Multi Energy CT with Photon Counting Detectors

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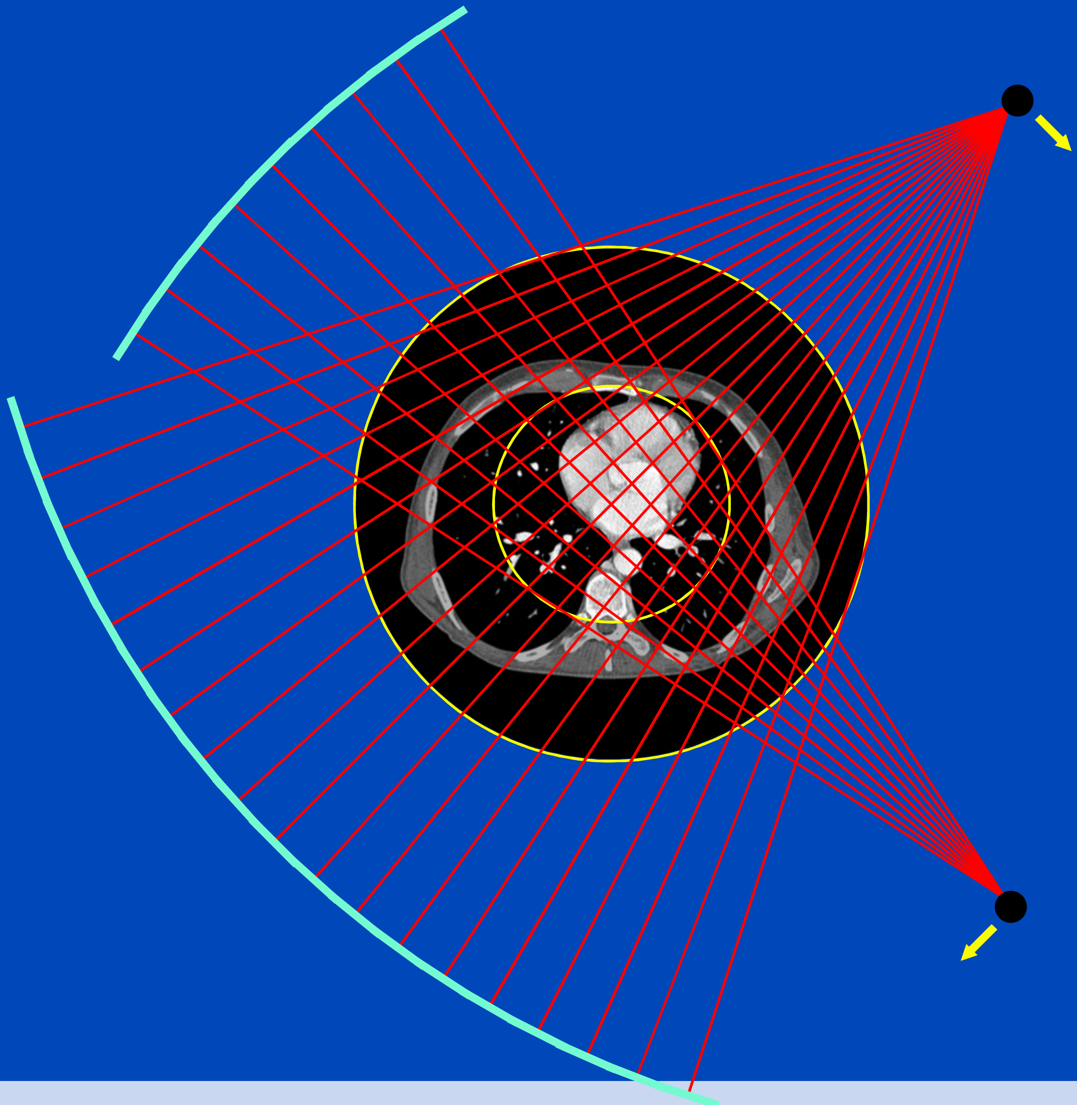
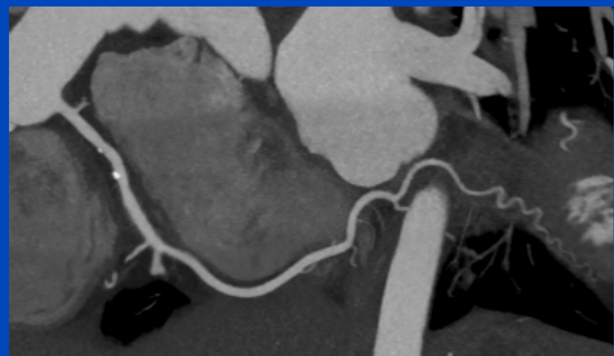
DEUTSCHES KREBSFORSCHUNGSZENTRUM
IN DER HELMHOLTZ-GEMEINSCHAFT

Dual-Source-CT (since 2005)



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

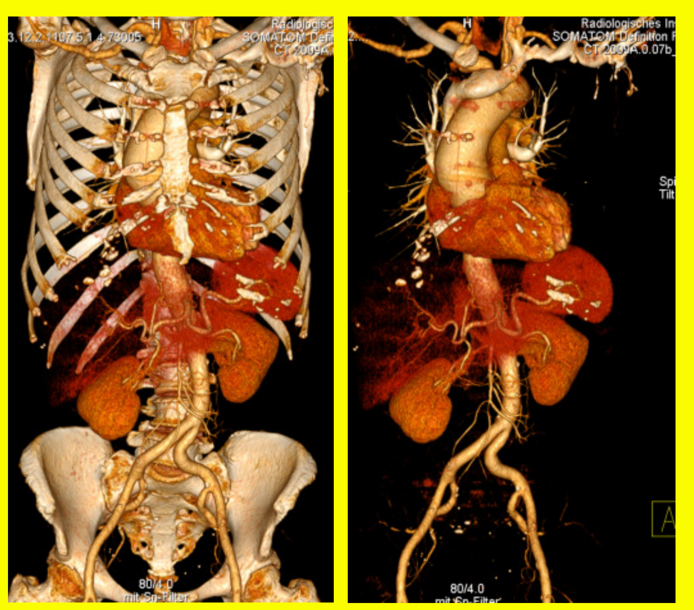


Turbo Flash, 70 kV, 0.55 mSv
63 ms temporal resolution
143 ms scan time

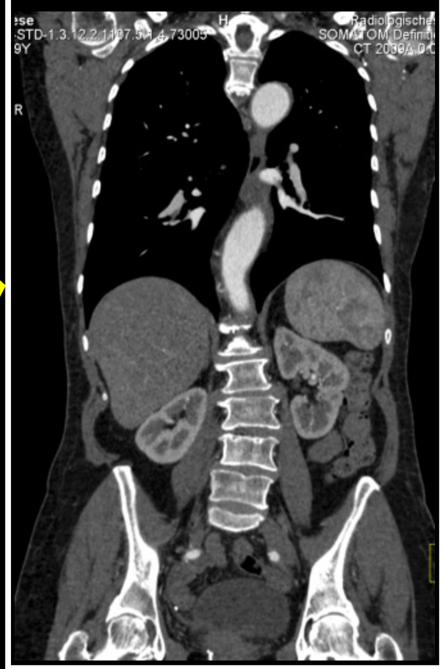
Examples

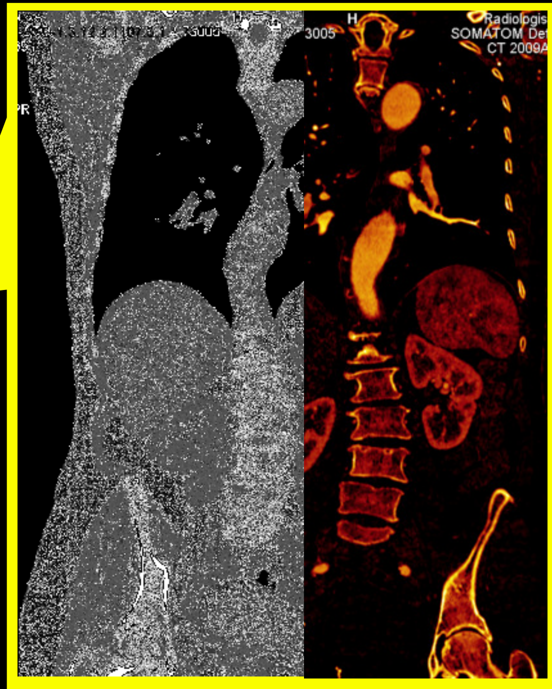
(Slide Courtesy of Siemens Healthcare)

DE bone removal



Single DECT Scan



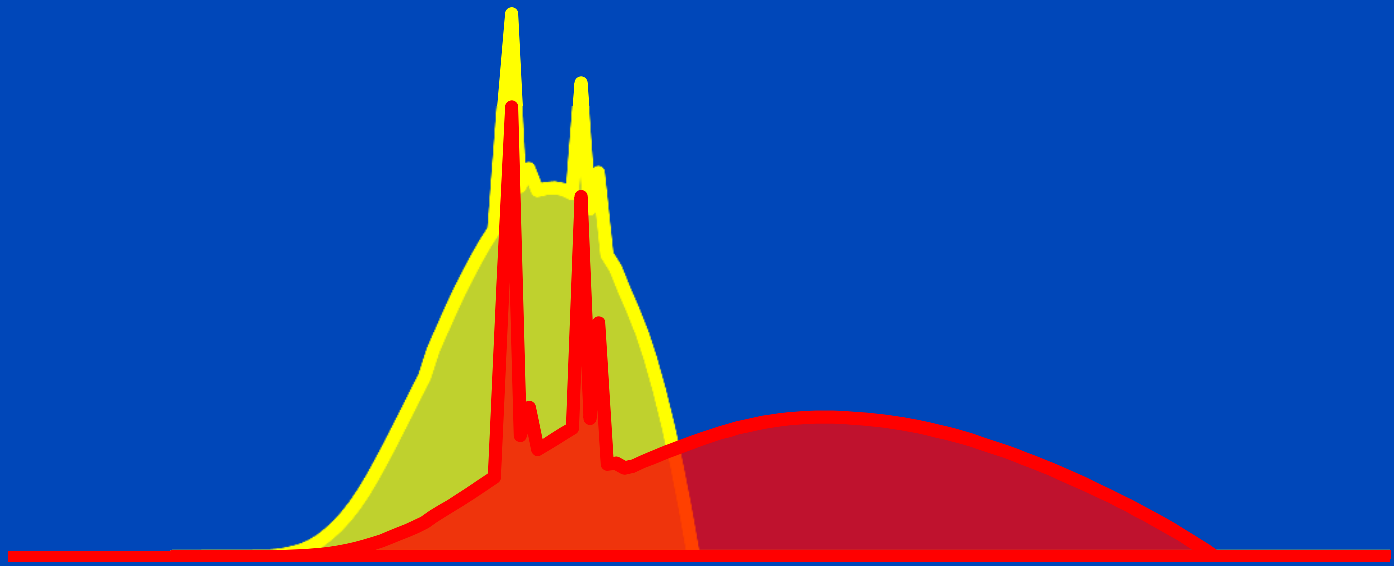


Virtual non-contrast and Iodine image

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

Courtesy of Friedrich-Alexander University Erlangen-Nürnberg

80 kV / 140 kV



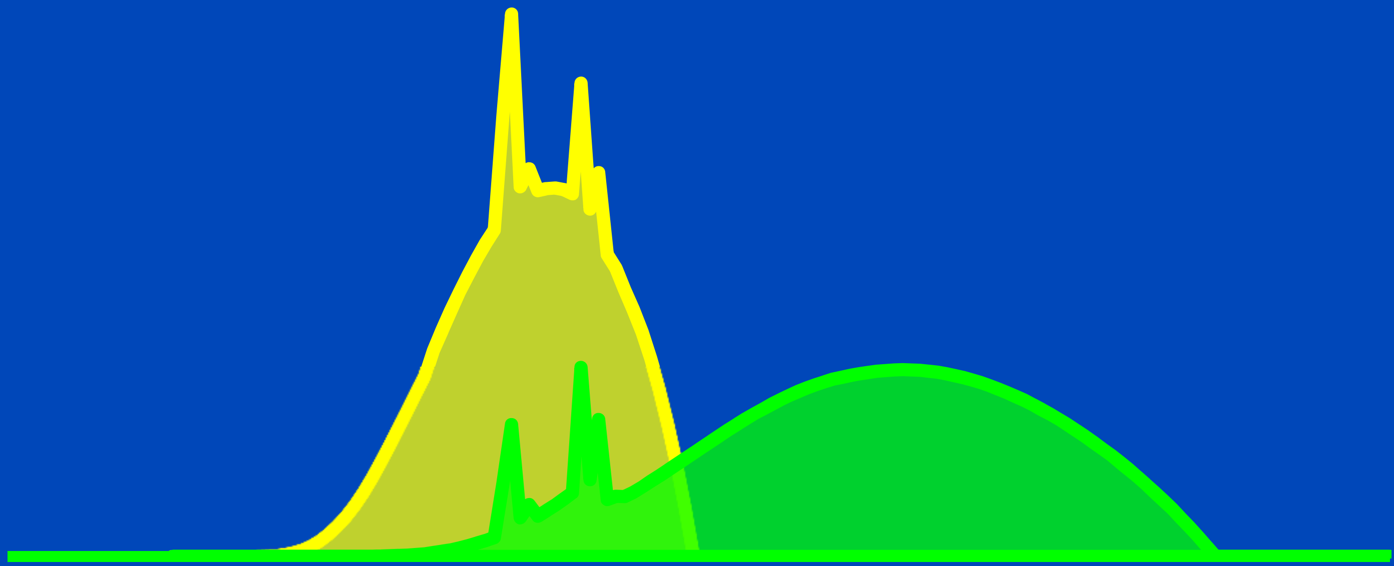
Used in
- Siemens‘ 1st generation DSCT

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

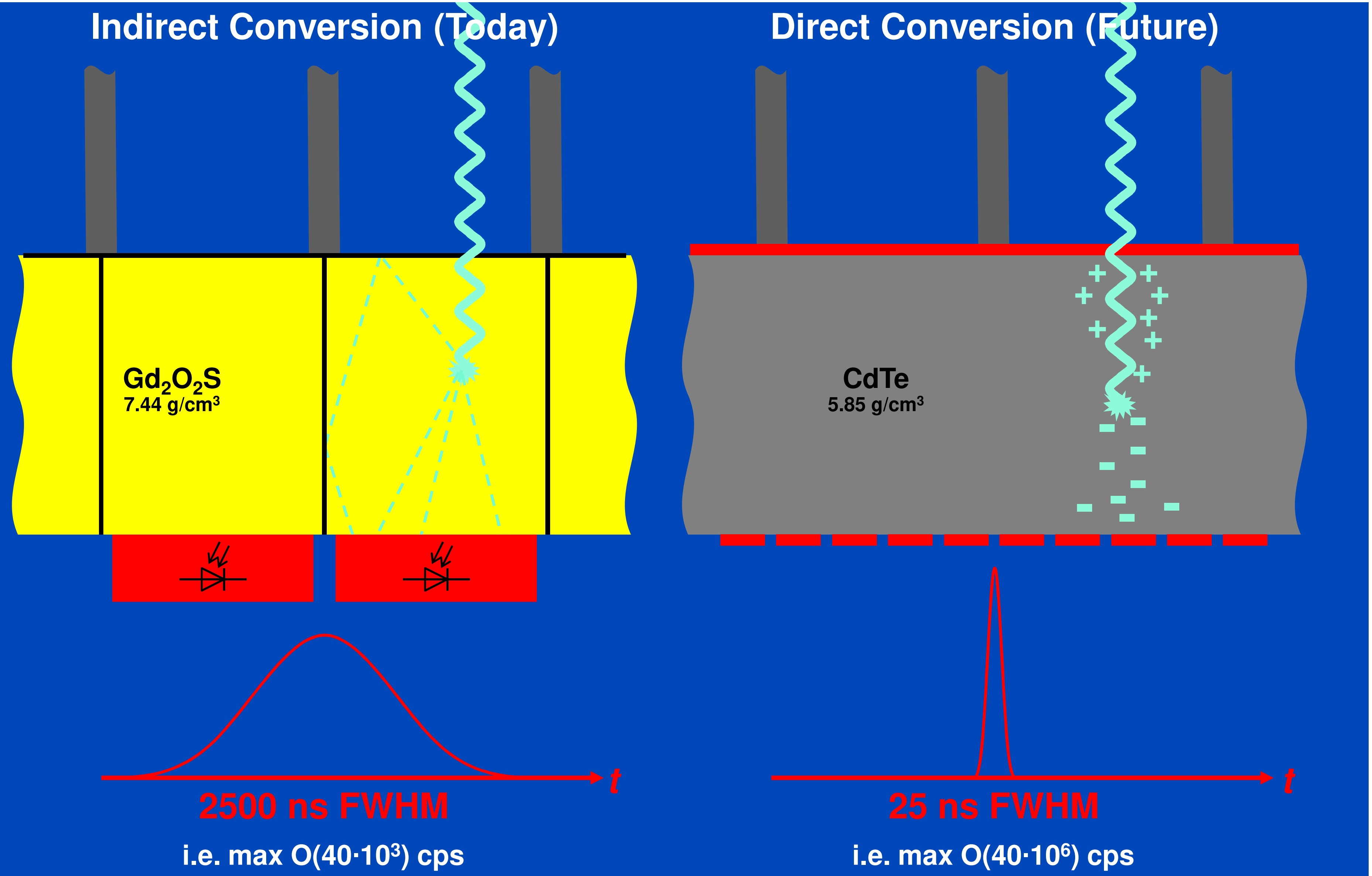
dkfz.

80 kV / 140 kV $Sn_{0.4\ mm}$

Used in
- **Siemens‘ 2nd generation DSCT**

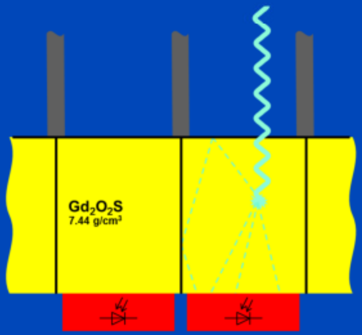


Spectra as seen after having passed a 32 cm water layer.

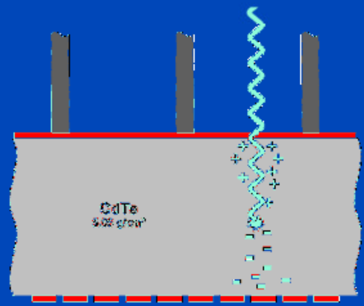


Requirements for CT: up to 10^9 x-ray photon counts per second per mm^2.
Hence, photon counting only achievable for direct converters.

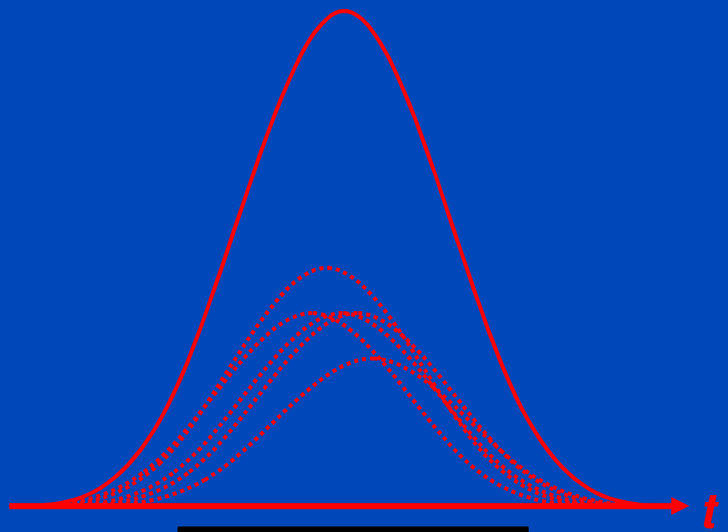
Energy Integrating (EI) vs. Photon Counting (PC) Detector Technology

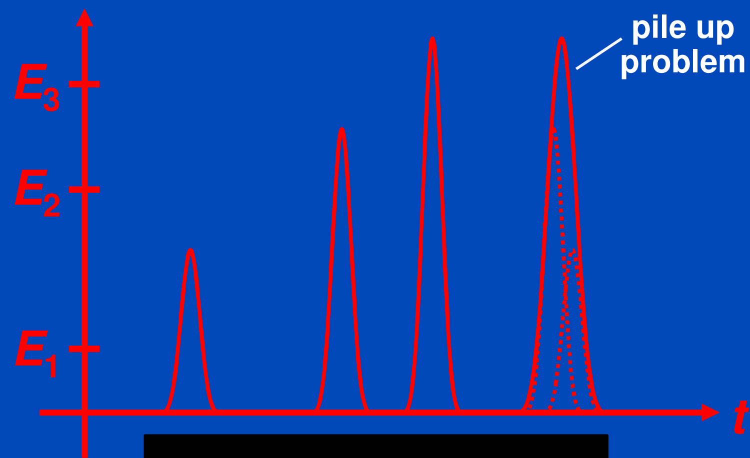


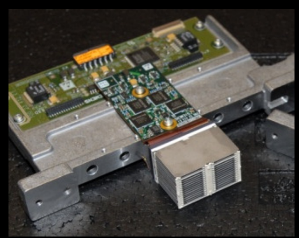
indirect conversion
= pile up only
= energy integrating



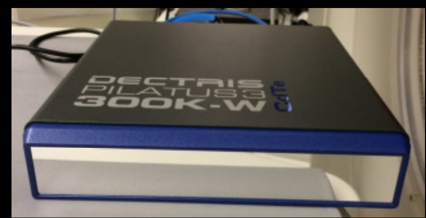
direct conversion
= photon counting
= energy selective







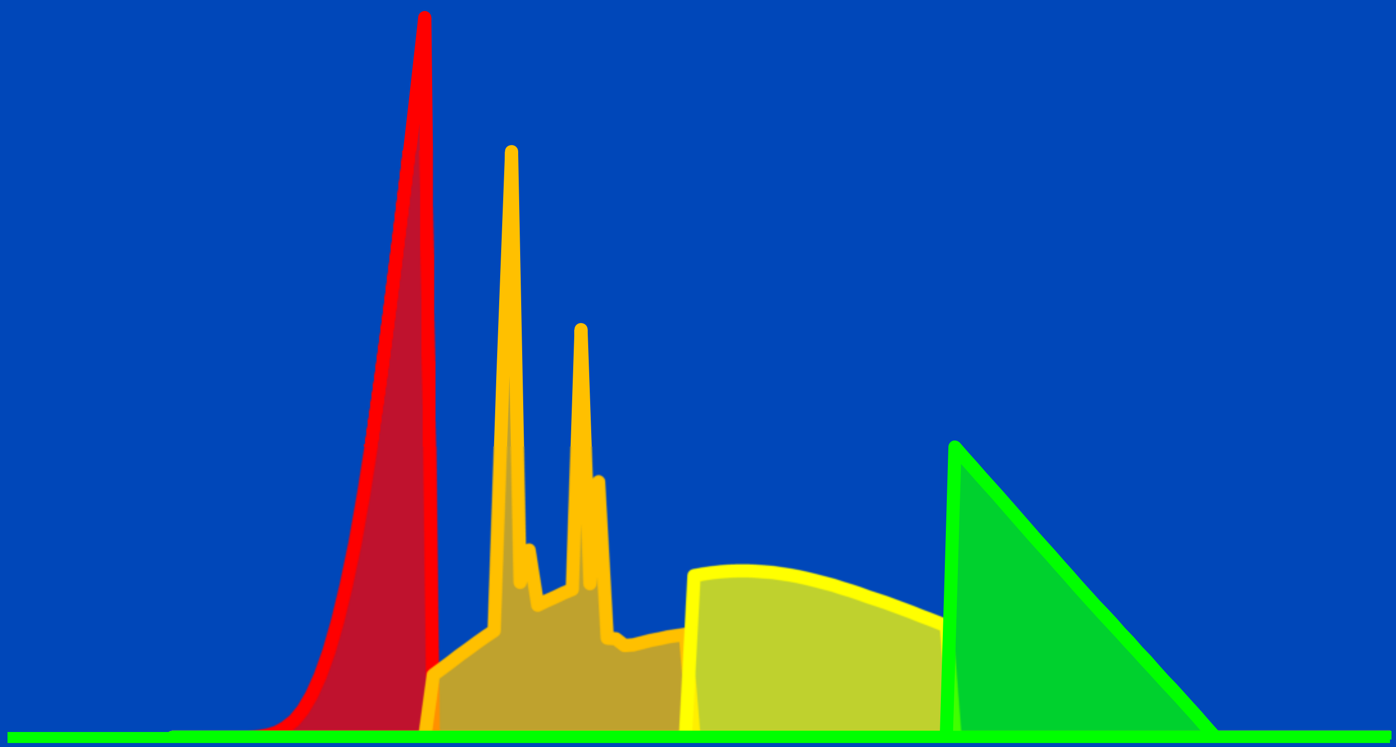
EI detector (Siemens)



PC detector (Dectris)

Energy-Selective Detectors: Improved Spectroscopy, Reduced Dose?

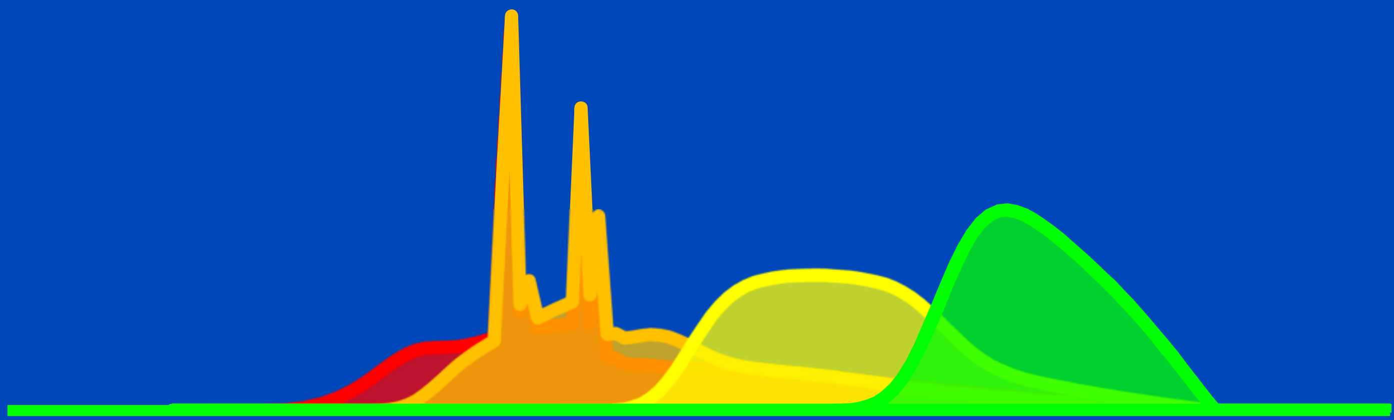
Ideally, bin spectra do not overlap, …



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

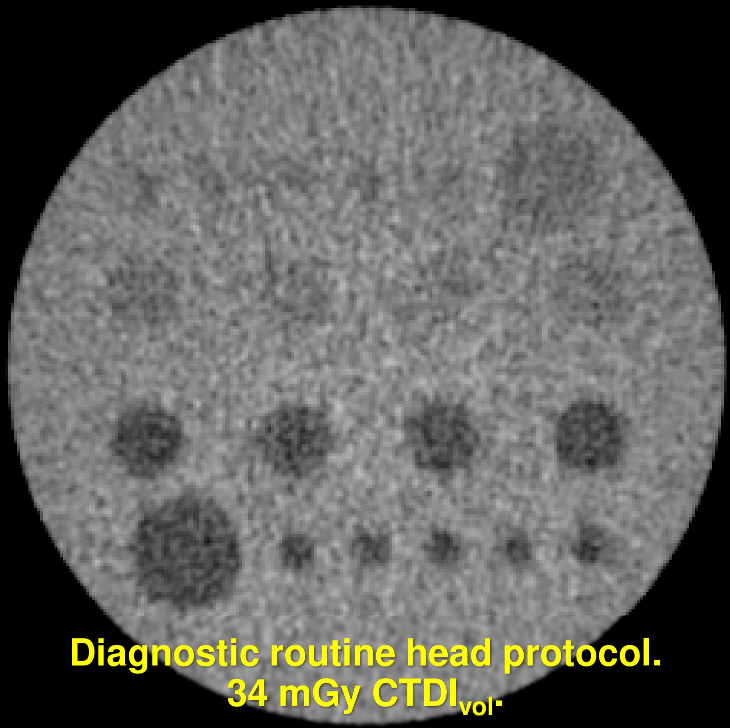
Energy-Selective Detectors: Improved Spectroscopy, Reduced Dose?

… realistically, however they do!



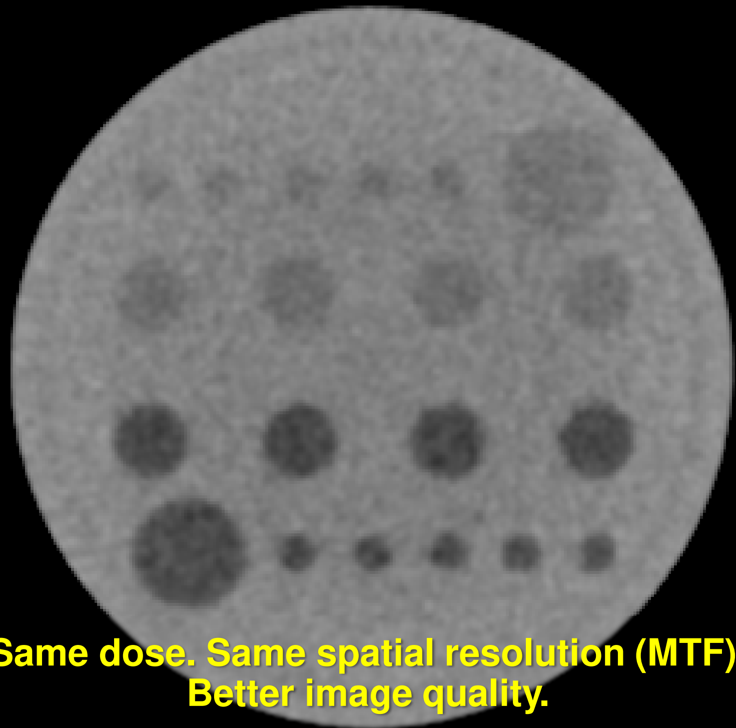
Spectra as seen after having passed a 32 cm water layer.

Diagnostic CT (Conventional Detector) of a Low Contrast Phantom

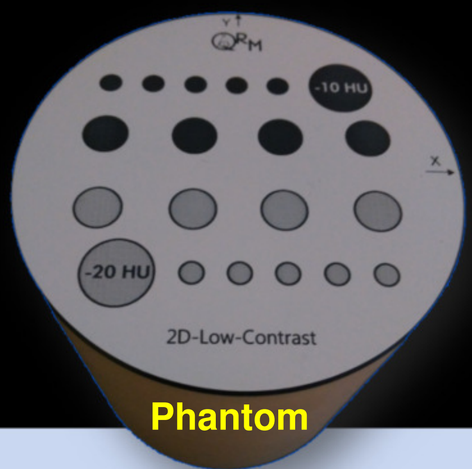


Diagnostic routine head protocol.
34 mGy $CTDI_{vol}$.

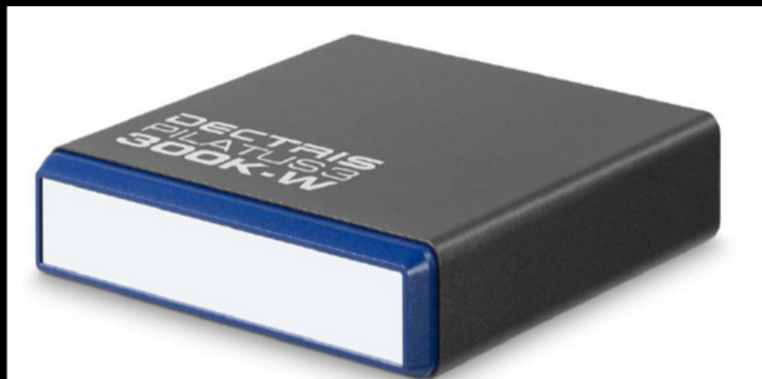
Photon Counting Detector CT of a Low Contrast Phantom



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



Phantom



Photon Counting Detector (Dectris)

C = 0 HU, W = 80 HU

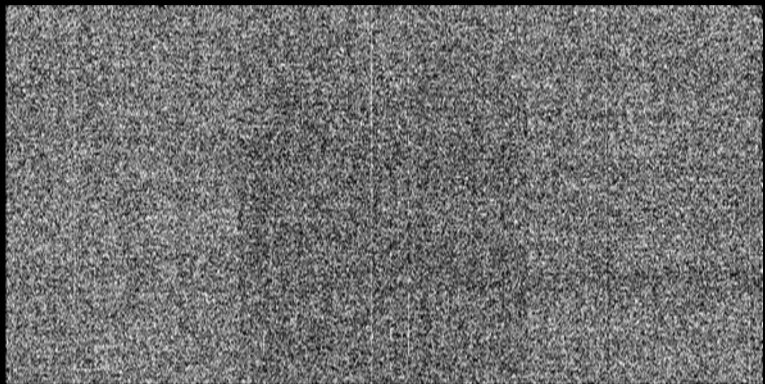
dkfz.

Dark Image of Photon Counter Shows Background Radiation

18 frames, 5 min integration time per frame

Events per Frame

Energy Integrating (Dexela)



C/W = 0 a.u./70 a.u.

Photon Counting (Dectris Santis)

C/W = 1 cnts/2 cnts

Dark current dominates.
Readout noise only.
Low flux events hidden!

No dark current.
No readout noise.
Low flux events visible!

DECTRIS

Santis: 1 mm CdTe, 150 μm pixel size, 4 thresholds.

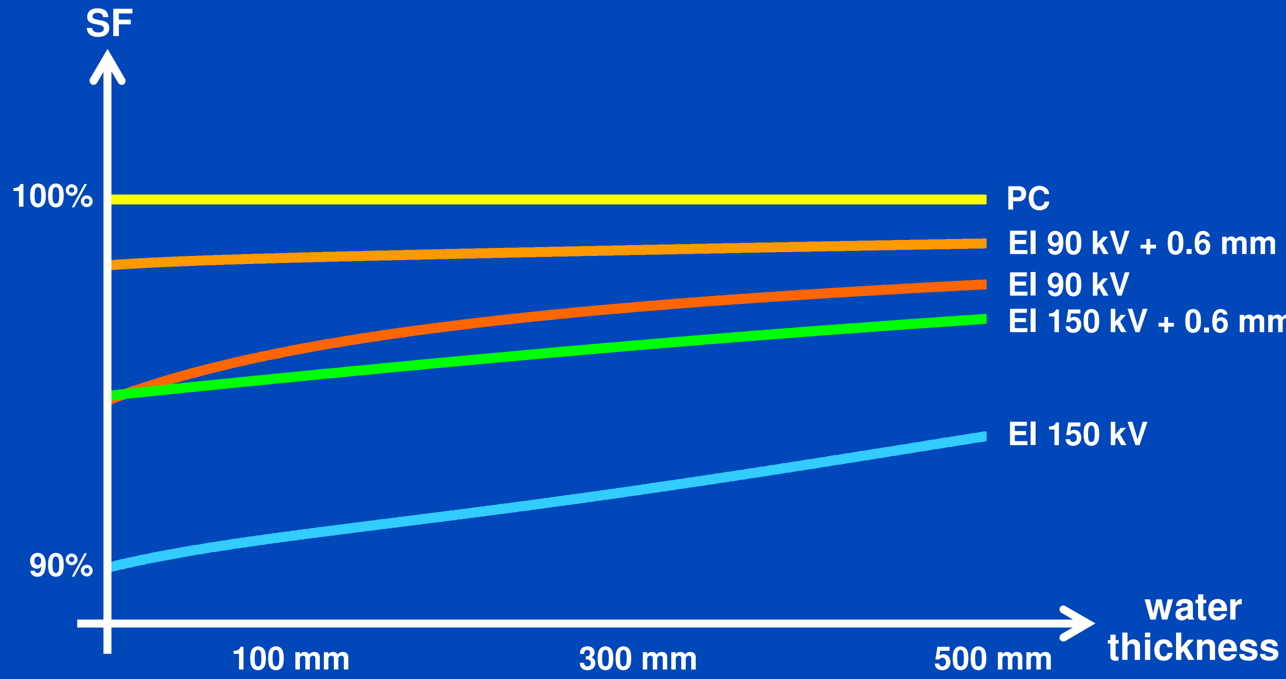
dkfz.

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
  – …

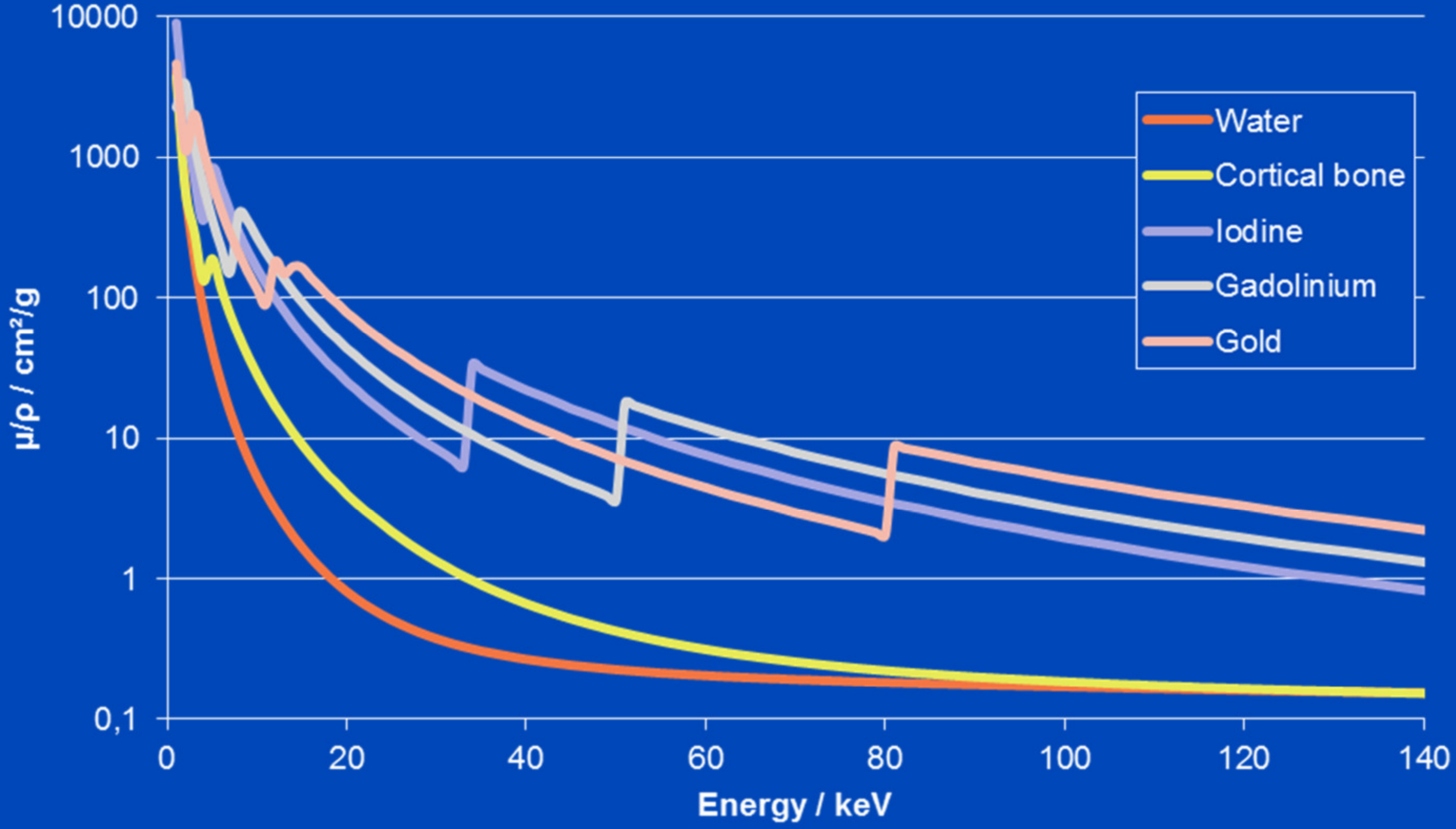
Swank Factor

- The Swank factor measures the relative SNR^2, and thus the relative dose efficiency between photon counting (PC) and energy integrating (EI).
- PC always has the better SNR!



$$\mathrm{SF} = \frac{\mathrm{SNR}_{\mathrm{EI}}^2}{\mathrm{SNR}_{\mathrm{PC}}^2} = \frac{\left(\int dE\, E N(E)\right)^2}{\left(\int dE\, N(E)\right)\left(\int dE\, E^2 N(E)\right)} = \frac{M_1^2}{M_0 M_2} \leq 1$$

Iodine Contrast

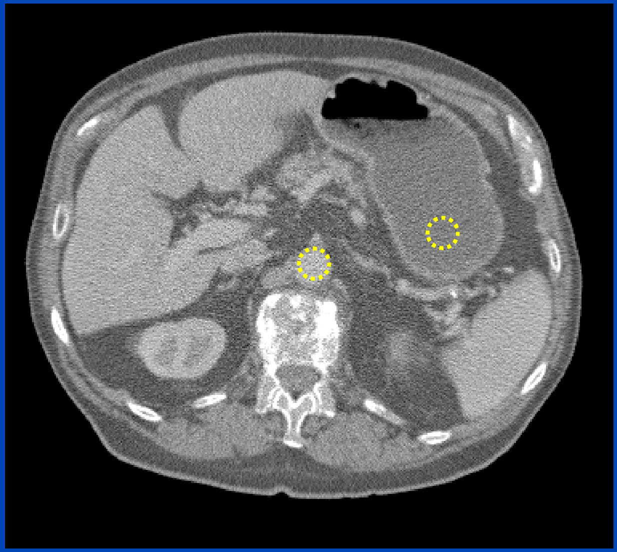


120 kV water transmission curves (gray) given in relative units on a non-logarithmic ordinate.

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

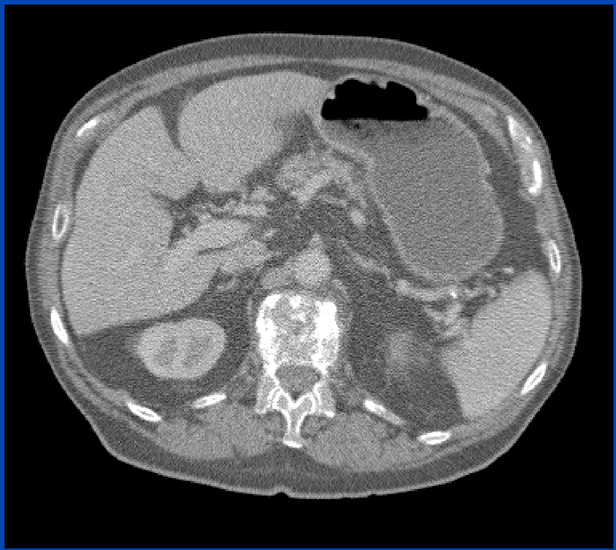
$$\mathrm{CNR}^2 = \frac{\left(\sum_b w_b C_b\right)^2}{\sum_b w_b^2 V_b}$$

- At the optimum this evaluates to

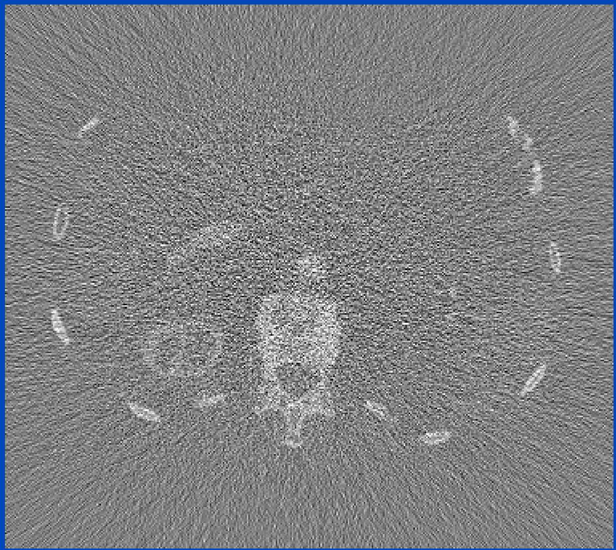
$$\mathrm{CNR}^2 = \sum_{b=1}^{B} \mathrm{CNR}_b^2$$

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

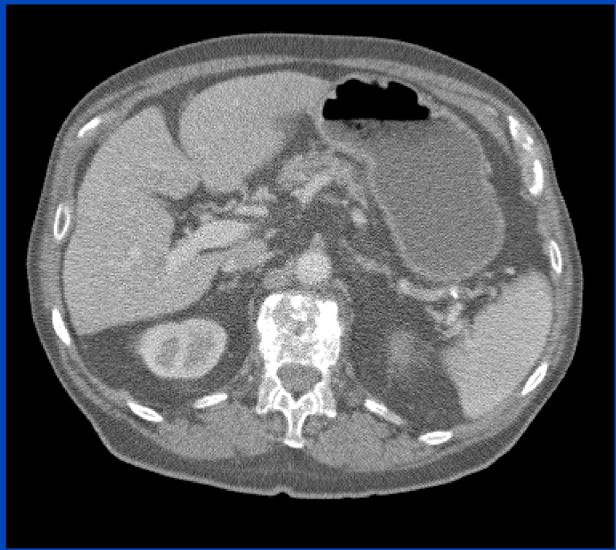
Energy Integrating



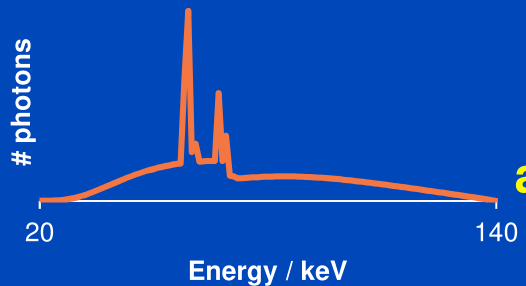
PC minus EI



Photon Counting

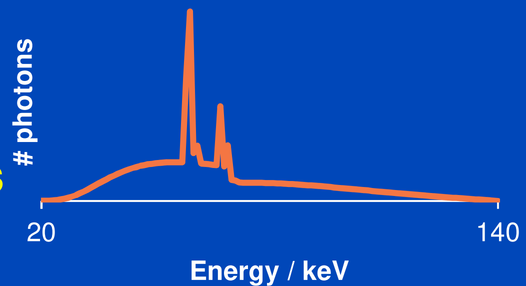


CNR = 2.11



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

CNR = 2.95

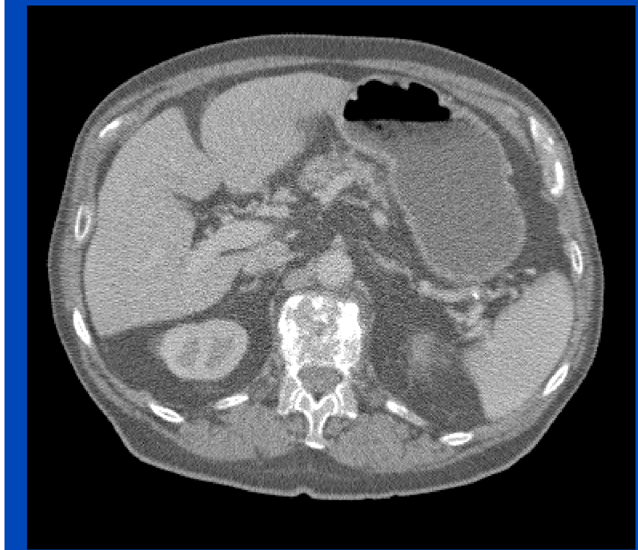


Images: *C* = 0 HU, *W* = 700 HU, difference image: *C* = 0 HU *W* = 350 HU, bins start at 20 keV

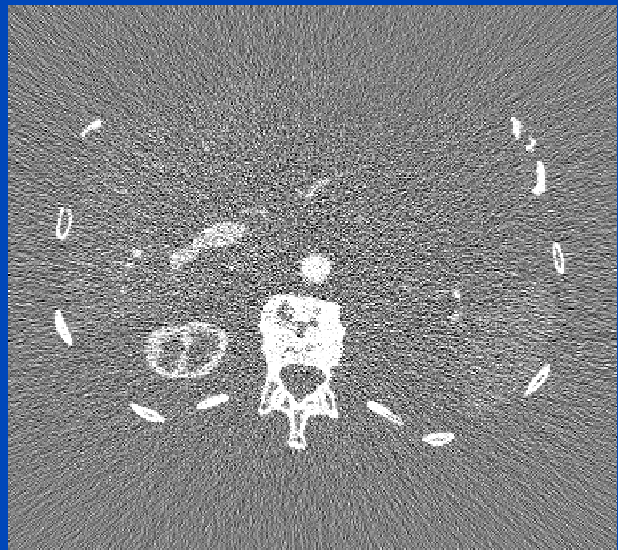
dkfz.

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

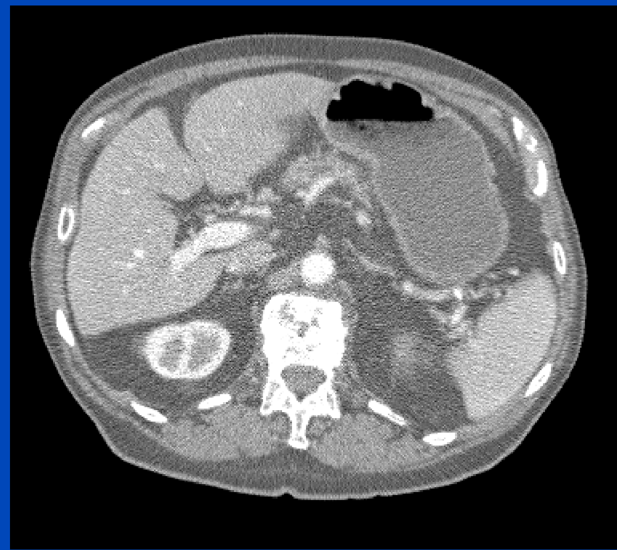
Energy Integrating



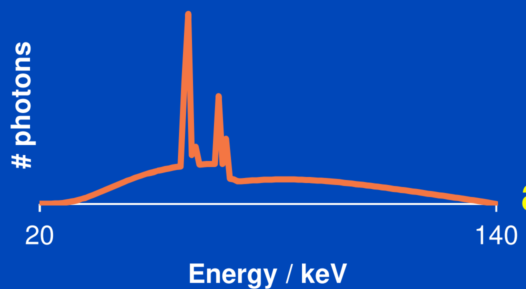
PC minus EI



Photon Counting

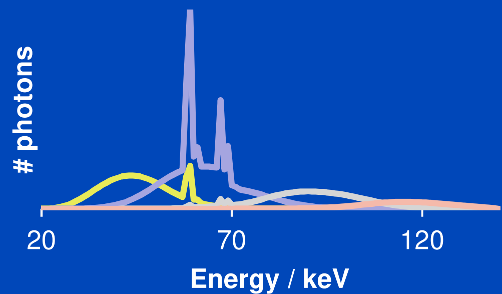


CNR = 2.11



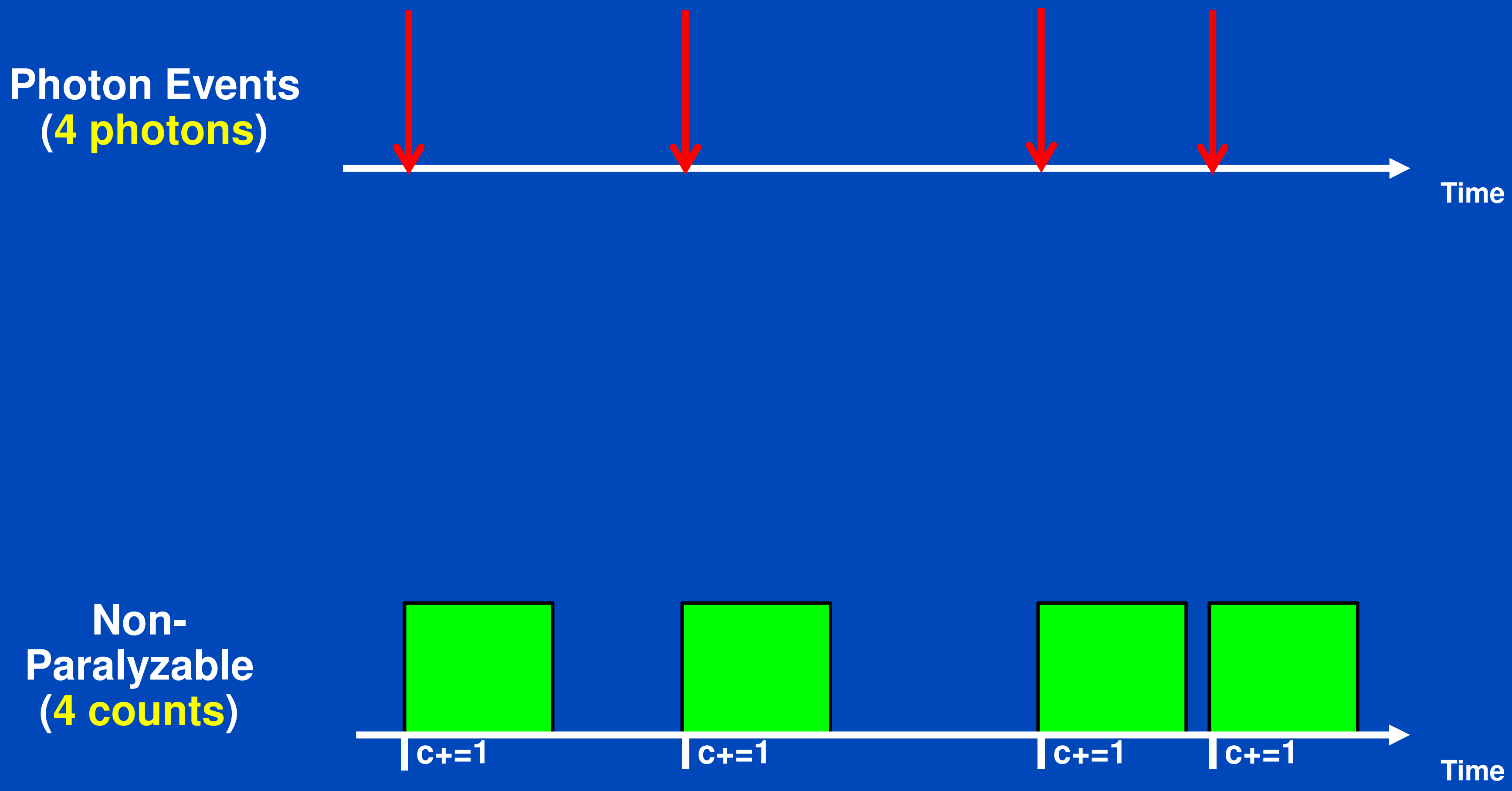
99% CNR improvement or
75% dose reduction achievable
due to improved Swank factor
and optimized energy weighting.

CNR = 4.19



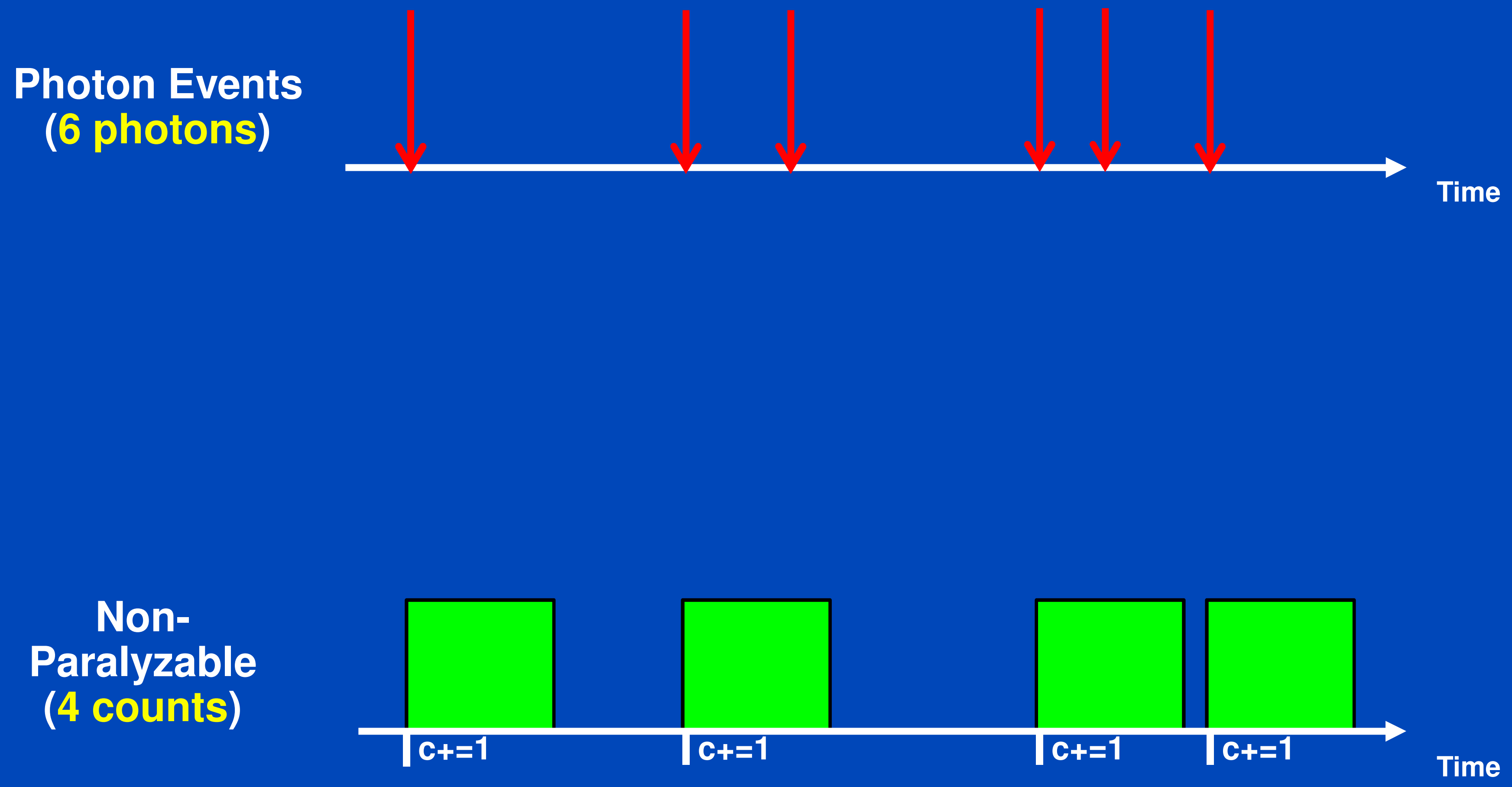
Images: *C* = 0 HU, *W* = 700 HU, difference image: *C* = 0 HU, *W* = 350 HU, bins start at 20 keV

Pulse Pile-Up: Medium Flux Rate



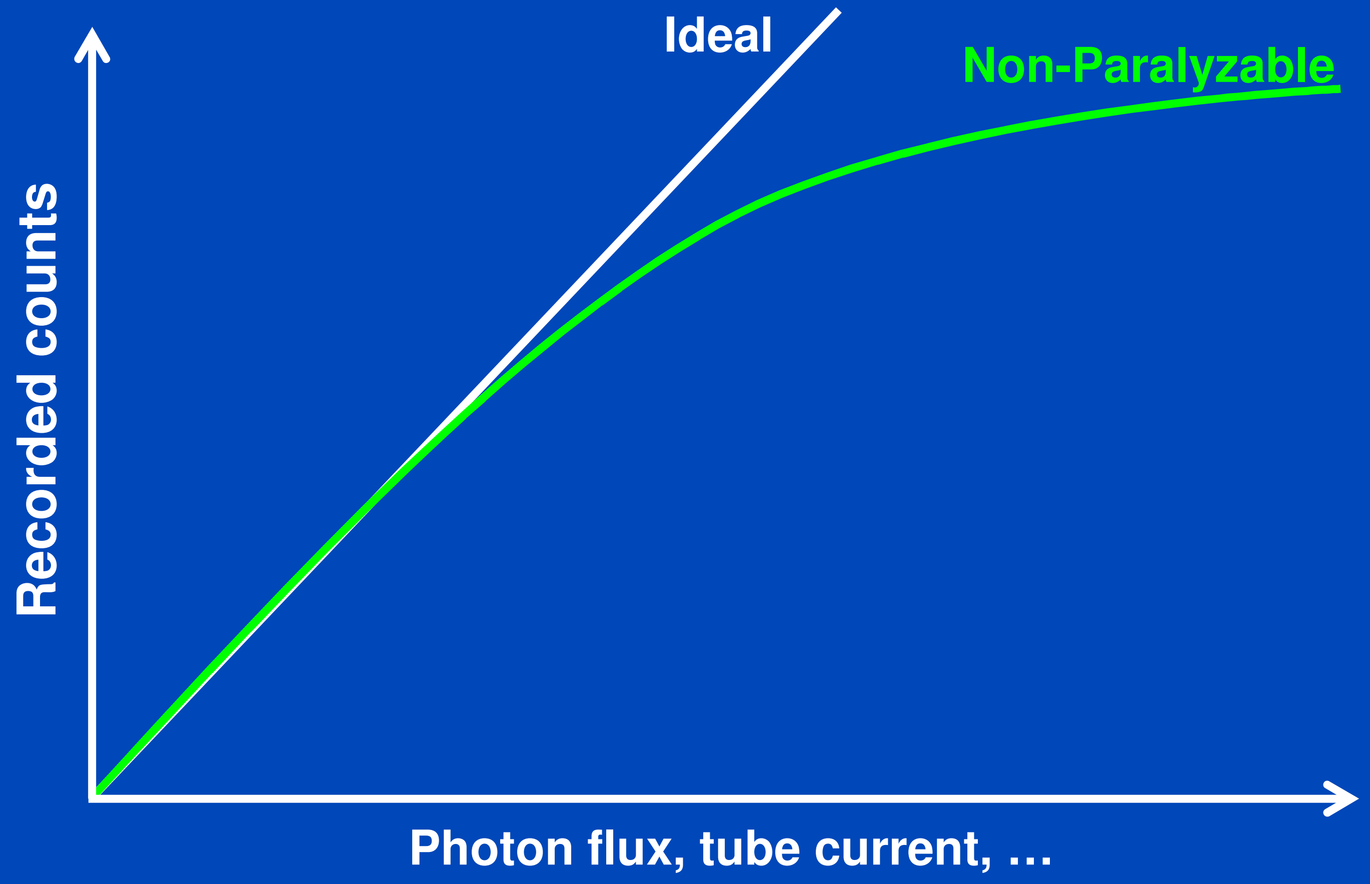
Boxes illustrate deadtime

Pulse Pile-Up: High Flux Rate



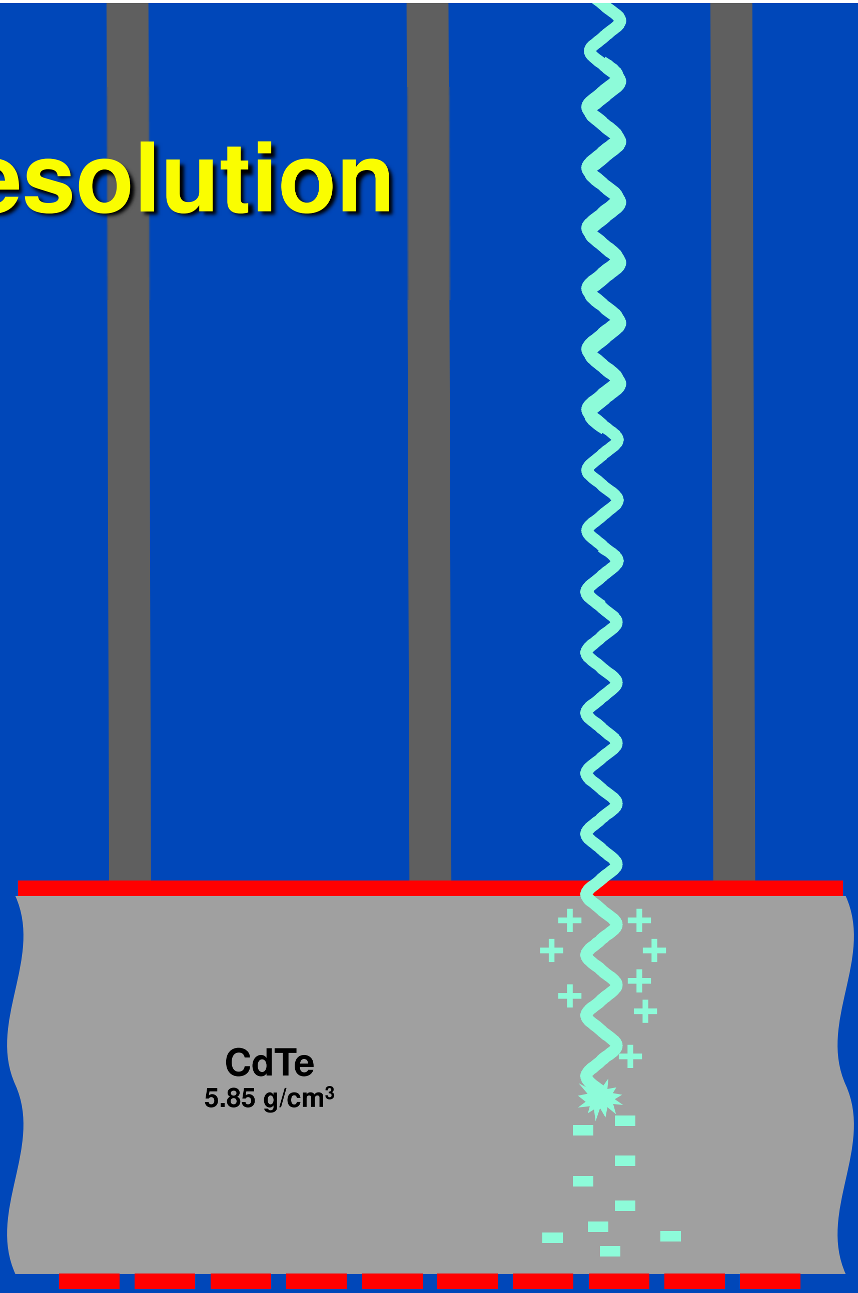
Boxes illustrate deadtime

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.



Ultra-High Resolution on Demand

Energy Integrating CT
(Somatom Flash)

Photon Counting CT
(Somatom Count. UHR-Mode)

MAYO
CLINIC

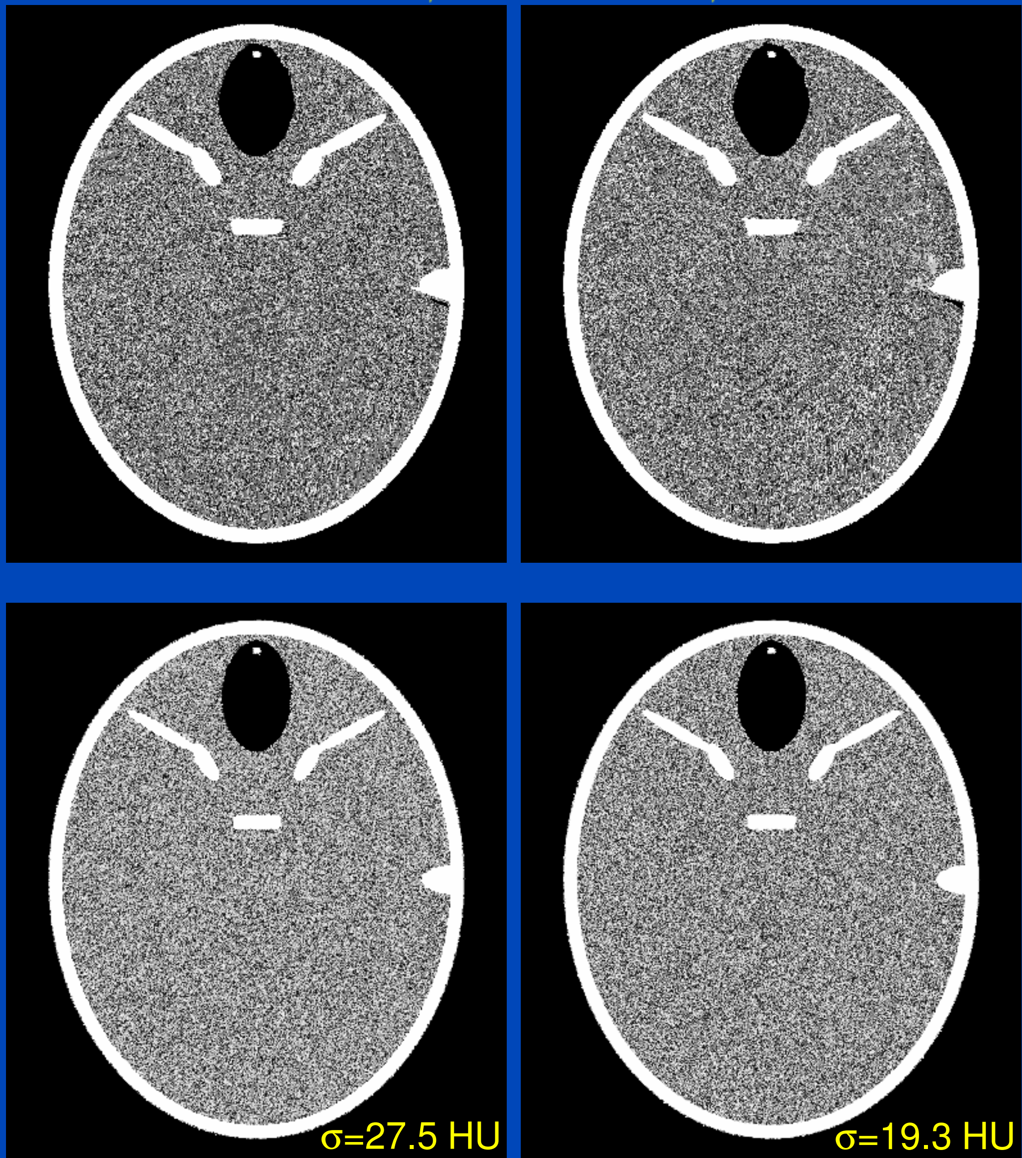
Courtesy of Cynthia McCollough, Mayo Clinic, Rochester, USA.

32×0.6 mm — $s(z) = \mathrm{II}^*_S(z), a(z) = \mathrm{II}^*_{S/2}(z)$

64×0.3 mm — $s(z) = \mathrm{II}^*_{S/2}(z), a(z) = \mathrm{II}^*_S(z)$

noise-free
C = 50 HU
W = 50 HU

with noise
C = 50 HU
W = 200 HU



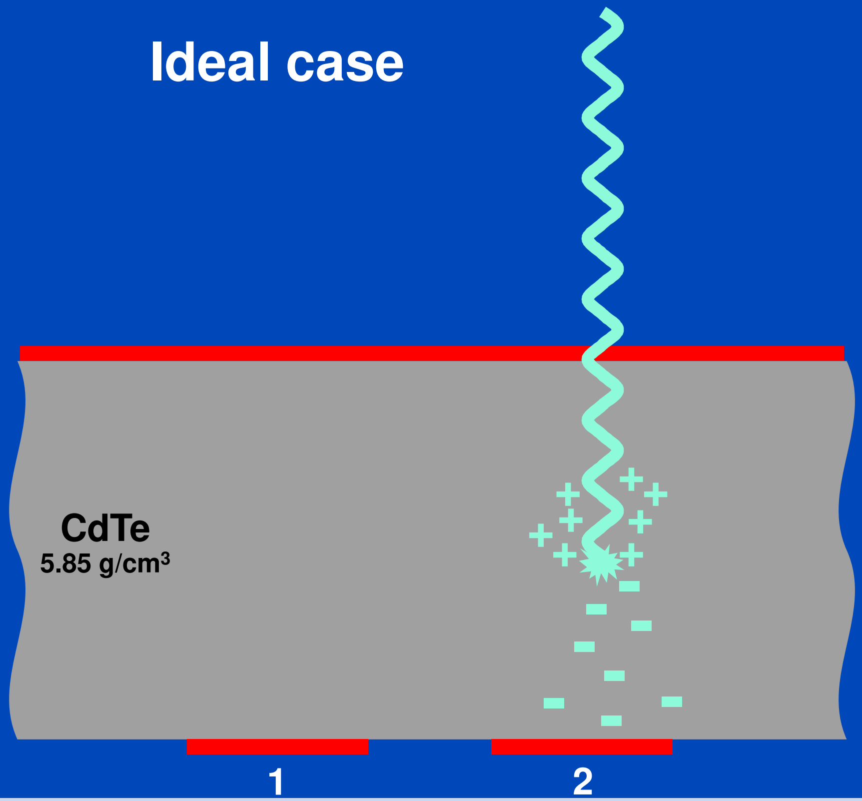
- To bin or not to bin?
- Simulated data
- $\mathrm{SSP}(z) = \mathrm{II}^{**}_{S,S/2}(z)$
- S_{eff} = 0.6 mm
- 1.78-fold dose usage with highres detector:

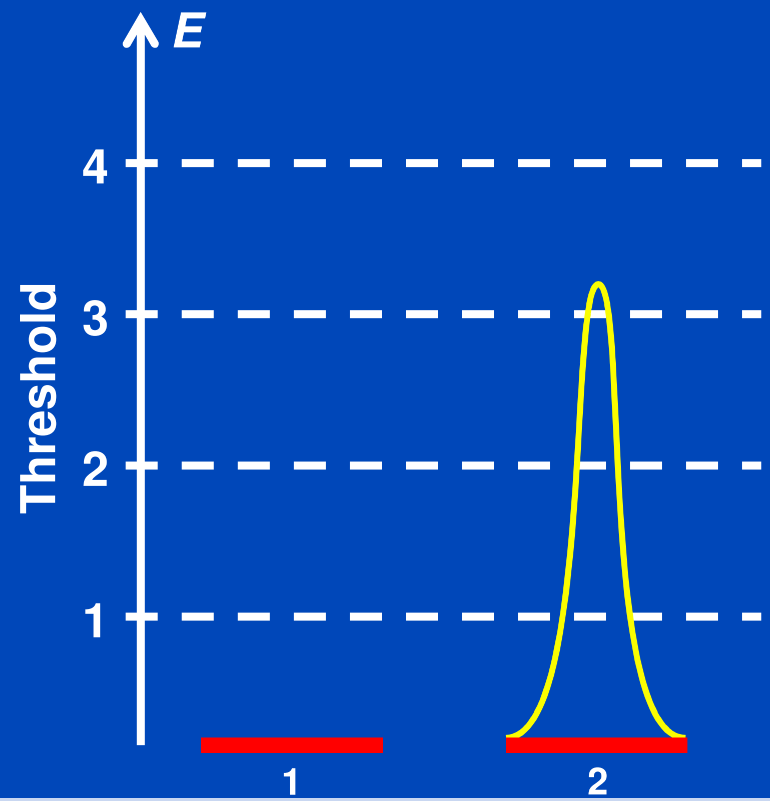
$$\underbrace{\left(\frac{27.5}{19.3}\right)^2}_{2.0} \frac{1+0.1/0.6}{1+\underset{\uparrow\ \text{0.1 mm septa}}{0.1}/0.3} \approx 1.78$$

- **44% dose reduction with highres detector**
- **Do not bin!**

Spectral Resolution

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

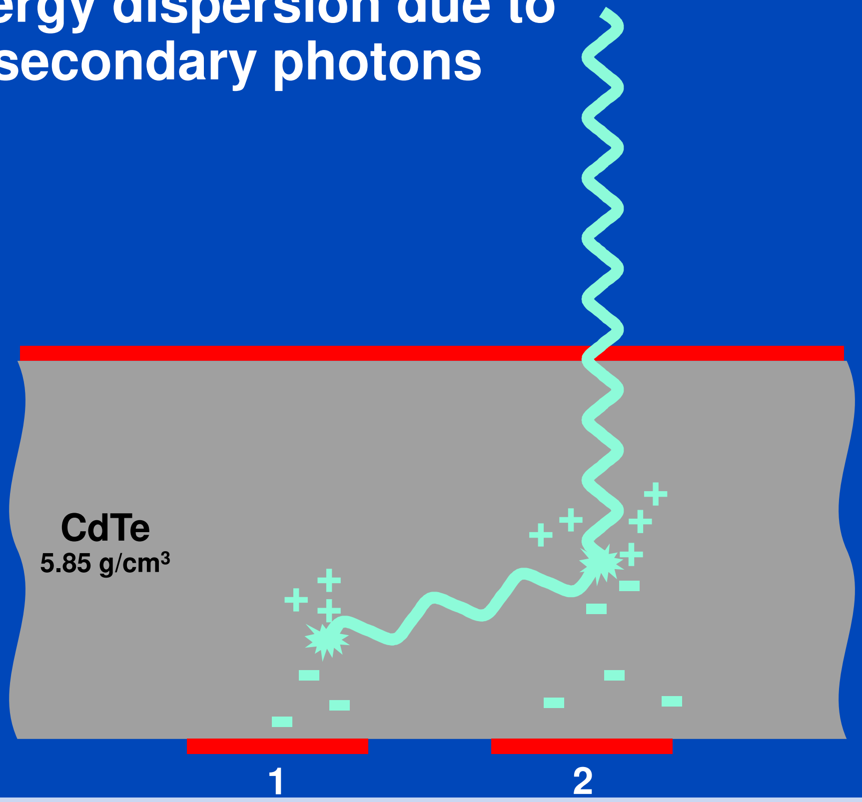


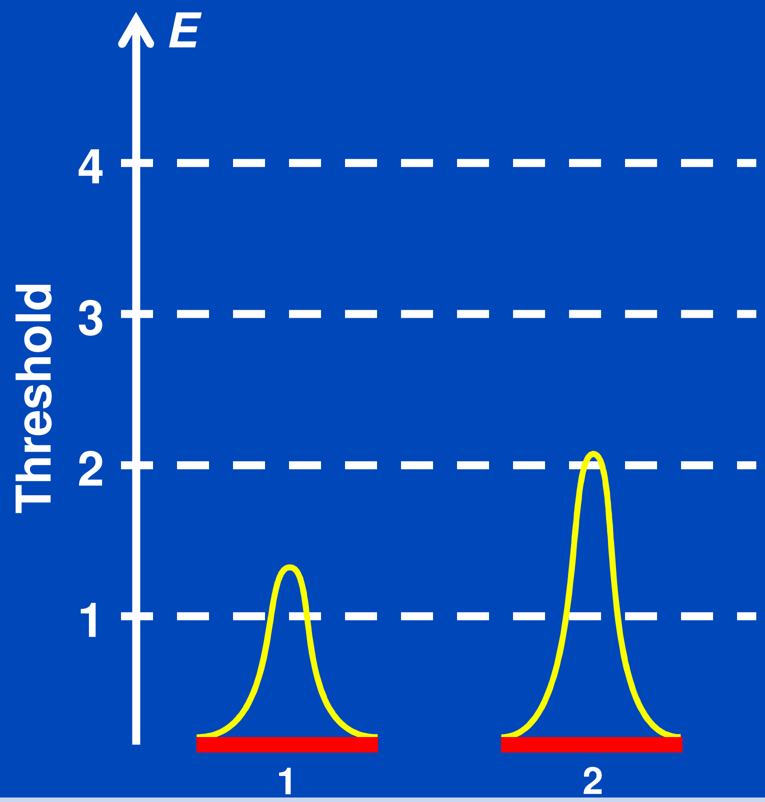


Spectral Resolution

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

Energy dispersion due to
secondary photons

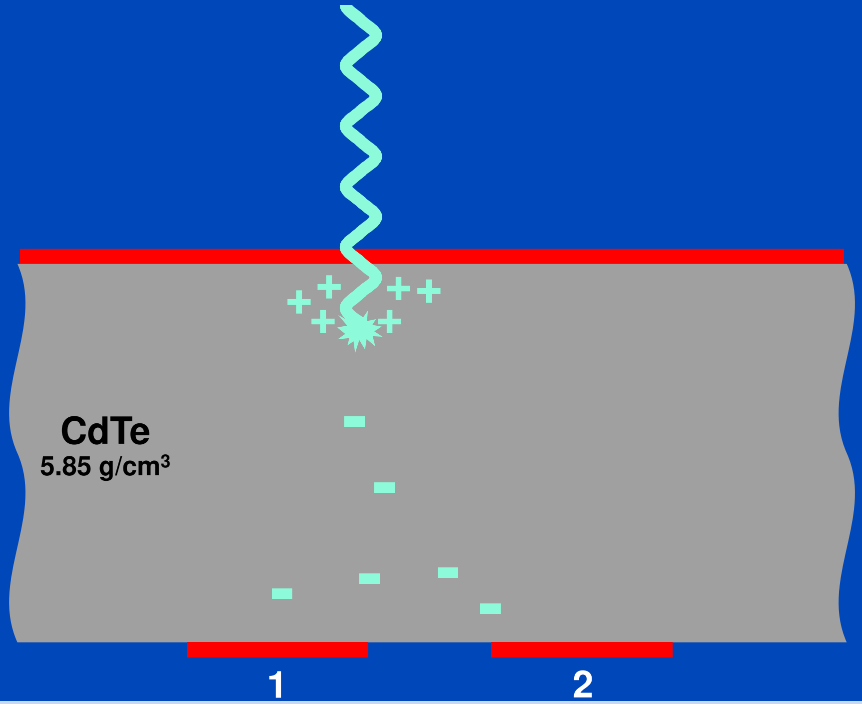


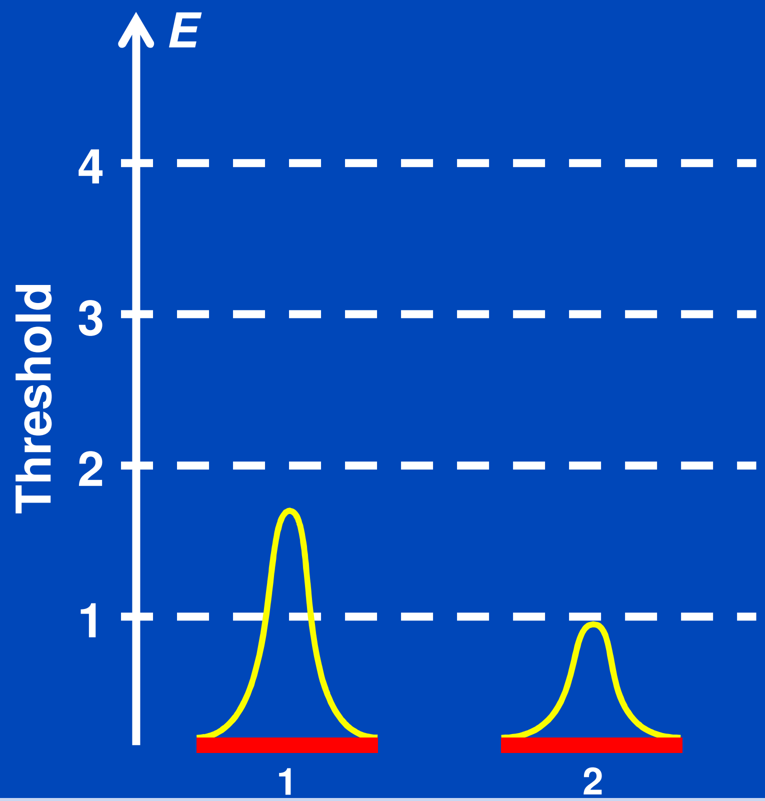


Spectral Resolution

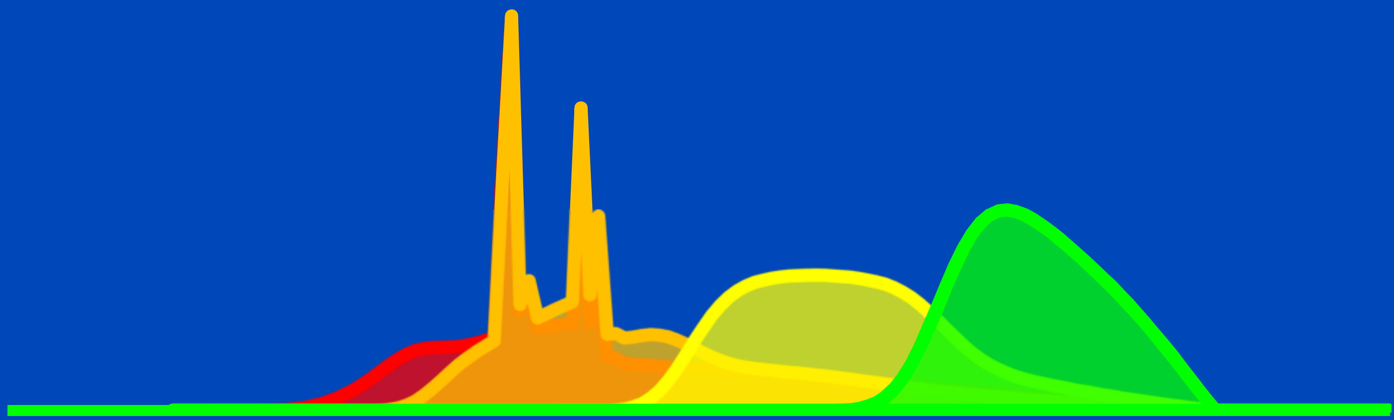
- Detection process in the sensor
- Photoelectric effect (e.g. 30 keV), charge sharing

Energy dispersion due to charge diffusion





Spectral Resolution

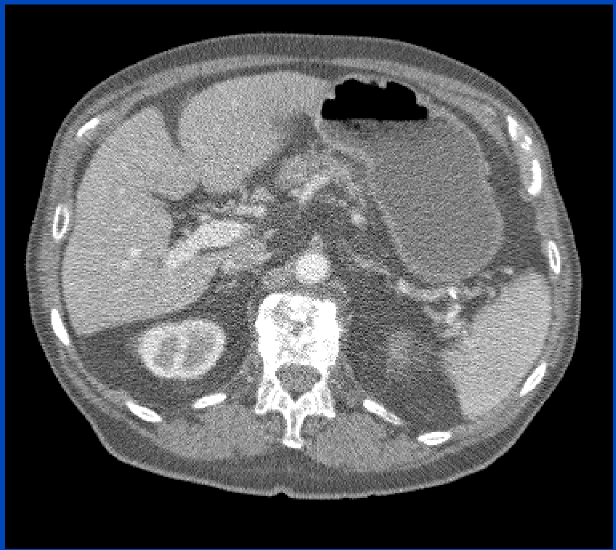


Spectra as seen after having passed a 32 cm water layer.

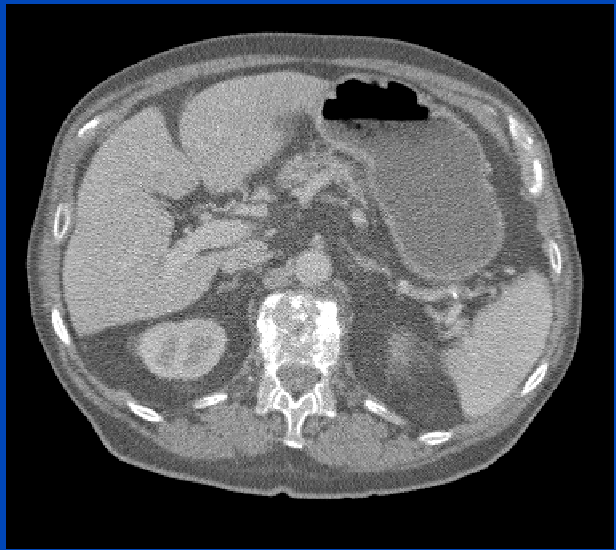
Photon Counting used for Spectral Imaging

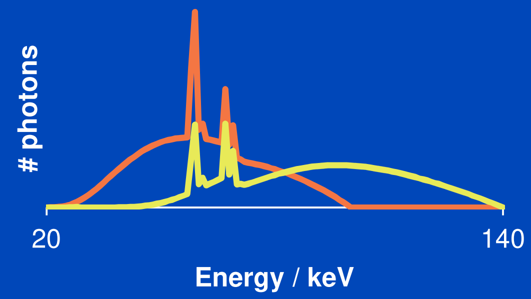
- DECT scan with 100 kV / 140 kV Sn
- Photon counting acquisition at 140 kV
- Same patient dose in both cases

100 kV



140 kV Sn

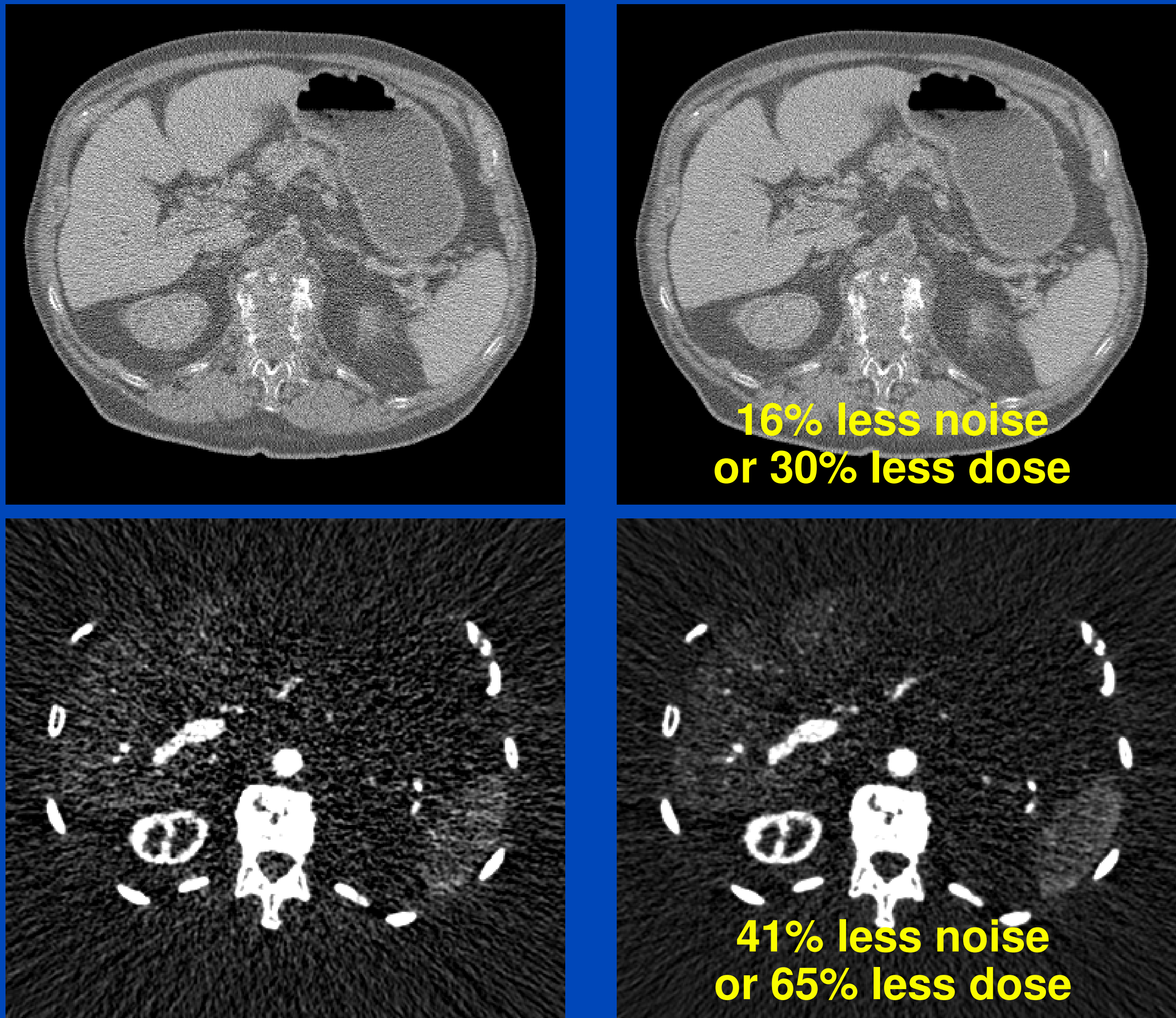




Images: *C* = 0 HU, *W* = 700 HU

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

dkfz.

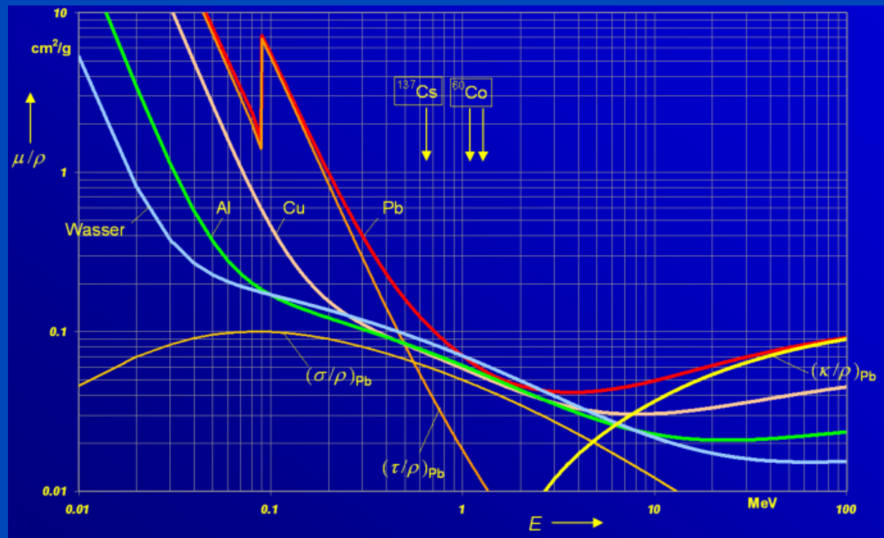
More than Dual Energy?

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

$$\mu(E) = \underbrace{\cancel{\eta(E)}}_{\text{Rayleigh}} + \underbrace{\tau(E)}_{\text{Photo}} + \underbrace{\sigma(E)}_{\text{Compton}} + \underbrace{\cancel{\kappa(E)}}_{\text{Pair}}$$

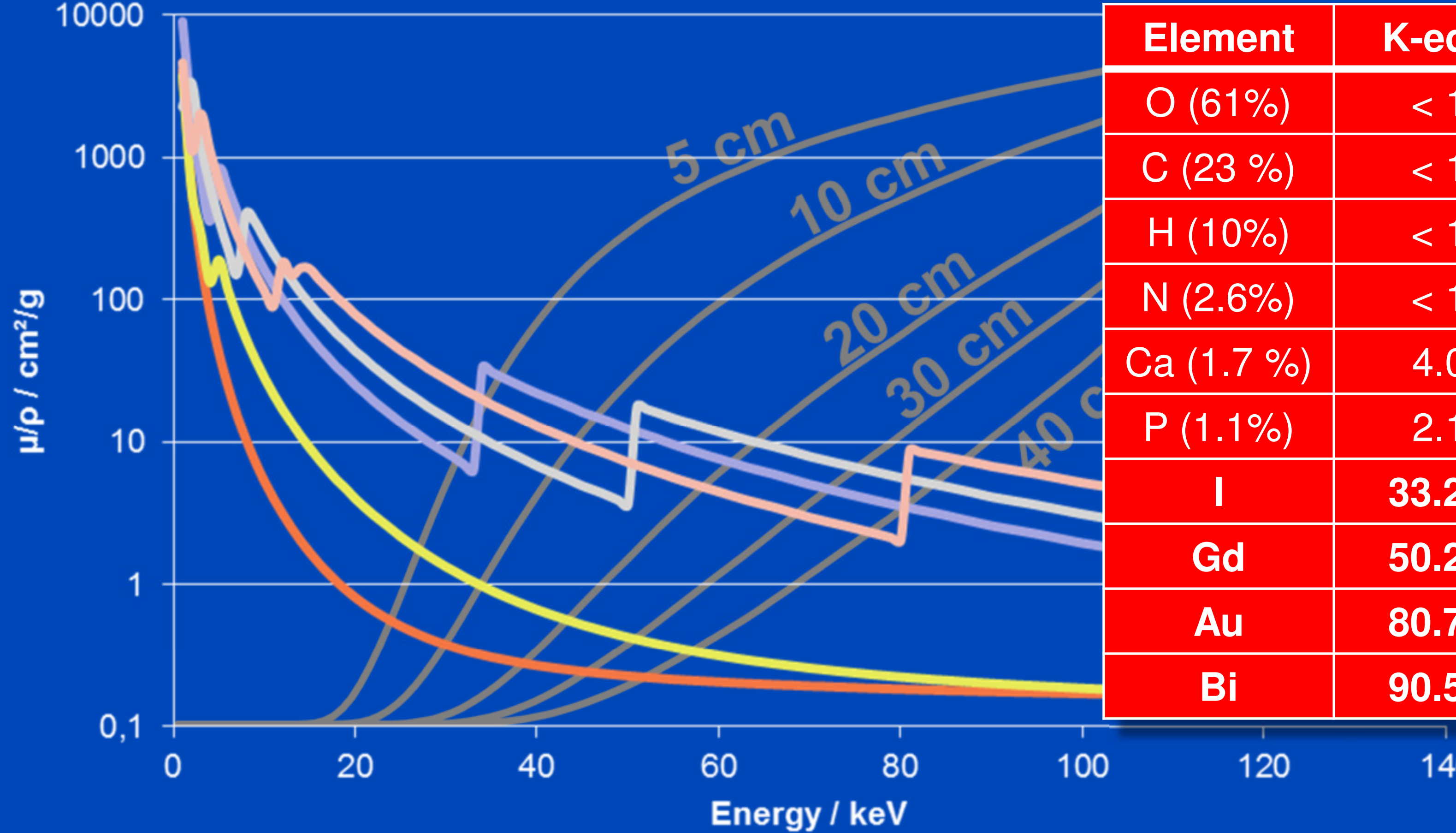
$$\tau(E) \propto \rho \frac{Z^3}{E^3}$$

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



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)?

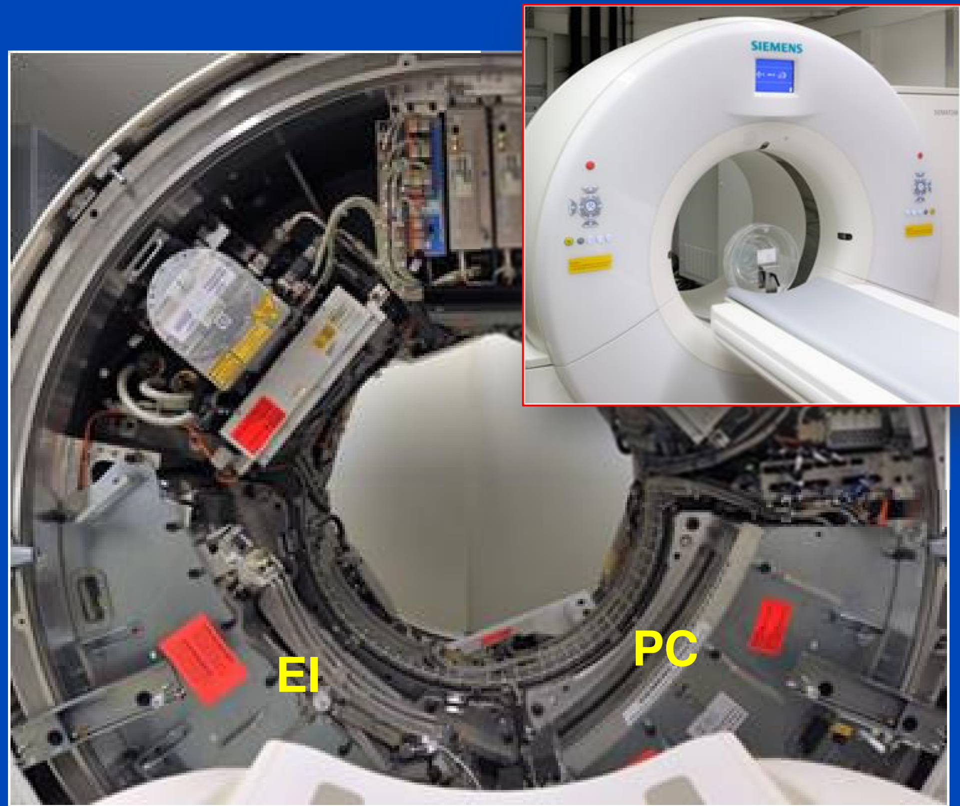
Macro

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.

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

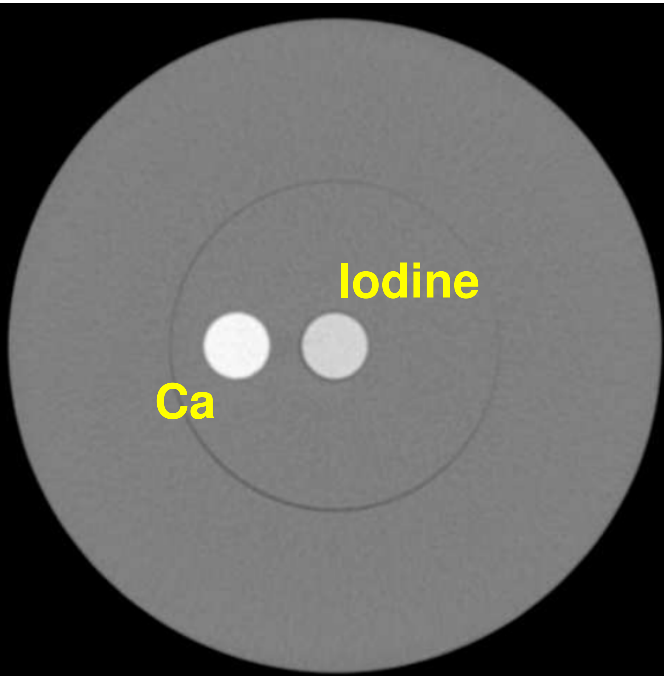
12	12	12	12
12	12	12	12
12	12	12	12
12	12	12	12

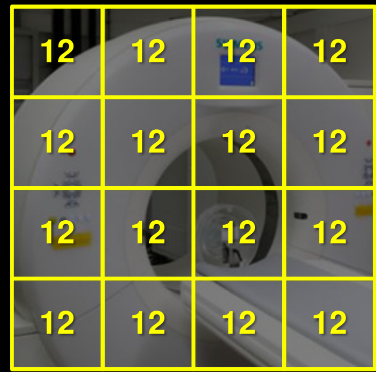
No FFS on thread B (photon counting detector).
4×4 subpixels of 225 µm size = 0.9 mm pixels (0.5 mm at isocenter).
The whole detector consists of 128×1920 subpixels = 32×480 macro pixels.

DECT

Ca-I Decomposition

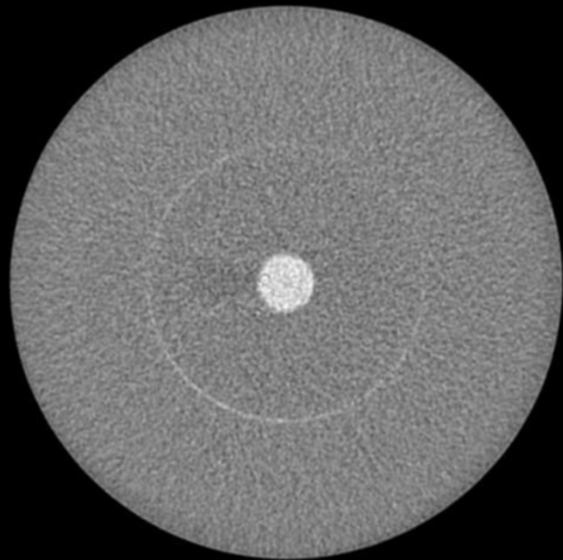
Macro mode
140 kV, 25/65 keV
***C* = 0 HU, *W* = 1200 HU**





Calcium image

Iodine image

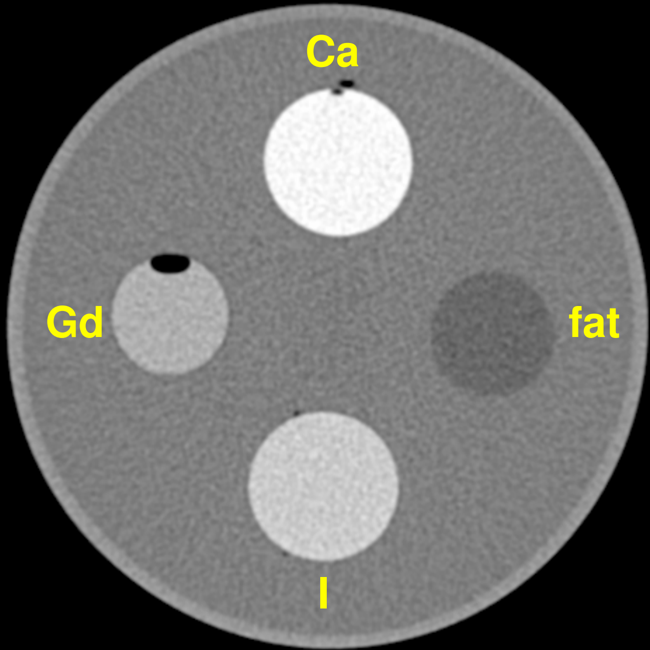


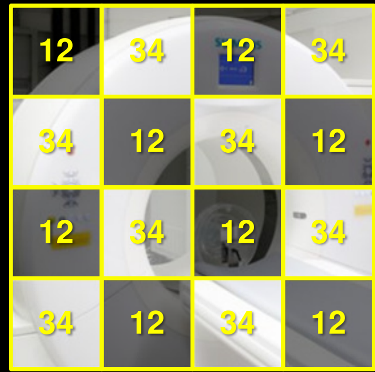
Courtesy of Siemens Healthcare

MECT

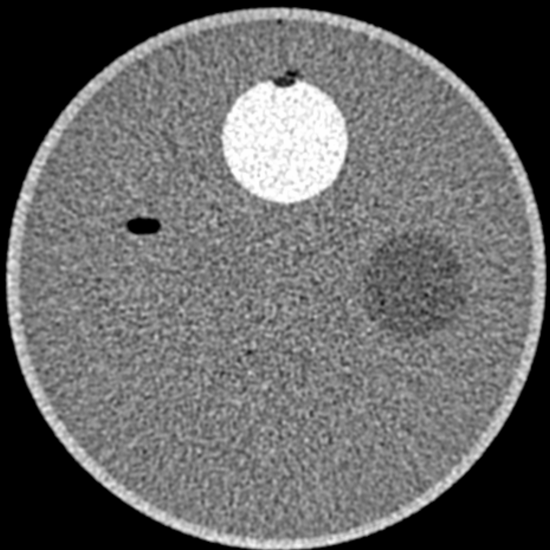
Ca-Gd-I Decomposition

Chess pattern mode
140 kV, 20/35/50/65 keV
C = 0 HU, W = 1200 HU

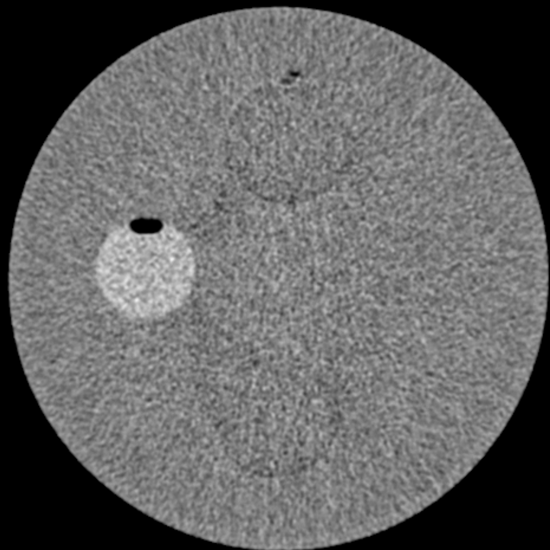




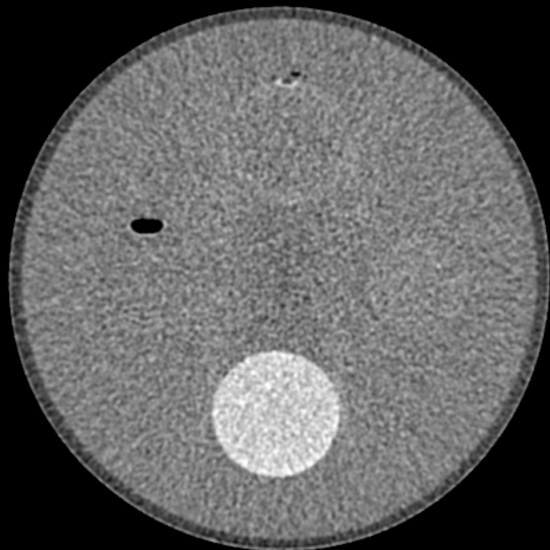
Calcium image



Gadolinium image

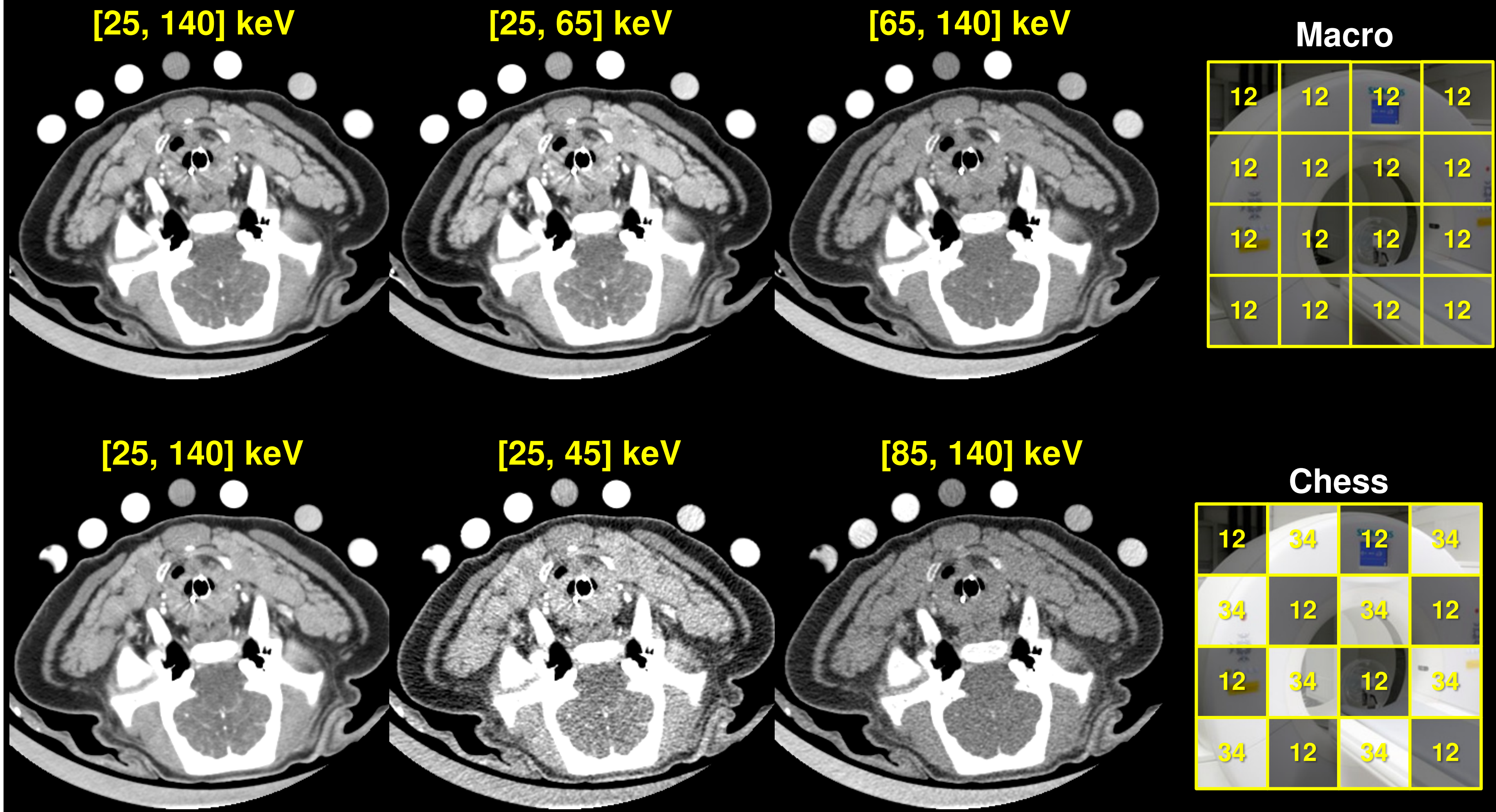


Iodine image



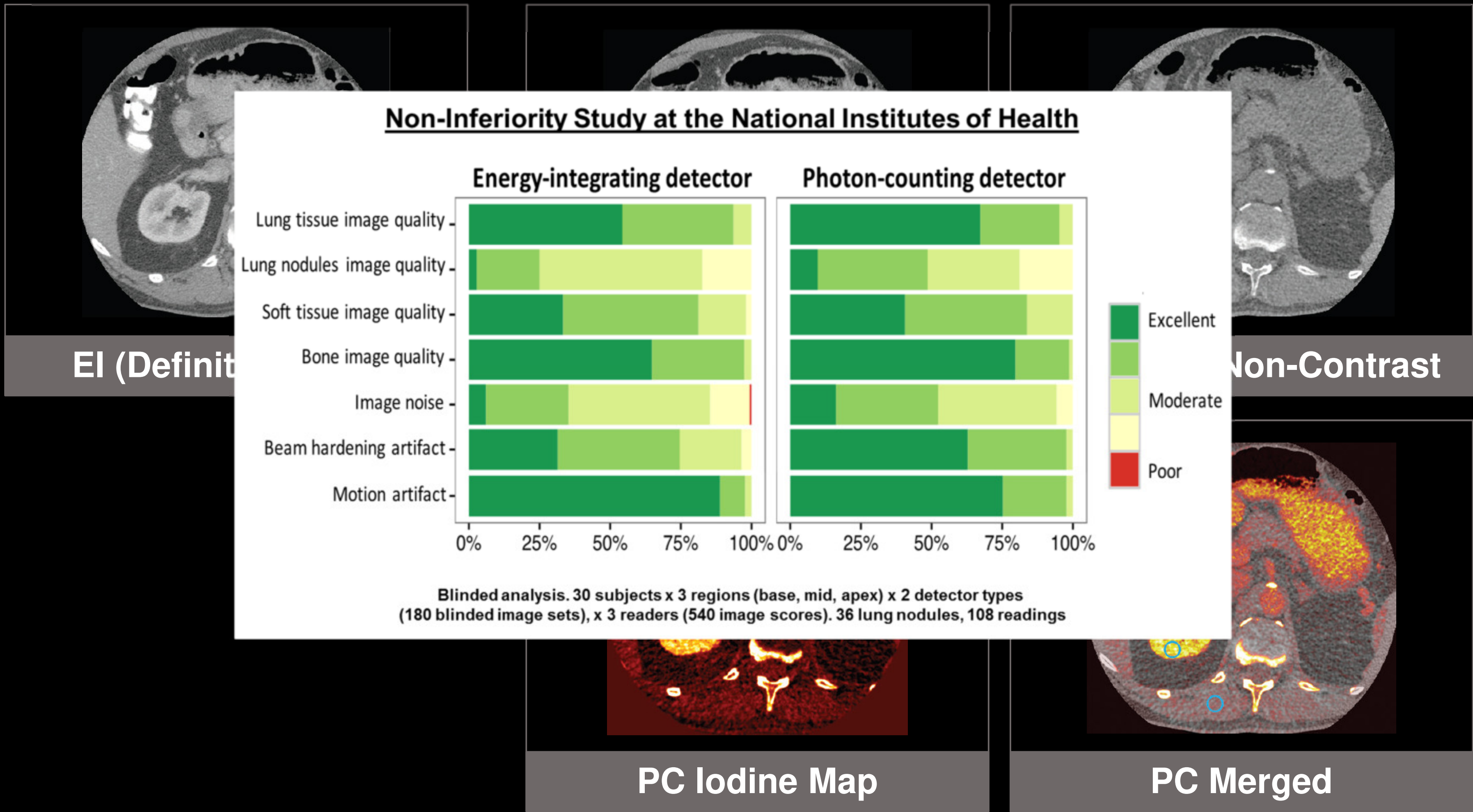
Courtesy of Siemens Healthcare

Preclinical Study
(40 kg swine, iodine contrast)



Courtesy of Mayo Clinic Rochester, USA, and of Siemens Healthcare, Forchheim, Germany

First Peer Reviewed Publication on CounT from NIH February 2016



Courtesy of National Institutes of Health, Berthesda, USA

Pourmorteza A et al., Abdominal Imaging with Contrast-enhanced Photon-counting CT: First Human Experience. Radiology. 2016 Apr;279(1):239-45

Ultra-High Resolution on Demand

Energy Integrating CT
(Somatom Flash)

Photon Counting CT
(Somatom Count. UHR-Mode)

MAYO CLINIC

Courtesy of Cynthia McCollough, Mayo Clinic, Rochester, USA.

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
  - energy bin weighting
  - no electronic noise
  - Swank factor = 1
  - smaller pixels
- Spectral information on demand
  - single energy
  - dual energy
  - multiple energy
  - virtual monochromatic
  - K-edge imaging
  - ...

Potential clinical impact


Thank You!
Job opportunities through DKFZ’s international PhD or Postdoctoral Fellowship programs (www.dkfz.de), or directly through Marc Kachelriess (marc.kachelriess@dkfz.de).
Parts of the reconstruction software were provided by RayConStruct® GmbH, Nürnberg, Germany.