21st Asia-Oceania Congress of Medical Physics (AOCMP 2021)

# Recent Advancement in Medical Physics
# Photon-Counting CT


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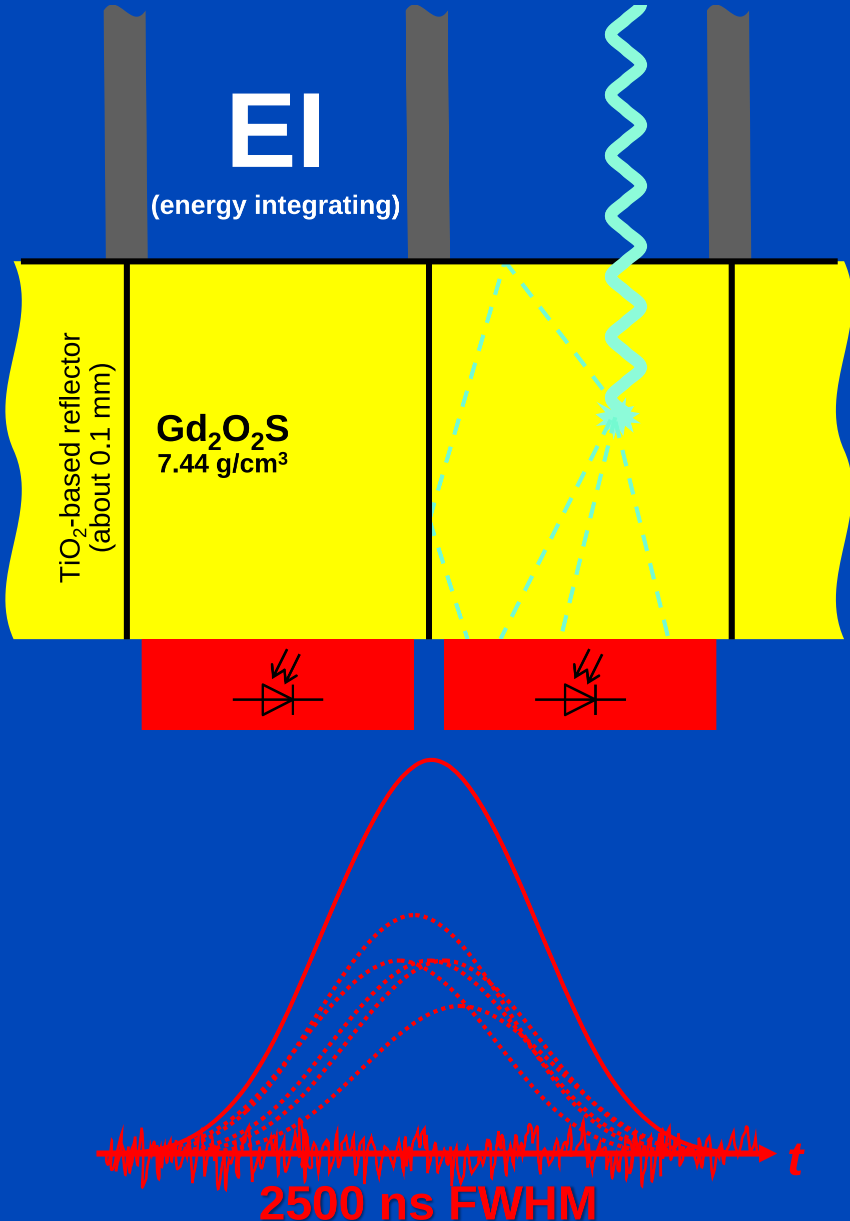
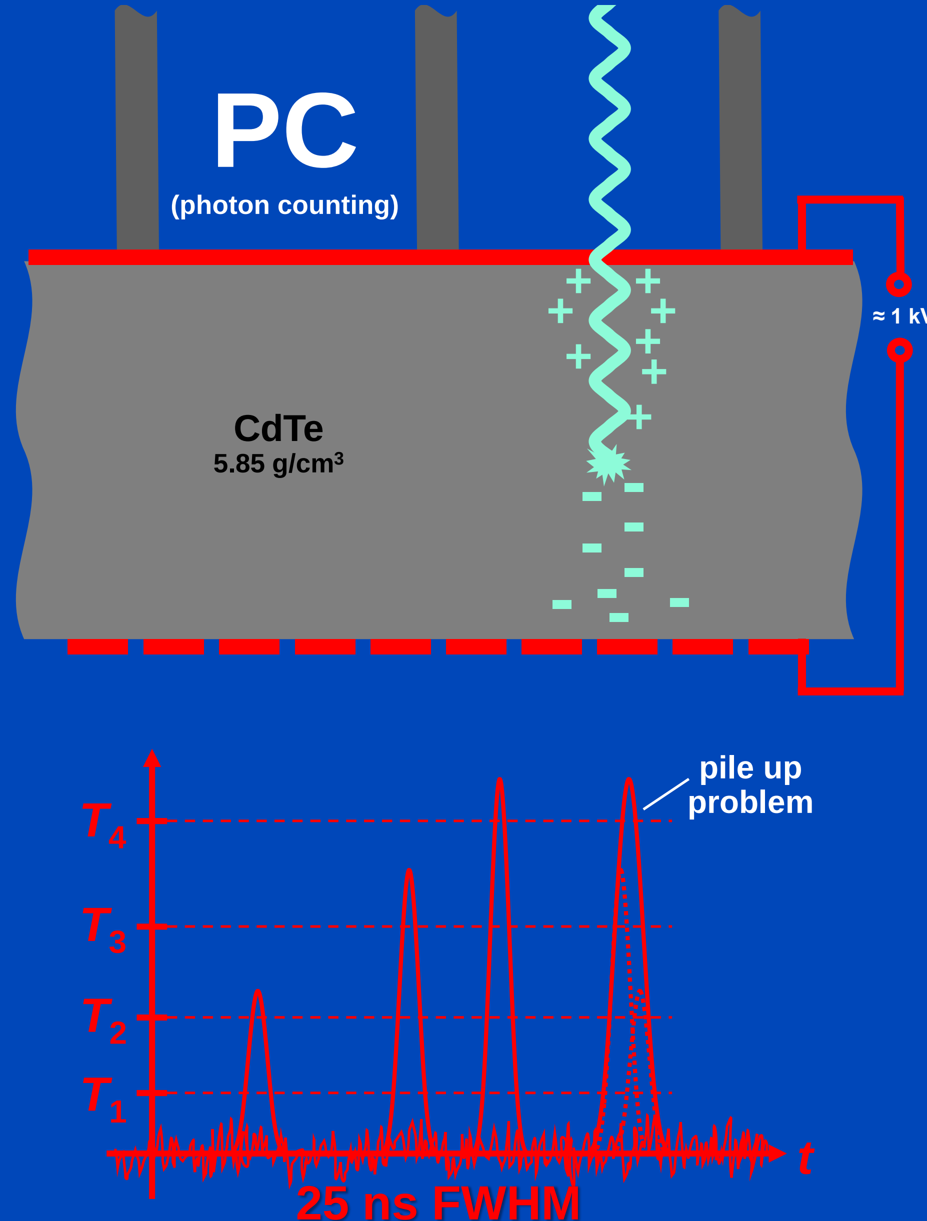
## Indirect Conversion (Today)




i.e. max O(40·10³) cps

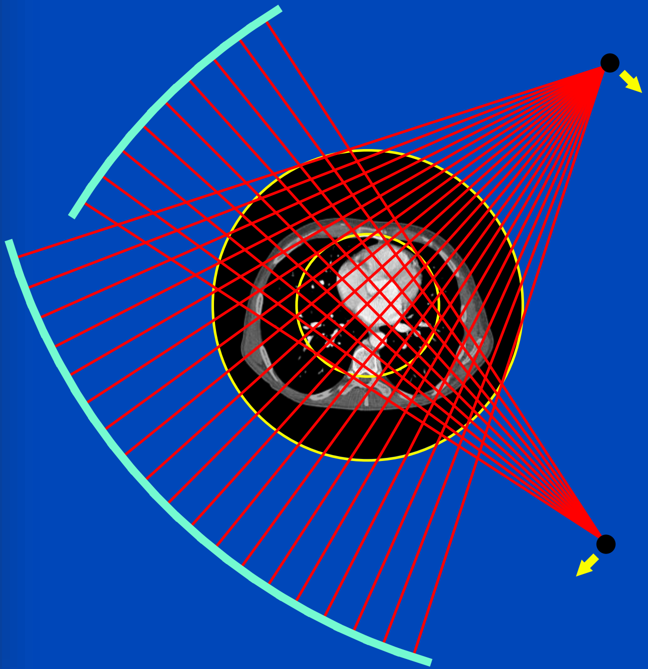
## Direct Conversion (Future)




i.e. max O(40·10⁶) 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.**

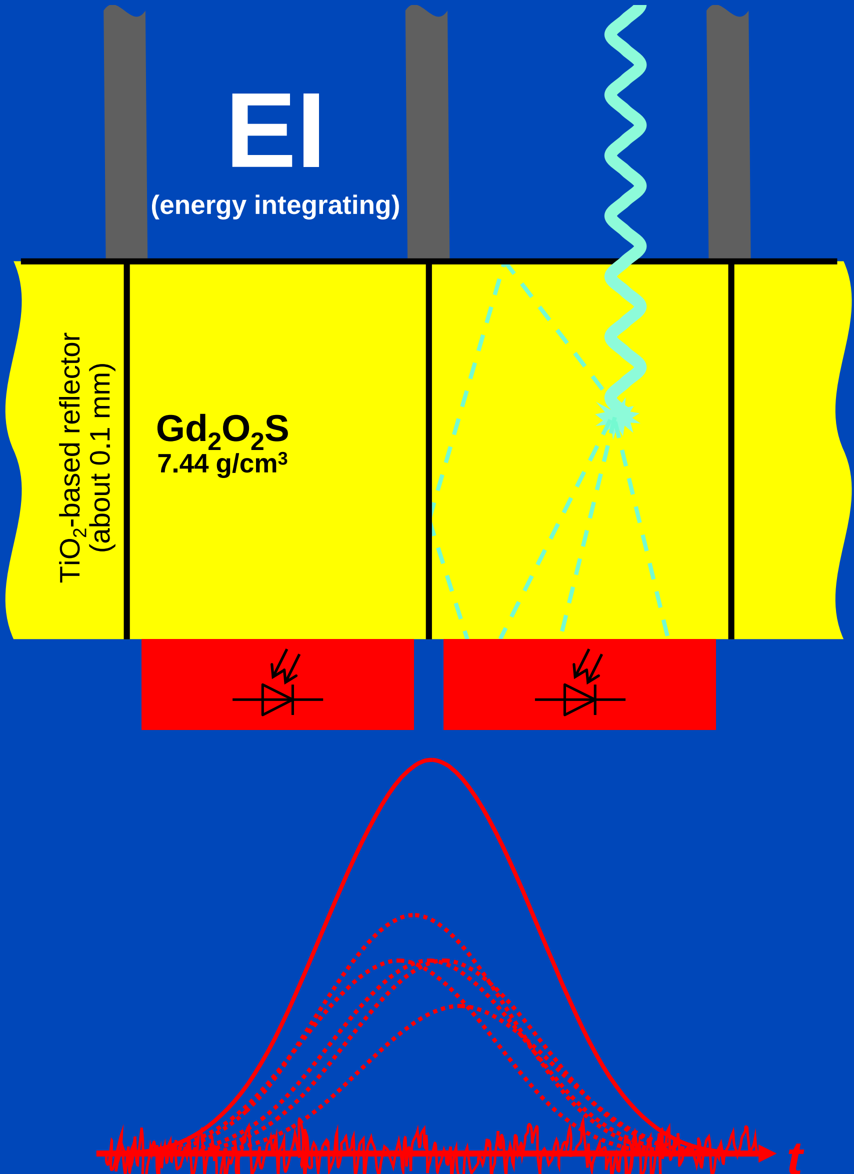
# Siemens Naeotom Alpha
# The World‘s First Photon-Counting CT



# Premium CT Systems 2021/2022

| Vendor | CT-System | Configuration | Collim, Cone | Rotation, FOM | Max. Power, Anode Angle | Max. mA @ low kV, patient-specific filters | Matrix | DECT |
|---|---|---|---|---|---|---|---|---|
| Canon | Aquilion ONE Genesis | 320 × 0.5 mm PUREViSION | 160 mm 15° | 0.275 s 50 cm | 100 kW, 10° MegaCool Vi | 600 mA @ 80 kV, none | 512 | 2 scans |
| Canon | Aquilion Precision | 160 × 0.25 mm PUREViSION | 40 mm 3.9° | 0.35 s 50 cm | 72 kW, 7° MegaCool | 600 mA @ 80 kV, none | 512, 1024, 2048 | 2 scans |
| GE | Revolution Apex | 256 × 0.625 mm GemStone Clarity | 160 mm 15° | 0.28 s 50 cm | 108 kW, 10° Quantix 160 | 1300 mA @ 70+80 kV, none | 512 | fast TVS or 2 scans |
| GE | Cardio-Graphe | 192 × 0.73 mm (focused FOM) | 140 mm 17° | 0.24 s 25 cm | 72 kW, 13° Dual MCS-2093 | 600 mA @ 80 kV, none | 512 | 2 scans |
| Philips | Brilliance iCT | 2 · 128 × 0.625 mm NanoPanel 3D | 80 mm 7.7° | 0.27 s 50 cm | 120 kW, 8° iMRC | 925 mA @ 80 kV, none | 512, 768, 1024 | 2 scans |
| Philips | Spectral CT 7500 | 2 · 128 × 0.625 mm NanoPanel Prism | 80 mm 7.7° | 0.27 s 50 cm | 120 kW, 8° iMRC | 925 mA @ 80 kV, none | 512, 768, 1024 | sandwich |
| Siemens | Somatom X.cite | 2 · 64 × 0.6 mm Stellar | 38.4 mm 3.7° | 0.3 s 50 cm | 105 kW, 8° Vectron | 1200 mA @ 70+80+90 kV, {0, 0.4, 0.7} mm Sn | 512, 768, 1024 | split filter or 2 scans |
| Siemens | Somatom Force | 2 · 2 · 96 × 0.6 mm Stellar | 57.6 mm 5.5° | 0.25 s 50/35 cm | 2 · 120 kW, 8° Vectron | 2 · 1300 mA @ 70+80+ 90 kV, {0, 0.6} mm Sn | 512, 768, 1024 | DSCT |
| **Siemens** | **Naeotom Alpha** | **2 · 144×0.4/120×0.2 mm Photon Counting!** | **57.6 mm 5.5°** | **0.25 s 50/36 cm** | **2 · 120 kW, 8° Vectron** | **2 · 1300 mA @ 90 kV, {0, 0.4} mm Sn** | **512, 768, 1024** | **DSPCCT** |

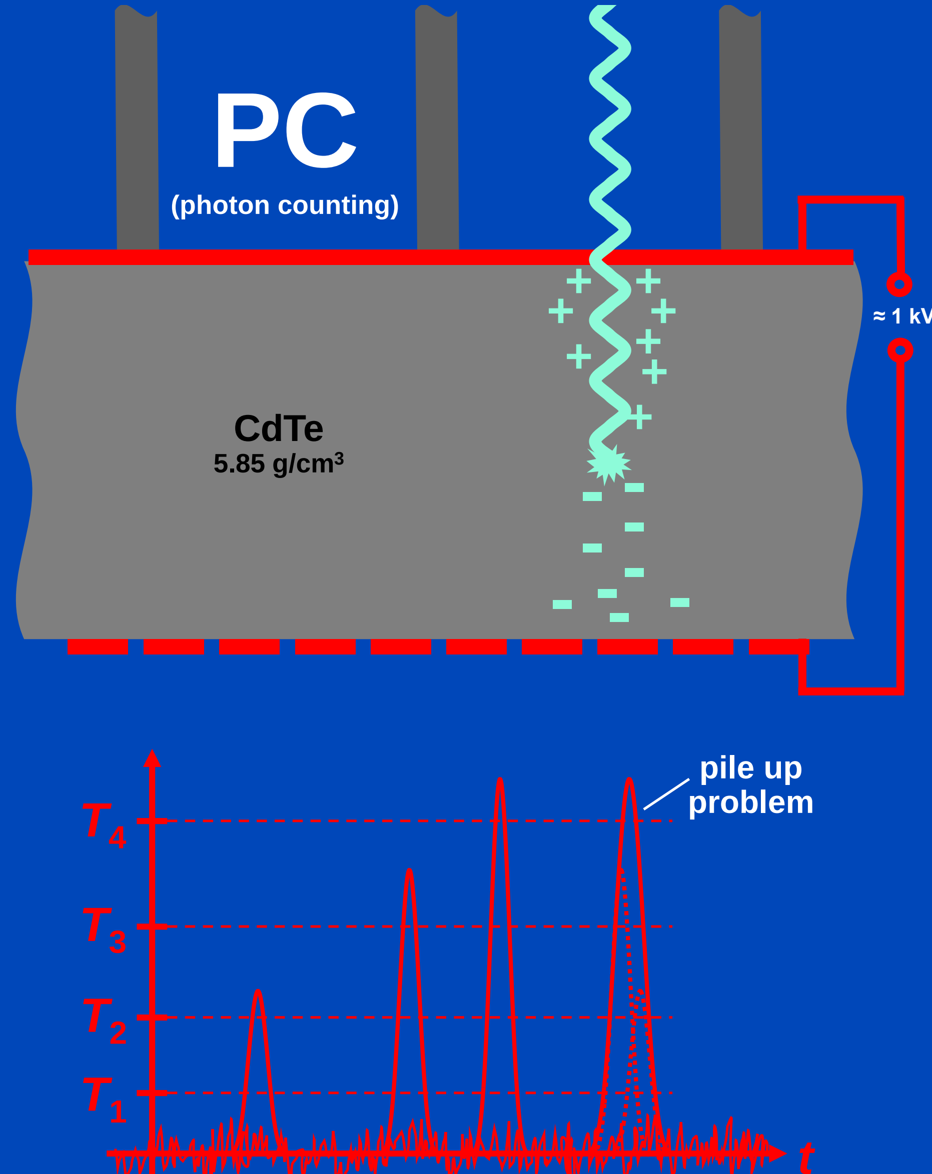
# Indirect Conversion (Today)




**2500 ns FWHM**

i.e. max $O(40 \cdot 10^3)$ cps

# Direct Conversion (Future)




**25 ns FWHM**

i.e. max $O(40 \cdot 10^6)$ cps

**Requirements for CT: up to $10^9$ x-ray photon counts per second per mm². Hence, photon counting only achievable for direct converters.**

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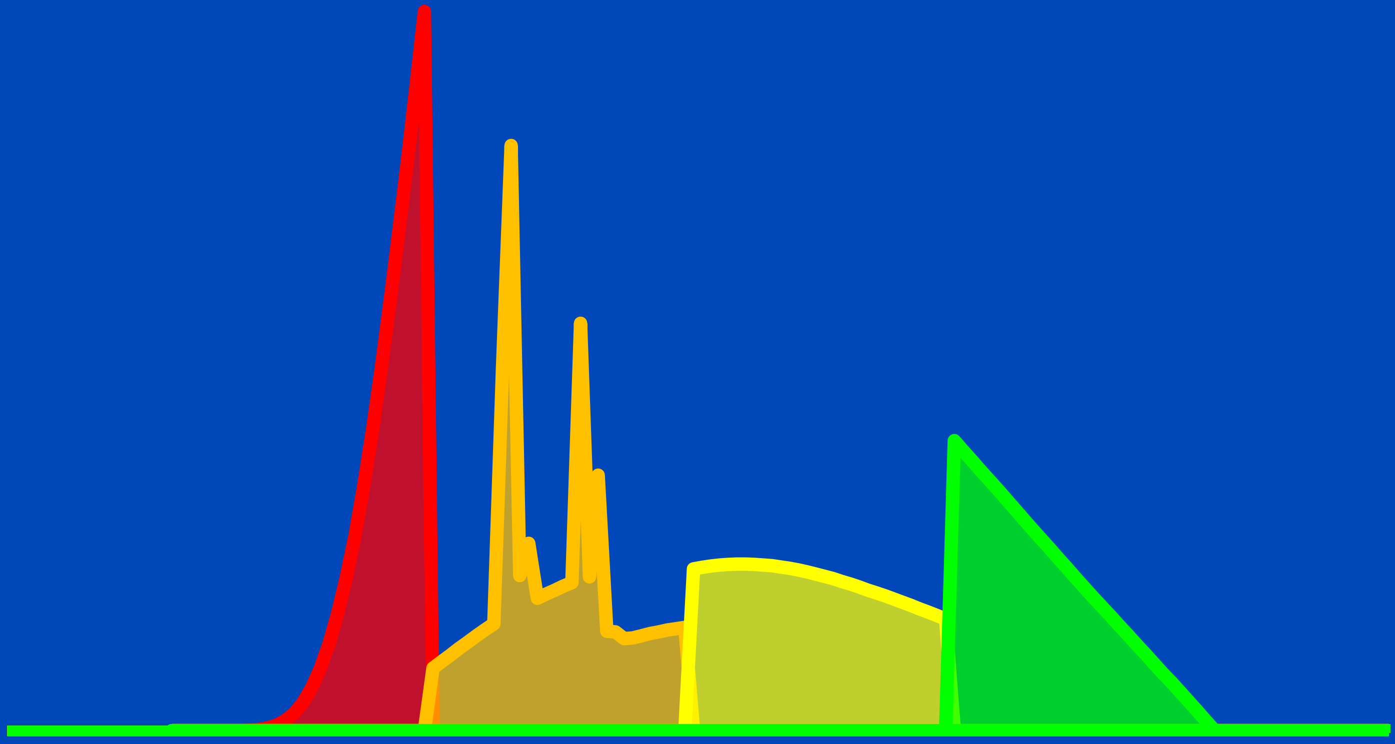
# Energy-Selective Detectors: Improved Spectroscopy, Reduced Dose?

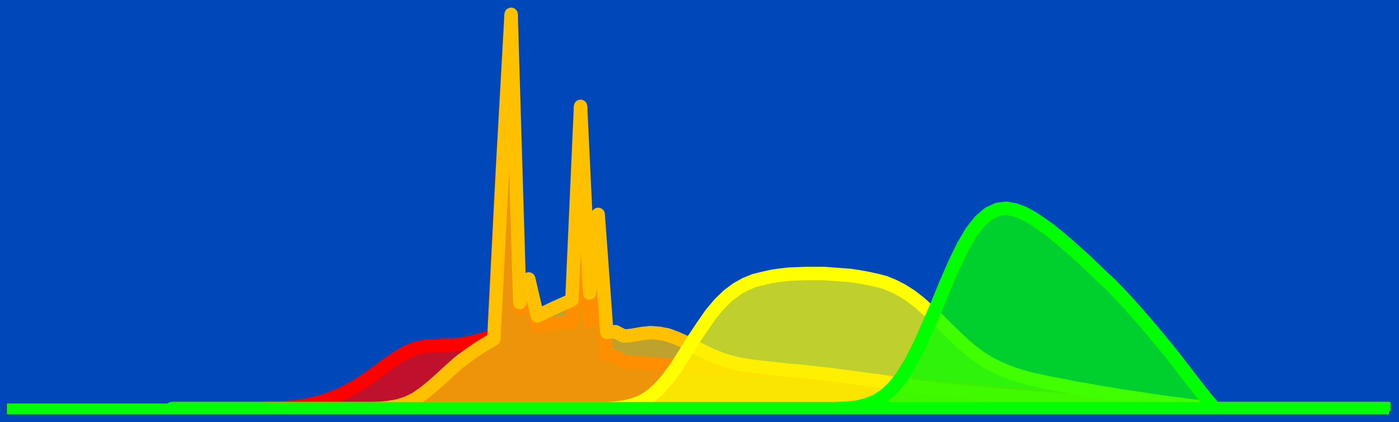
Ideally, bin spectra do not overlap, …



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

# Energy-Selective Detectors: Improved Spectroscopy, Reduced Dose?

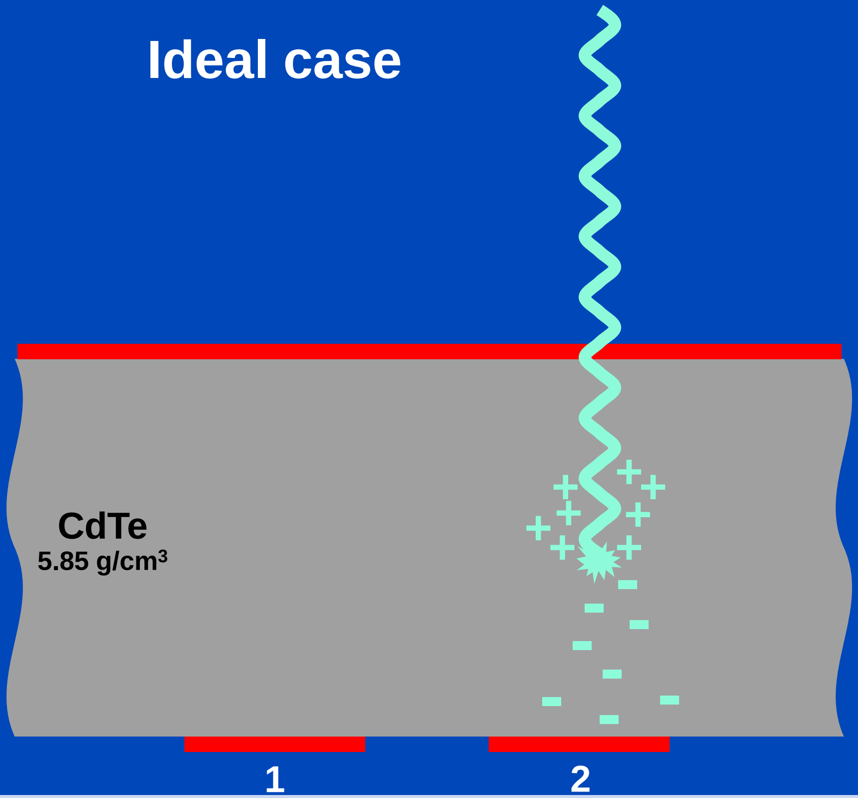
… realistically, however they do!

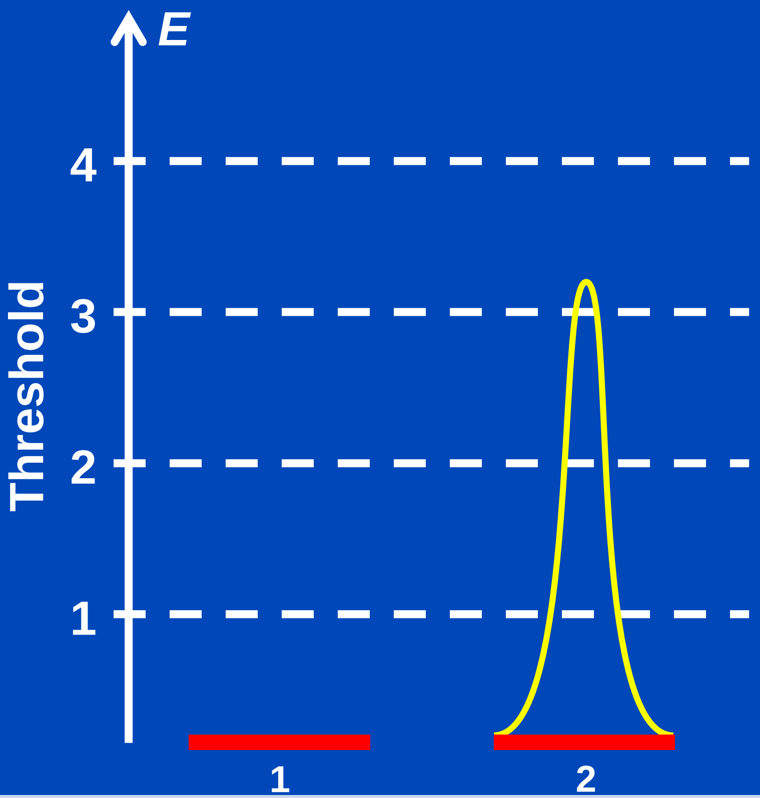


Spectra as seen after having passed a 32 cm water layer.

# Photon Events

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

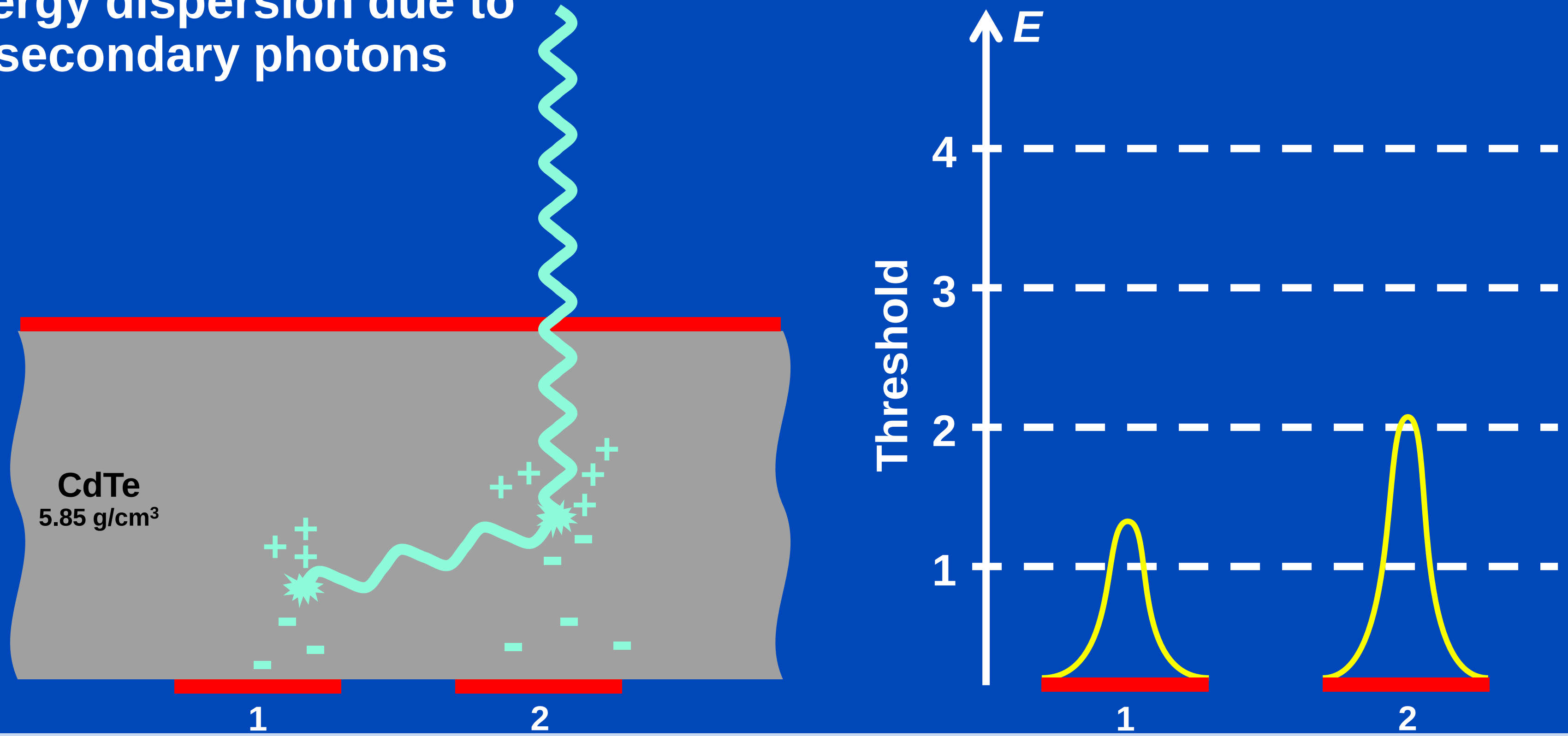


# Photon Events

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

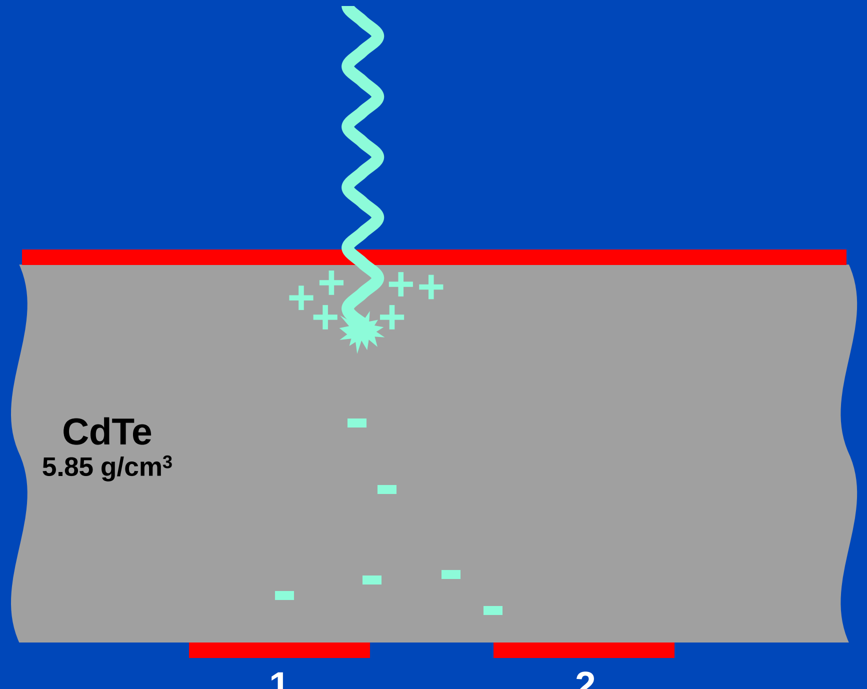
Energy dispersion due to secondary photons

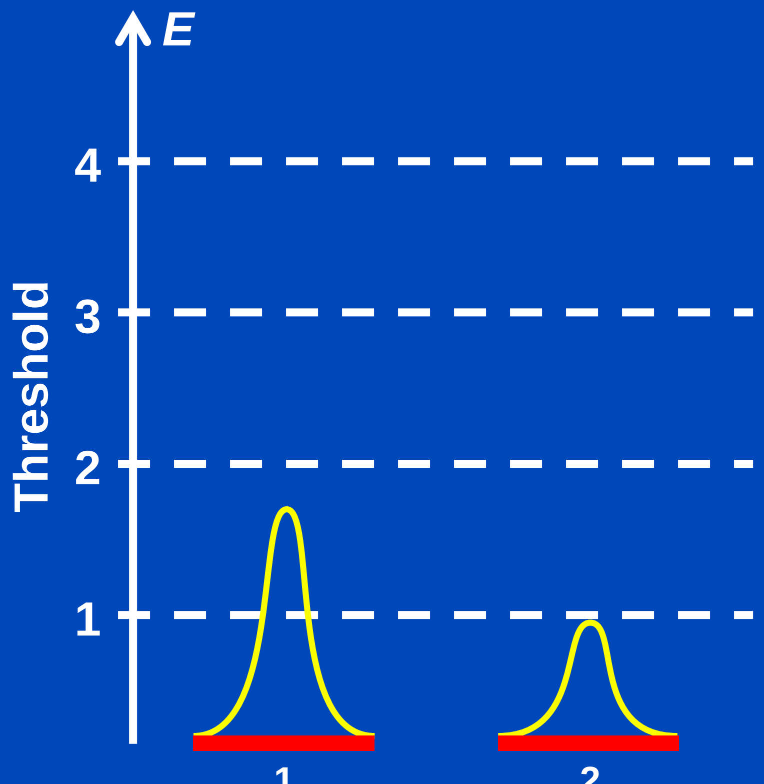


# Photon Events

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

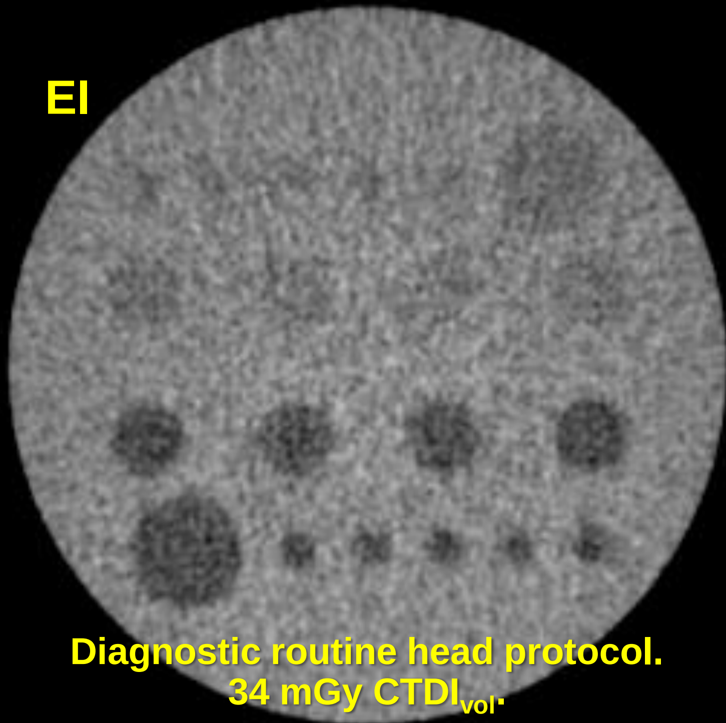
Energy dispersion due to charge diffusion



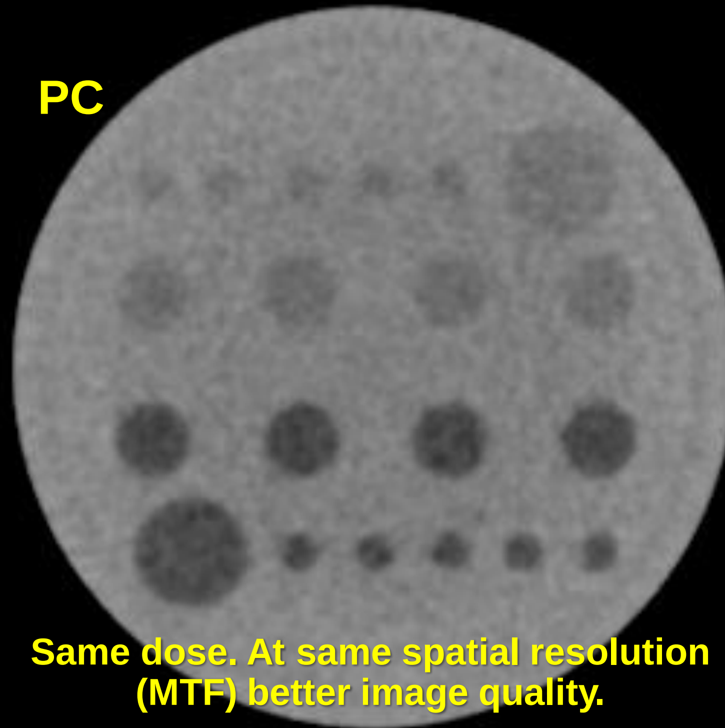



## Diagnostic CT (Conventional Detector) of a Low Contrast Phantom

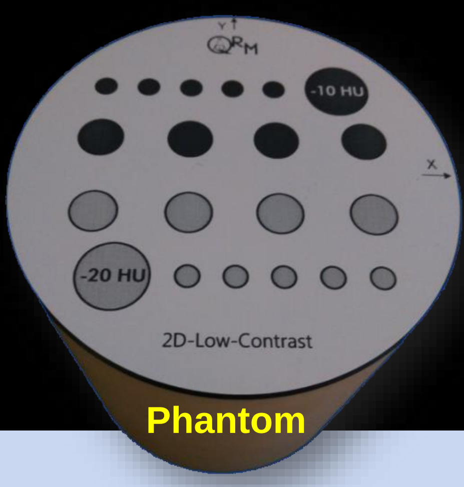
## Photon Counting Detector CT of a Low Contrast Phantom



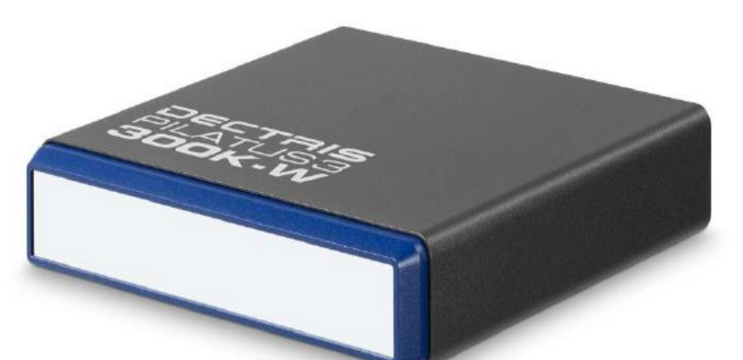

Diagnostic routine head protocol.
34 mGy $CTDI_{vol}$.




Same dose. At same spatial resolution (MTF) better image quality.




Phantom




Photon Counting Detector

*C* = 0 HU, *W* = 80 HU

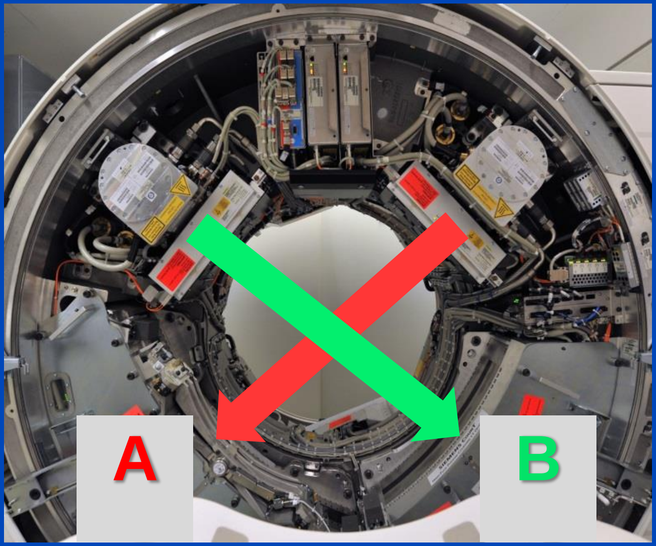

SIEMENS
SOMATOM CounT
@ dkfz.

# Siemens CounT CT System

**Gantry from a clinical dual source scanner**

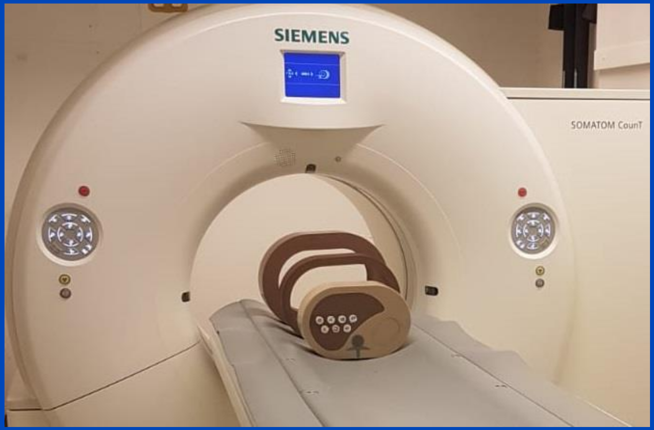
**A: conventional CT detector (50.0 cm FOV)**

**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 |




Experimental CT, not commercially available.

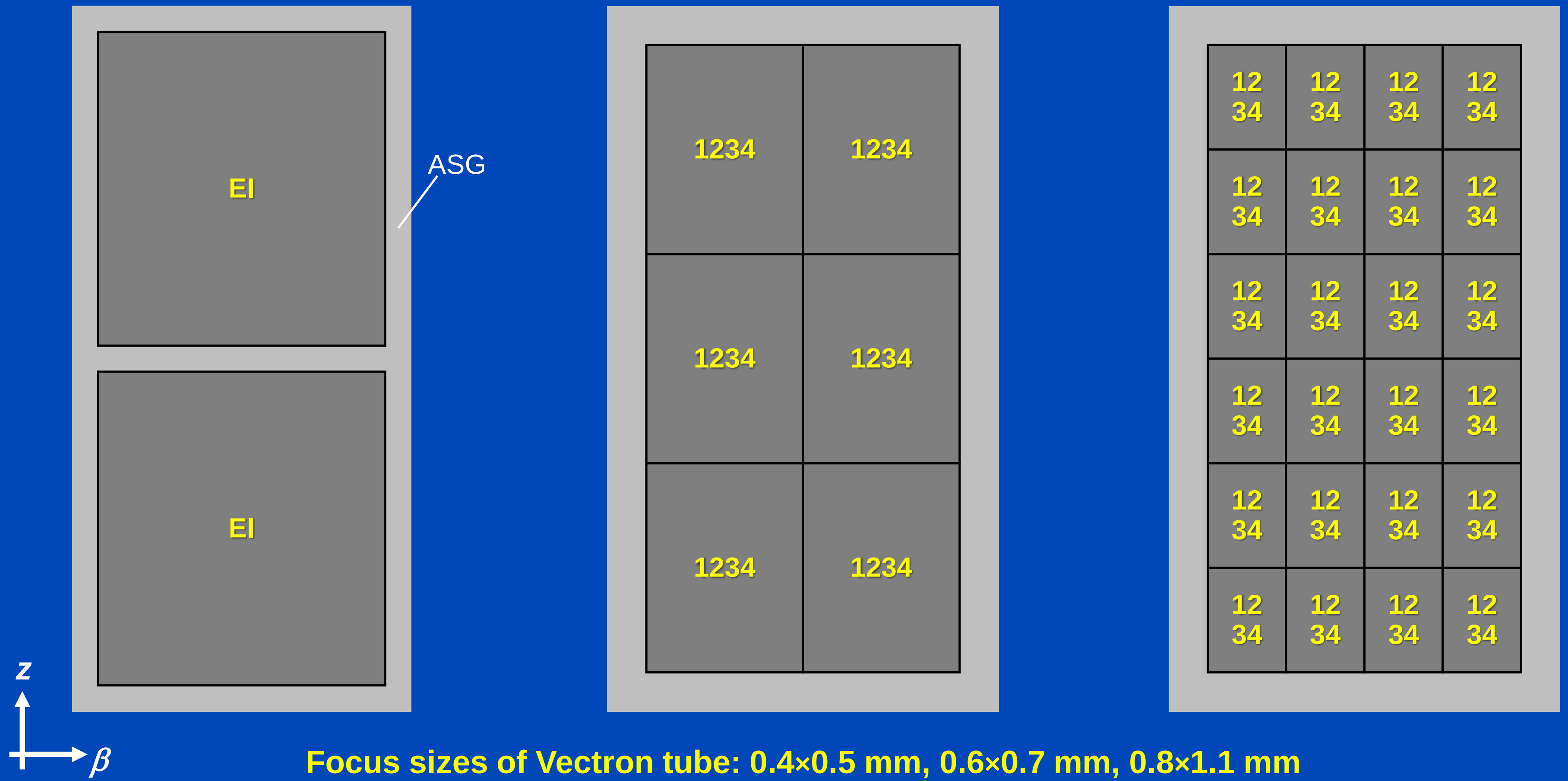
# Siemens Naeotom Alpha
# The World‘s First Photon-Counting CT

# Detector Pixel Force vs. Alpha[1]

**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 of Vectron tube: 0.4×0.5 mm, 0.6×0.7 mm, 0.8×1.1 mm**

[1]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

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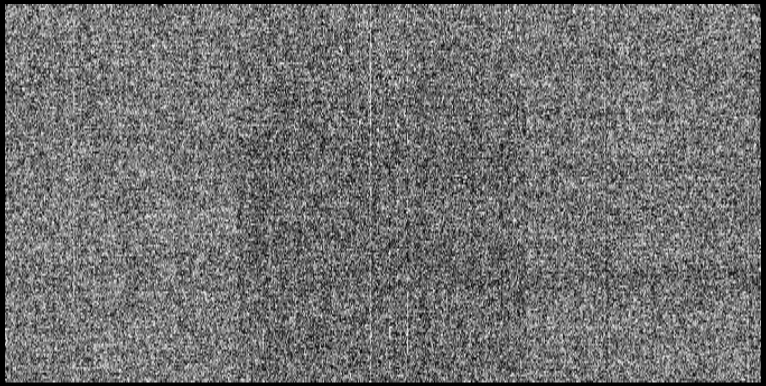
# Advantages of Photon Counting CT

- 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)



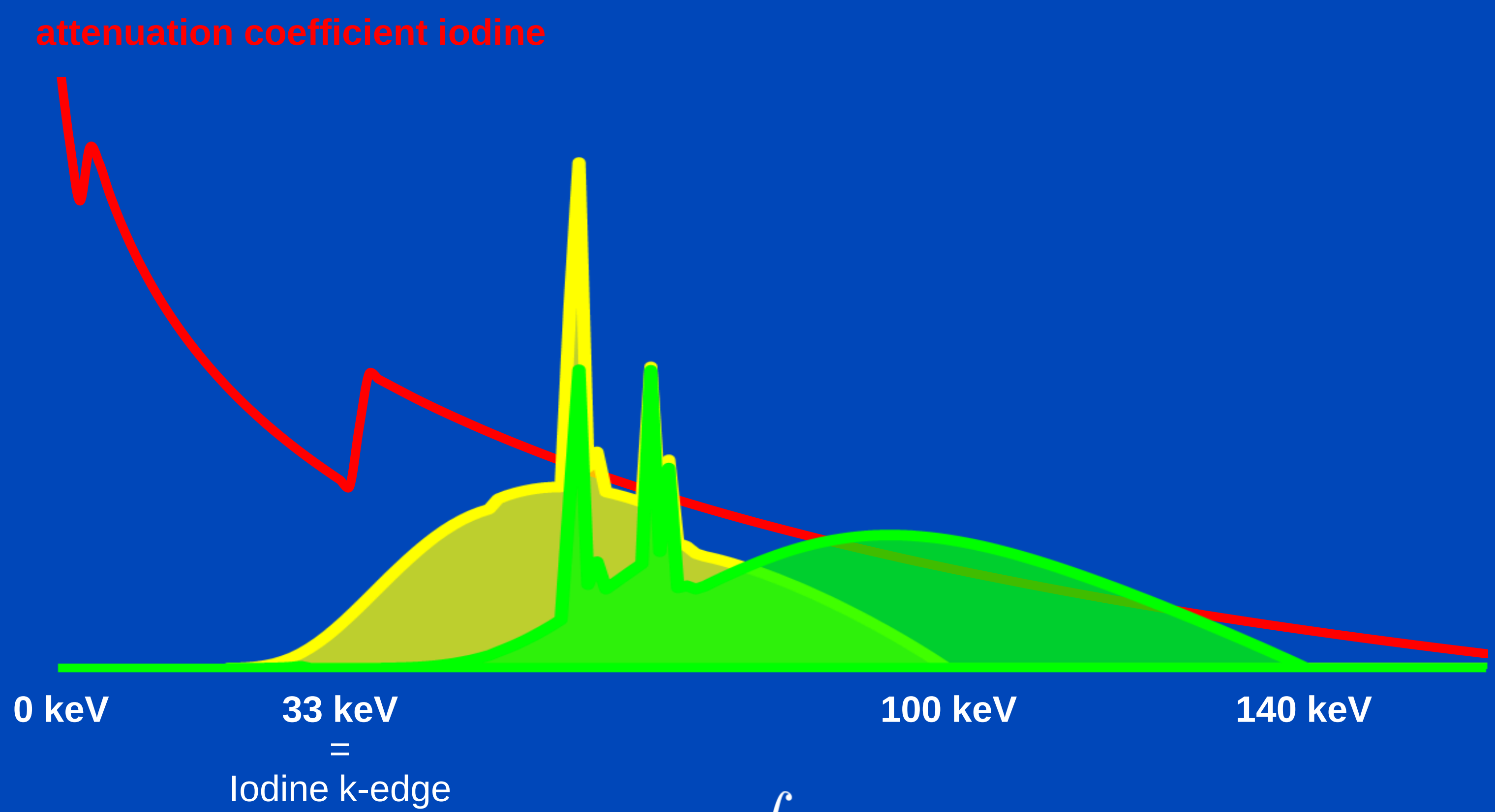
**Readout noise only. Single events hidden!**

**PC** (Dectris)

**No readout noise. Single events visible!**

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

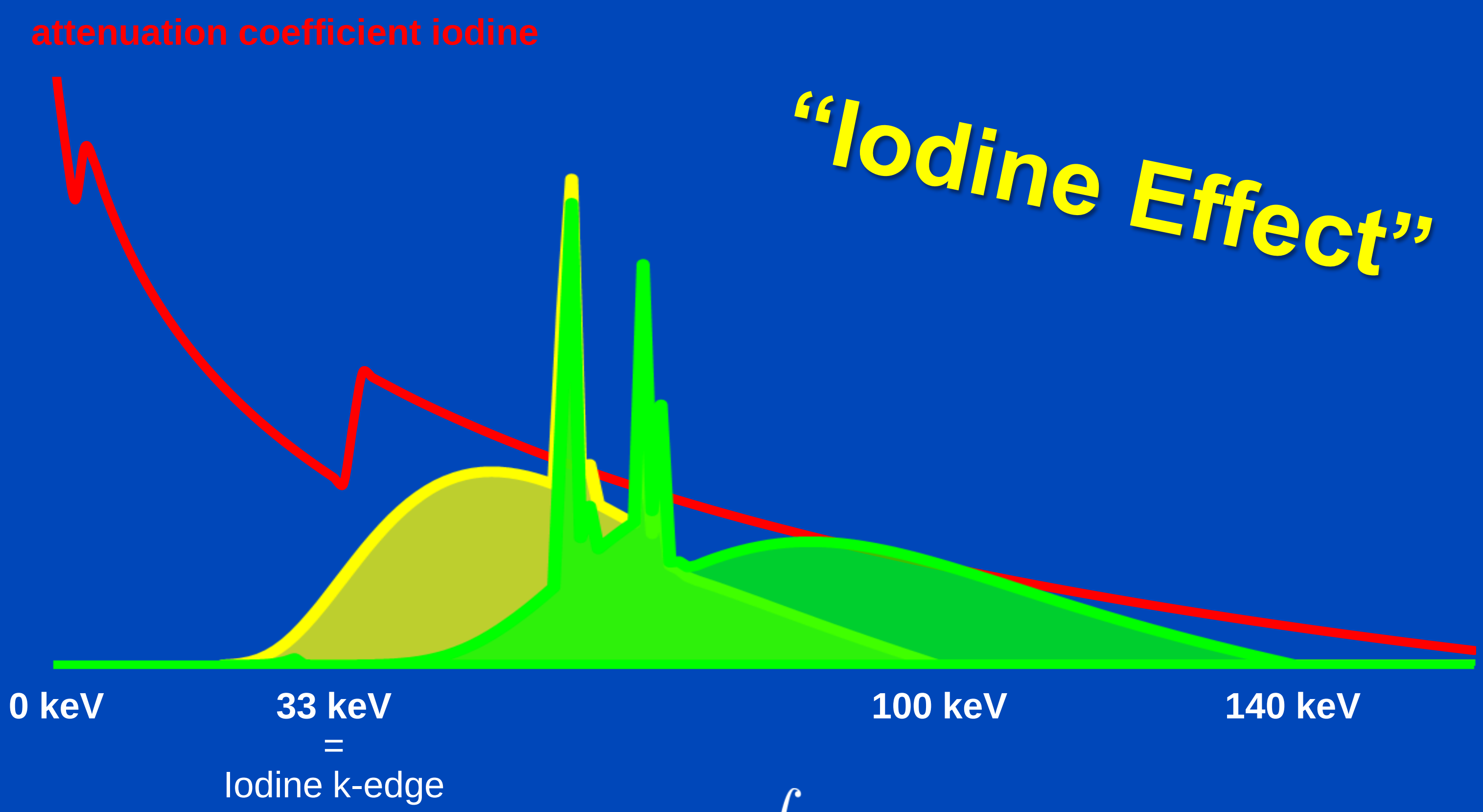
# Energy Integrating
## (Detected Spectra at 100 kV and 140 kV)




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

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

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




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

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

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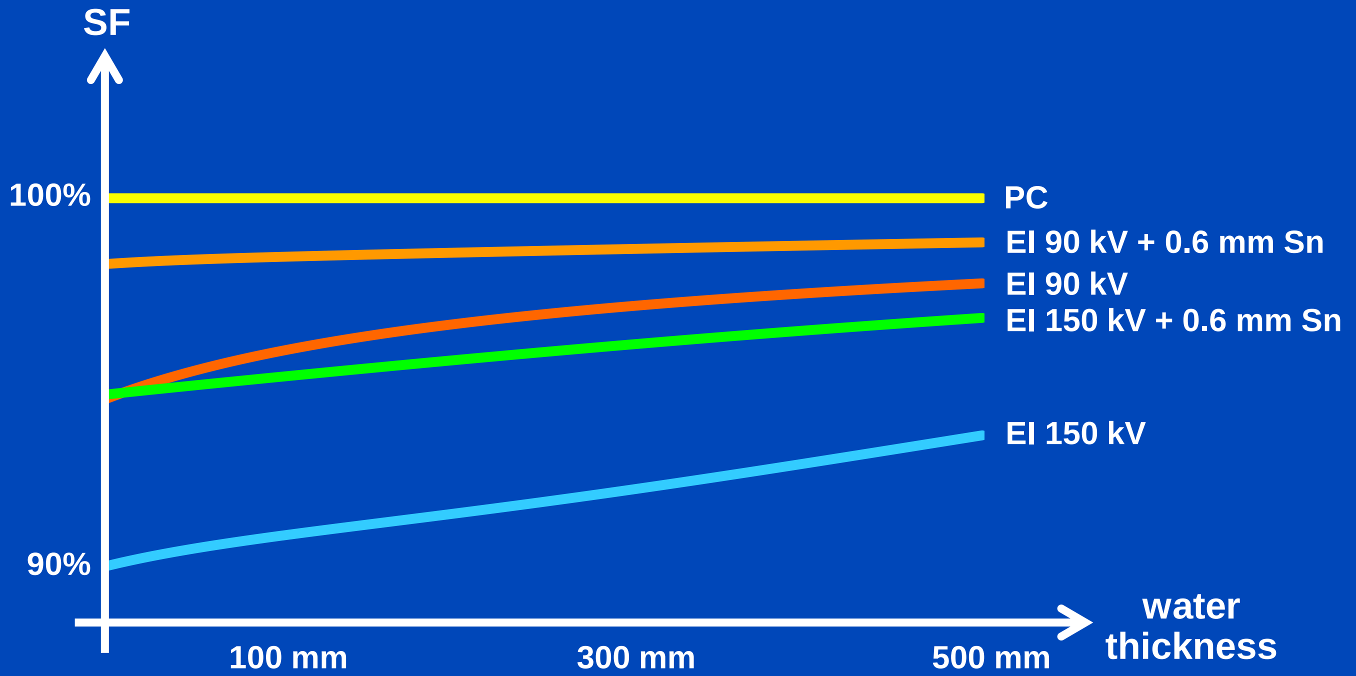
# Swank Factor

- 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.



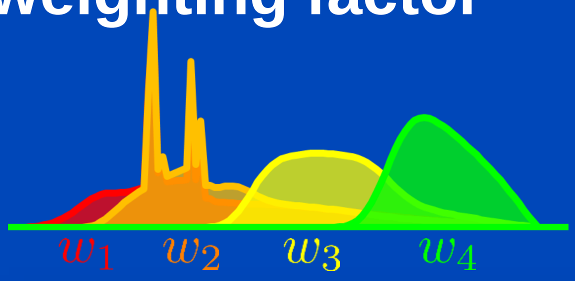

$$\mathrm{SF} = \frac{\mathrm{SNR}_{\mathrm{EI}}^2}{\mathrm{SNR}_{\mathrm{PC}}^2} = \frac{\left(\int dE\, E N(E)\right)^2}{\left(\int dE\, N(E)\right)\left(\int dE\, E^2 N(E)\right)} \leq 1$$

due to Schwarz' inequality

# Photon Counting used to Maximize CNR

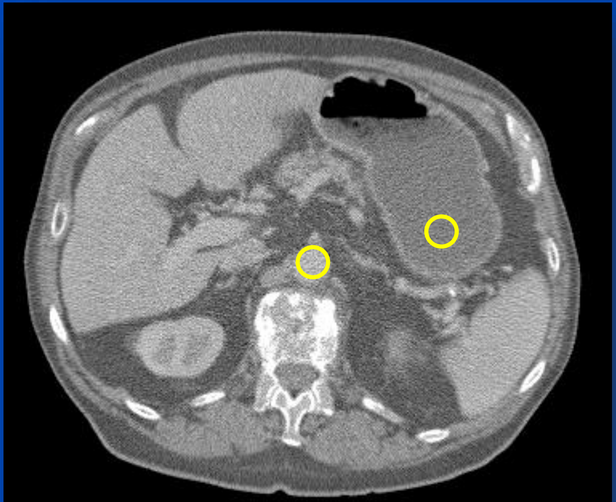
- With PC, energy bin sinograms can be weighted individually, i.e. by a weighted summation
- To optimize the CNR the optimal bin weighting factor $w_b$ is given by (weighting after log):

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




- 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.

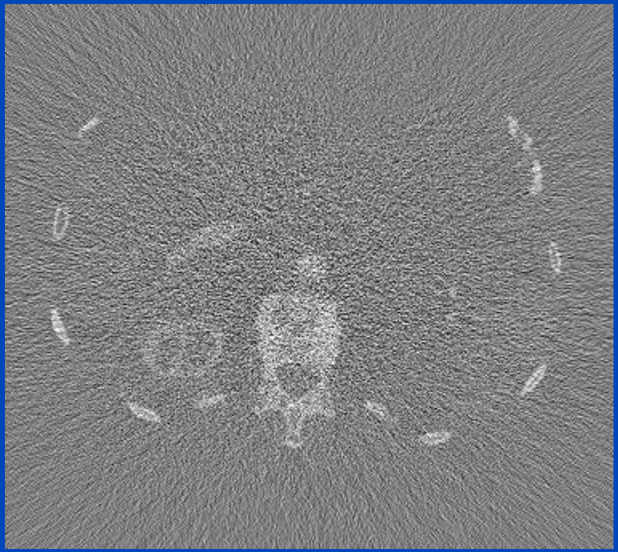
- At the optimum this evaluates to

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

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

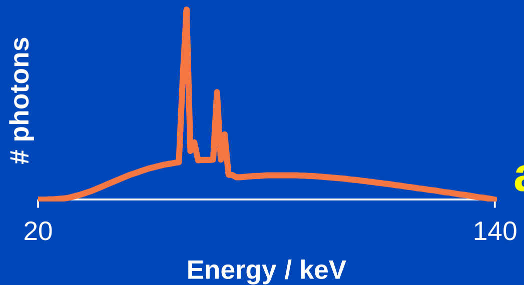
**Energy Integrating**

**PC minus EI**



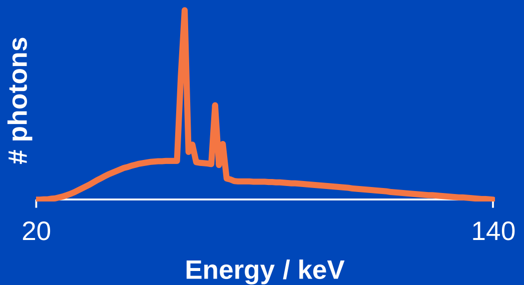
**Photon Counting**

**CNR = 2.11**




**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

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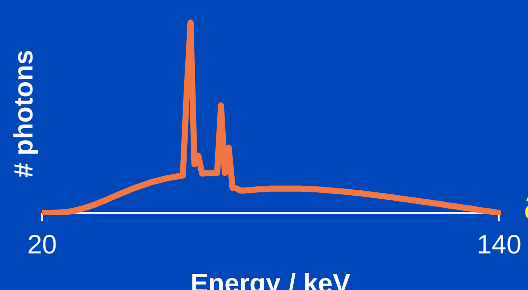
# Energy Integrating vs. Photon Counting

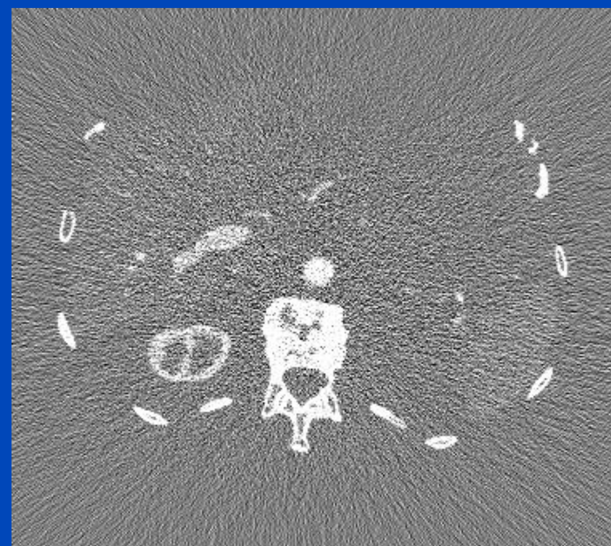
## with 4 bins from 20 to 140 keV

**Energy Integrating**

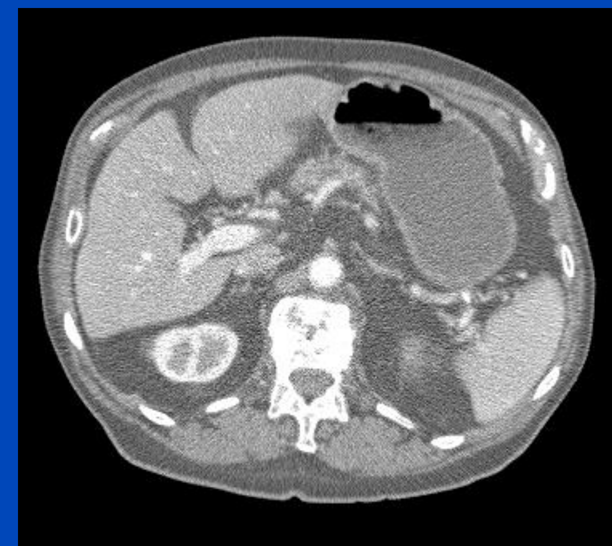
**CNR = 2.11**



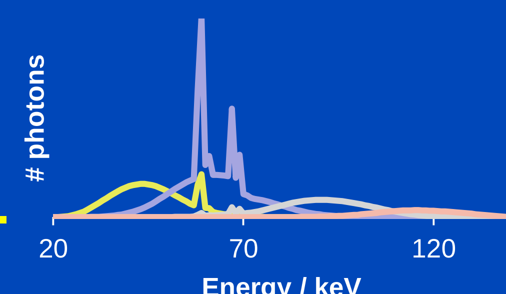

**PC minus EI**



**Photon Counting**



**CNR = 4.19**



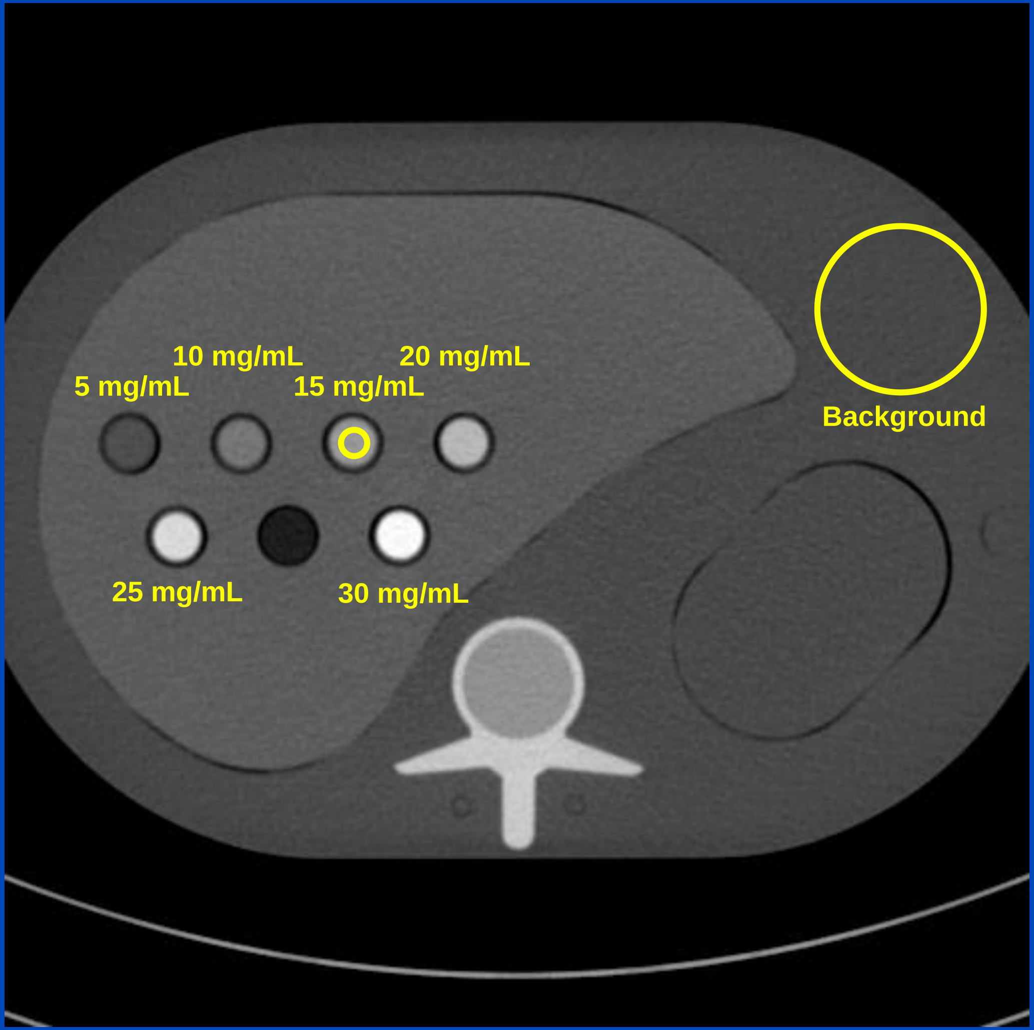

**99% CNR improvement or**
**75% dose reduction achievable**
**due to improved Swank factor**
**and optimized energy weighting.**

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

# Iodine CNRD Assessment

- Measure contrast between 2 ROIs, noise and dose
- Calculate CNRD
- Calculate dose reduction as

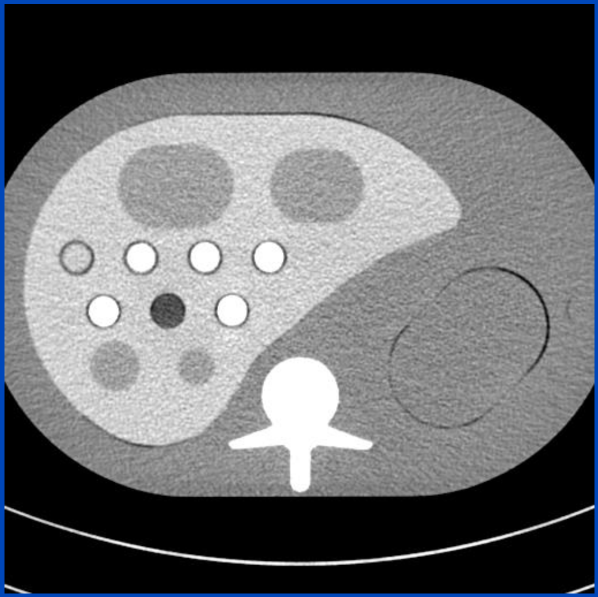
$$\mathrm{DR} = 1 - \frac{\mathrm{CNRD}_{\mathrm{EI}}^2}{\mathrm{CNRD}_{\mathrm{PC}}^2}$$



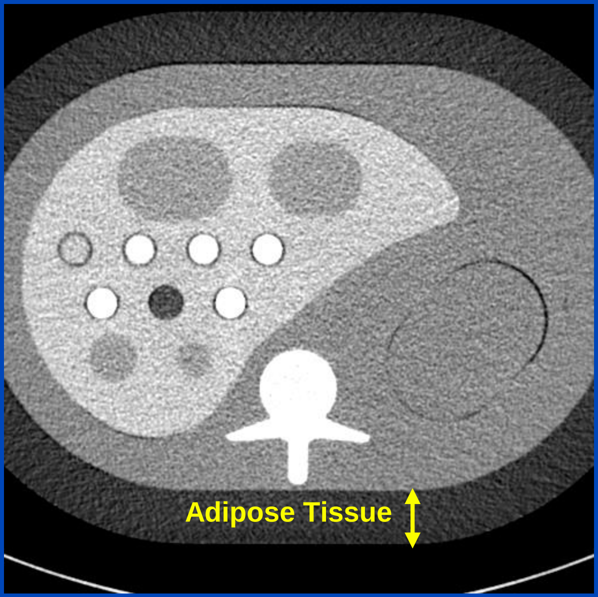

*C* = 180 HU, *W* = 600 HU

# Iodine CNRD Assessment

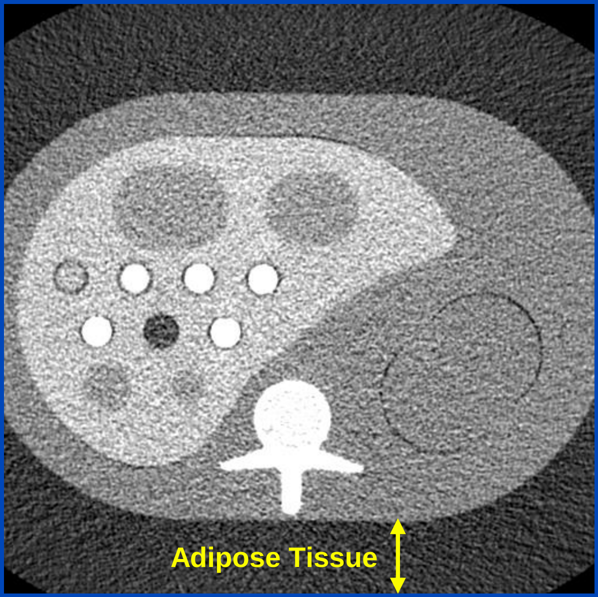
Small (200 × 300 mm)



Medium (250 × 350 mm)



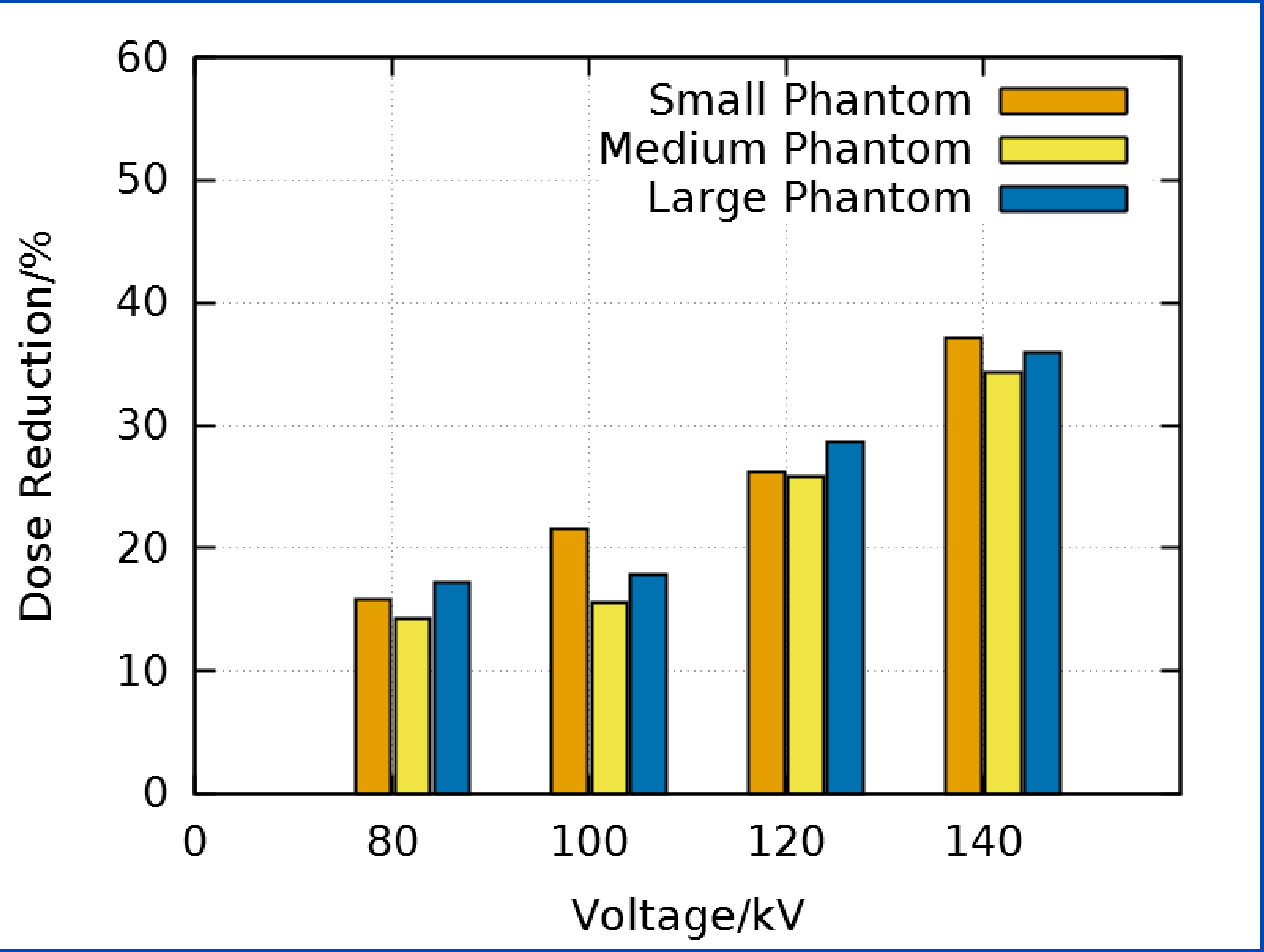

Large (300 × 400 mm)




80 kV, *C* = 0 HU, *W* = 400 HU

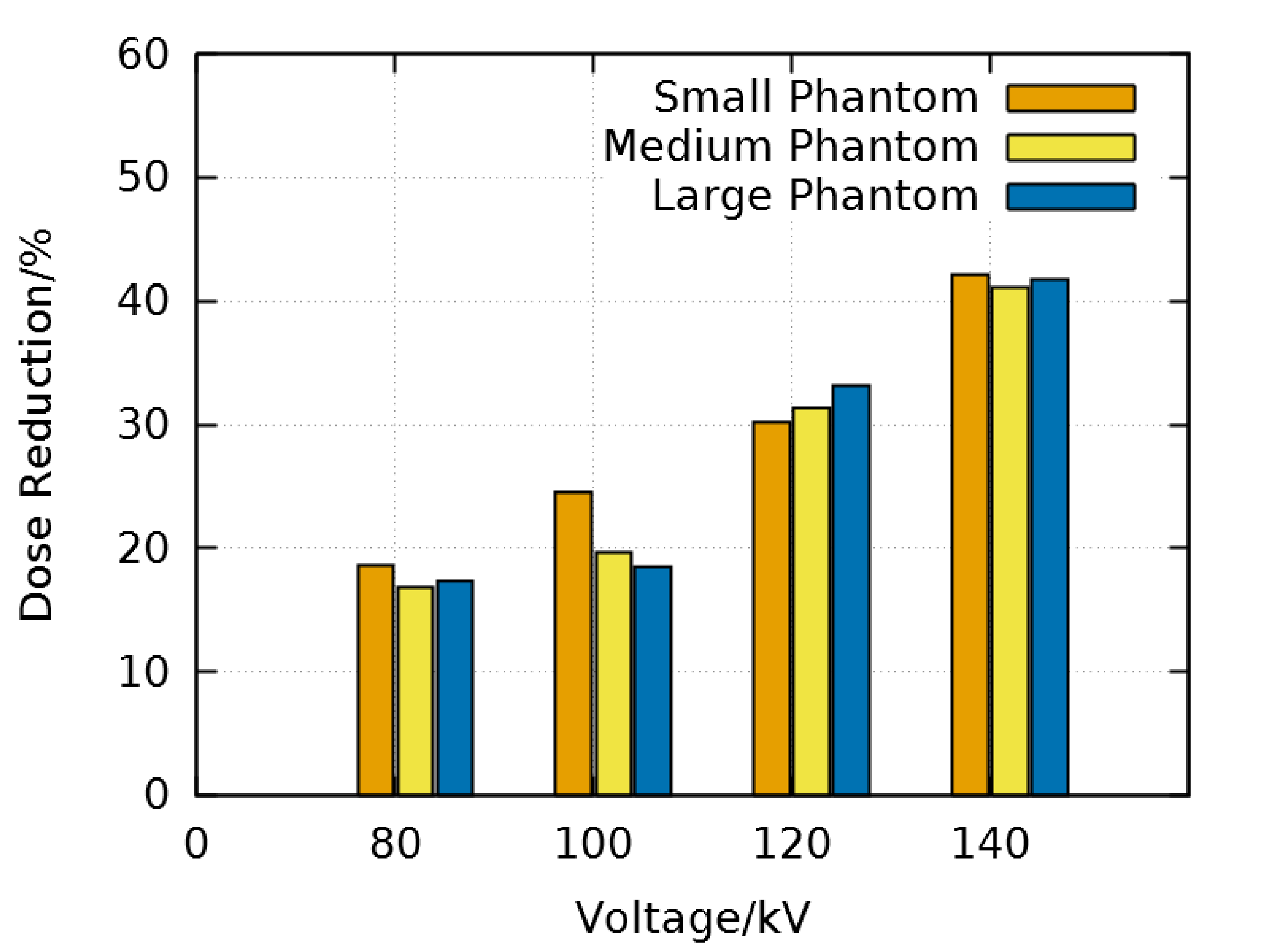
# PC with 1 Bin vs. EI

## Potential Dose Reduction



# PC with 2 Bins vs. EI

## Potential Dose Reduction



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

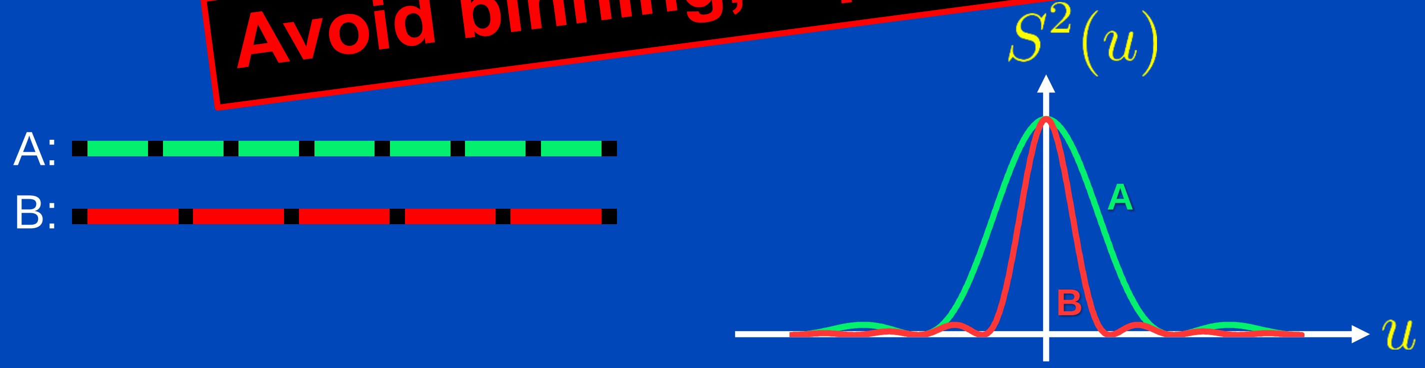
*This nice phrase was coined by Norbert Pelc.*

- **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:**

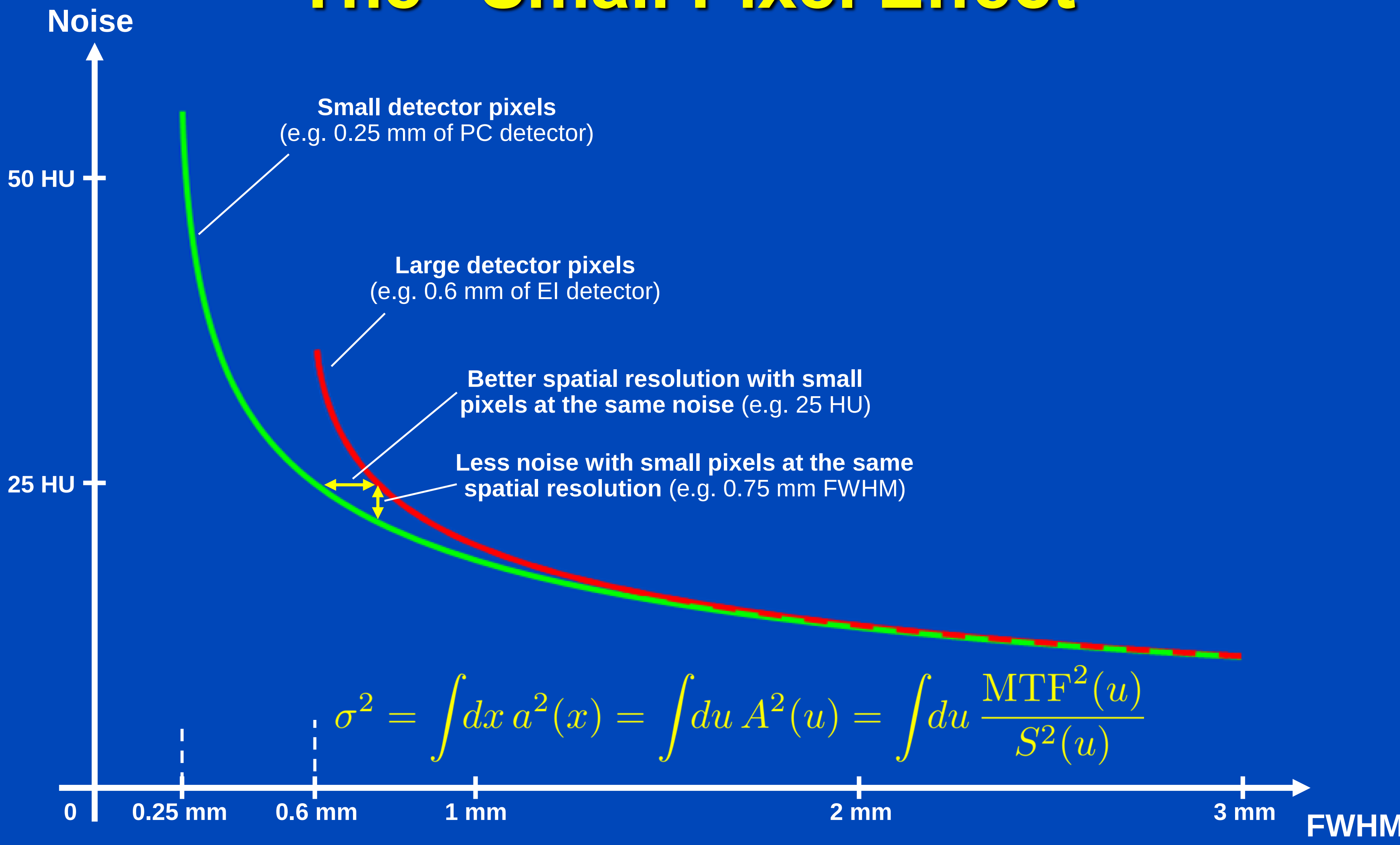
**Avoid binning, if possible!**




- **We have** $S_{\mathrm{A}}(u) > S_{\mathrm{B}}(u)$ **and thus** $\sigma_{\mathrm{A}}^2 < \sigma_{\mathrm{B}}^2$.
- **I.e. a desired PSF/MTF is often best achieved with smaller detectors. This is the “small pixel effect”.**


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

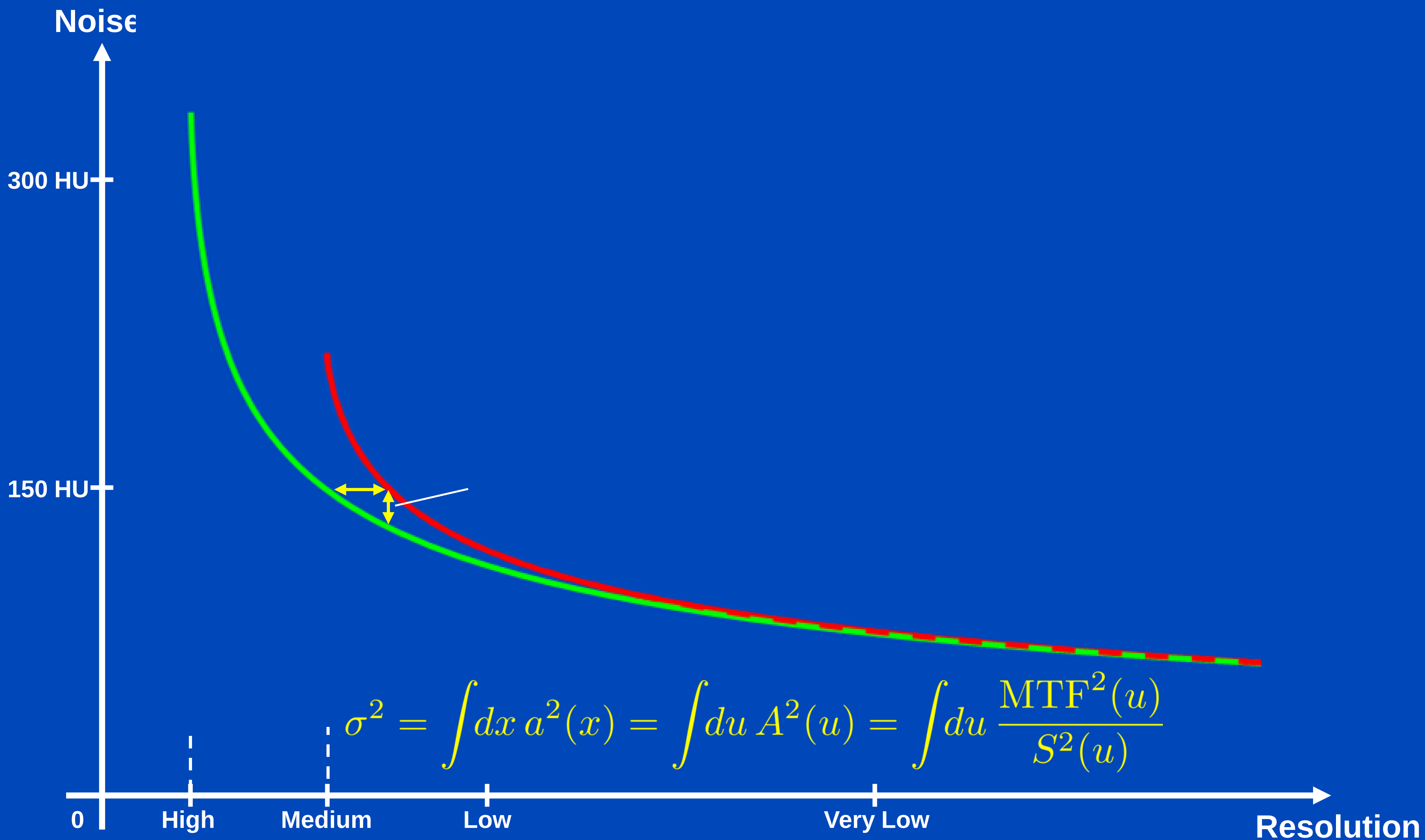
# The “Small Pixel Effect”




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

**Kachelrieß, Kalender. Med. Phys. 32(5):1321-1334, May 2005**

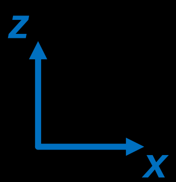
# The “Small Pixel Effect”

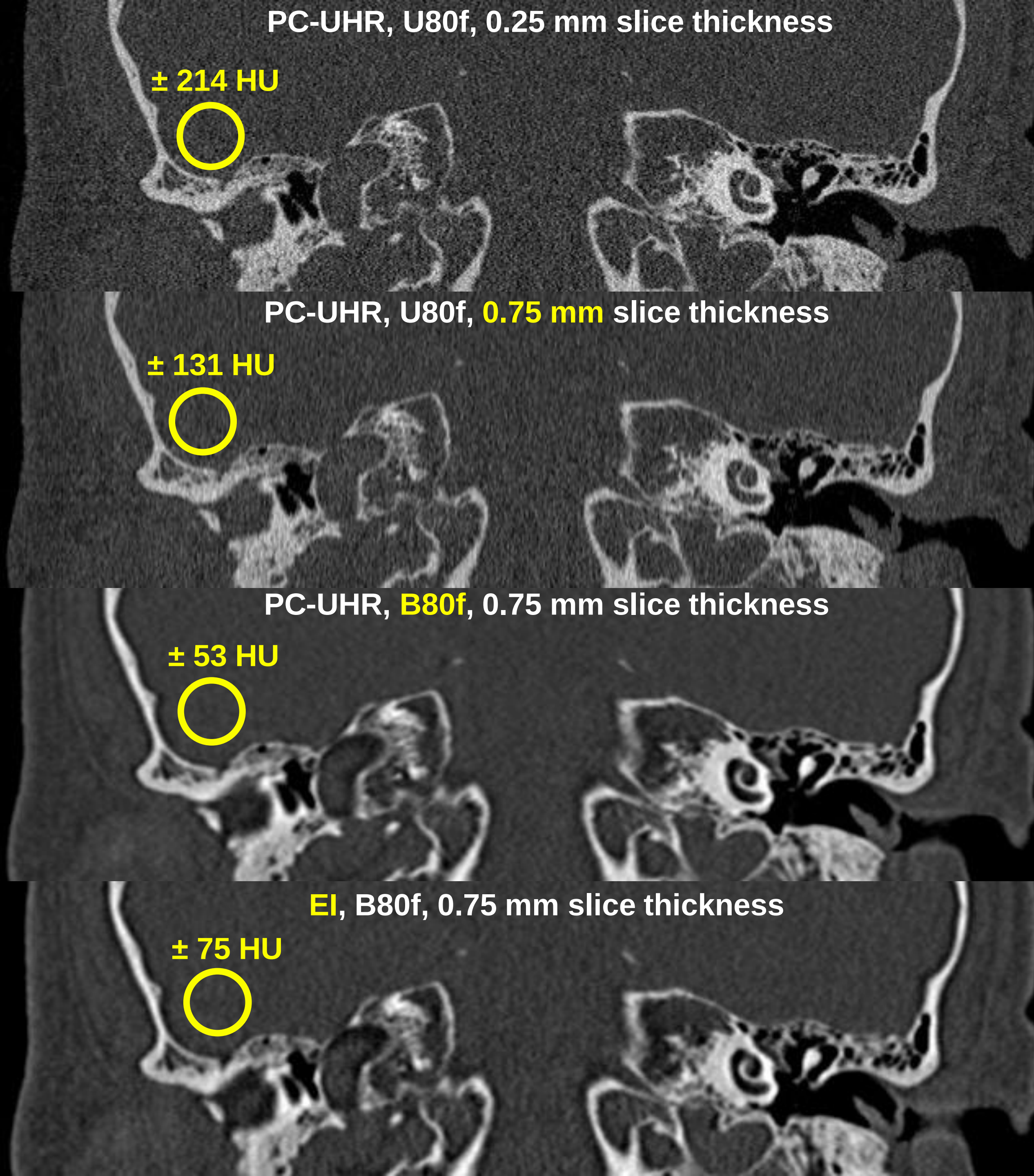



$$\sigma^2 = \int dx\, a^2(x) = \int du\, A^2(u) = \int du\, \frac{\mathrm{MTF}^2(u)}{S^2(u)}$$

Kachelrieß, Kalender. Med. Phys. 32(5):1321-1334, May 2005

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

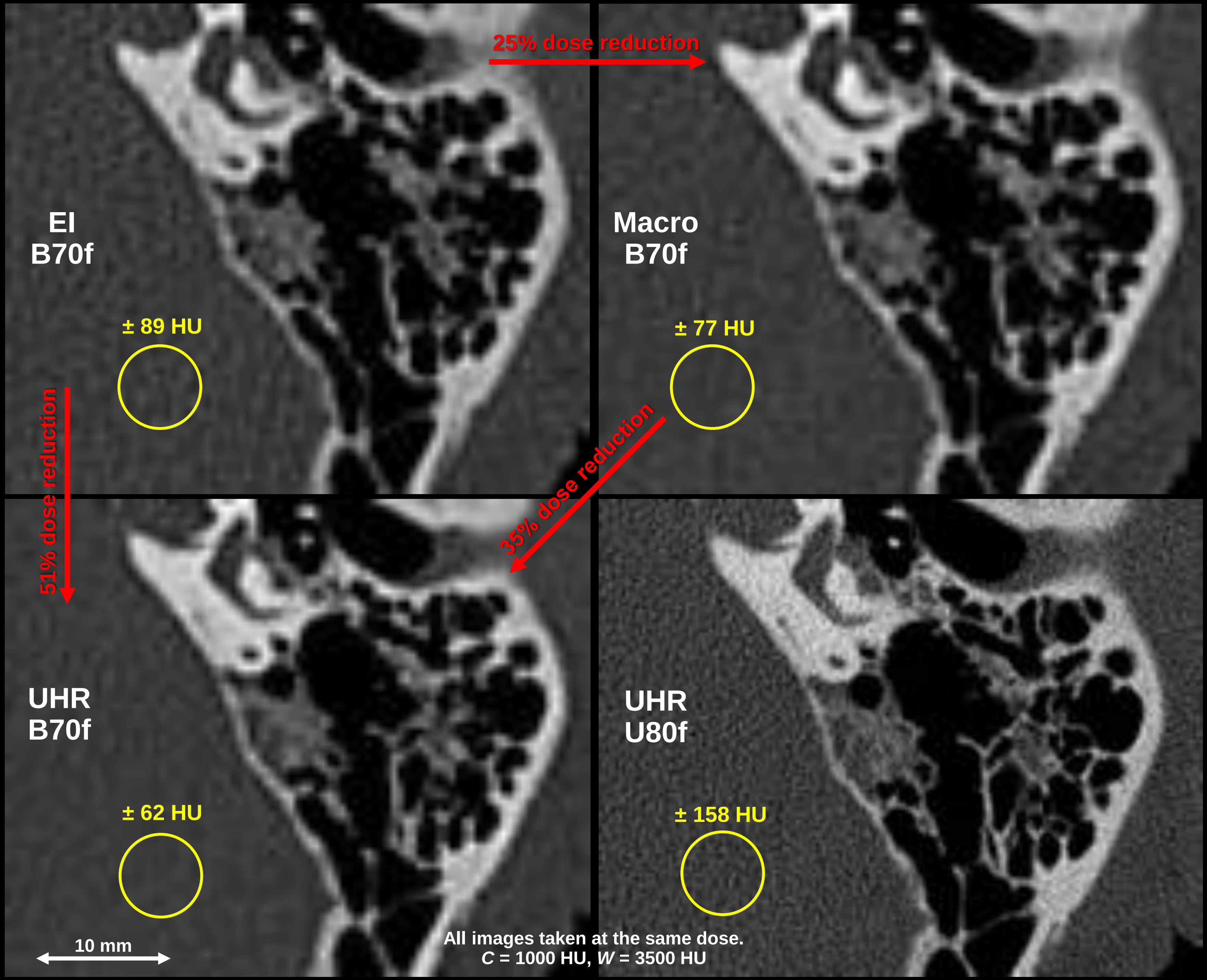
10% MTF: 19.1 lp/cm
10% MTF:17.2 lp/cm
xy FWHM: 0.48 mm
z FWHM: 0.67 mm
$CTDI_{vol}$: 16.0 mGy

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
UHR
B70f
± 62 HU
UHR
U80f
± 158 HU
10 mm
All images taken at the same dose.
C = 1000 HU, W = 3500 HU

# Acquisitions at same noise


*C* = 1000 HU
*W* = 3500 HU

**Acquisition with EI:**

- Tube voltage of 120 kV
- Tube current of 350 mAs
- Resulting dose of $CTDI_{vol\ 32\ cm}$ = **26.4 mGy**

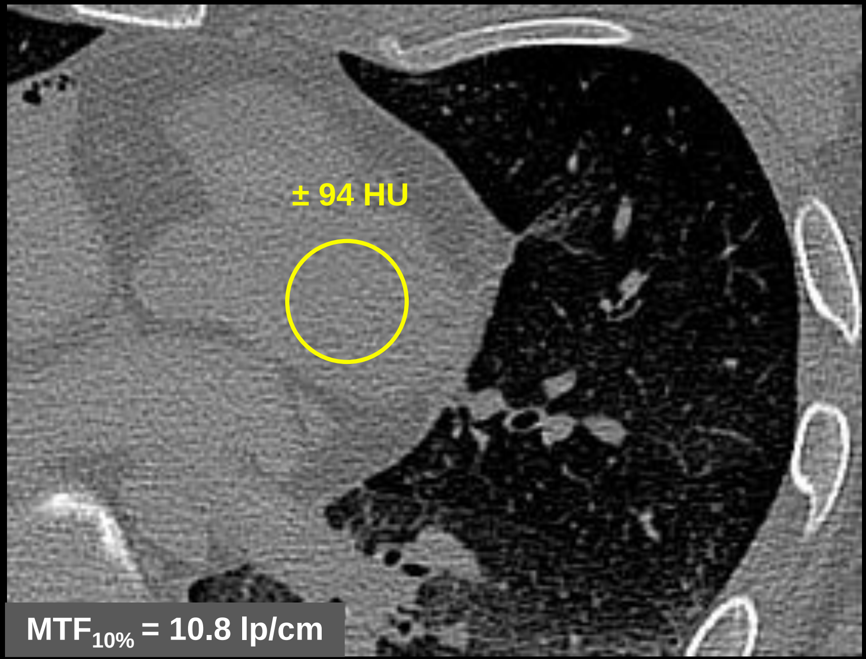
**Acquisition with PC (UHR):**

- Tube voltage of 120 kV
- Tube current of 200 mAs
- Resulting dose of $CTDI_{vol\ 32\ cm}$ = **16.1 mGy**

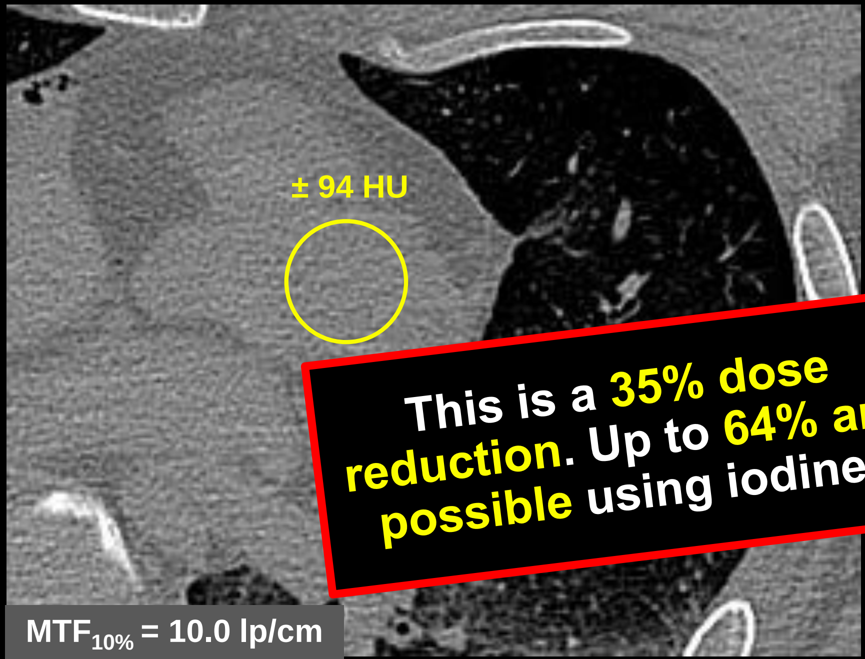
**This is a 39% reduction of dose!**

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)**




**This is a 35% dose reduction. Up to 64% are possible using iodine.**

**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**

dkfz.

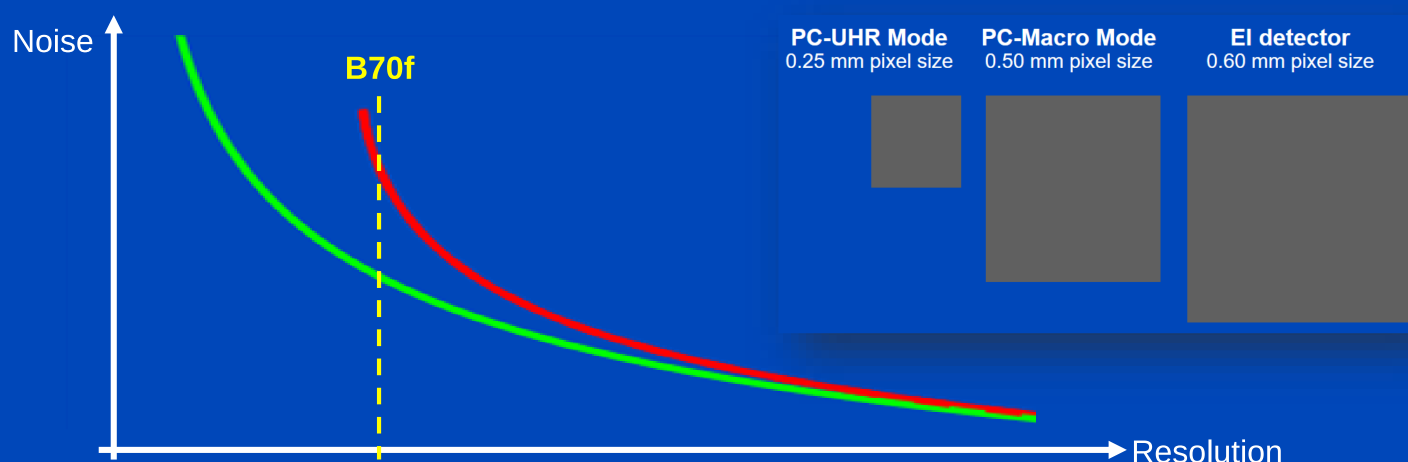
# X-Ray Dose Reduction of B70f

| 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. PC**
**(“small pixel effect only”)**

| 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% |

**PC vs. EI**
**(“small pixel effect” and “iodine effect”)**




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

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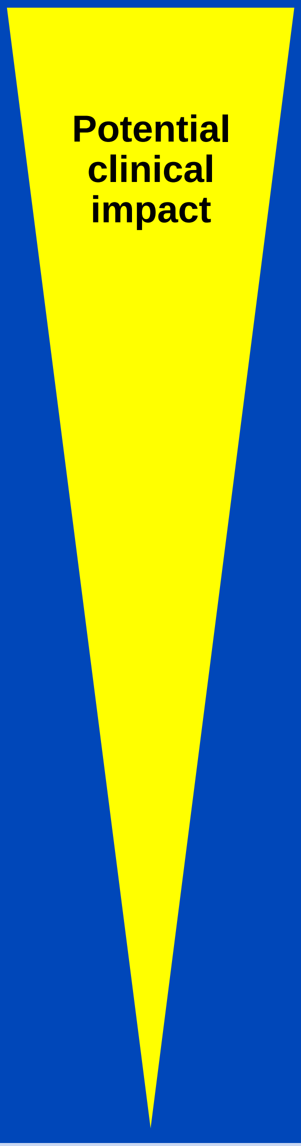
# Potential Advantages of PCCT

- **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
  - …



# Thank You!

**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.**