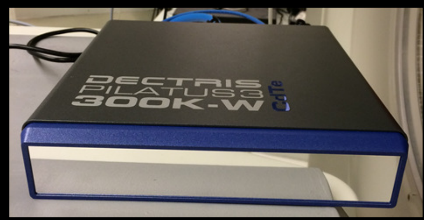
direct conversion
= photon counting
= energy selective







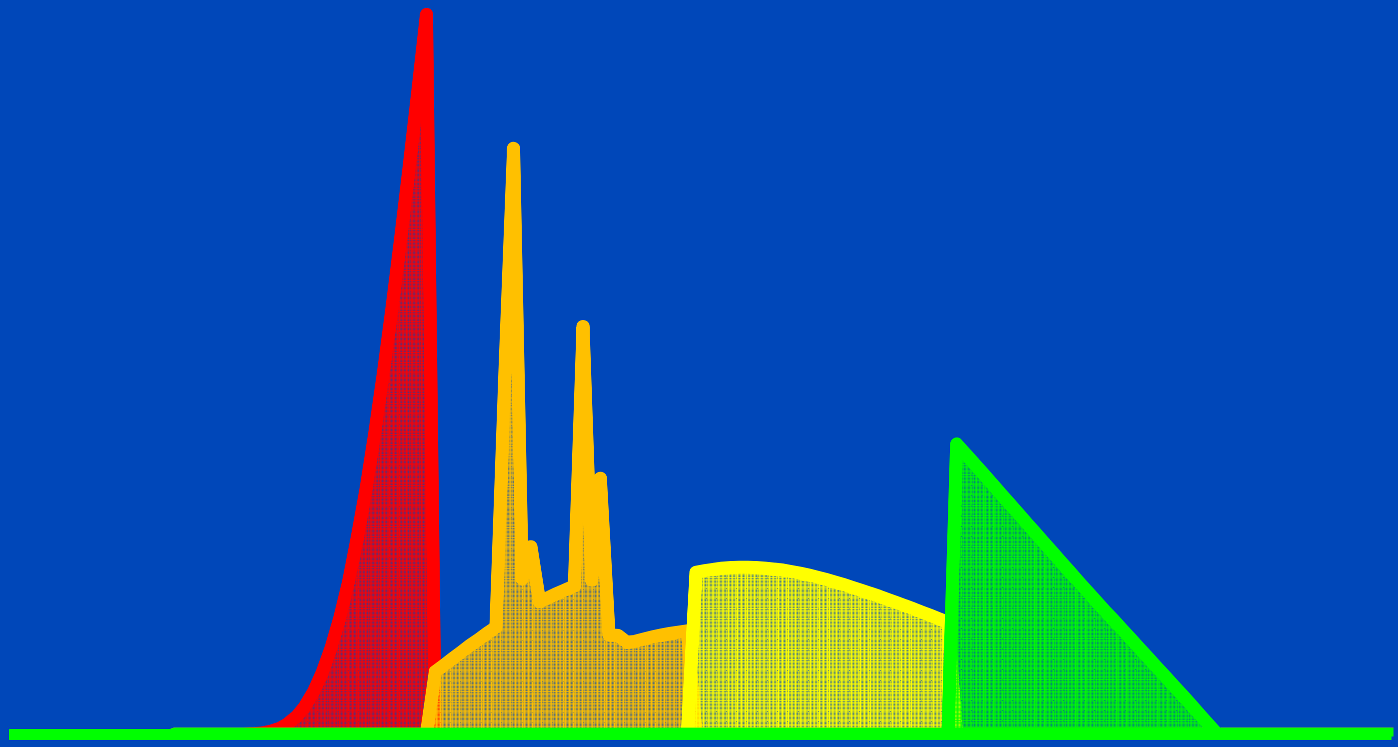
EI detector (Siemens)



PC detector (Dectris)

Energy-Selective Detectors: Improved Spectroscopy, Reduced Dose?

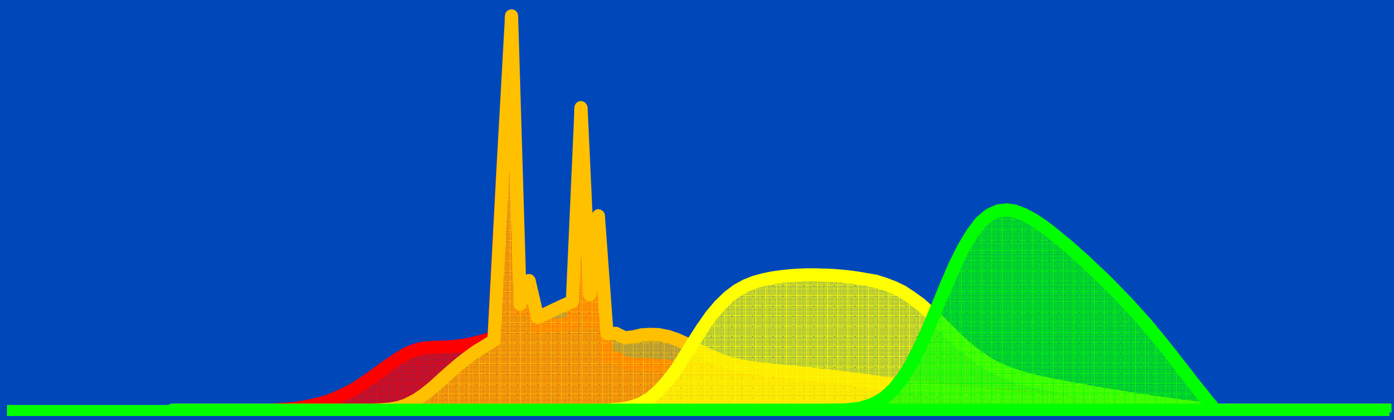
Ideally, bin spectra do not overlap, …



Spectra as seen after having passed a 32 cm water layer.

Energy-Selective Detectors: Improved Spectroscopy, Reduced Dose?

… realistically, however they do!



Spectra as seen after having passed a 32 cm water layer.

Dark Image (X-Rays are Off!) Shows Background Radiation

220 frames, 1 min integration time per frame

No dark current. <u>No readout noise</u>. Single events visible!

Events per Frame, C = 1 cnts, W = 2 cnts

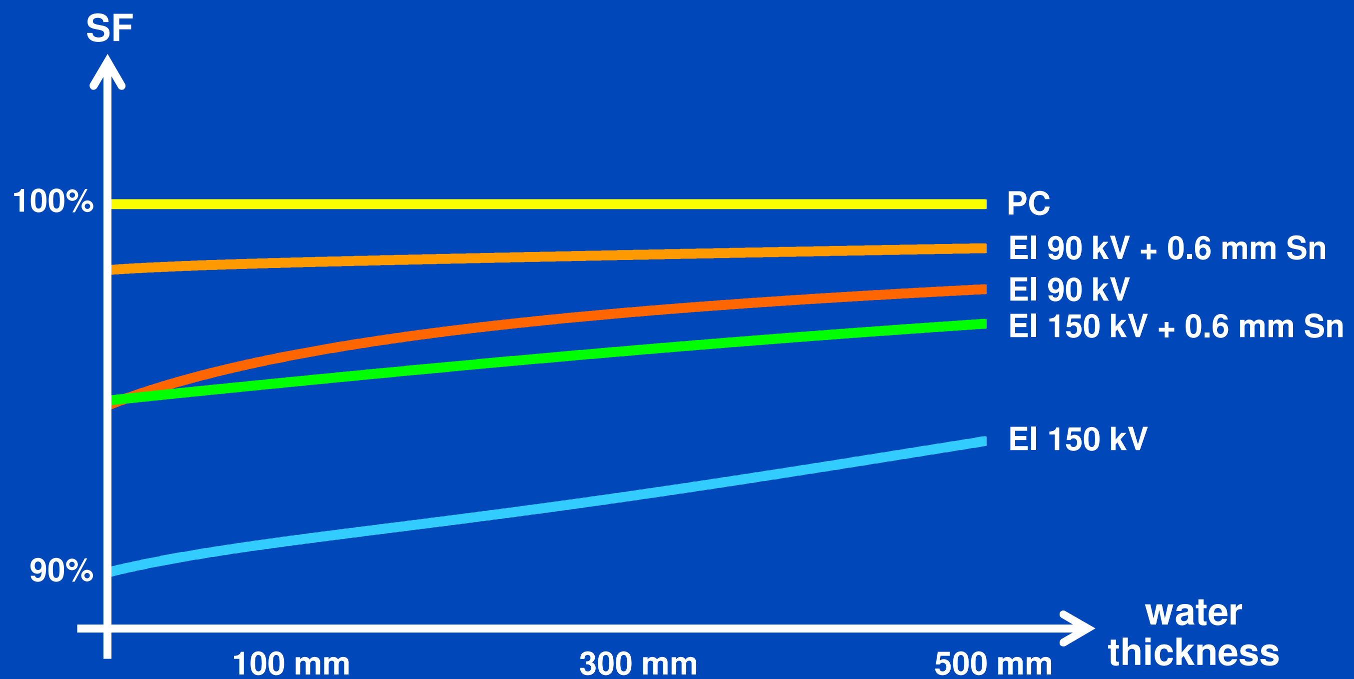
Accumulated Signal, C = 5 cnts, W = 10 cnts

Electronic Noise

- Photon counting detectors have no electronic noise.
- Extreme low dose situations will benefit
  - Pediadric scans at even lower dose
  - Obese patients with less noise
  - ...

Swank Factor

- The Swank factor measures the relative SNR², and thus the relative dose efficiency between photon counting (PC) and energy integrating (EI).
- EI always has the lower SNR.

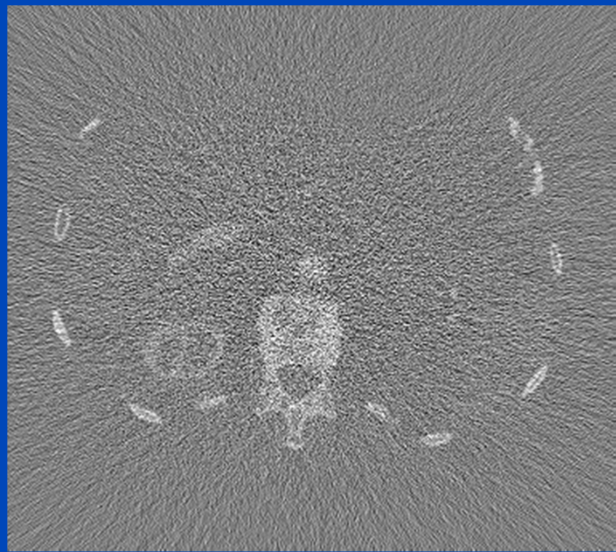


$$\mathrm{SF}=\frac{\mathrm{SNR}_{\mathrm{EI}}^2}{\mathrm{SNR}_{\mathrm{PC}}^2}=\frac{\left(\int dE\, s(E)\mathrm{E}N(E)\right)^2}{\left(\int dE\,\mathrm{E}N(E)\right)\left(\int dE\, s^2(E)\mathrm{E}N(E)\right)}=\frac{M_1^2}{M_0M_2}\leq 1$$

Energy Integrating vs. Photon Counting
with 1 bin from 20 to 140 keV

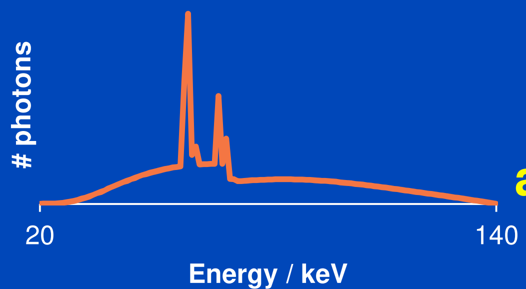
Energy Integrating

PC minus EI

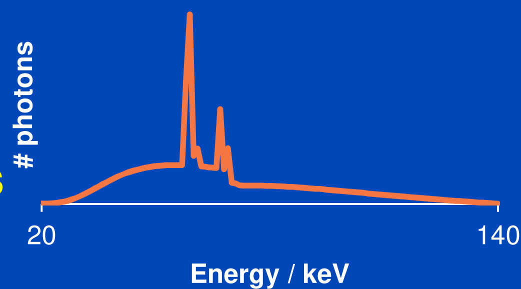


Photon Counting

CNR = 2.11



CNR = 2.95

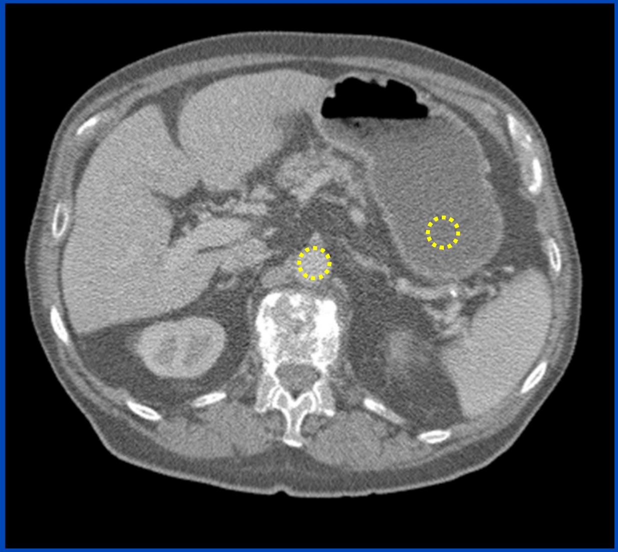


40% CNR improvement or 49% dose reduction achievable due to improved Swank factor and more weight on low energies (iodine contrast benefits).

Images: *C* = 0 HU, *W* = 700 HU, difference image: *C* = 0 HU *W* = 350 HU, bins start at 20 keV

Photon Counting
Enables Energy Bin Weighting

- With PC energy bins can be weighted individually.
- To optimize the CNR the optimal bin weighting factor is given by (weighting after log):



$$w_b \propto \frac{C_b}{V_b}$$

- The resulting CNR is

$$\mathrm{CNR}^2 = \frac{\left(\sum_b w_b C_b\right)^2}{\sum_b w_b^2 V_b}$$

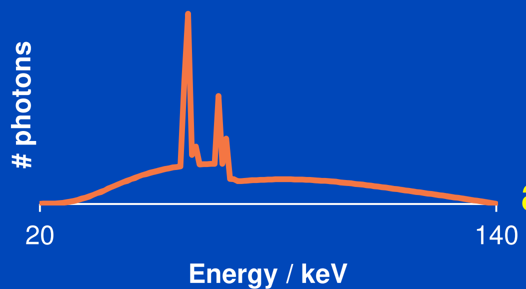
- At the optimum this evaluates to

$$\mathrm{CNR}^2 = \sum_{b=1}^{B} \mathrm{CNR}_b^2$$

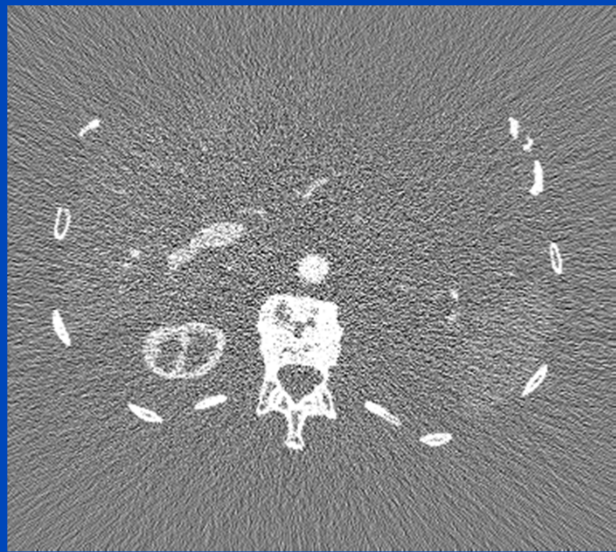
Energy Integrating vs. Photon Counting
with 4 bins from 20 to 140 keV

Energy Integrating

CNR = 2.11

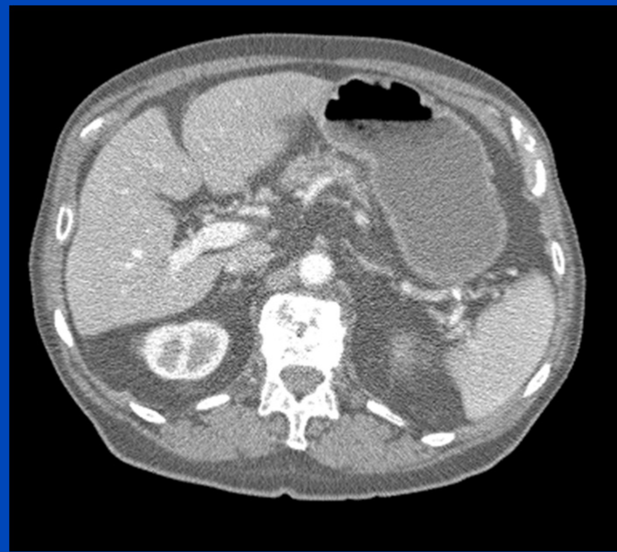


PC minus EI

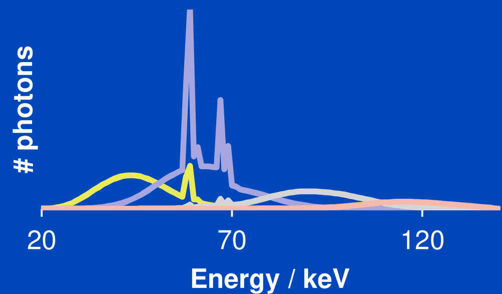


99% CNR improvement or 75% dose reduction achievable due to improved Swank factor and optimized energy weighting.

Photon Counting

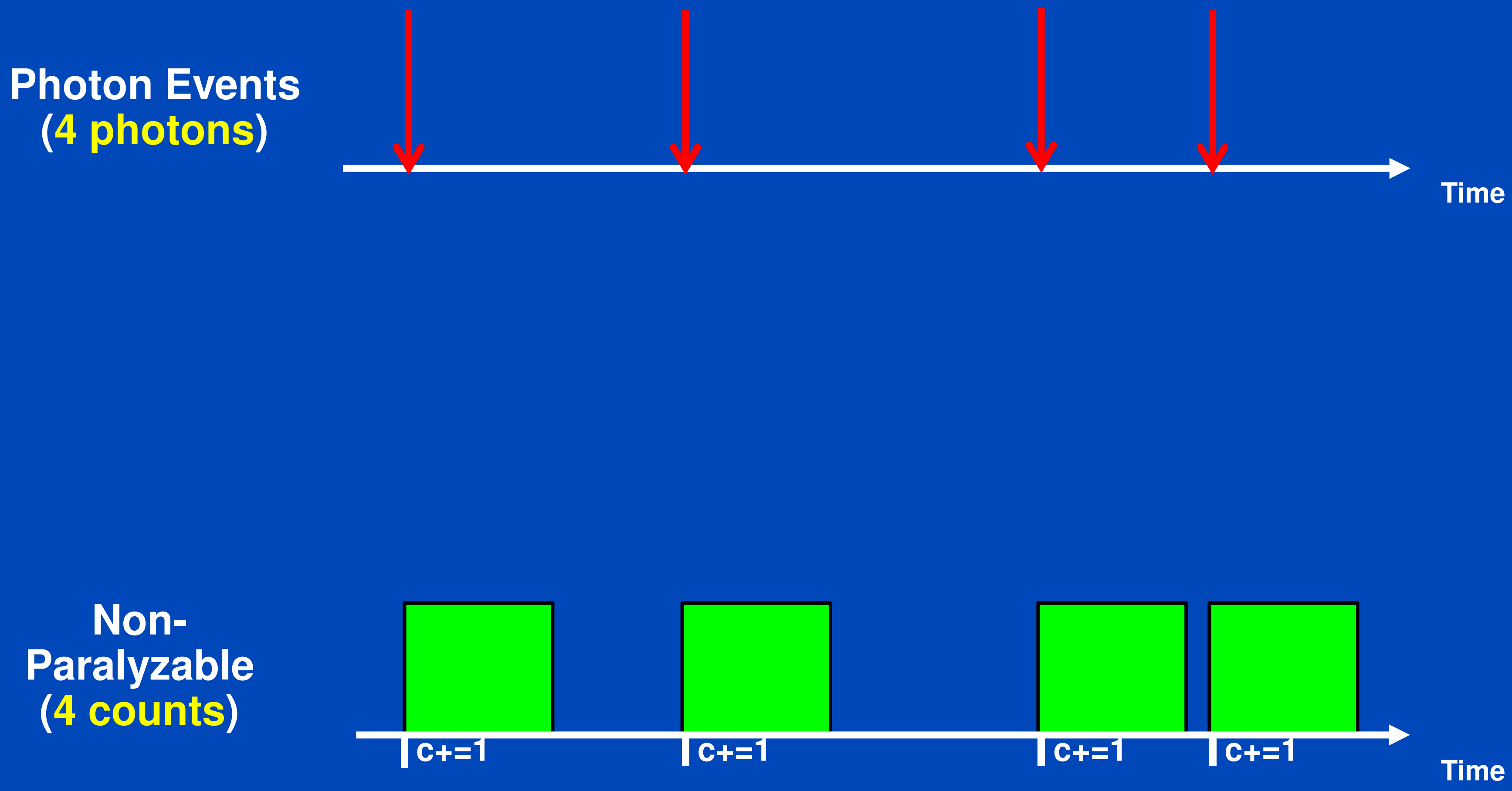


CNR = 4.19



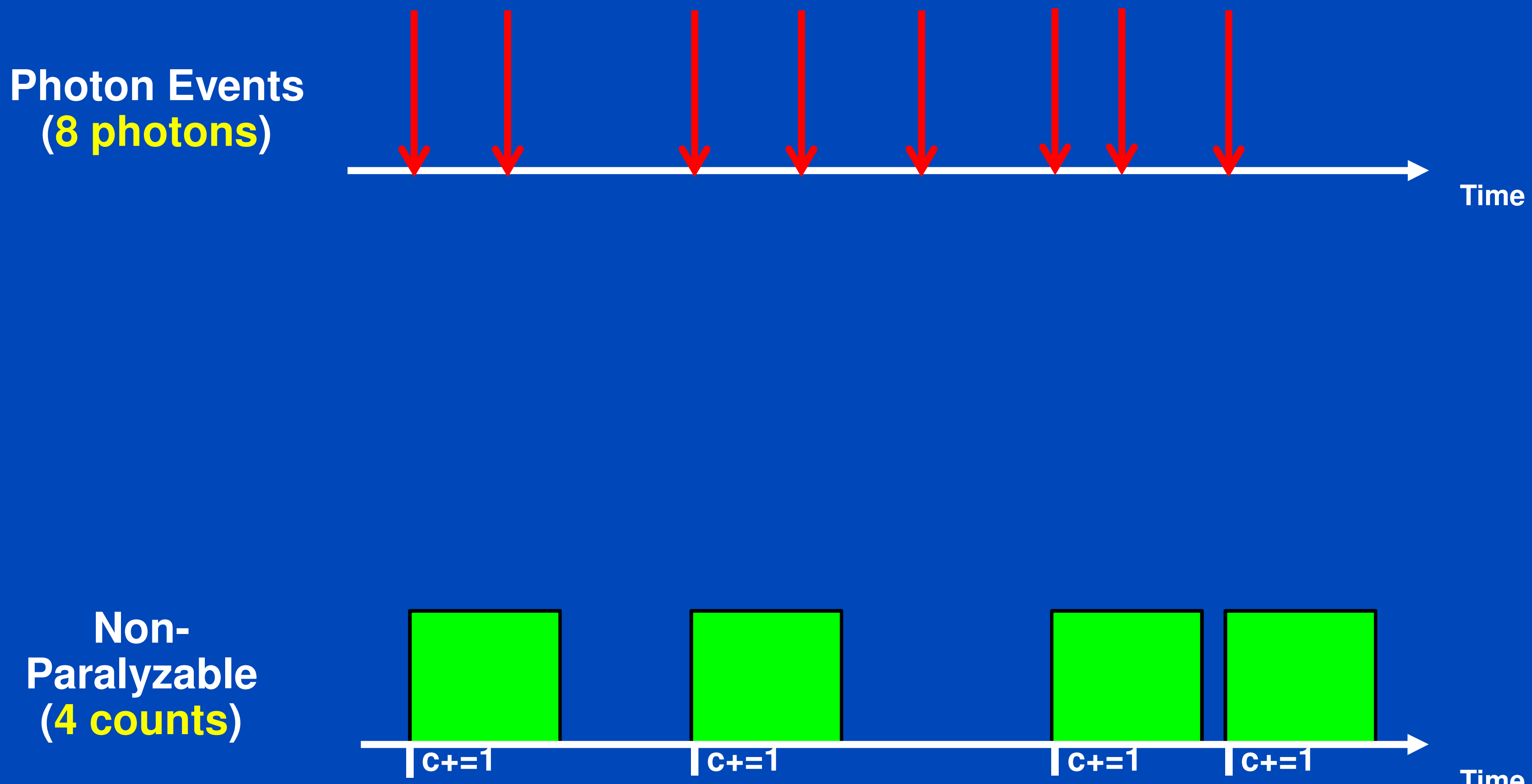
Images: *C* = 0 HU, *W* = 700 HU, difference image: *C* = 0 HU *W* = 350 HU, bins start at 20 keV

Pulse Pile-Up: Low Flux Rate



Boxes illustrate deadtime

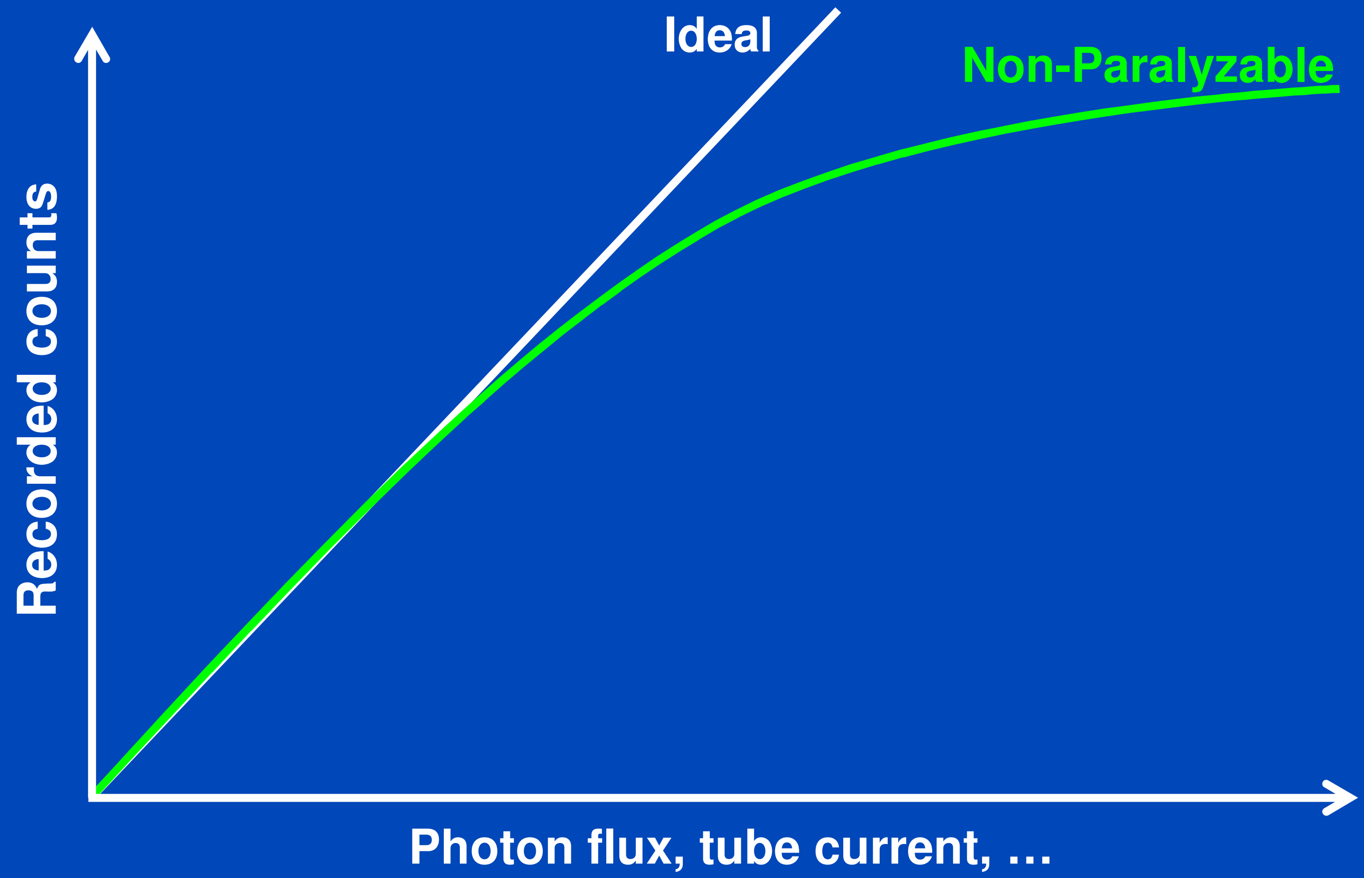
Pulse Pile-Up: High Flux Rate



Boxes illustrate deadtime

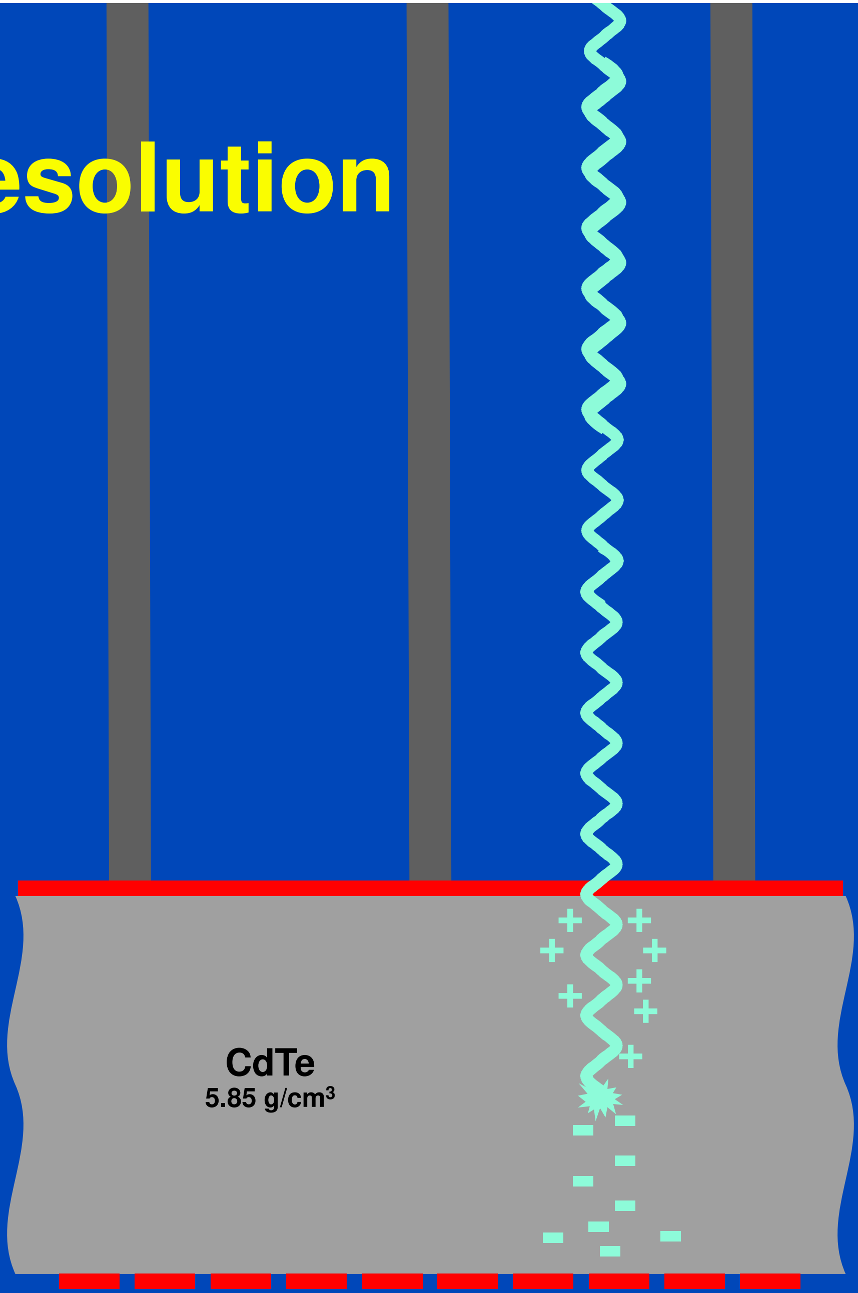
dkfz.

Pulse Pile-Up: Recorded Counts



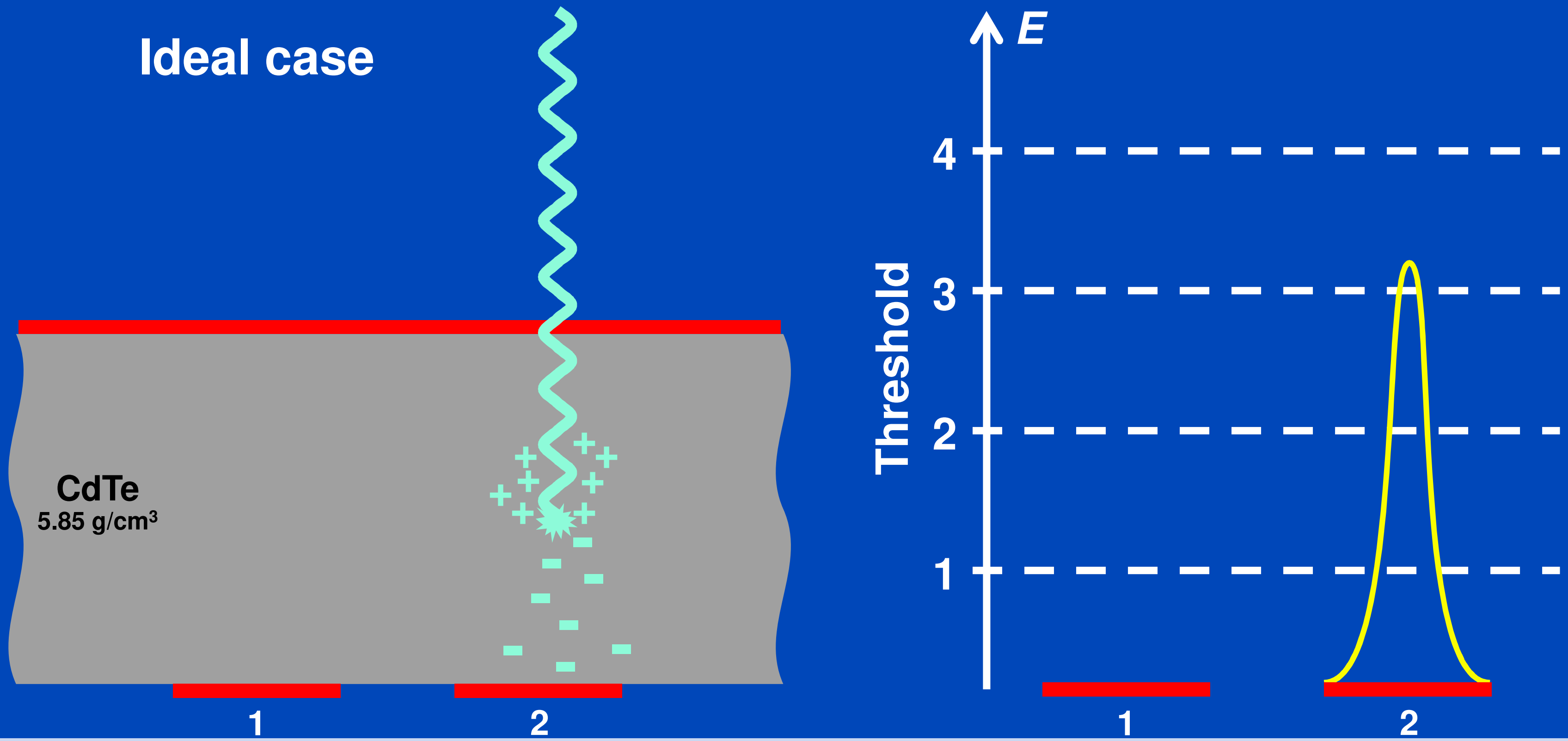
Spatial Resolution

- Small electrodes are necessary to avoid pile-up.
- High bias voltages (around 300 V) limit charge diffusion and thus blurring in the non-structured semiconductor layer.
- Thus, higher spatial resolution is achievable.



Spectral Resolution

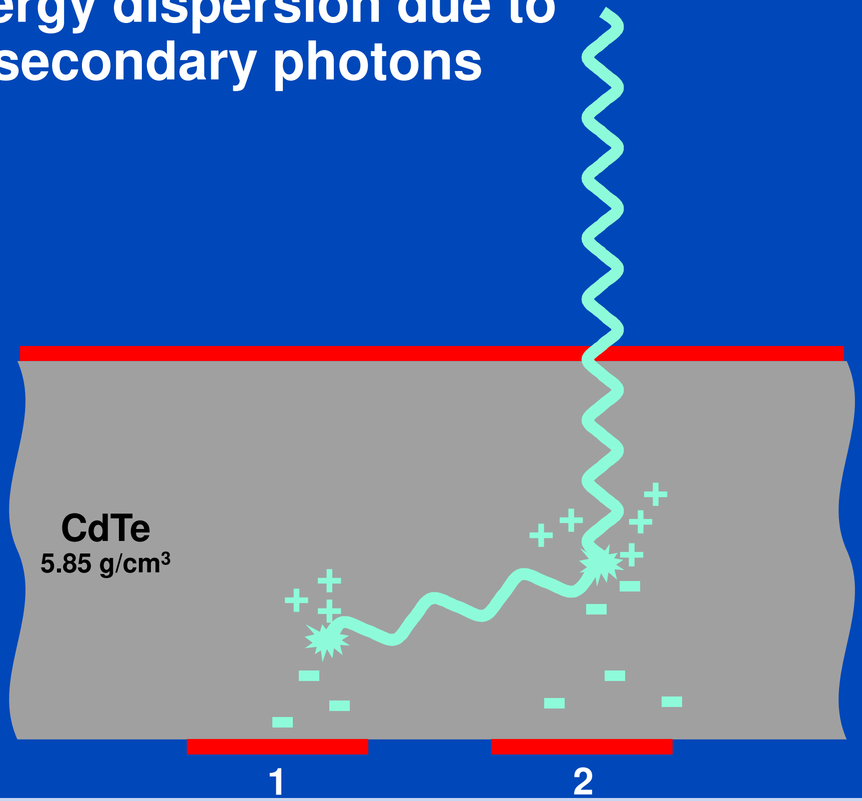
- Detection process in the sensor
- Photoelectric effect (e.g. 80 keV)

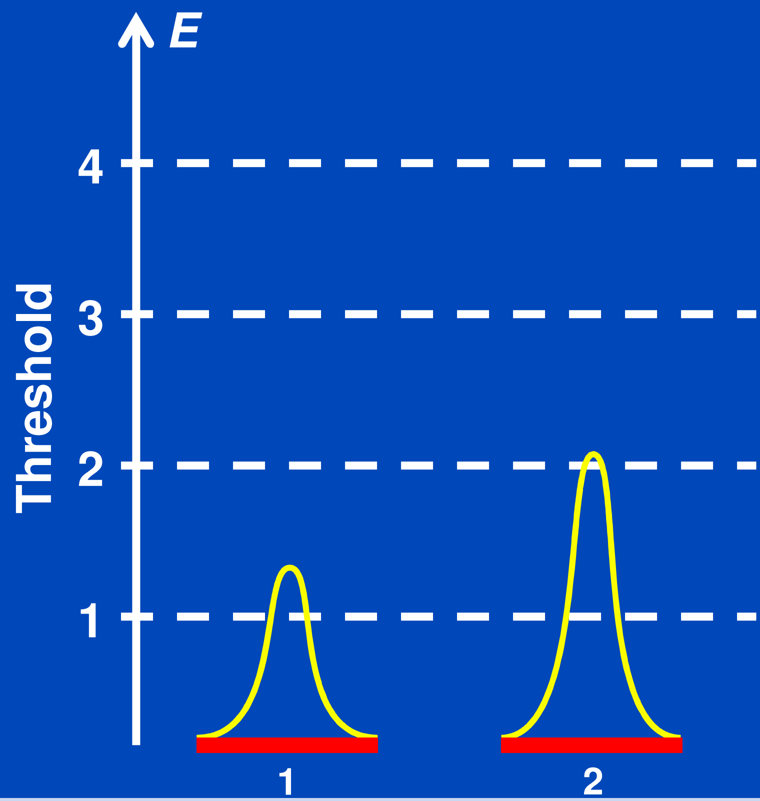


Spectral Resolution

- Detection process in the sensor
- Compton scattering or K-fluorescence (e.g. 80 keV)

Energy dispersion due to secondary photons

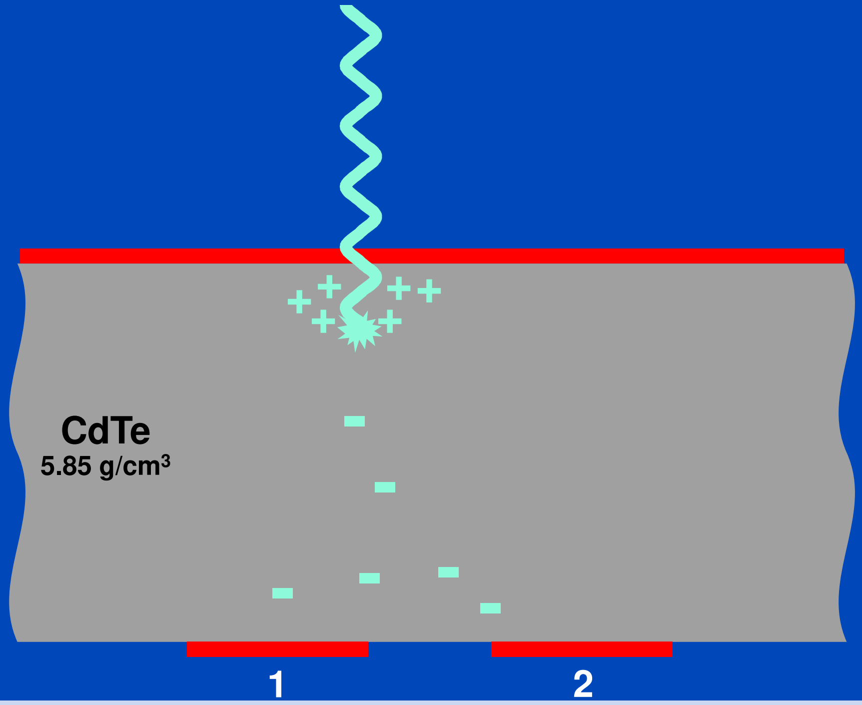


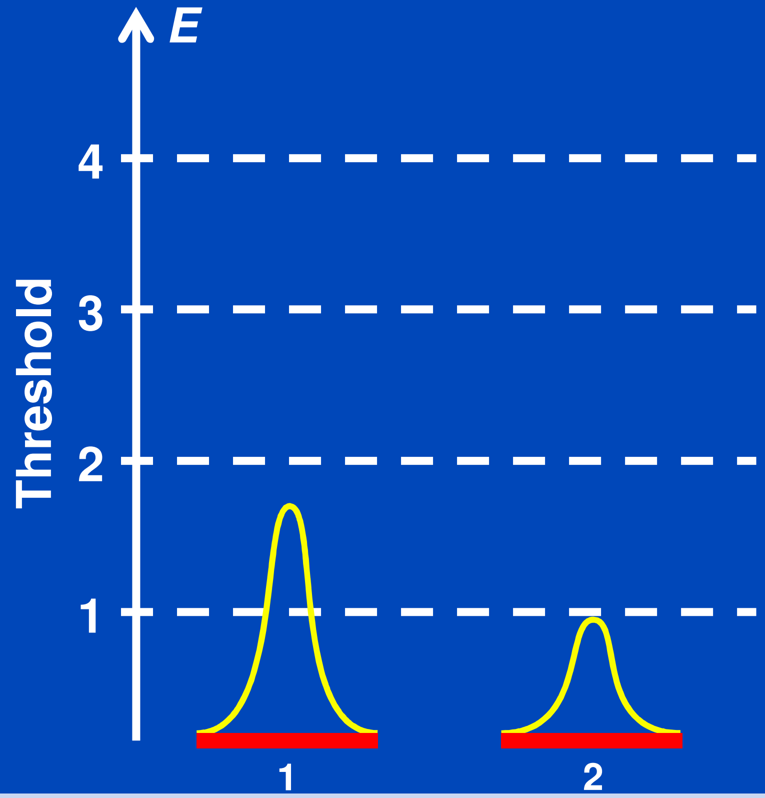


Spectral Resolution

- Detection process in the sensor
- Photoelectric effect (e.g. 30 keV), charge sharing

Energy dispersion due to charge diffusion





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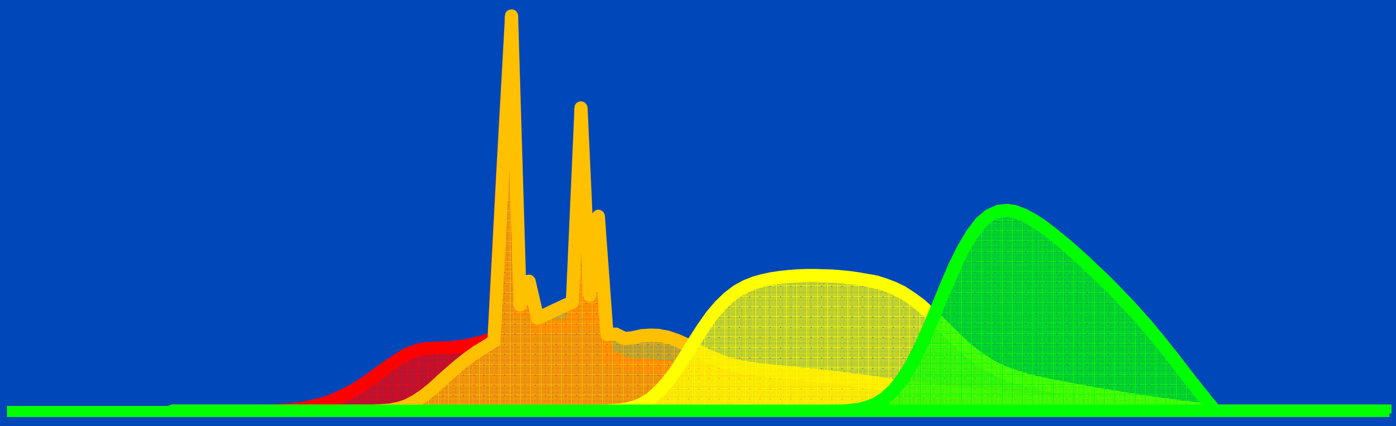
Spectral Resolution

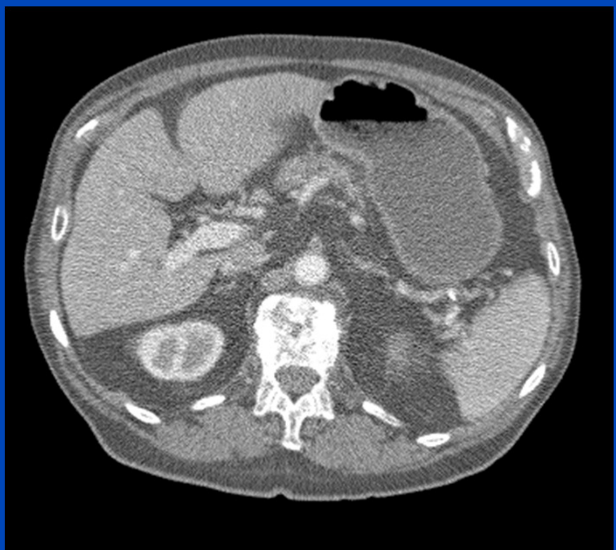


Spectra as seen after having passed a 32 cm water layer.

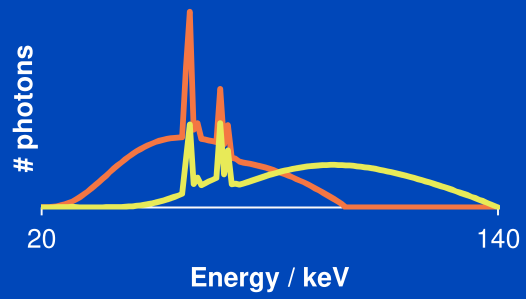
Photon Counting used for Spectral Imaging

- DECT scan with 100 kV / 140 kV Sn
- Photon counting acquisition at 140 kV
- Same patient dose in both cases

100 kV



140 kV Sn



Images: *C* = 0 HU, *W* = 700 HU

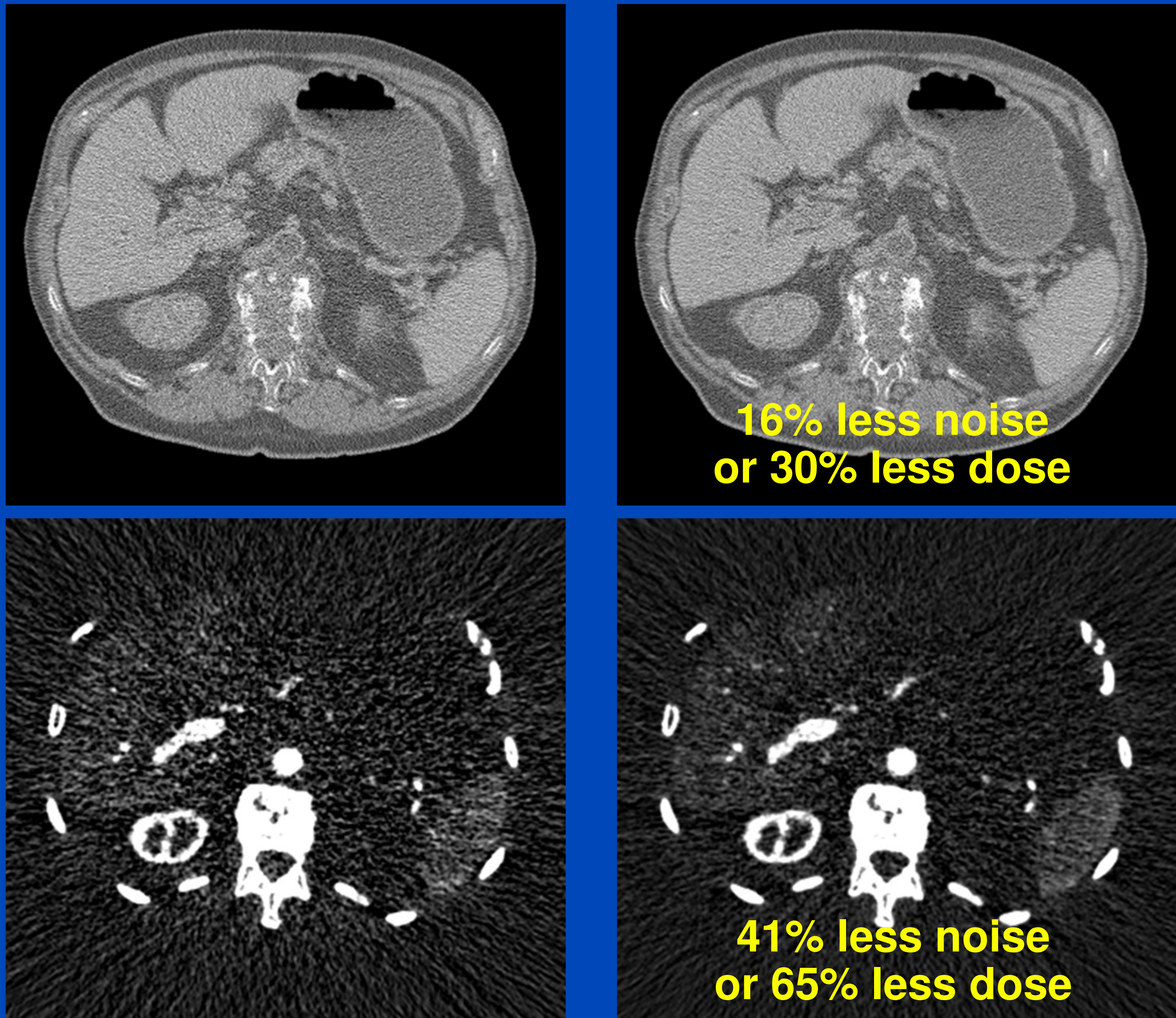
Energy Integrating vs. Photon Counting

Energy Integrating
100 kV / 140 kV Sn

Photon Counting 140 kV
4×30 keV Gaussian bins

Water

Iodine



Water image: C = 0 HU, W = 700 HU, iodine image: C = 0 HU, W = 2000 HU, bins start at 20 keV

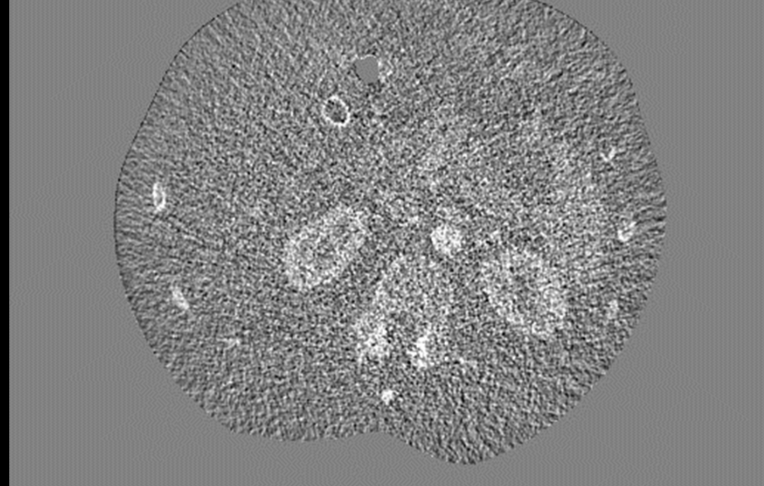
Decomposition Increases Noise

80 kV

VNC

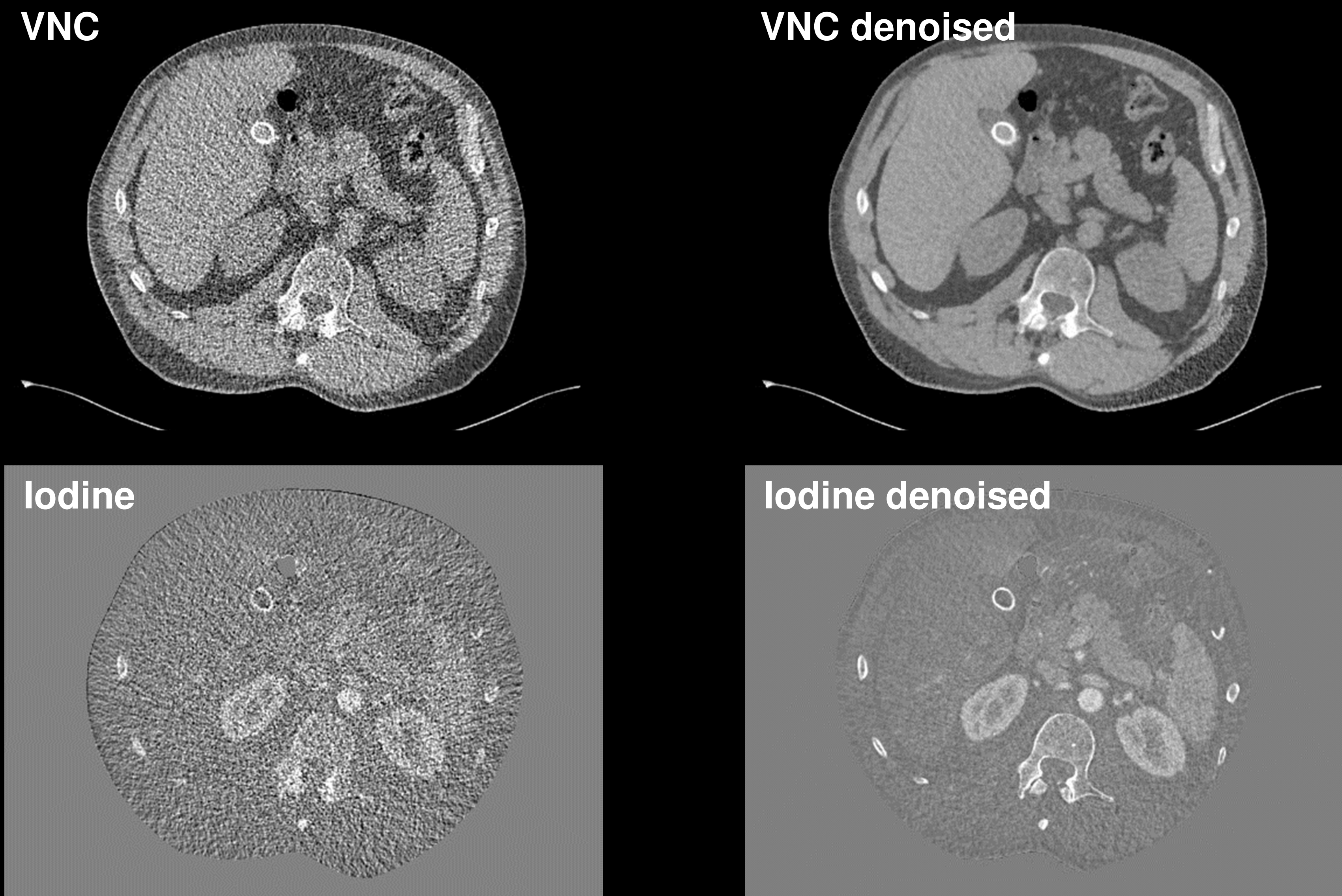
140 kV

Iodine



C = 0 HU, W = 700 HU

Denoising is Mandatory!



dkfz.

More than Dual Energy?

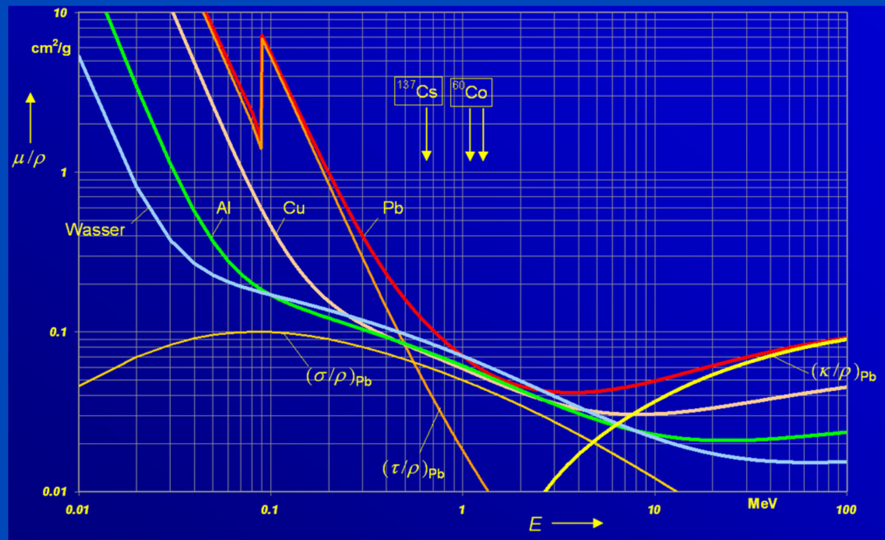
- Ways to remove the spectral overlap?
- Lower noise, less dose?
- Improve contrast-to-noise ratio at unit dose?
- Distinguish more than three materials?

$$\mu(E) = \cancel{\eta(E)} + \tau(E) + \sigma(E) + \cancel{\kappa(E)}$$

Rayleigh Photo Compton Pair

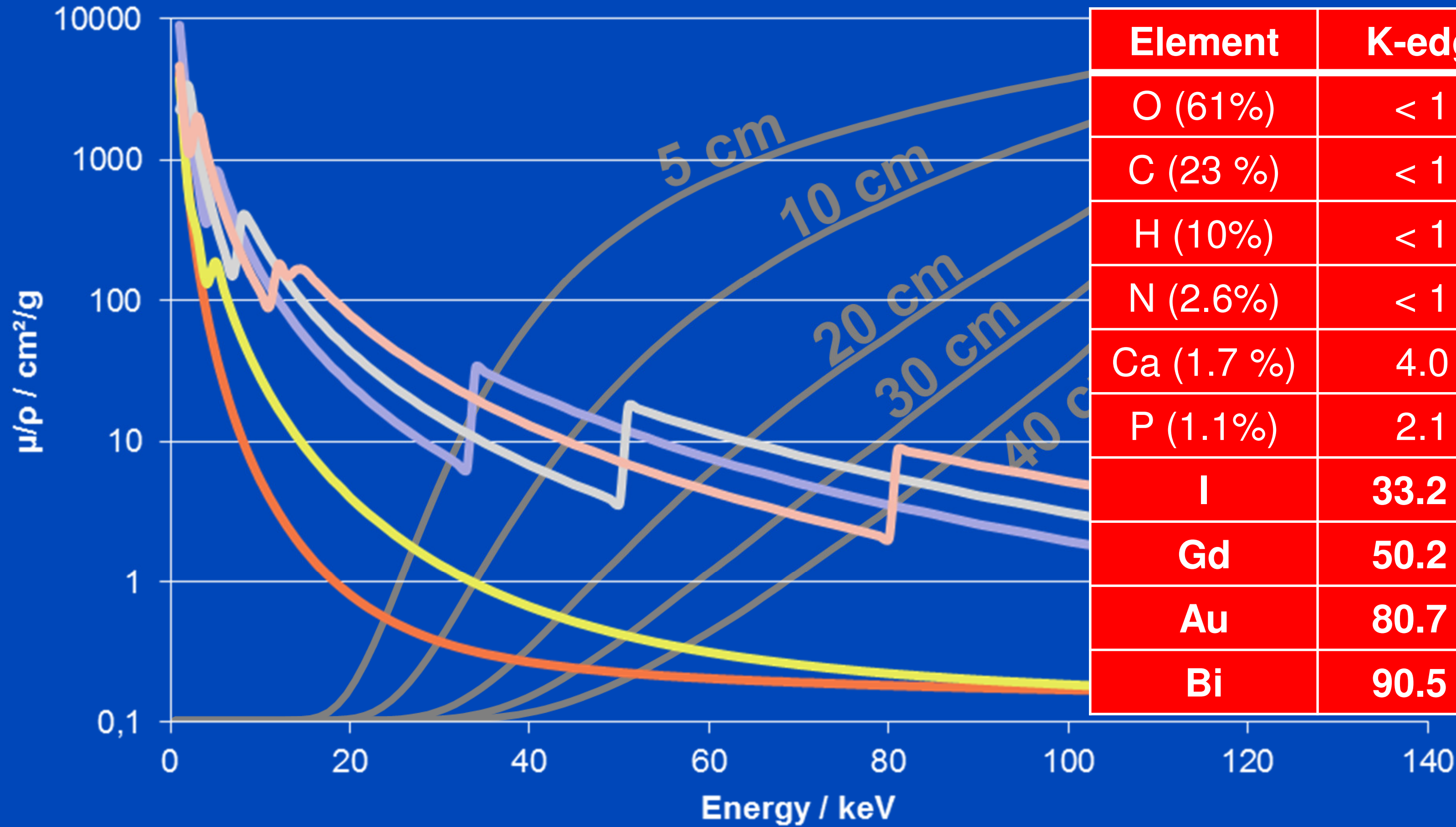
$$\tau(E) \propto \rho \frac{Z^3}{E^3}$$

$$\sigma(E) \propto \rho \frac{Z}{A} f(E)$$



K-Edges: More than Dual Energy CT?

$$\mu(\boldsymbol{r},E) = f_1(\boldsymbol{r})\psi_1(E) + f_2(\boldsymbol{r})\psi_2(E) + \underbrace{f_3(\boldsymbol{r})\psi_3(E)}_{\text{Iff new contrast agents become available}} + \dots$$



Element	K-edge
O (61%)	< 1 keV
C (23 %)	< 1 keV
H (10%)	< 1 keV
N (2.6%)	< 1 keV
Ca (1.7 %)	4.0 keV
P (1.1%)	2.1 keV
I	**33.2 keV**
Gd	**50.2 keV**
Au	**80.7 keV**
Bi	**90.5 keV**

120 kV water transmission curves (gray) given in relative units on a non-logarithmic ordinate.

Future, Photon Counting (≥ 2020)?

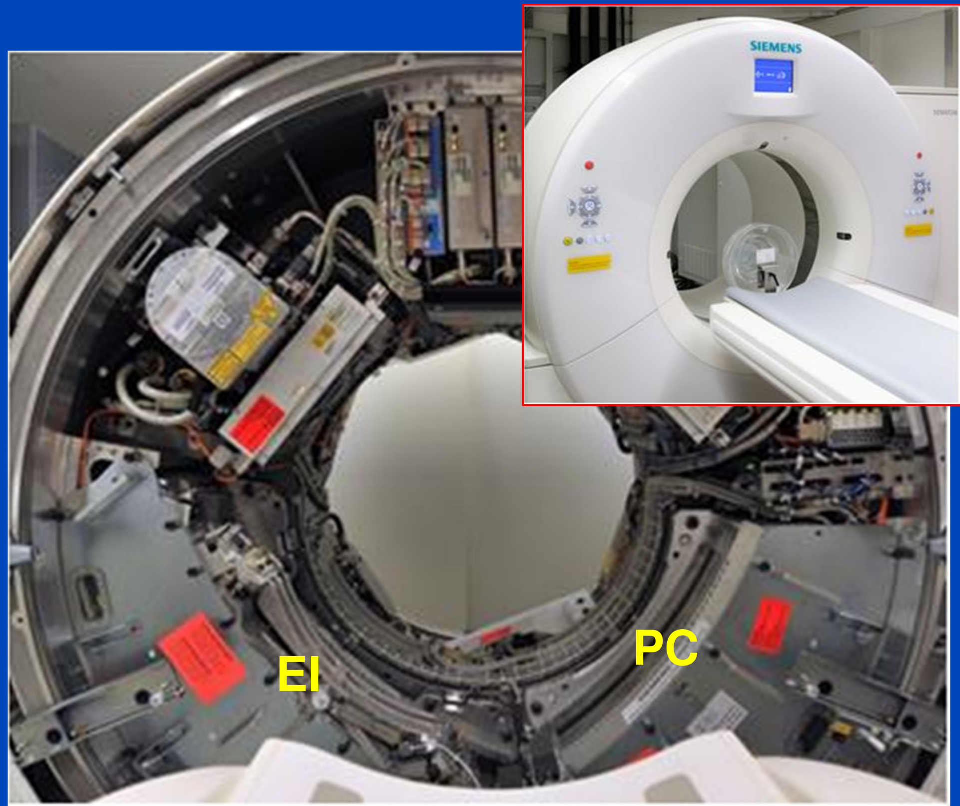
Macro

12	12	12	12
12	12	12	12
12	12	12	12
12	12	12	12

Chess

12	34	12	34
34	12	34	12
12	34	12	34
34	12	34	12

4×4 subpixels of 225 µm size = 0.9 mm pixels
(0.5 mm at isocenter)



This photon-counting whole-body CT prototype, installed at the Mayo Clinic and at the NIH, is a DSCT system. However, it is restricted to run in single source mode. The second source is used for data completion and for comparisons with EI detectors.

Photo courtesy of Siemens Healthcare, Forchheim, Germany.

Readout Modes of the Siemens CounT

Macro Mode
1×2 readouts
16 mm z-coverage

12	12	12	12
12	12	12	12
12	12	12	12
12	12	12	12

Chess Mode
2×2 readouts
16 mm z-coverage

12	34	12	34
34	12	34	12
12	34	12	34
34	12	34	12

Sharp Mode
5×1 readouts
12 mm z-coverage

1	1	1	1
1	1	1	1
1	1	1	1
1	1	1	1

2	2	2	2
2	2	2	2
2	2	2	2
2	2	2	2

UHR Mode
4×2 readouts
8 mm z-coverage

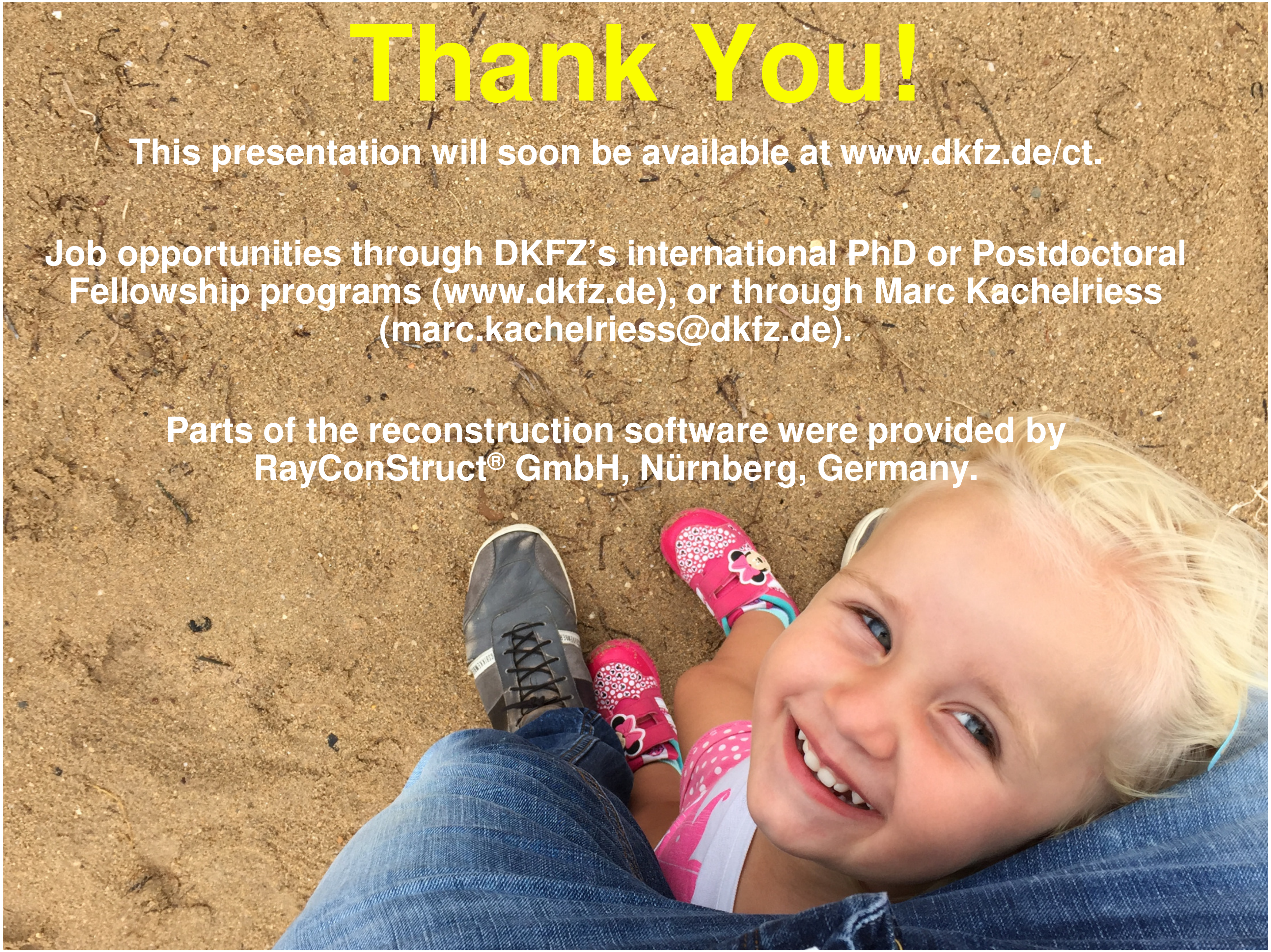
12	12	12	12
12	12	12	12
12	12	12	12
12	12	12	12

No FFS on thread B (photon counting detector).
The whole detector consists of 128×1920 subpixels = 32×480 macro pixels.

Potential Advantages of Photon Counting Detectors in CT

- Higher spatial resolution due to
  - smaller pixels
  - lower cross-talk between pixels
- Lower dose/noise due to
  - Swank factor = 1
  - energy bin weighting
  - zero electronic noise
- Spectral information on demand
  - single energy
  - dual energy
  - multiple energy
  - virtual monochromatic
  - K-edge imaging
  - ...

Potential clinical impact


Thank You!
This presentation will soon be available at www.dkfz.de/ct.
Job opportunities through DKFZ's international PhD or Postdoctoral Fellowship programs (www.dkfz.de), or through Marc Kachelriess (marc.kachelriess@dkfz.de).
Parts of the reconstruction software were provided by RayConStruct® GmbH, Nürnberg, Germany.