Photon Counting CT

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dkfz. DEUTSCHES KREBSFORSCHUNGSZENTRUM IN DER HELMHOLTZ-GEMEINSCHAFT

Diagnostic PCCT Systems
(as of August 2022)

	Detector material	Detector pixel size at iso	Field of Measurement	Bins	Approved product?	Publications (scopus search)	
Canon	CdZnTe		50 cm	5	no		
GE	Si (edge on)				no		
Philips	CdZnTe	275 × 275 µm	50 cm	5	no	≈22	
Siemens CounT	dual source GOS/CdTe	2· 250 × 250 µm	50 / 27.5 cm	4	no	≈50	
Siemens CounT+	CdTe	2· 150 × 176 µm	50 cm	4	no	≈11	**> 100**
Siemens Naeotom Alpha	dual source CdTe	2· 150 × 176 µm	50 / 36 cm	4	yes ≈40 installed	≈40	

Edge on design (Si only)

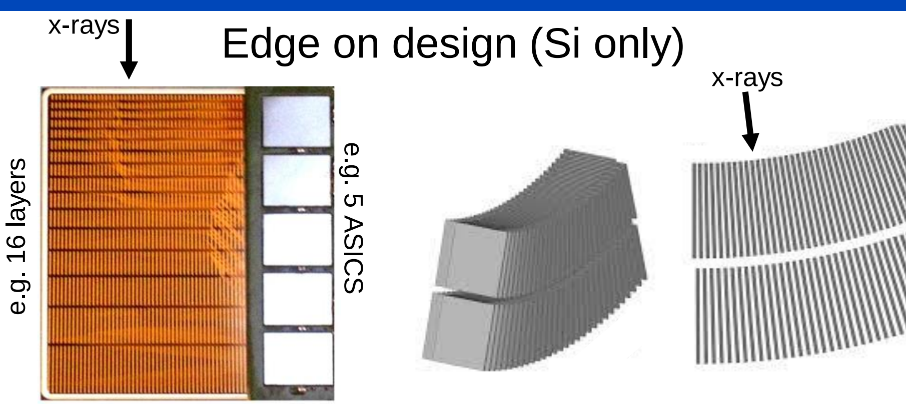

Image courtesy of KTH Royal Institute of Technology

Face on design (all others)

Image courtesy of Siemens Healthineers

The additional factor 2 in the detector pixel size column indicates that some scan modes may use binning.

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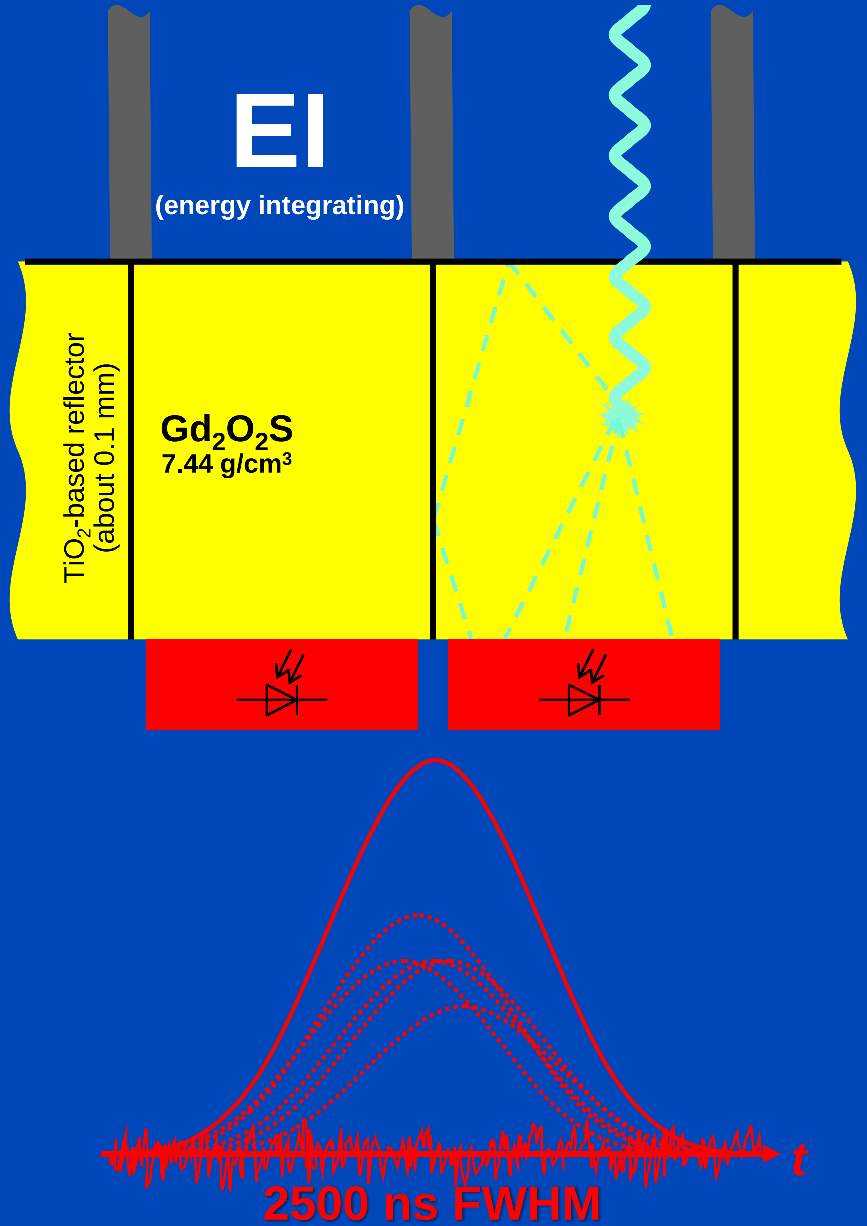
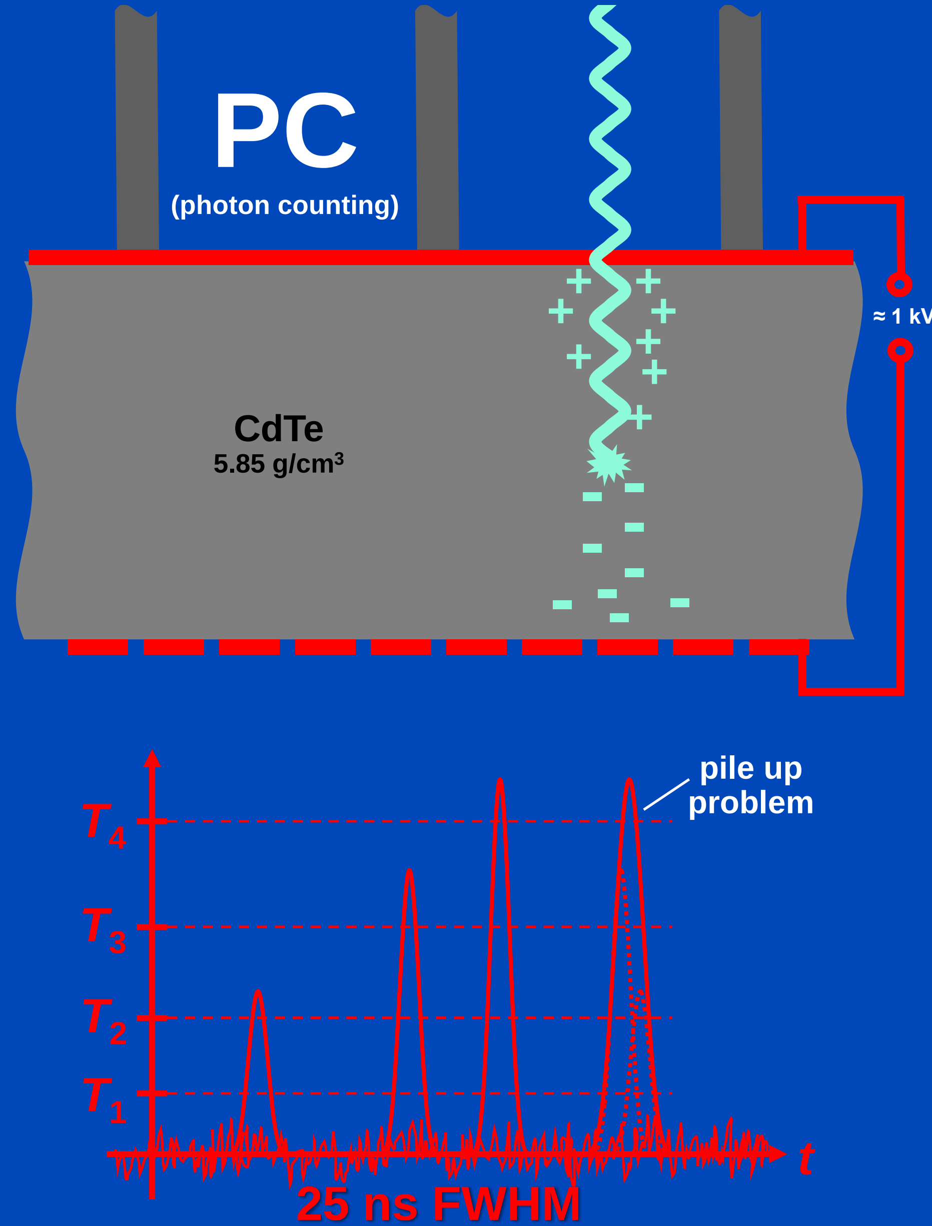
Indirect Conversion (Today)



i.e. max O(40·10^3) cps

Direct Conversion (Future)

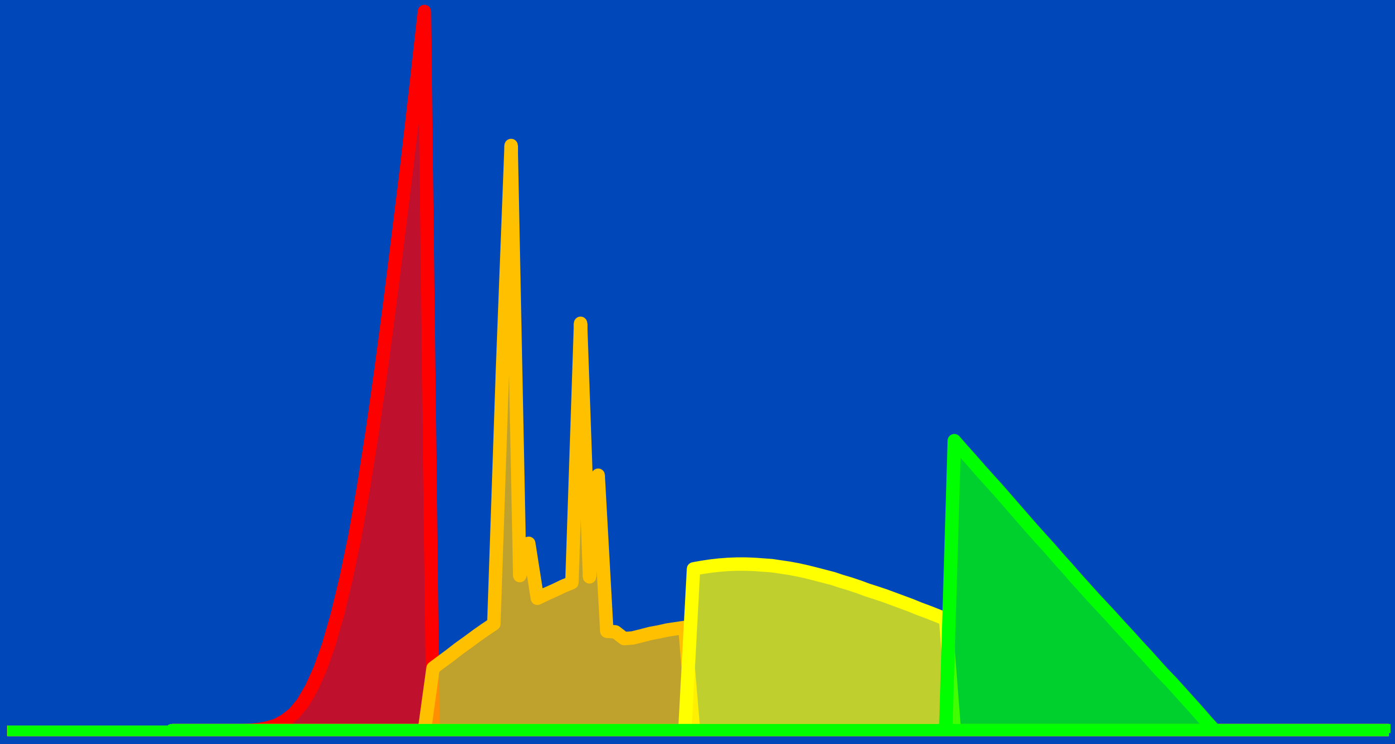


i.e. max O(40·10^6) cps

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

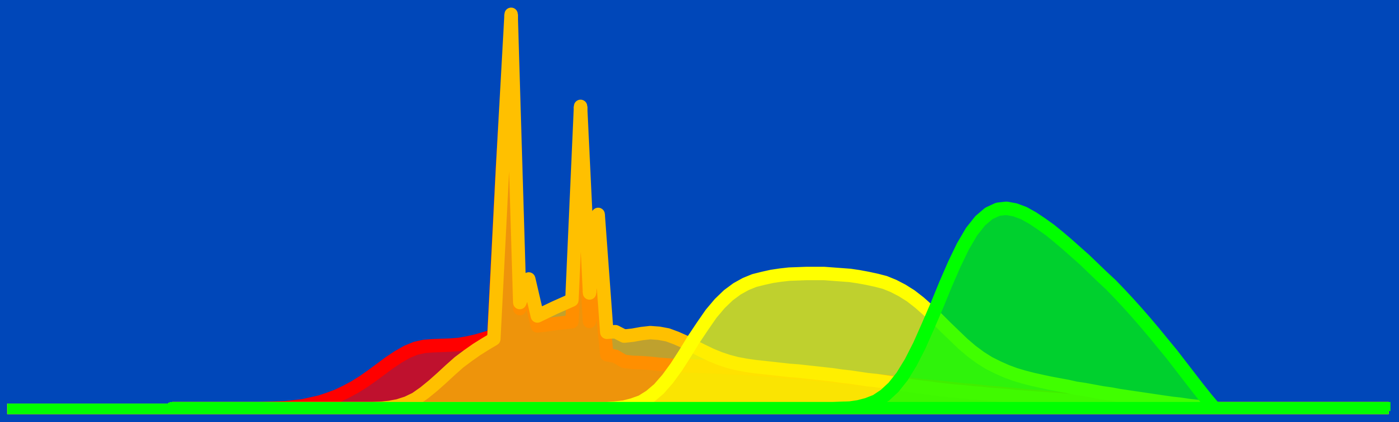
Energy-Selective Detectors: Improved Spectroscopy, Reduced Dose?

Ideally, bin spectra do not overlap, …



Energy-Selective Detectors: Improved Spectroscopy, Reduced Dose?

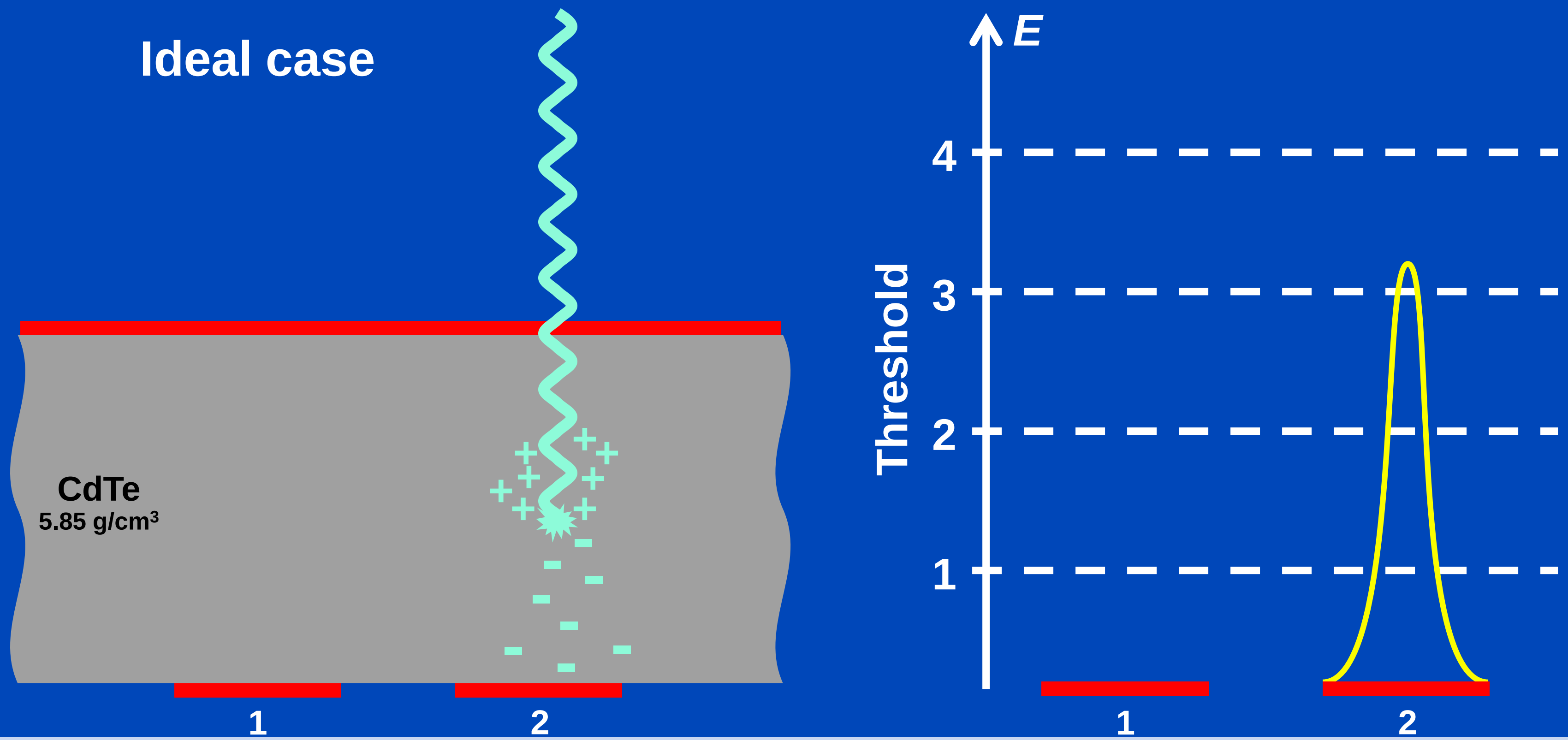
… realistically, however, they do!



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

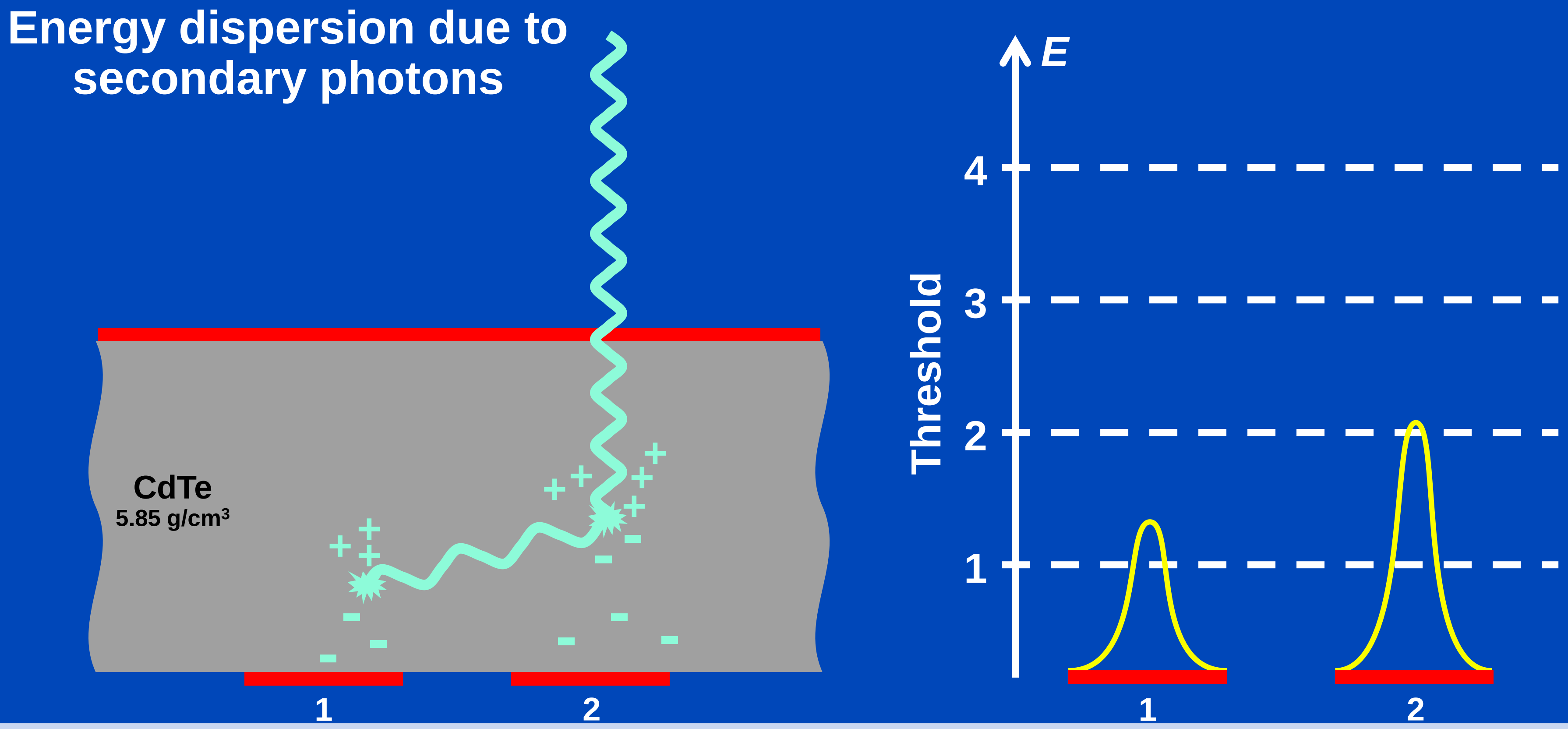
Photon Events

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



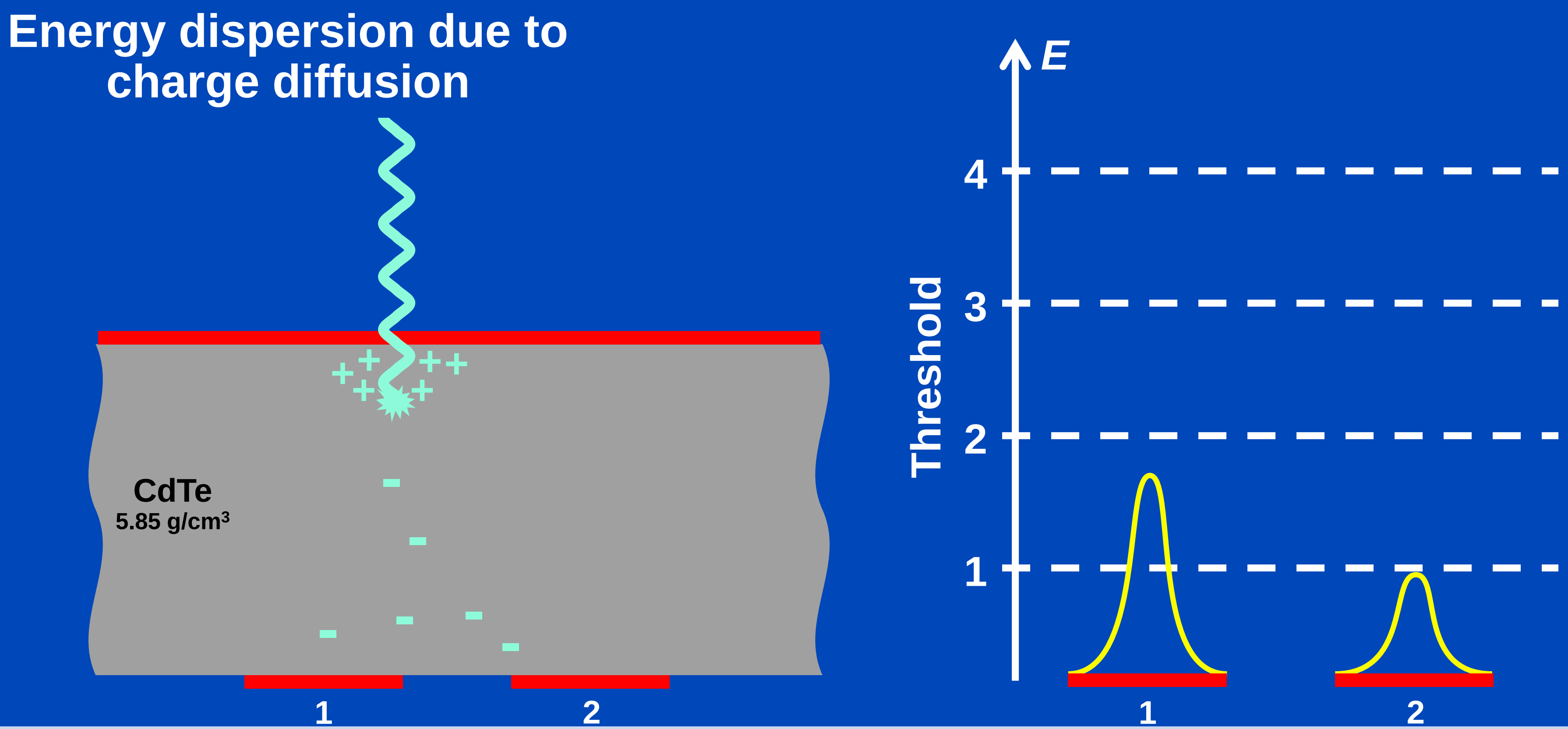
Photon Events

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



Photon Events

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



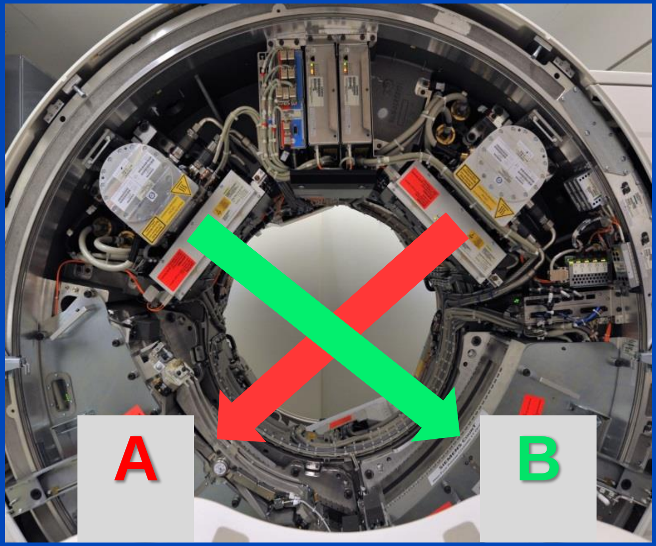

SIEMENS
SOMATOM CounT
@ dkfz.

Siemens CounT CT System

Gantry from a clinical dual source scanner

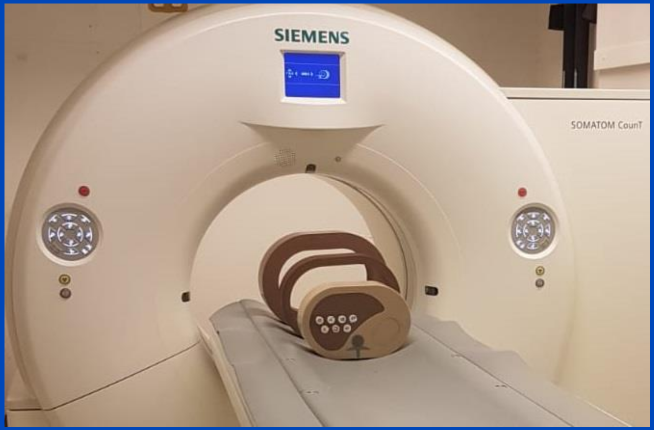
A: conventional CT detector (50.0 cm FOV)

B: Photon counting detector (27.5 cm FOV)



Readout Modes of the CounT

PC-UHR Mode	PC-Macro Mode	EI detector
0.25 mm pixel size	0.50 mm pixel size	0.60 mm pixel size



Readout Modes of the Siemens CounT

Macro Mode
0.9 × 1.1 mm focus
2 readouts
16 mm z-coverage

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

Chess Mode
0.9 × 1.1 mm focus
4 readouts
16 mm z-coverage

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

Sharp Mode
0.9 × 1.1 mm focus
5 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
0.7 × 0.7 mm focus
8 readouts
8 mm z-coverage

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

1.6 mm CdTe sensor. No FFS on detector B (photon counting detector). 4×4 subpixels of 225 µm size = 0.9 mm pixels (0.5 mm at isocenter). An additional 225 µm gap (e.g. for anti scatter grid) yields a pixel pitch of 1.125 mm.The whole detector consists of 128×1920 subpixels = 32×480 macro pixels.

This whole-body PCCT prototype, that had been installed at the Mayo Clinic, at the NIH and at the DKFZ, 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.

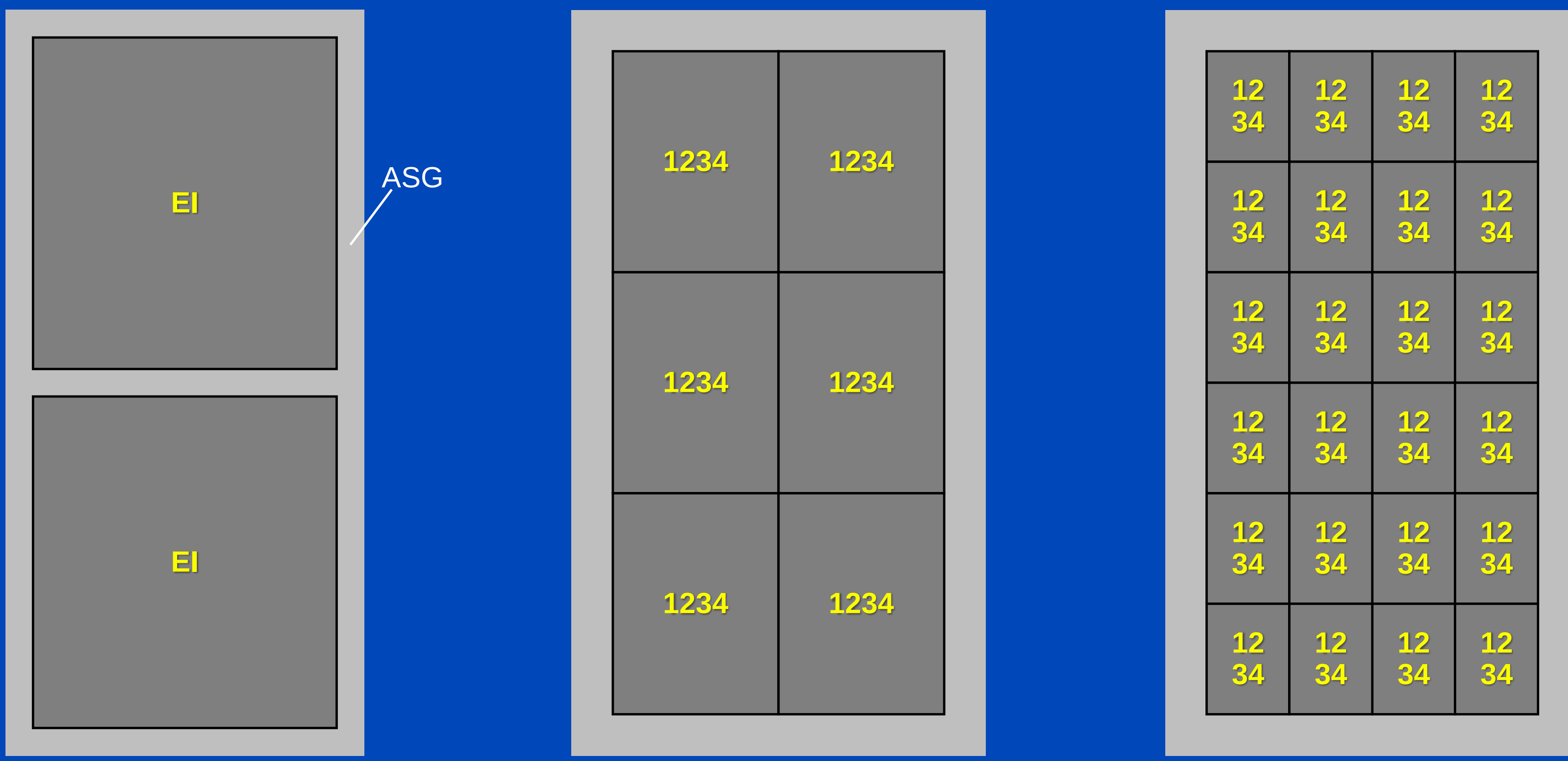
Siemens Naeotom Alpha
The World‘s First Photon-Counting CT

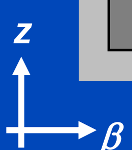
Detector Pixel Force vs. Alpha

Force
920 × 96 detector pixels
pixel size 0.52 × 0.56 mm at iso
avg. sampling 0.56 × 0.6 mm at iso
57.6 mm z-coverage

Alpha (Quantum Plus)
1376 × 144 macro pixels
pixel size 0.3 × 0.352 mm at iso
avg. sampling 0.344 × 0.4 mm at iso
57.6 mm z-coverage

Alpha (UHR)
2752 × 120 pixels
pixel size 0.15 × 0.176 mm at iso
avg. sampling 0.172 × 0.2 mm at iso
24 mm z-coverage





Focus sizes (Vectron): 0.181×0.226 mm, 0.271×0.7316 mm, 0.362×0.497 mm at iso
which are 0.4×0.5 mm, 0.6×0.7 mm, 0.8×1.1 mm at focal spot

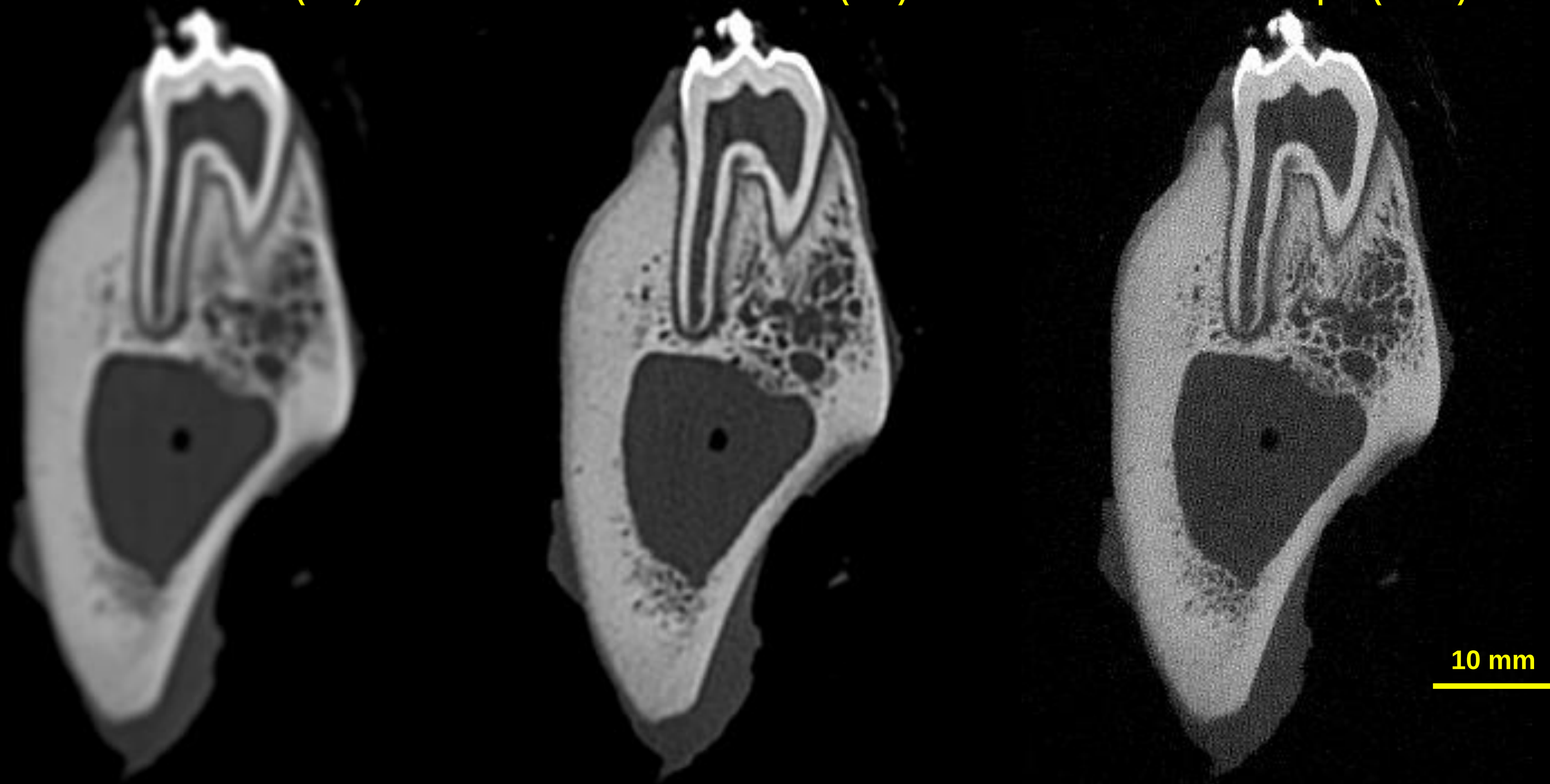
ASG information taken from [J. Ferda et al. Computed tomography with a full FOV photon-counting detector in a clinical setting, the first experience. European Journal of Radiology 137:109614, 2021]

Evolution of Spatial Resolution

similar to
2005: Somatom Flash (B70)

similar to
2014: Somatom CounT (U70)

scanned at
2021: Naeotom Alpha (Br98u)

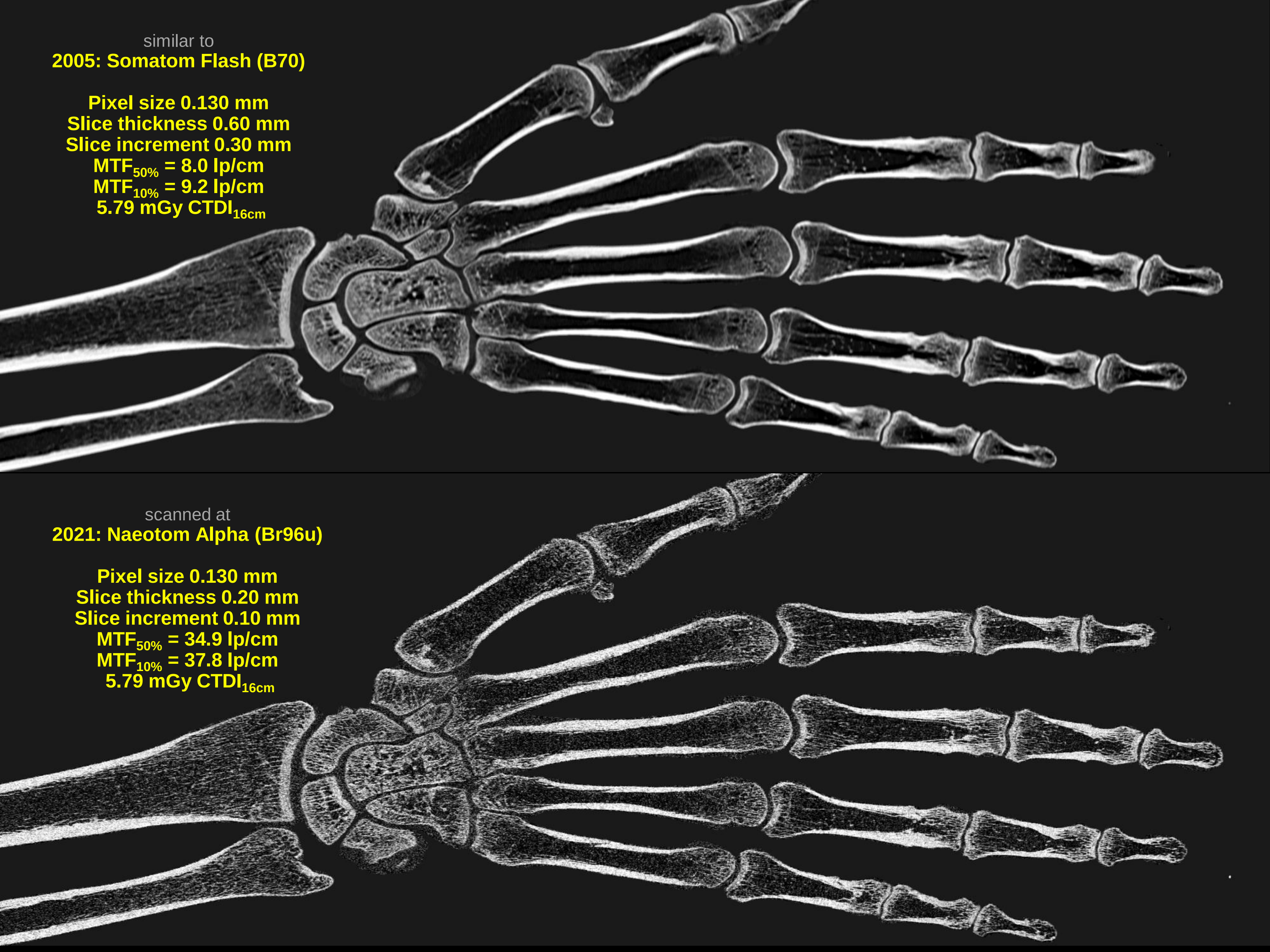


Pixel size 0.181 mm
Slice thickness 0.60 mm
Slice increment 0.30 mm
$MTF_{50\%}$ = 8.0 lp/cm
$MTF_{10\%}$ = 9.2 lp/cm

Pixel size 0.181 mm
Slice thickness 0.20 mm
Slice increment 0.10 mm
$MTF_{50\%}$ = 12.1 lp/cm
$MTF_{10\%}$ = 16.0 lp/cm

Pixel size 0.181 mm
Slice thickness 0.20 mm
Slice increment 0.10 mm
$MTF_{50\%}$ = 39.0 lp/cm
$MTF_{10\%}$ = 42.9 lp/cm

All measurements at Naeotom Alpha, Siemens Healthineers. QIR Reconstructions such that the maximum spatial resolution of Flash, CounT and Alpha is demonstrated on the same sample. *C* = 1200 HU, *W* = 4000 HU


similar to
2005: Somatom Flash (B70)
Pixel size 0.130 mm
Slice thickness 0.60 mm
Slice increment 0.30 mm
$MTF_{50\%}$ = 8.0 lp/cm
$MTF_{10\%}$ = 9.2 lp/cm
5.79 mGy $CTDI_{16cm}$
scanned at
2021: Naeotom Alpha (Br96u)
Pixel size 0.130 mm
Slice thickness 0.20 mm
Slice increment 0.10 mm
$MTF_{50\%}$ = 34.9 lp/cm
$MTF_{10\%}$ = 37.8 lp/cm
5.79 mGy $CTDI_{16cm}$

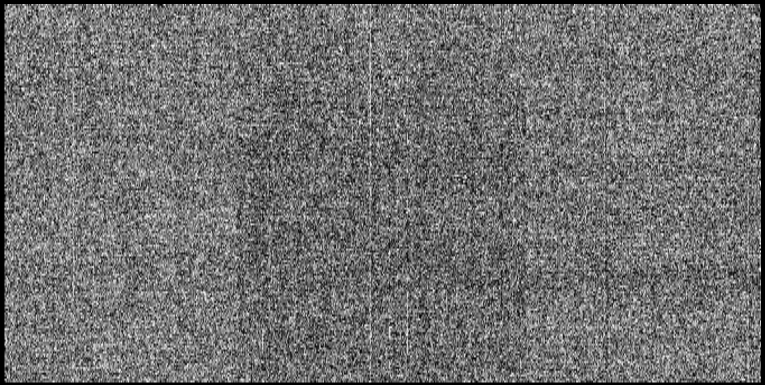
Advantages of Photon Counting CT

- No reflective gaps between detector pixels
  - Higher geometrical efficiency
  - Less dose
- No electronic noise
  - Less dose for infants
  - Less noise for obese patients
- Counting
  - Swank factor = 1 = maximal
  - “Iodine effect“ due to higher weights on low energies
- Energy bin weighting
  - Lower dose/noise
  - Improved iodine CNR
- Smaller pixels (to avoid pileup)
  - Higher spatial resolution
  - “Small pixel effect” i.e. lower dose/noise at conventional resolution
- Spectral information on demand
  - Dual Energy CT (DECT)
  - Multi Energy CT (MECT)

No 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**
  - **…**

EI (Dexela)



Readout noise only. Single events hidden!

PC (Dectris)

No readout noise. Single events visible!

18 frames, 5 min integration time per frame, x-ray off

Expected Value and Variance

- **Transmitted number of photons *N*:**

$$N(E) = N_0(E)e^{-p\psi(E)}$$

- **Poisson distribution:** $\mathrm{E}N(E) = \mathrm{Var}N(E)$
- **Detected signal *S* with sensitivity *s*(*E*):**

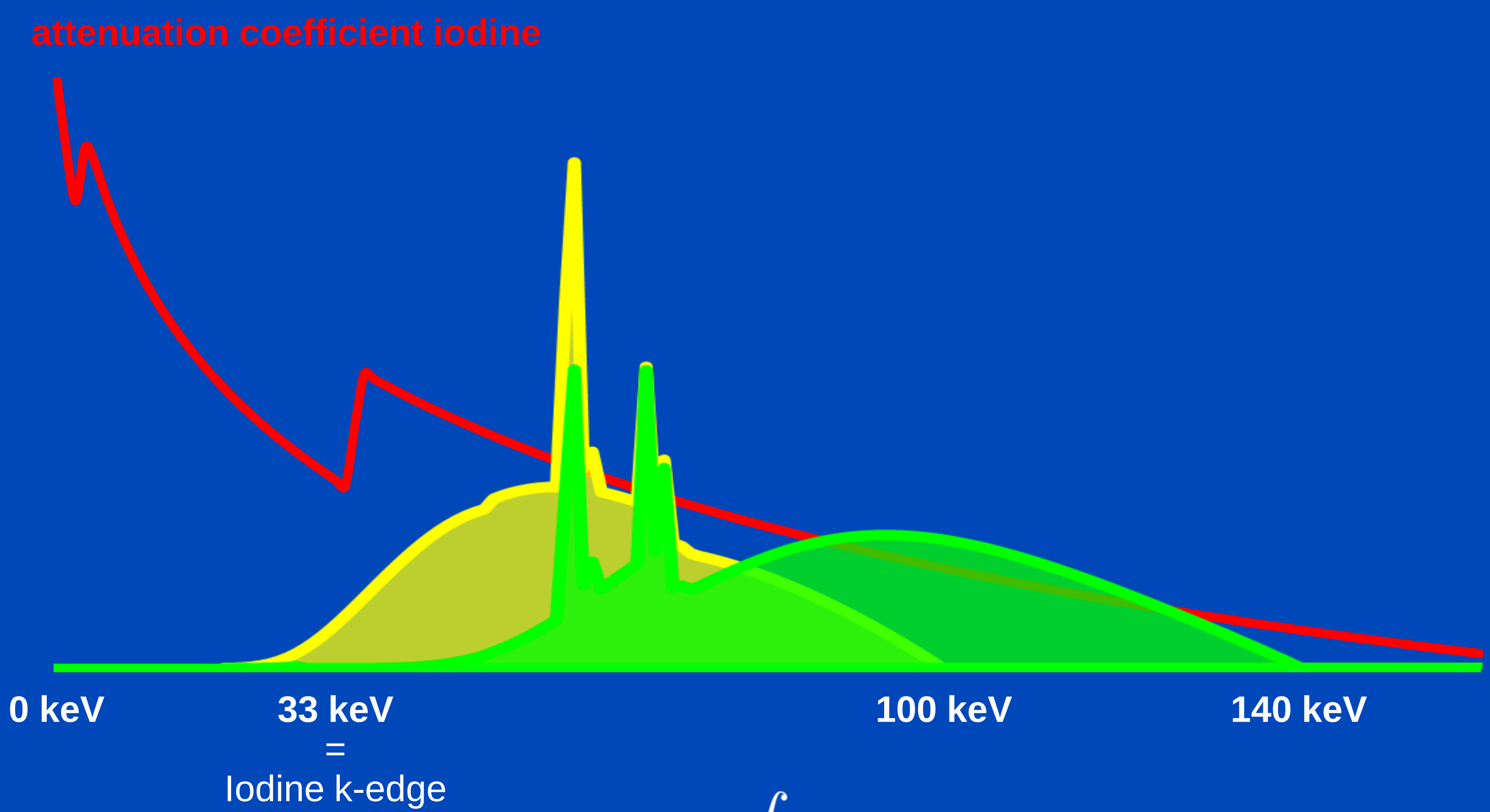
$$S = \int dE\, s(E)N(E)$$

- **Expected value and variance of the signal *S*:**

$$\mathrm{E}S = \int dE\, s(E)\mathrm{E}N(E) \text{ and } \mathrm{Var}S = \int dE\, s^2(E)\mathrm{E}N(E)$$

- **Detector sensitivity: PC** $s(E) = 1$**, but EI** $s(E) \propto E$ **!**

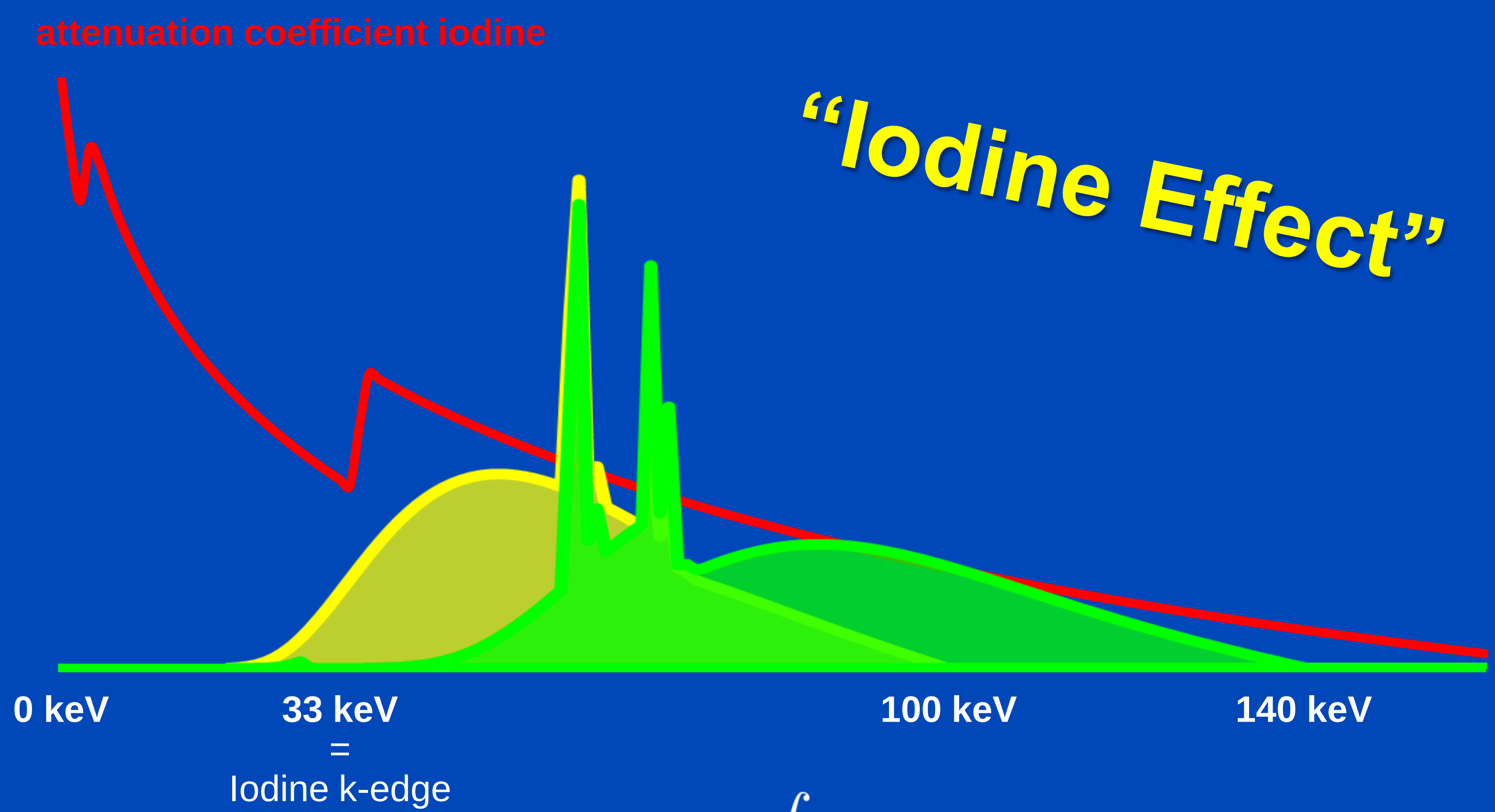
Energy Integrating
(Detected Spectra at 100 kV and 140 kV)



$$\text{Signal}_{\text{EI}} = \int dE\, E\ \ N(E)$$

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

Photon Counting
(Detected Spectra at 100 kV and 140 kV)

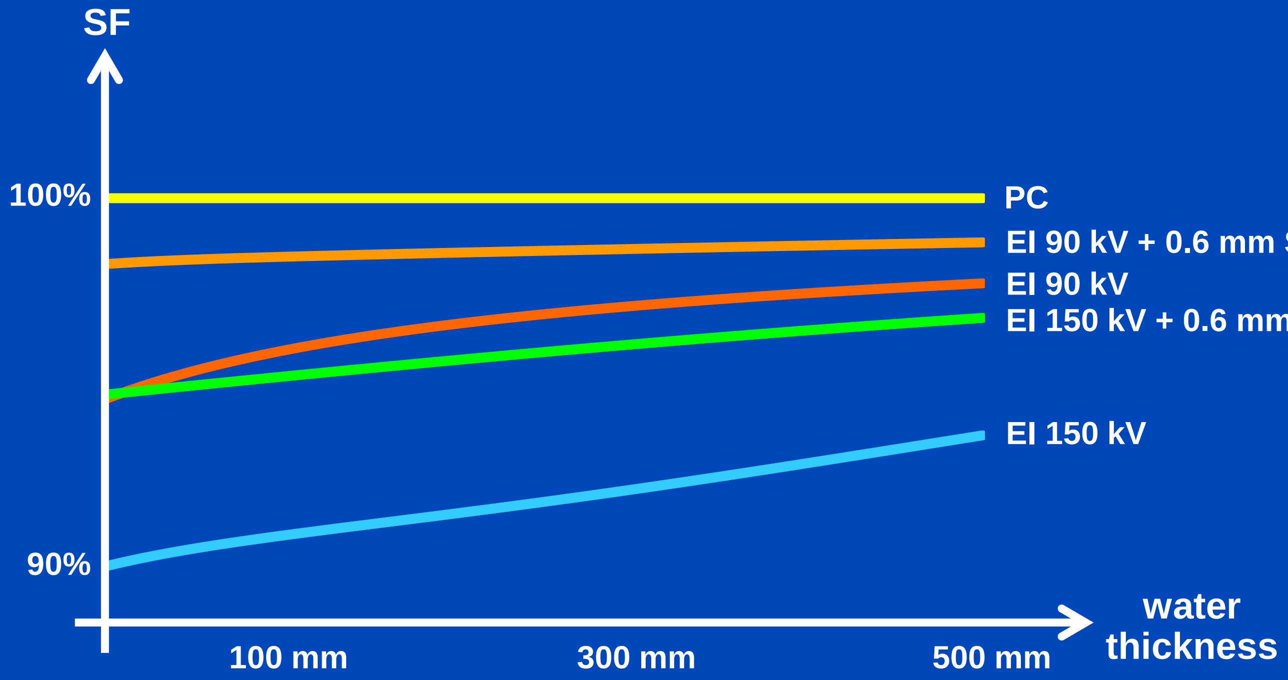


$$\mathrm{Signal}_{\mathrm{PC}} = \int dE\, 1\;\; N(E)$$

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

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 highest SNR.



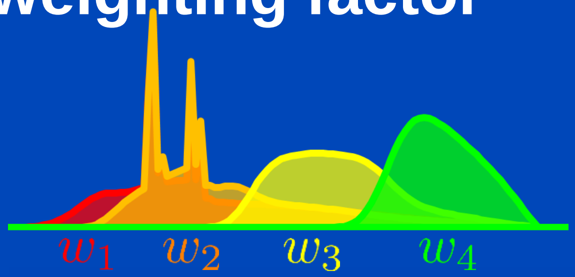
$$\mathrm{SF} = \frac{\mathrm{SNR}_{\mathrm{EI}}^2}{\mathrm{SNR}_{\mathrm{PC}}^2} = \frac{\left(\int dE\, E N(E)\right)^2}{\left(\int dE\, N(E)\right)\left(\int dE\, E^2 N(E)\right)} \leq 1$$

due to Schwarz' inequality

Photon Counting used to Maximize CNR

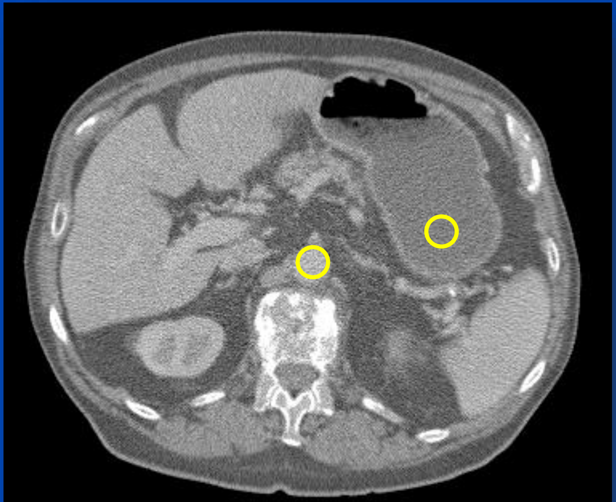
- With PC, energy bin sinograms can be weighted individually, i.e. by a weighted summation
- To optimize the CNR the optimal bin weighting factor w_b is given by (weighting after log):

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



- The resulting CNR is

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


The two ROIs are used to measure the CNR.

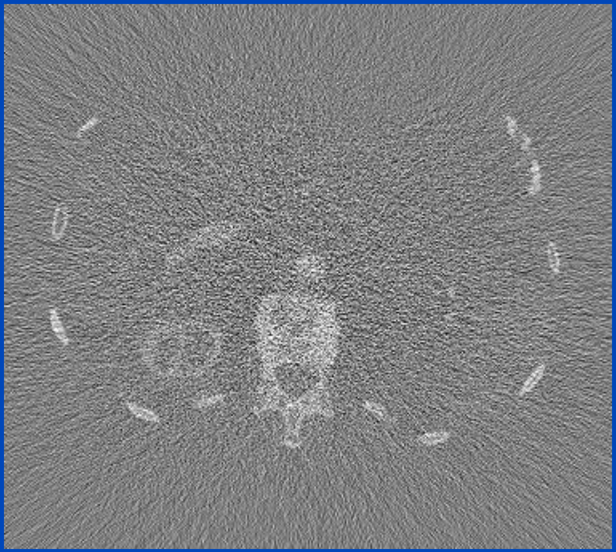
- At the optimum this evaluates to

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

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

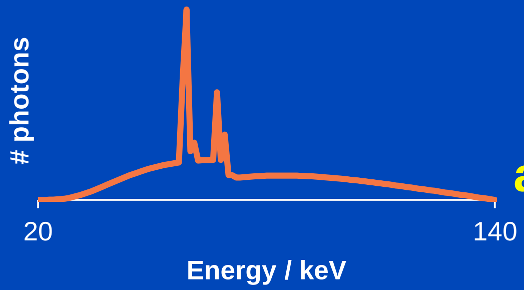
Energy Integrating

PC minus EI

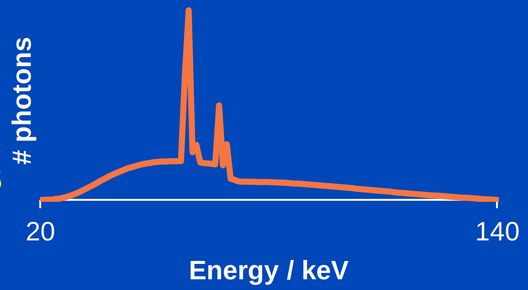


Photon Counting

CNR = 2.11



CNR = 2.95



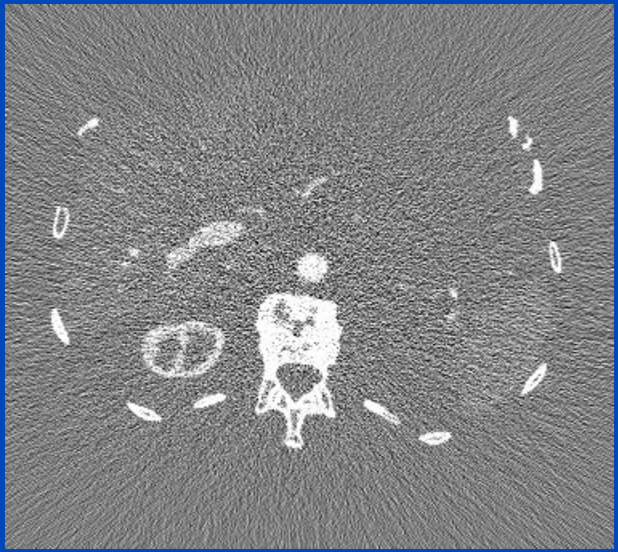
40% CNR improvement or 49% dose reduction achievable due to improved Swank factor and more weight on low energies (iodine contrast benefits).

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

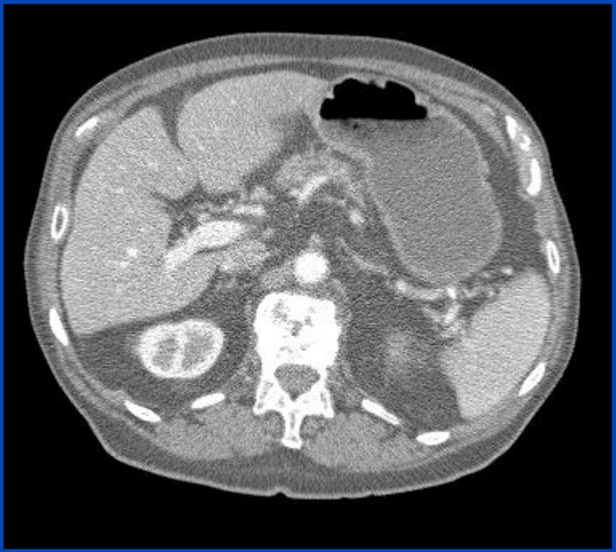
Energy Integrating vs. Photon Counting
with 4 bins from 20 to 140 keV

Energy Integrating

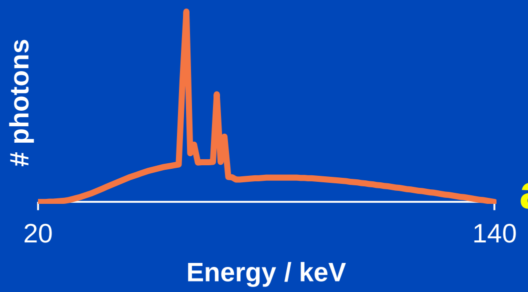
PC minus EI



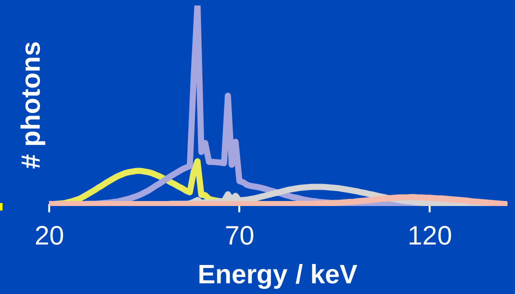
Photon Counting



CNR = 2.11



CNR = 4.19



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

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

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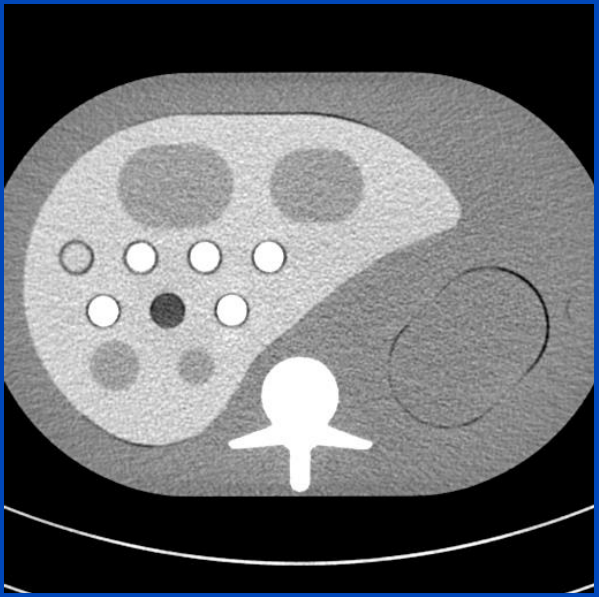
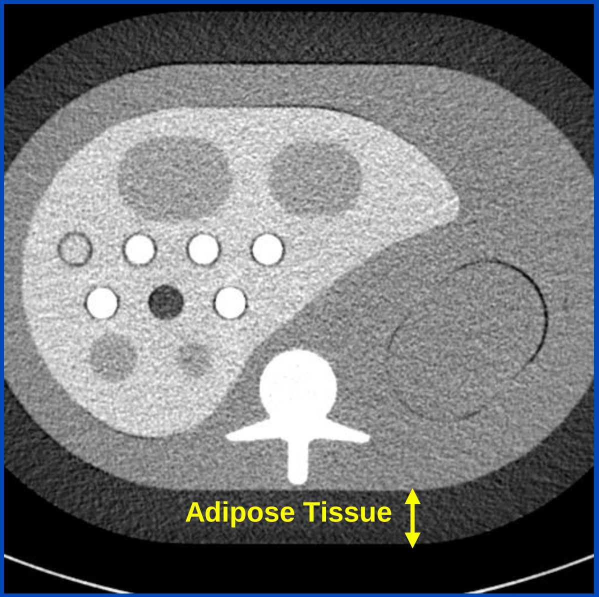
Iodine CNRD Assessment

Reconstruction Examples @ 80 kV

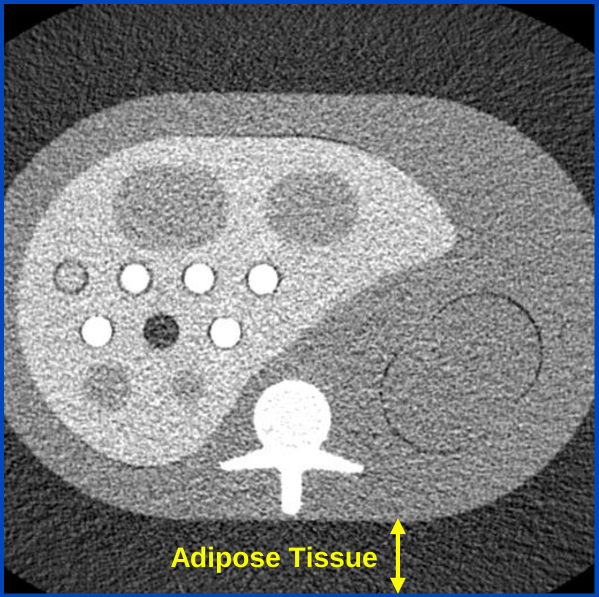
Small (200 × 300 mm)



Medium (250 × 350 mm)



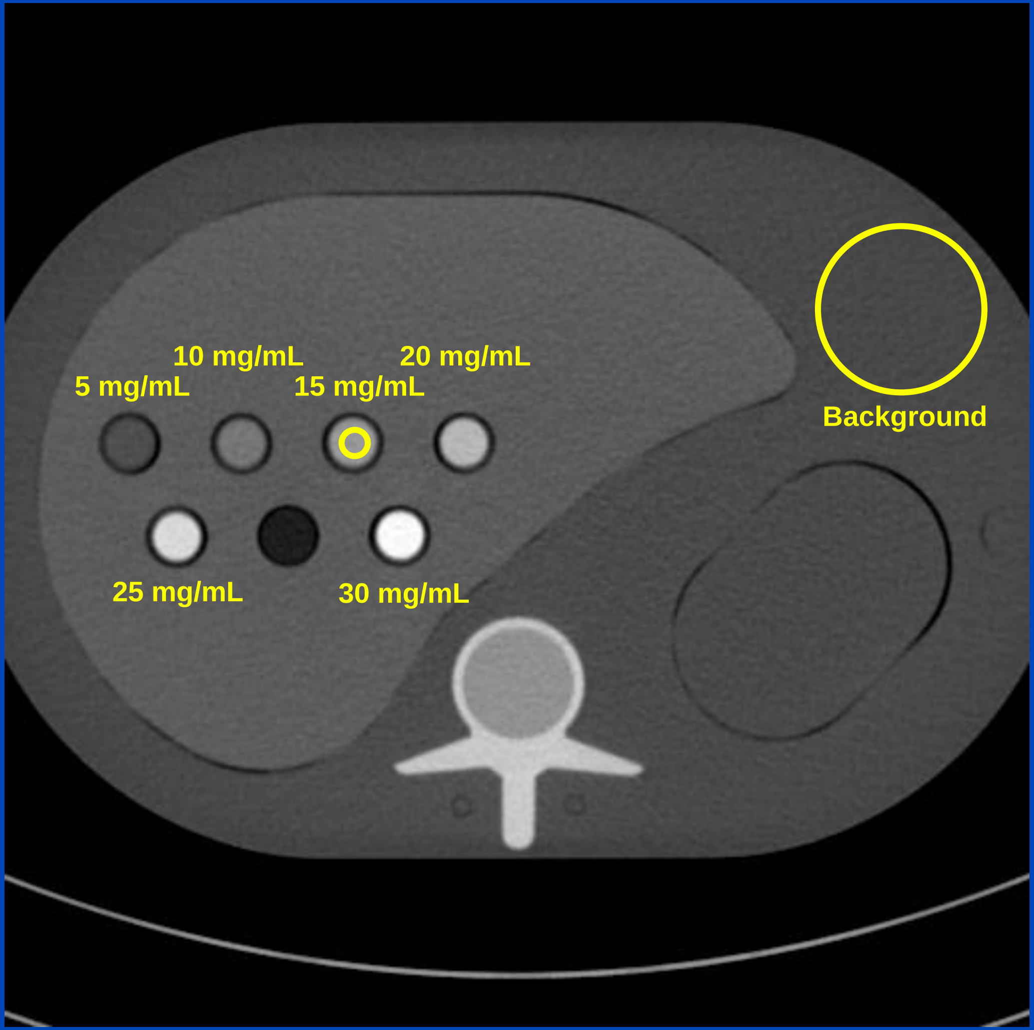
Large (300 × 400 mm)



C/W=0 HU/400HU

Iodine CNRD Assessment

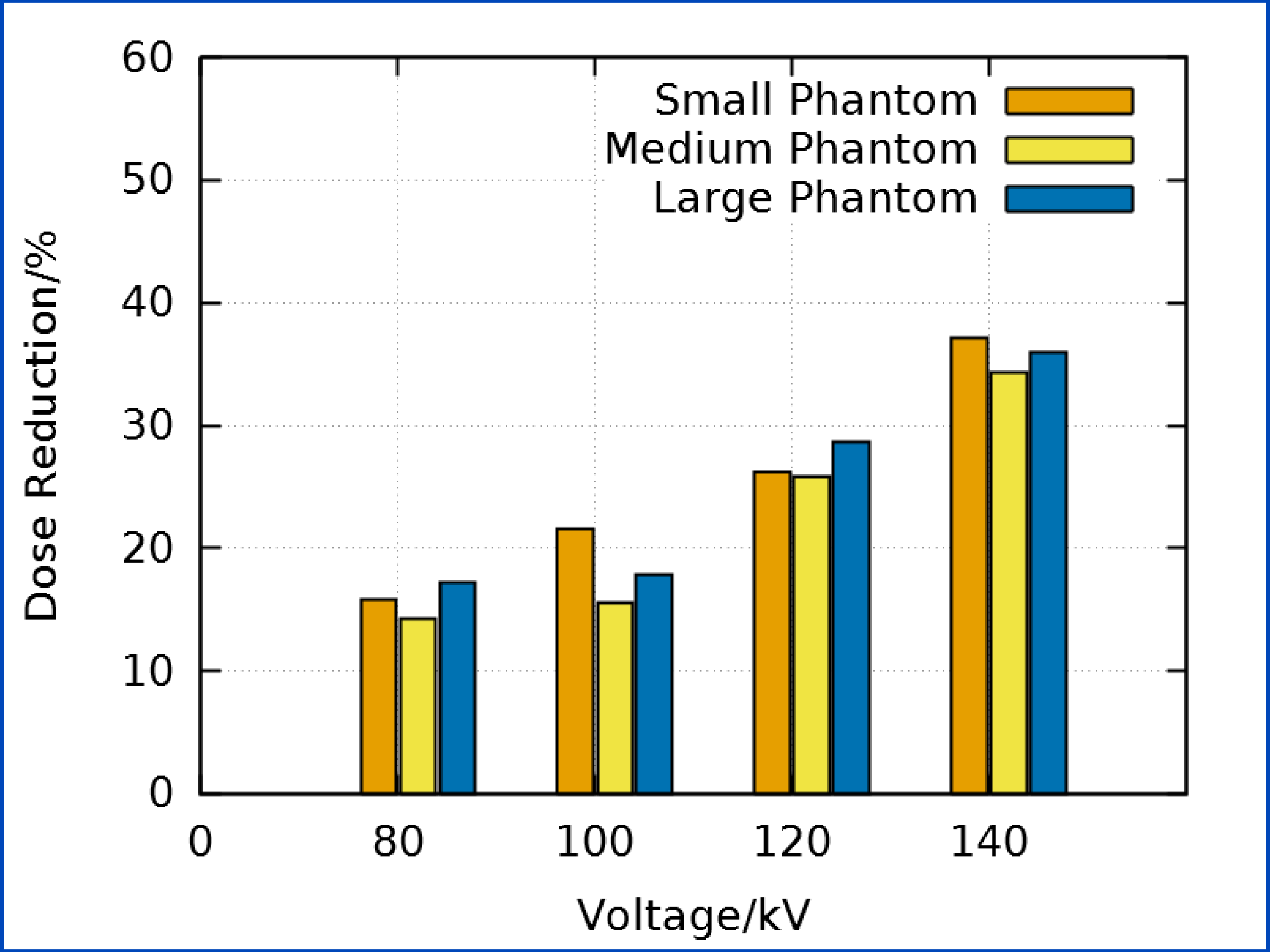
Regions of Interest



C/W=180 HU/600HU

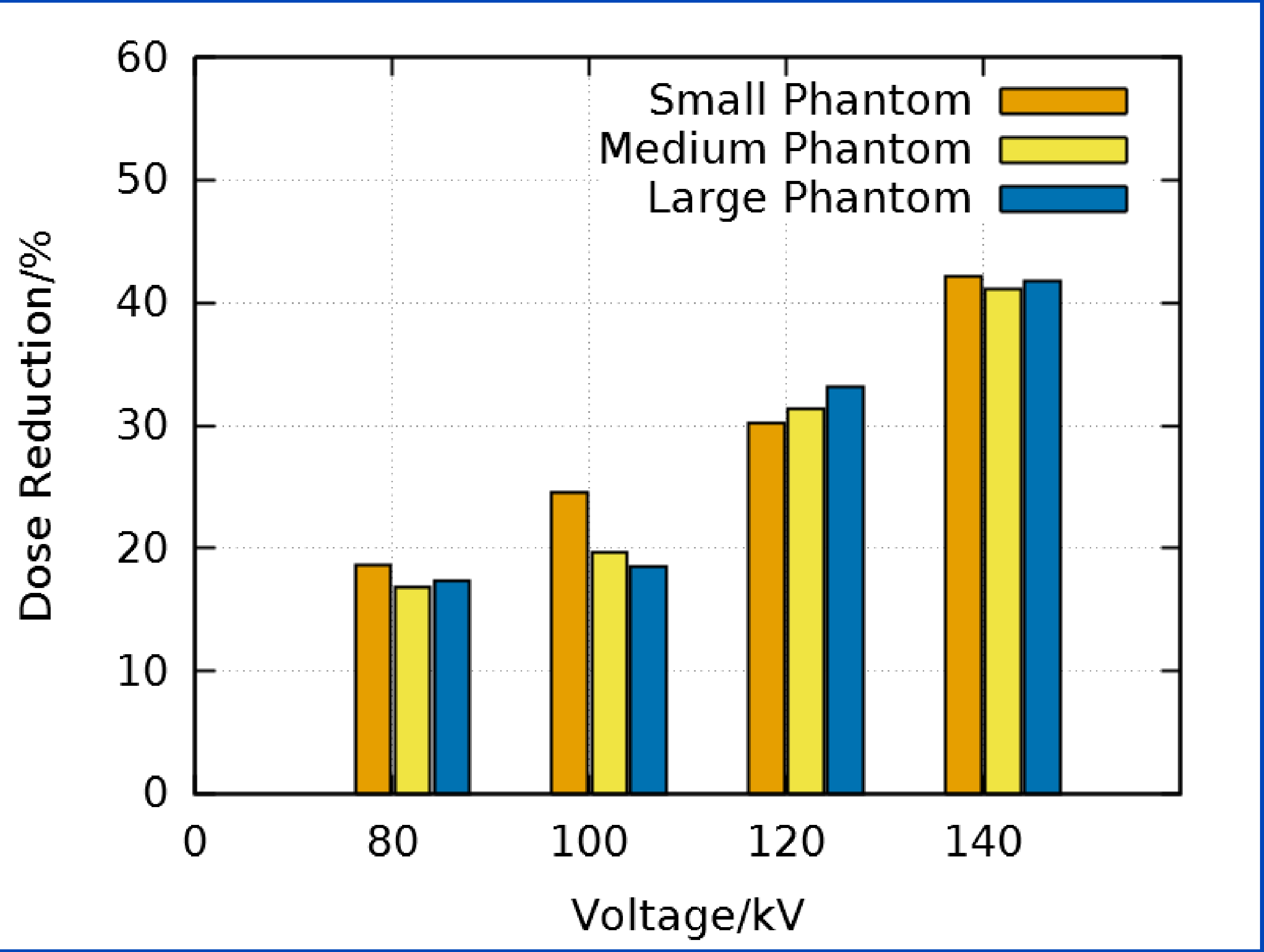
PC with 1 Bin vs. EI

Potential Dose Reduction



PC with 2 Bins vs. EI

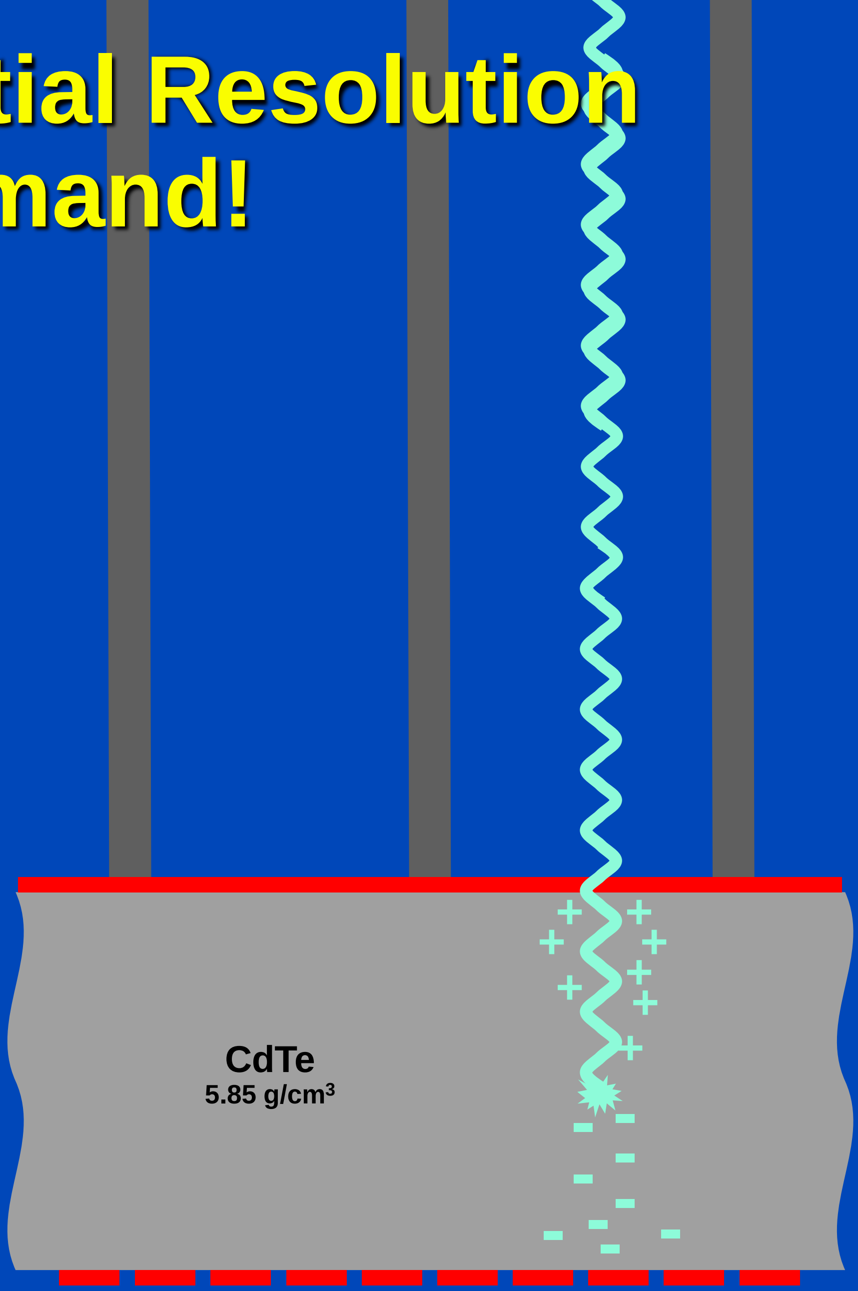
Potential Dose Reduction



dkfz.

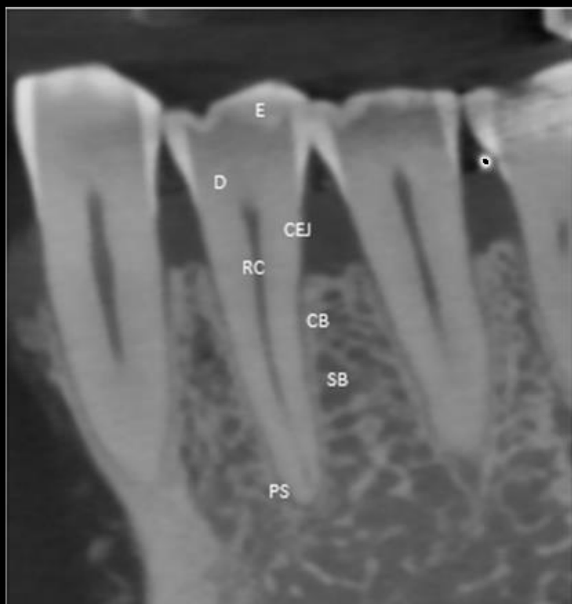
Ultra-High Spatial Resolution on Demand!

- 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.



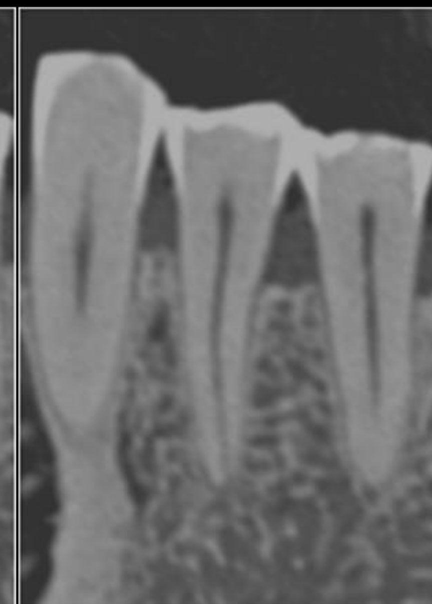
Dental Imaging

SOMATOM CounT



DVT
8 mGy, 102 kV

PCCT
8.5 mGy, 120 kV



PCCT
38 mGy, 120 kV

E: enamel, CEJ: cemento-enamel-junction, RC: root canal, CB: cortical bone, SB: spongious bone, PS: peridontal space

DVT: Veraview X800, Morita, Japan, PCCT: Somatom CounT, Siemens, Germany

Dose values are 16 cm CTDI values.

Slice positions between DVT and PCCT do not match exactly.

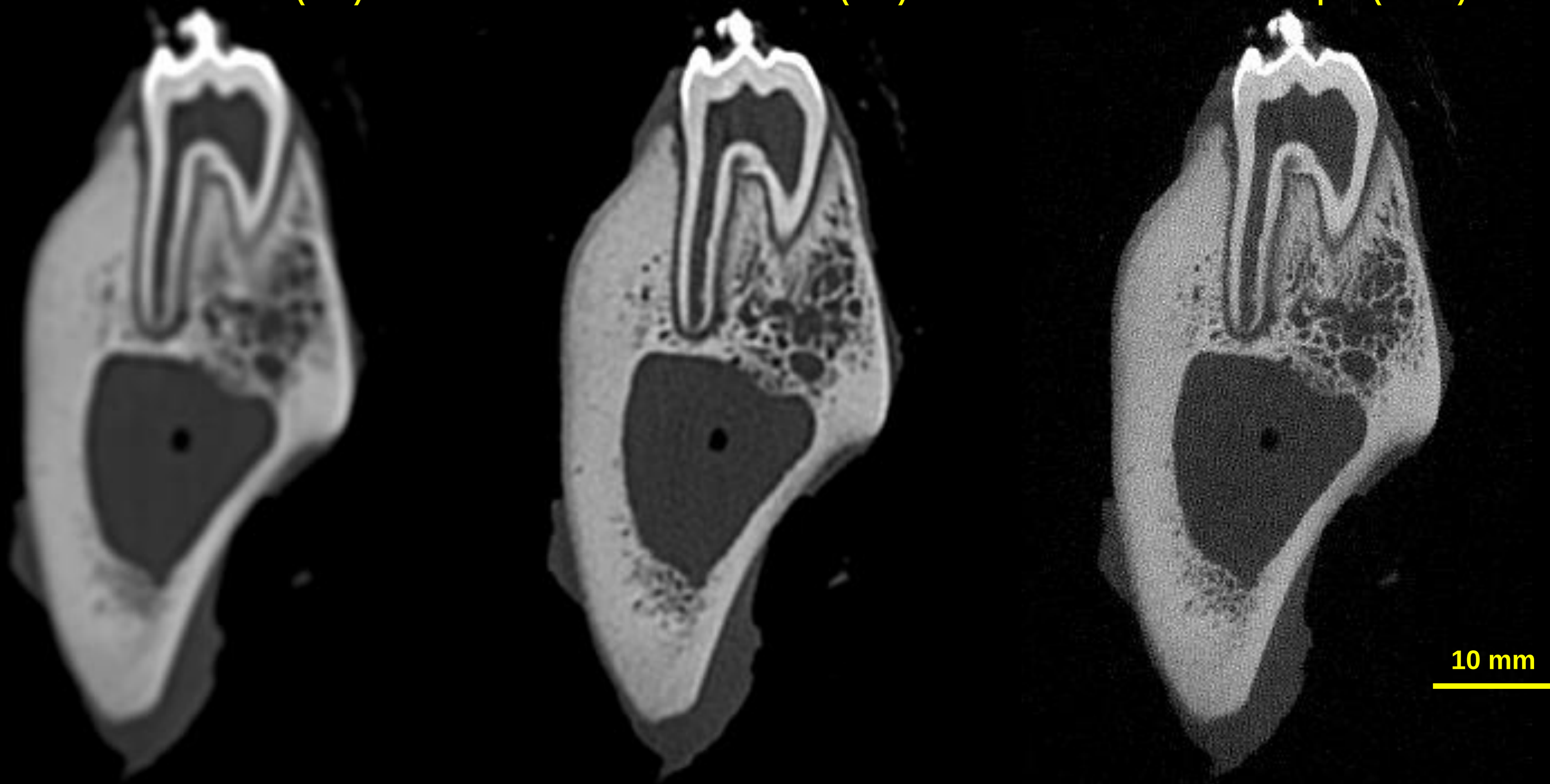
Collaboration with Dr. Maurice Rütters, Section of Periodontology, Department of Operative Dentistry, University Hospital Heidelberg.

Evolution of Spatial Resolution

similar to
2005: Somatom Flash (B70)

similar to
2014: Somatom CounT (U70)

scanned at
2021: Naeotom Alpha (Br98u)



Pixel size 0.181 mm
Slice thickness 0.60 mm
Slice increment 0.30 mm
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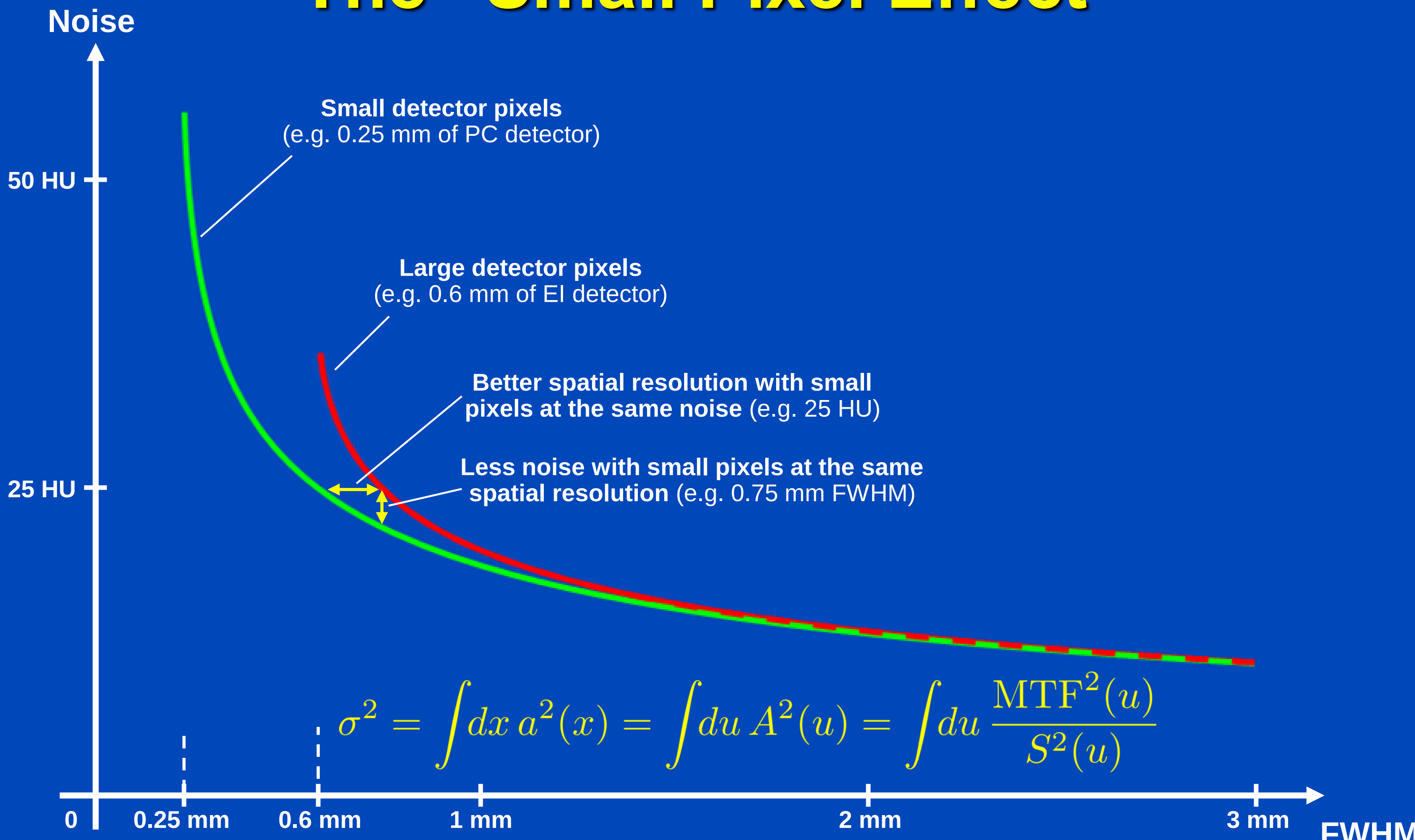
All measurements at Naeotom Alpha, Siemens Healthineers. QIR reconstructions such that the maximum spatial resolution of Flash, CounT and Alpha is demonstrated on the same sample. *C* = 1200 HU, *W* = 4000 HU

The “Small Pixel Effect”



$$\sigma^2 = \int dx\, a^2(x) = \int du\, A^2(u) = \int du\, \frac{\mathrm{MTF}^2(u)}{S^2(u)}$$

To Bin or not to Bin?
(the continuous view)

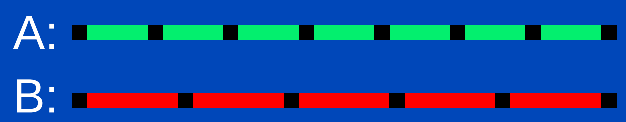
This nice phrase was coined by Norbert Pelc.

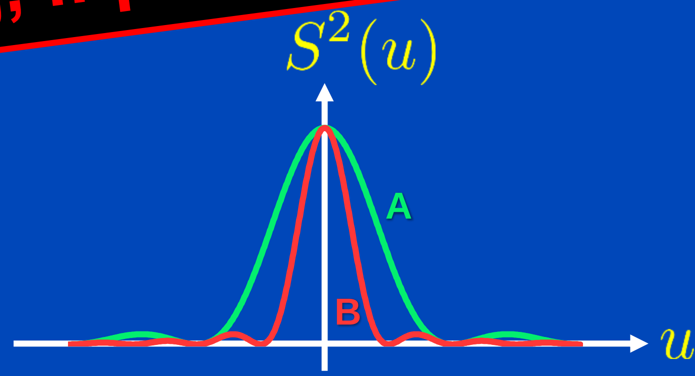
- We have $\mathrm{PSF}(x) = s(x) * a(x)$ and $\mathrm{MTF}(u) = S(u)A(u)$.
- From Rayleigh's theorem we find noise is

$$\sigma^2 = \int dx\, a^2(x) = \int du\, A^2(u) = \int du\, \frac{\mathrm{MTF}^2(u)}{S^2(u)}$$

- Compare Small (A) with la pixels:

Avoid binning, if possible!





- We have $S_{\mathrm{A}}(u) > S_{\mathrm{B}}(u)$ and thus $\sigma_{\mathrm{A}}^2 < \sigma_{\mathrm{B}}^2$.
- This means that a desired PSF/MTF is often best achieved with smaller detectors.

Kachelrieß, Kalender. Med. Phys. 32(5):1321-1334, May 2005
Baek, Pineda, and Pelc. PMB 58:1433-1446, 2013

To Bin or not to Bin?

(the discrete view, NN)

- Let detector B be the 2-binned version of detector A:

$$B_{2n} = \tfrac{1}{2}(A_{2n} + A_{2n+1}) \qquad \mathrm{Var}B = \tfrac{1}{2}\mathrm{Var}A$$

- Let us now do an upsampling of the detector B such that each of B´s pixels becomes two pixels with the same value and with th

30% more noise variance may be compensated by 30% more x-ray dose. Any alternative? Yes:

Avoid binning, if possible!

In 2D binning implies 69% more noise variance or dose. Again, the answer is: „not to bin“.

- To ol etector we n

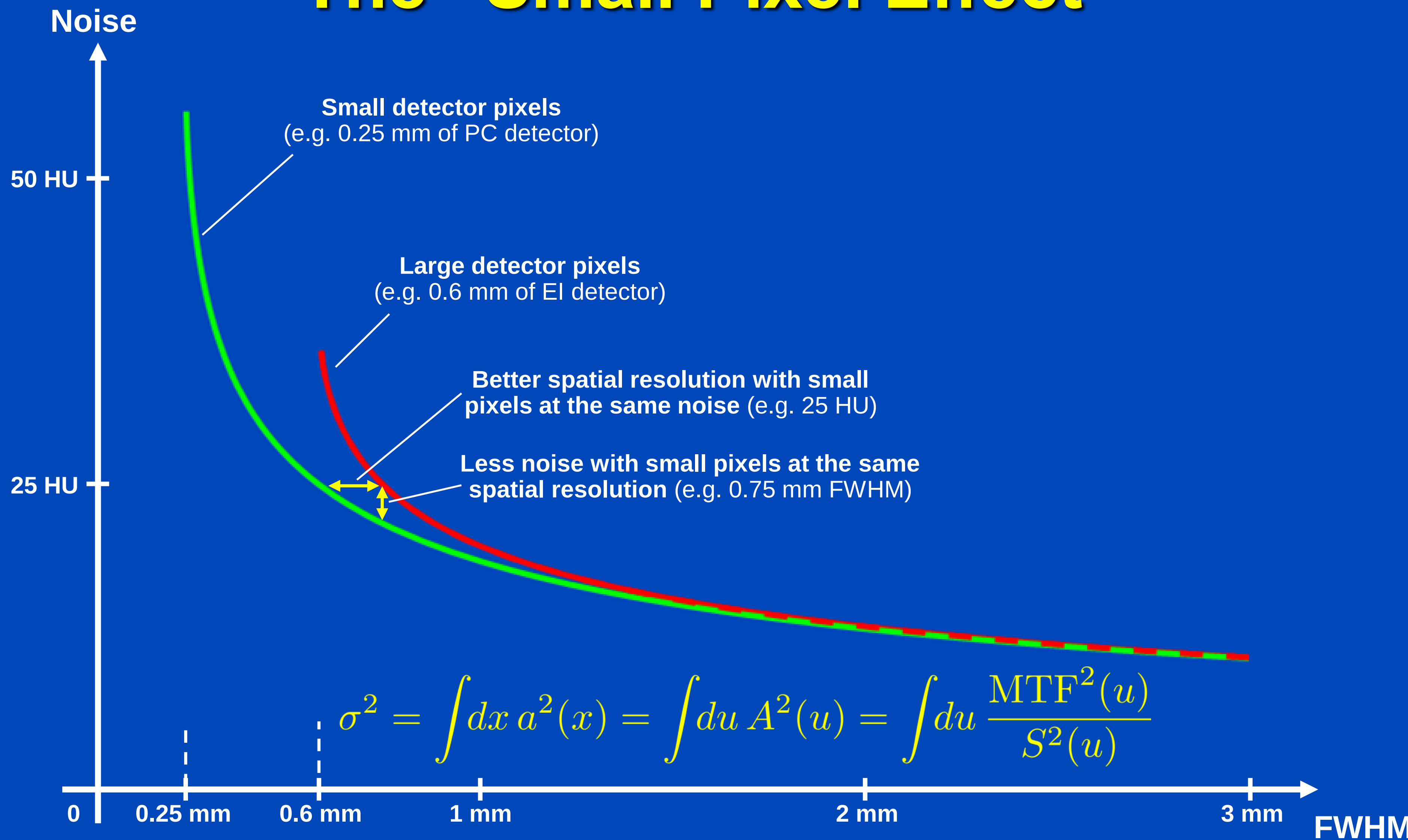
$$\boldsymbol{a} = \tfrac{1}{4}\left(1,\ 2,\ 1\right)$$

- Noise propagation yields 20% more noise (variance) for the binned detector:

$$\mathrm{Var}\hat{A} = \tfrac{6}{16}\mathrm{Var}A = \tfrac{3}{8}\mathrm{Var}A$$

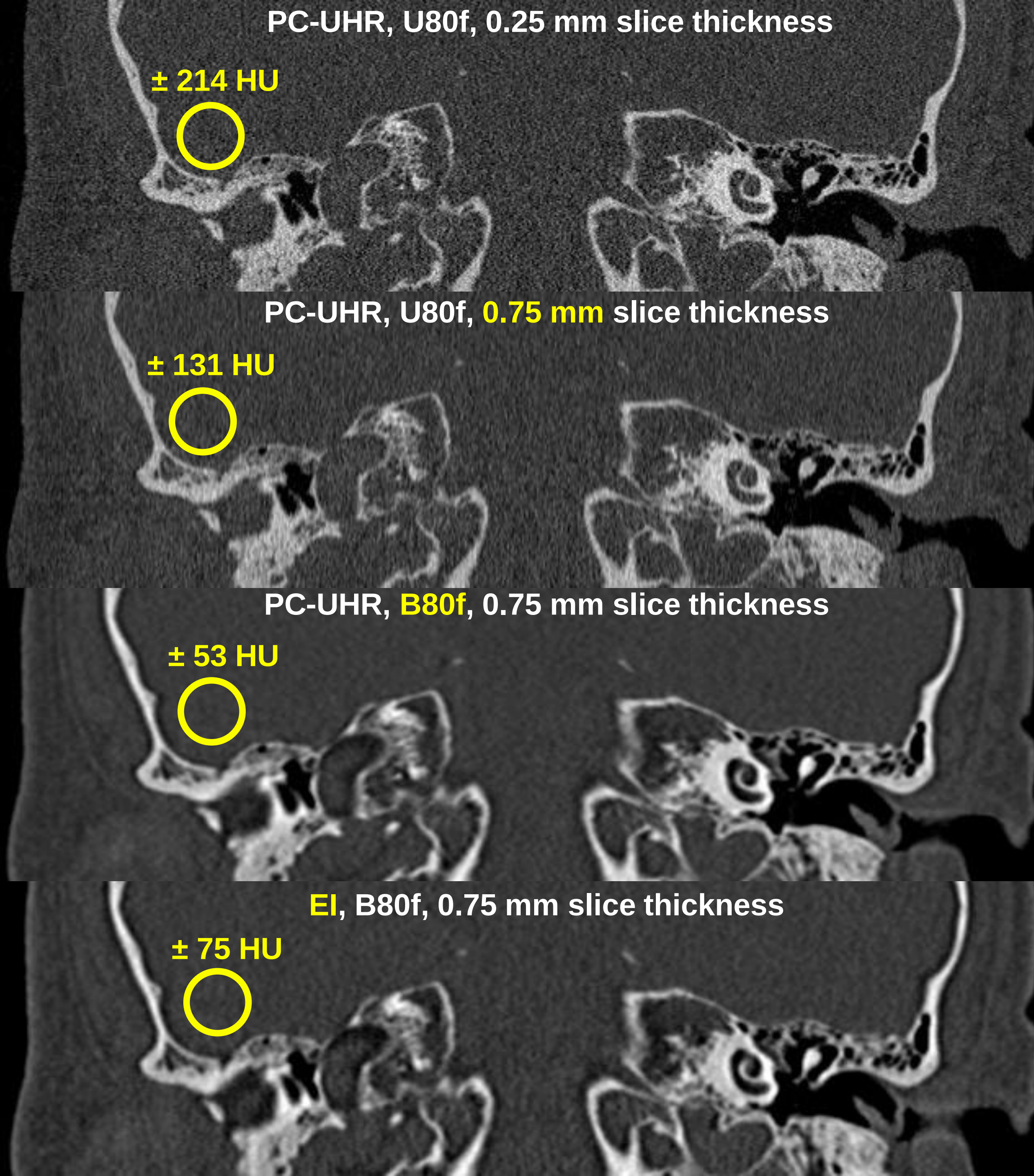
$$\mathrm{Var}\hat{B} = \tfrac{1}{2}\mathrm{Var}A = \tfrac{4}{3}\mathrm{Var}\hat{A} = 1.3\,\mathrm{Var}\hat{A}$$

The "Small Pixel Effect"



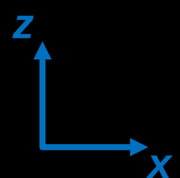
$$\sigma^2 = \int dx\, a^2(x) = \int du\, A^2(u) = \int du\, \frac{\mathrm{MTF}^2(u)}{S^2(u)}$$

All images reconstructed with 1024^2 matrix and 0.15 mm slice increment.
C = 1000 HU
W = 3500 HU



10% MTF: 19.1 lp/cm
10% MTF:17.2 lp/cm
xy FWHM: 0.48 mm
z FWHM: 0.40 mm
$CTDI_{vol}$: 16.0 mGy

10% MTF: 19.1 lp/cm
10% MTF:17.2 lp/cm
xy FWHM: 0.48 mm
z FWHM: 0.67 mm
$CTDI_{vol}$: 16.0 mGy

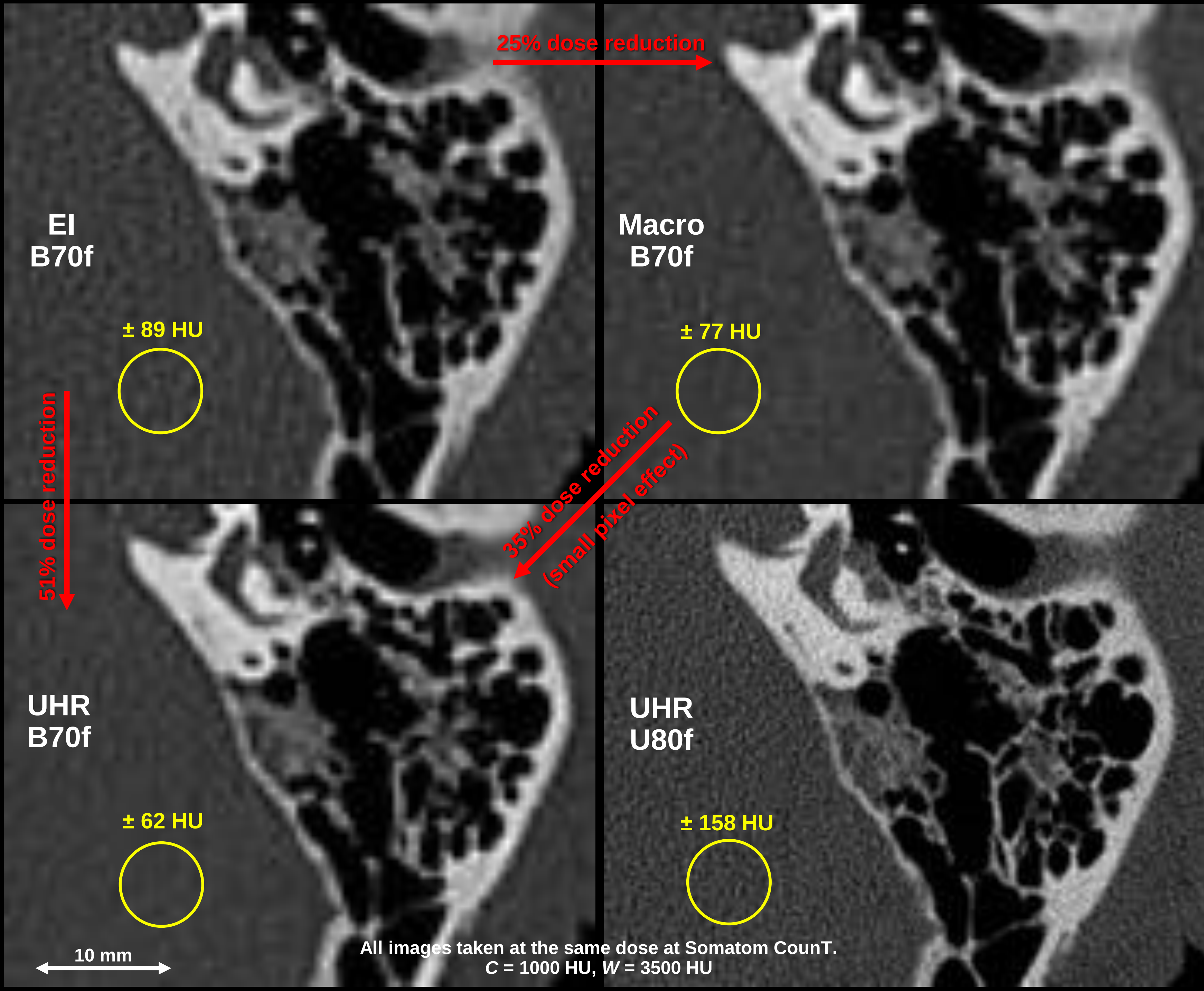


10% MTF: 9.3 lp/cm
10% MTF:10.5 lp/cm
xy FWHM: 0.71 mm
z FWHM: 0.67 mm
$CTDI_{vol}$: 16.0 mGy

10% MTF: 9.3 lp/cm
10% MTF:10.5 lp/cm
xy FWHM: 0.71 mm
z FWHM: 0.67 mm
$CTDI_{vol}$: 16.0 mGy

Data courtesy of the Institute of Forensic Medicine of the University of Heidelberg and of the Division of Radiology of the German Cancer Research Center (DKFZ)

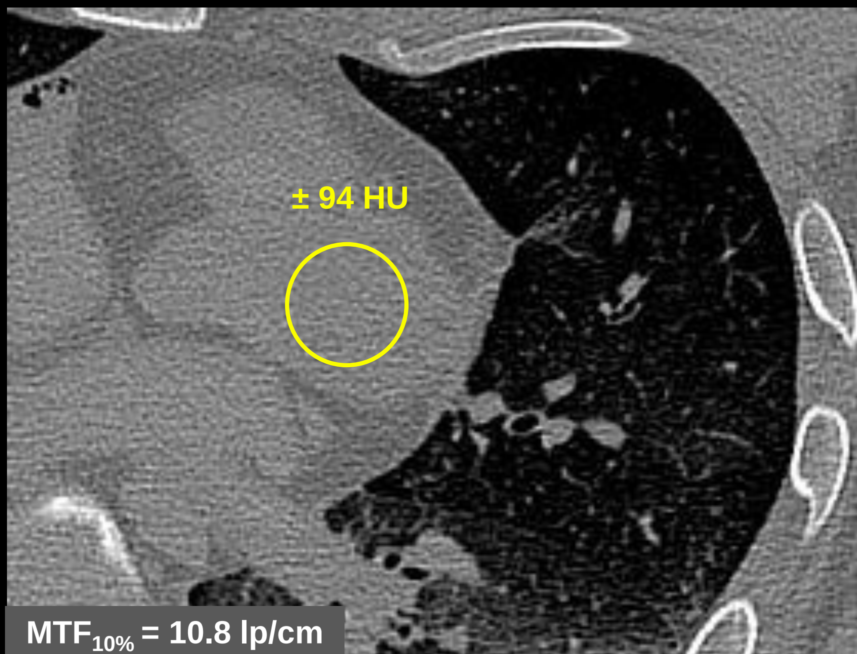
dkfz.



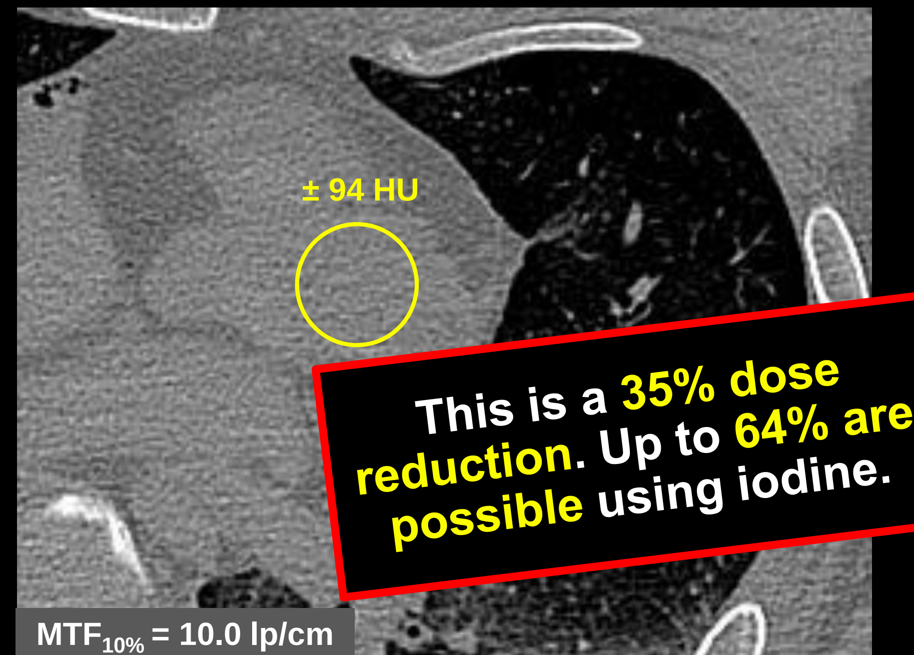
25% dose reduction
EI
B70f
± 89 HU
Macro
B70f
± 77 HU
51% dose reduction
35% dose reduction
(small pixel effect)
UHR
B70f
± 62 HU
UHR
U80f
± 158 HU
10 mm
All images taken at the same dose at Somatom CounT.
C = 1000 HU, W = 3500 HU

L. Klein, C. Amato, S. Heinze, M. Uhrig, H.-P. Schlemmer, M. Kachelrieß, and S. Sawall. **Effects of Detector Sampling on Noise Reduction in a Clinical Photon Counting Whole-Body CT**. Investigative Radiology, vol. 55(2):111-119, February 2020.

Energy Integrating Detector (B70f)



Photon Counting Detector (B70f)



Acquisition with EI:
- **Tube voltage of 120 kV**
- **Tube current of 300 mAs**
- **Resulting dose of $CTDI_{vol\ 32\ cm}$ = 22.6 mGy**

Acquisition with UHR:
- **Tube voltage of 120 kV**
- **Tube current of 180 mAs**
- **Resulting dose of $CTDI_{vol\ 32\ cm}$ = 14.6 mGy**

***C* = 50 HU, *W* = 1500 HU**

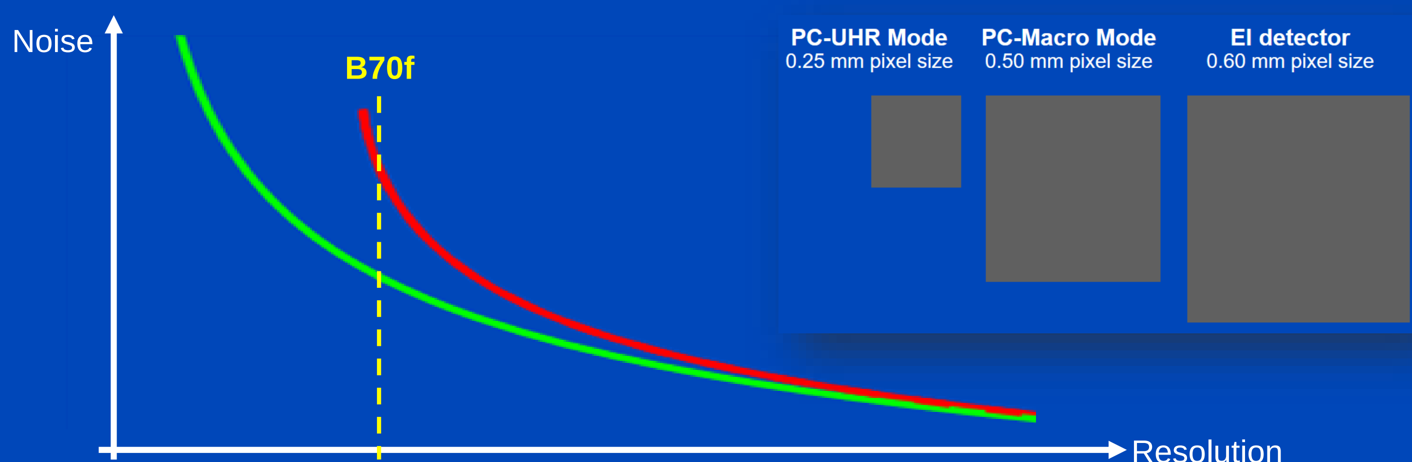
X-Ray Dose Reduction of B70f

PC vs. PC
("small pixel effect only")

UHR vs. Macro	80 kV	100 kV	120 kV	140 kV
S	**23%** ± 12%	**34%** ± 10%	**35%** ± 11%	**25%** ± 10%
M	**32%** ± 10%	**32%** ± 8%	**35%** ± 8%	**34%** ± 9%
L	**35%** ± 10%	**29%** ± 15%	**27%** ± 9%	**31%** ± 11%

PC vs. EI
("small pixel effect" and "iodine effect")

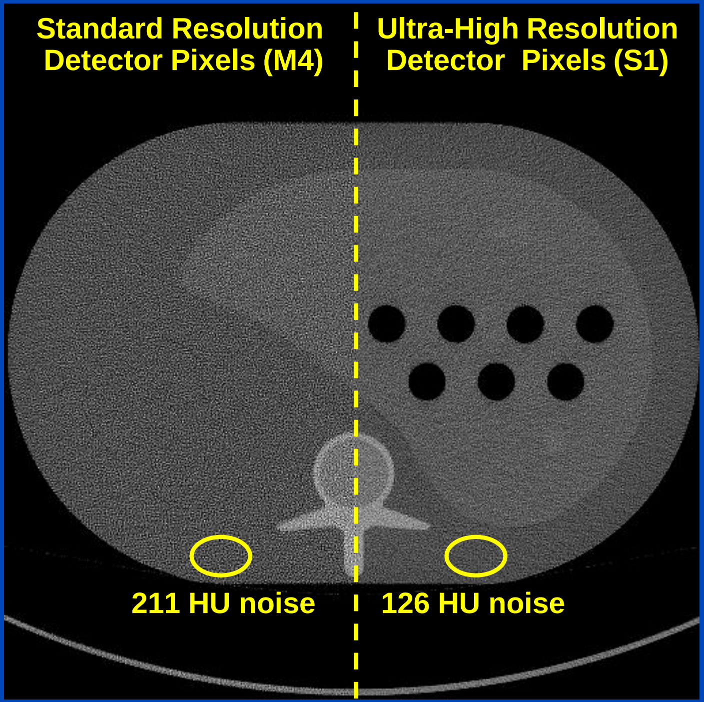
UHR vs. EI	80 kV	100 kV	120 kV	140 kV
S	**33%** ± 9%	**52%** ± 5%	**57%** ± 7%	**57%** ± 6%
M	**41%** ± 8%	**47%** ± 7%	**60%** ± 6%	**62%** ± 4%
L	**48%** ± 8%	**43%** ± 10%	**54%** ± 6%	**63%** ± 5%



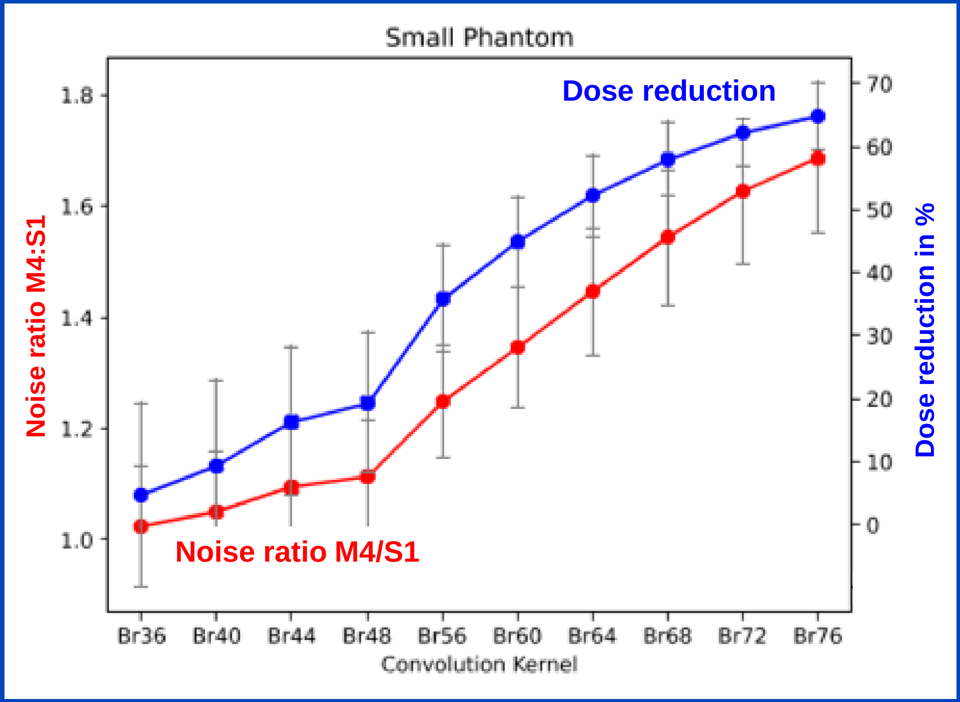
Klein, Kachelrieß, Sawall et al. Invest. Radiol. 55(2), Feb 2020

Lateral Small Pixel Effect at Naeotom Alpha

PC vs. PC
("small pixel effect only")



Analytical recon. *C* = 500 HU, *W* = 1500 HU



These results are valid for analytical and iterative recon.

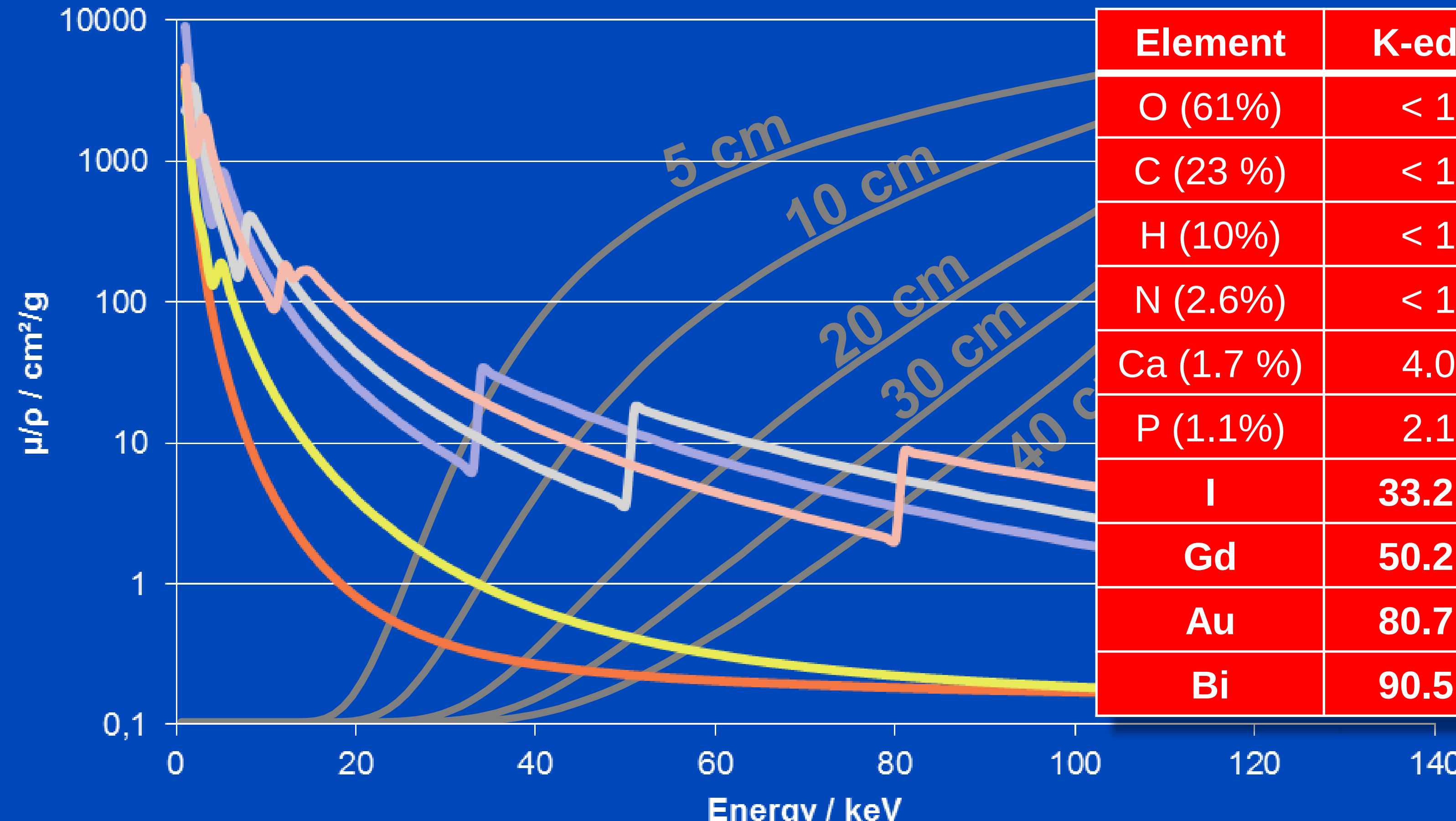
To disable the longitudinal small pixel effect, we reconstructed rather thick slices (1 mm thickness).

$$\text{DoseReduction} = 1 - \frac{1}{\text{NoiseRatio}^2}$$

K-Edges: More than Dual Energy CT?

$$\mu(\boldsymbol{r},E) = f_1(\boldsymbol{r})\psi_1(E) + f_2(\boldsymbol{r})\psi_2(E) + \cancel{f_3(\boldsymbol{r})\psi_3(E) + \ldots}$$

Apart from special applications, e.g.
Iodine k-edge imaging of the breast



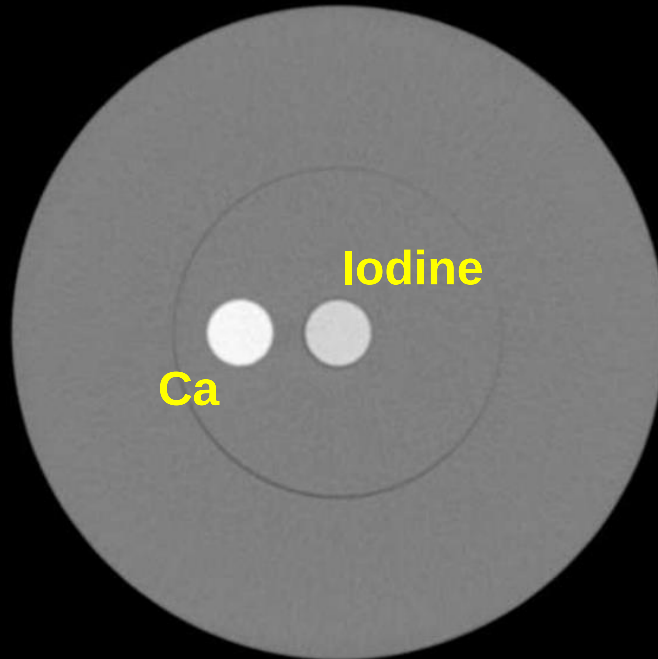
Element	K-edge
O (61%)	< 1 keV
C (23 %)	< 1 keV
H (10%)	< 1 keV
N (2.6%)	< 1 keV
Ca (1.7 %)	4.0 keV
P (1.1%)	2.1 keV
I	**33.2 keV**
Gd	**50.2 keV**
Au	**80.7 keV**
Bi	**90.5 keV**

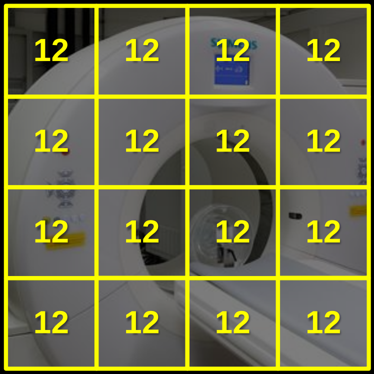
Gray curves: 120 kV water transmission on a non-logarithmic ordinate individually normalized to 1 at 140 keV.

DECT

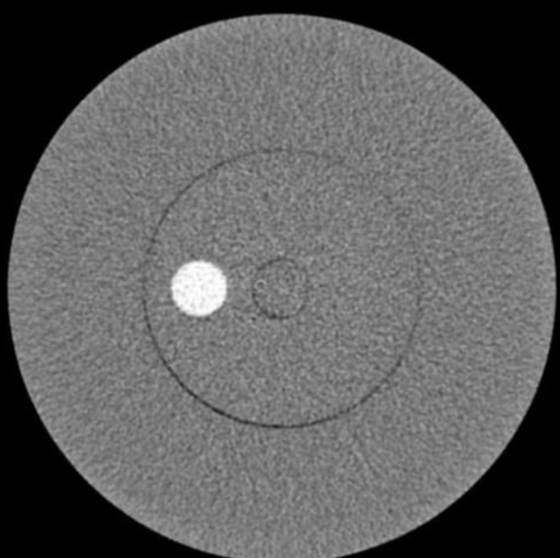
Ca-I Decomposition

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

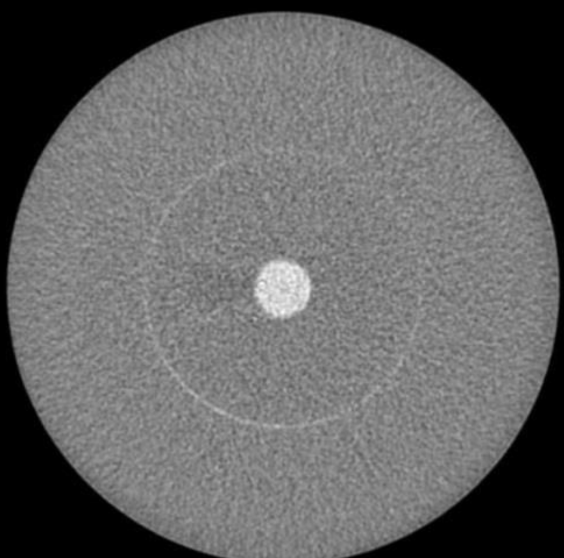




Calcium image



Iodine image

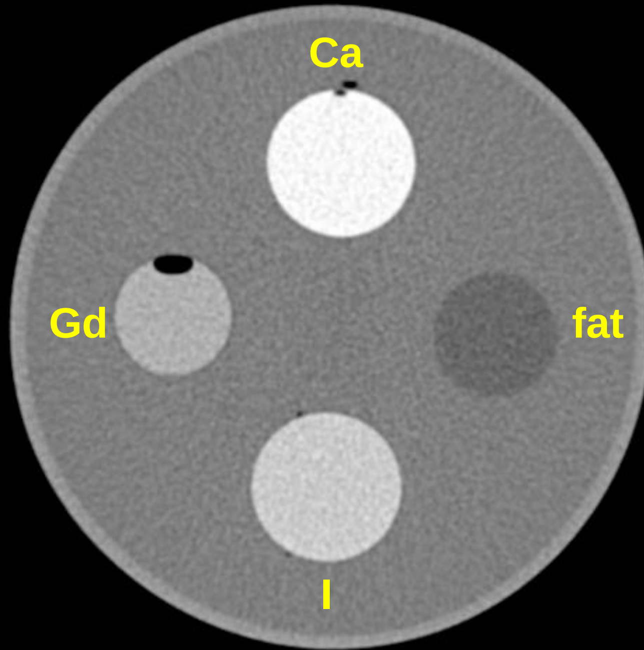


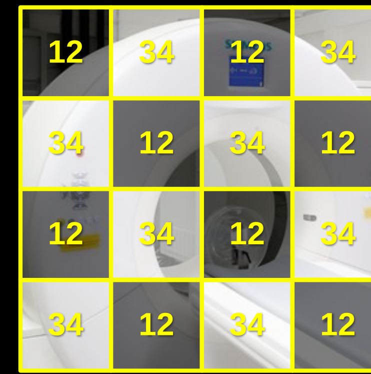
Courtesy of Siemens Healthcare

MECT

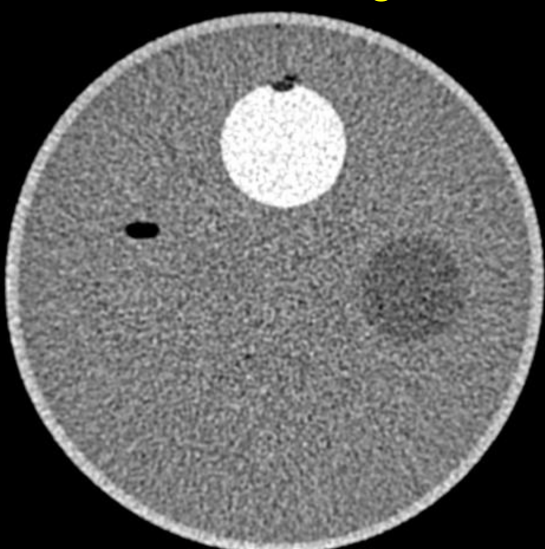
Ca-Gd-I Decomposition

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

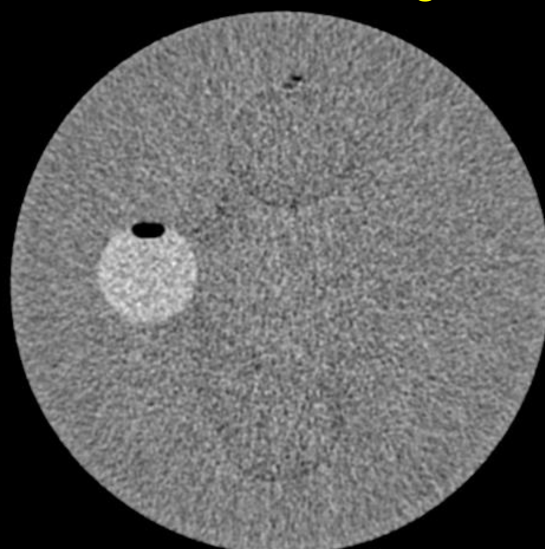




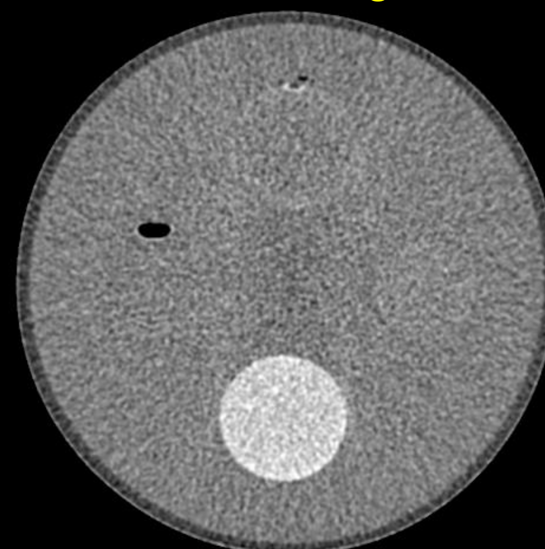
Calcium image



Gadolinium image

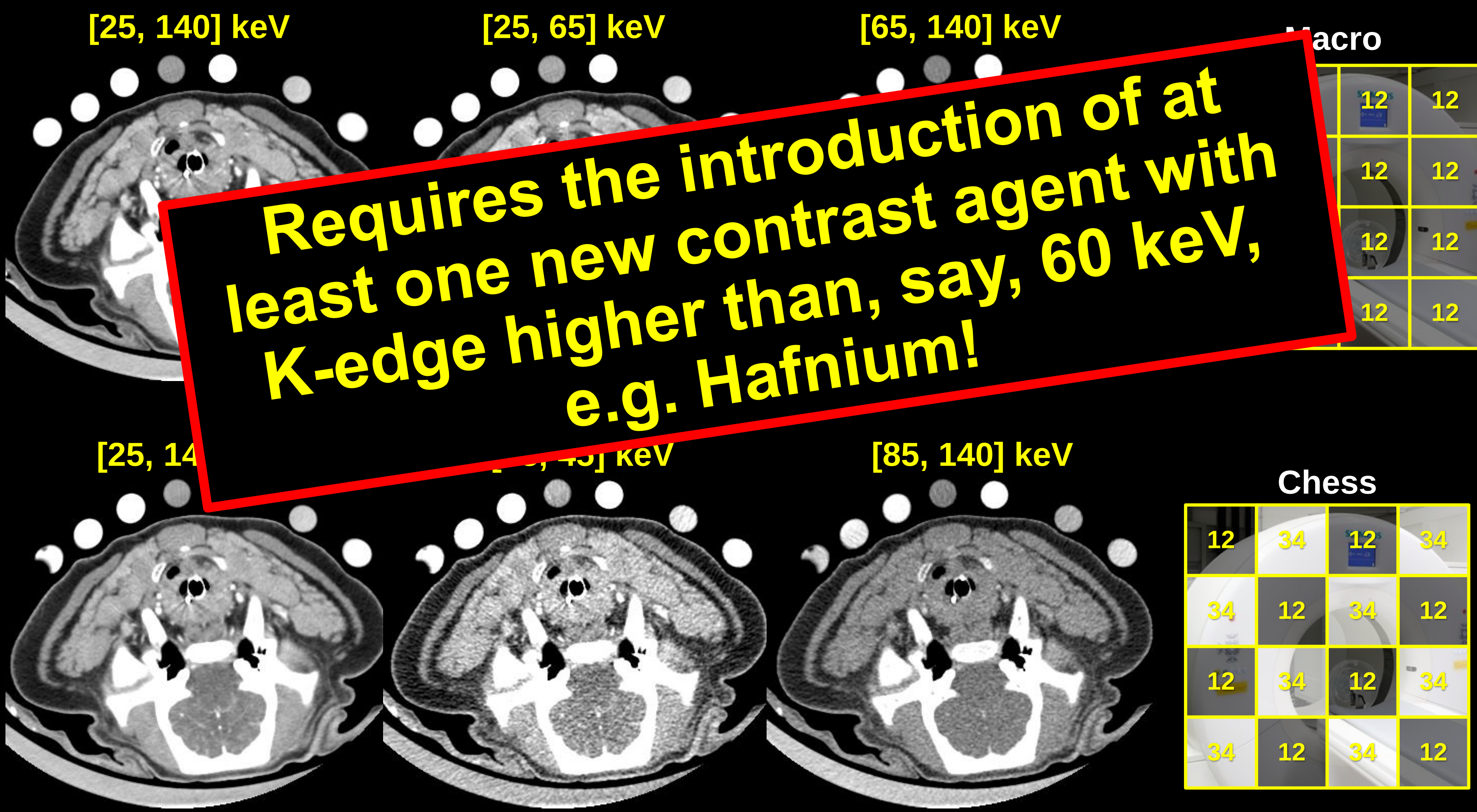


Iodine image



Courtesy of Siemens Healthcare

Preclinical Study
(40 kg swine, iodine contrast)



Requires the introduction of at least one new contrast agent with K-edge higher than, say, 60 keV, e.g. Hafnium!

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

Potential Advantages of PCCT

- Everything retrospectively on demand
  - Spatial resolution
  - Spectral information
  - Virtual tube voltage setting
- 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!

This presentation will soon be available at www.dkfz.de/ct.

Job opportunities through DKFZ's international PhD or Postdoctoral Fellowship programs (marc.kachelriess@dkfz.de).

Parts of the reconstruction software were provided by RayConStruct® GmbH, Nürnberg, Germany.