Detektorabtastung und Dosisreduktion in der photonenzählenden Computertomographie – Eine Phantom- und Ex-Vivo-Studie

Master's thesis

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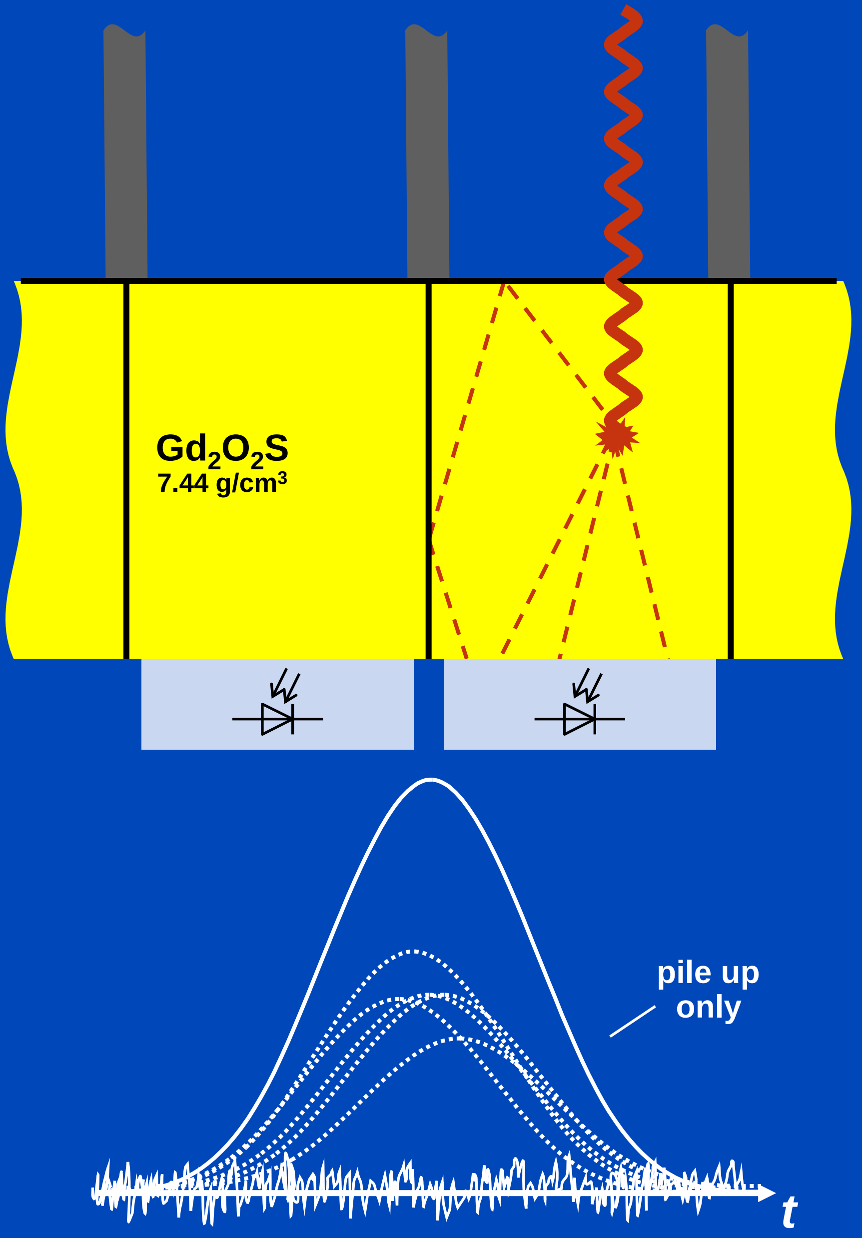
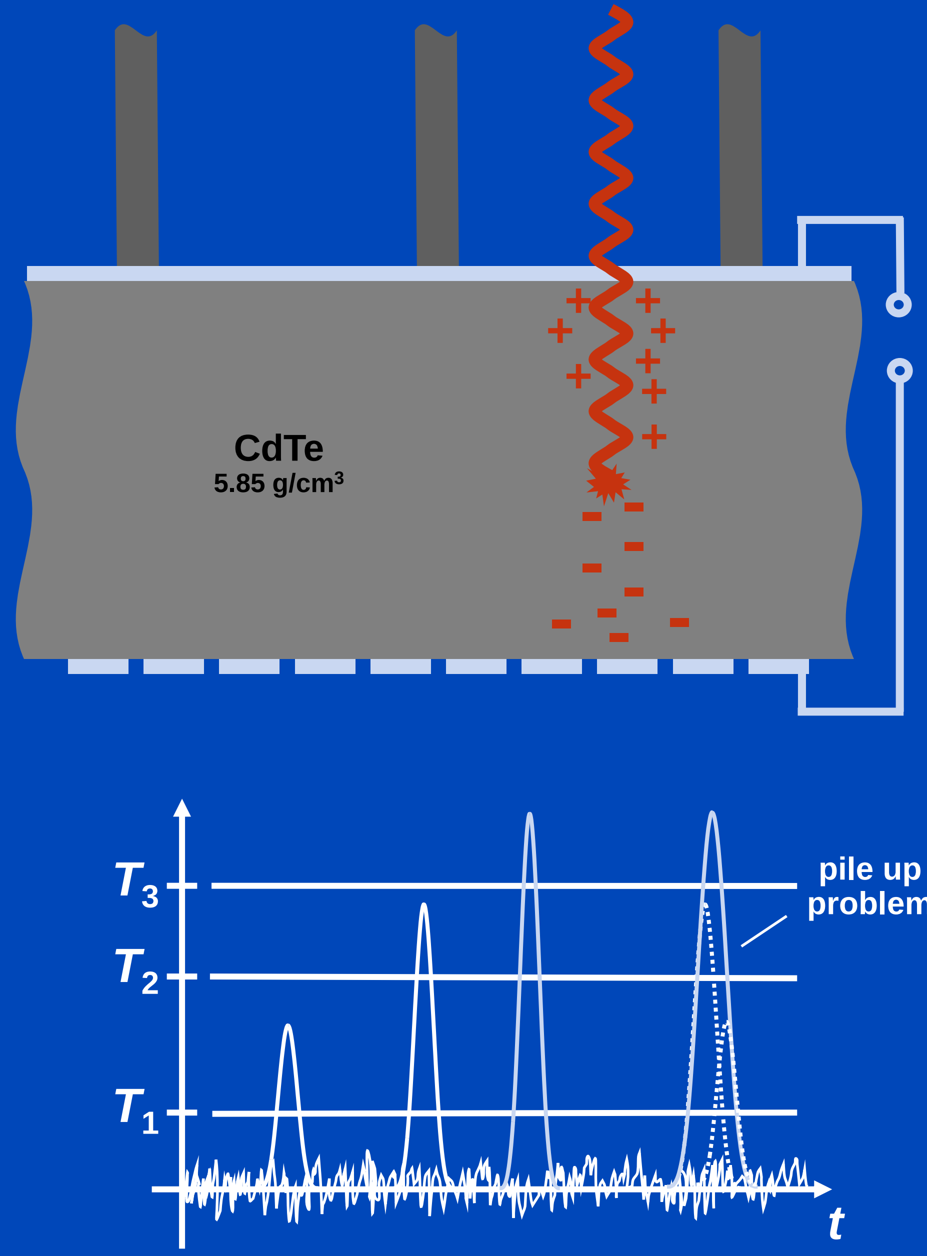
dkfz. DEUTSCHES KREBSFORSCHUNGSZENTRUM IN DER HELMHOLTZ-GEMEINSCHAFT

Indirect Conversion (Today)



Direct Conversion (Future)

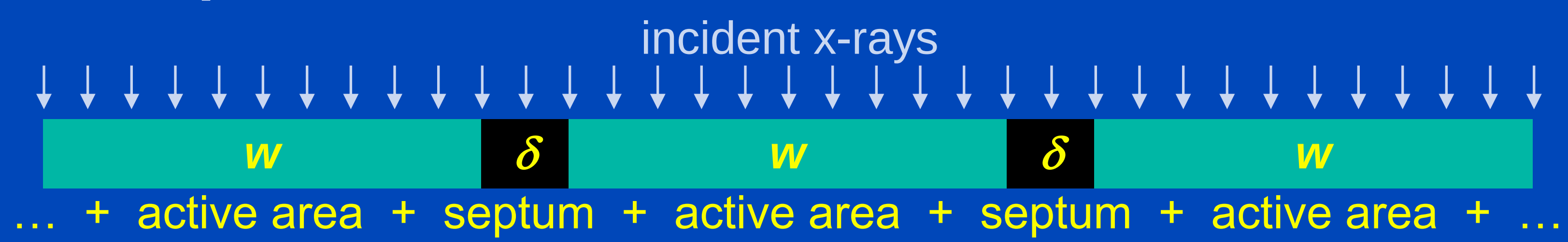


System Model

- **True projection *f*(*x*)**
- **Presampling function *s*(*x*), normalized to unit area**
- **Algorithm *a*(*x*), normalized to unit area**
- **Observed projection *g*(*x*) with**

$$g(x) = f(x) * s(x) * a(x) = f(x) * \mathrm{PSF}(x)$$

- **Example:**



$$s(x) = \mathrm{II}_d^*(x)$$

w = detector pixel width
δ = dead space between pixels

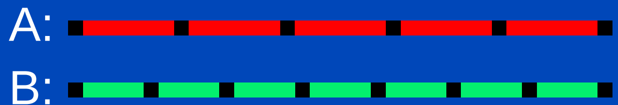
Kachelrieß, Kalender. Med. Phys. 32(5):1321-1334, May 2005

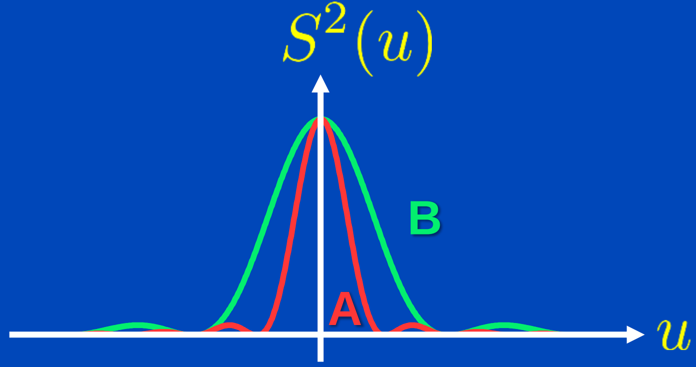
To Bin or not to Bin?

- **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 large (A) with small (B) detector pixels:**

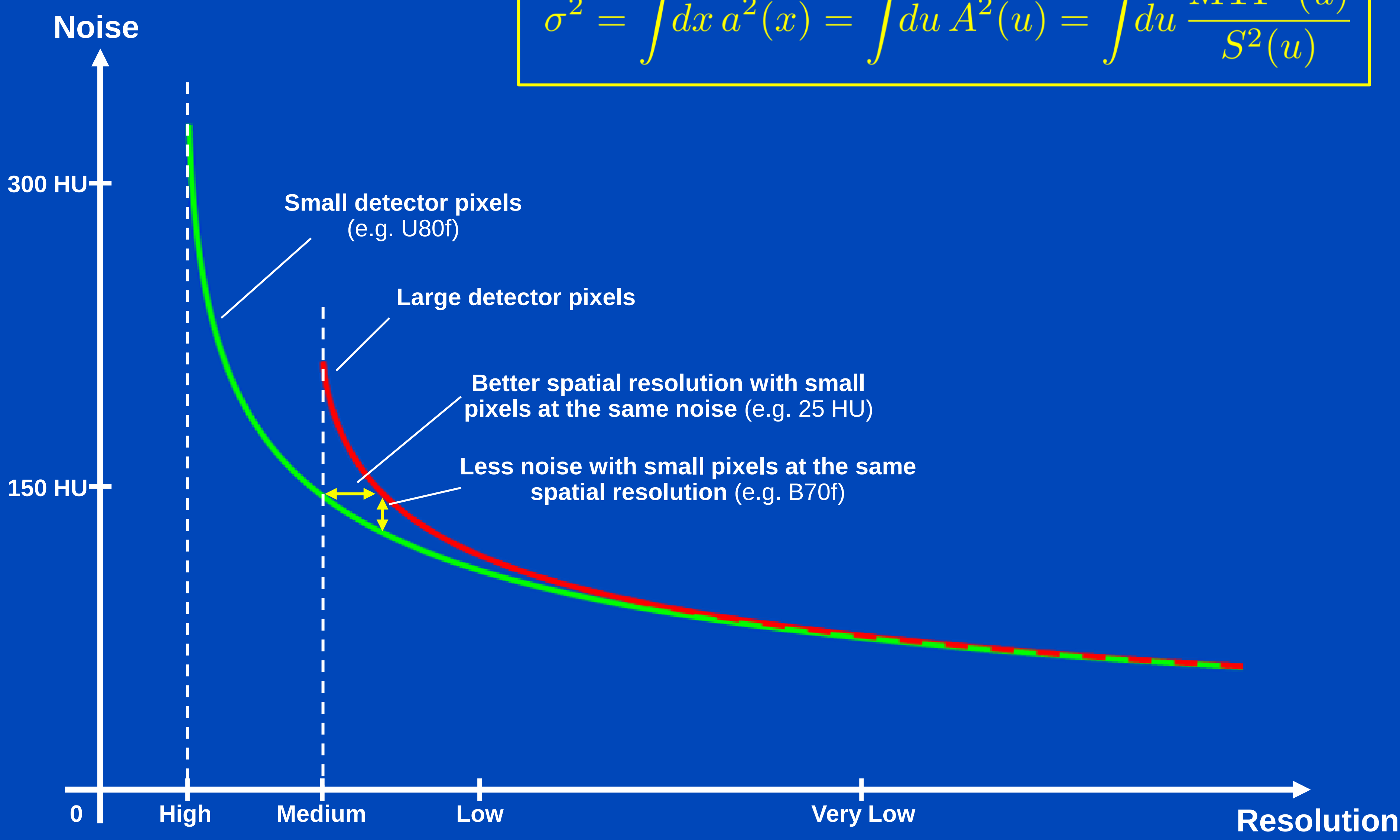




- **We have** $S_{\mathrm{B}}(u) > S_{\mathrm{A}}(u)$ **and thus** $\sigma_{\mathrm{B}}^2 < \sigma_{\mathrm{A}}^2$**.**
- **This means that a desired PSF/MTF is often best achieved with smaller detectors.**

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

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

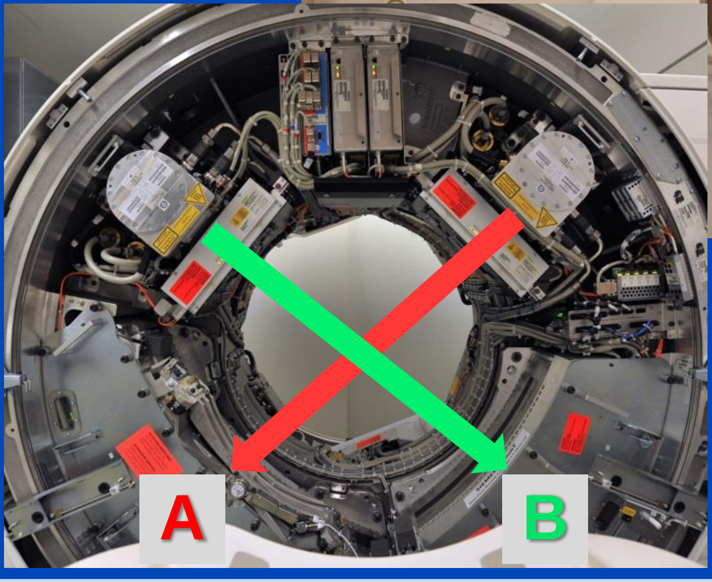


CounT CT System at the DKFZ

Gantry from a clinical dual source scanner

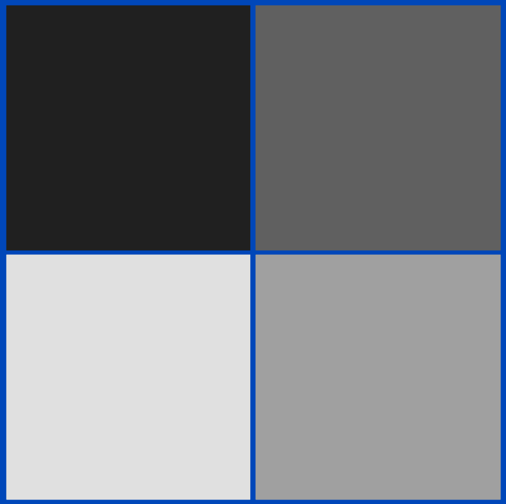
A: conventional CT detector (50.0 cm FOV)

B: Photon counting detector (27.5 cm FOV)

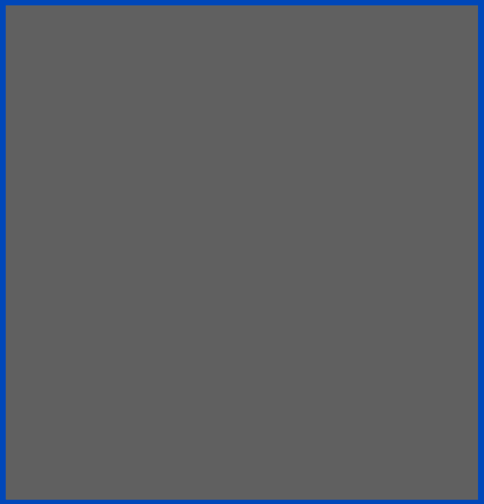


Readout Modes of the CounT

PC-UHR Mode
0.25 mm pixel size



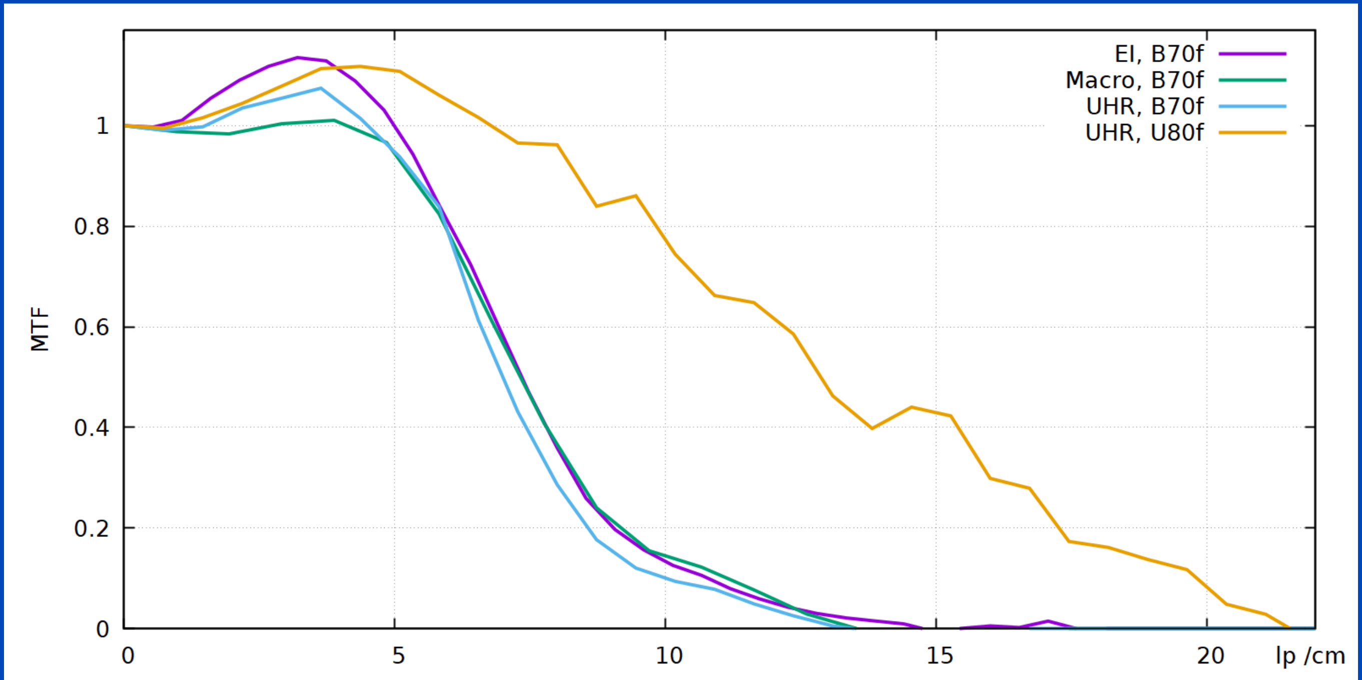
PC-Macro Mode
0.50 mm pixel size



EI detector
0.60 mm pixel size

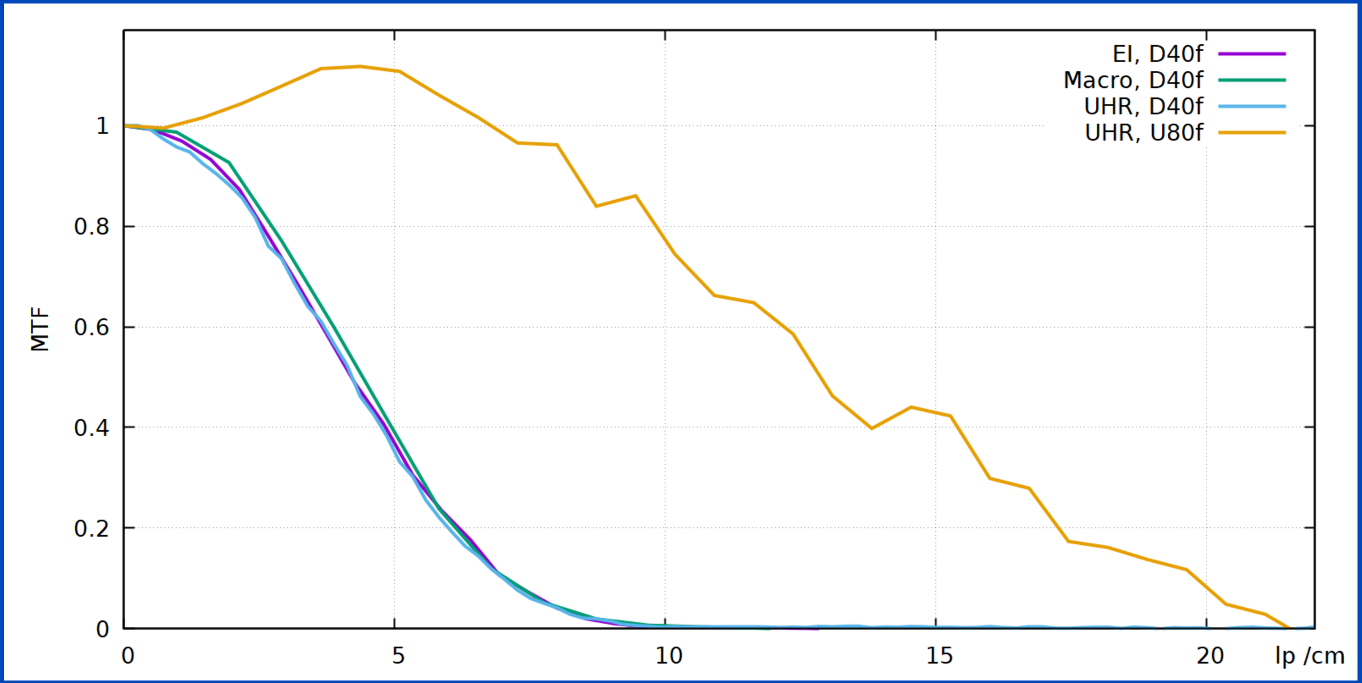
Reconstruction

	Pixel size	Kernel	$MTF_{10\%}$
EI	0.60 mm	B70f	10.8 lp/cm
Macro	0.50 mm	B70f	11.1 lp/cm
UHR	0.25 mm	B70f	10.0 lp/cm
UHR-U80f	0.25 mm	U80f	19.8 lp/cm



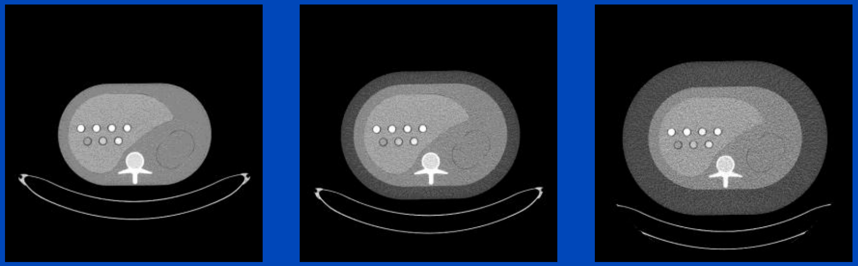
Reconstruction

	Pixel size	Kernel	$MTF_{10\%}$
EI	0.60 mm	D40f	7.0 lp/cm
Macro	0.50 mm	D40f	7.1 lp/cm
UHR	0.25 mm	D40f	7.0 lp/cm
UHR-U80f	0.25 mm	U80f	19.8 lp/cm



Materials and Methods

- Abdomen phantoms of three different sizes (S, M, L) with iodine inserts of different concentrations
  - Small: 20 cm × 30 cm
  - Medium: 25 cm × 35 cm
  - Large: 30 cm × 40 cm



- Animal and human cadavers
- Tube voltages: 80 kV, 100 kV, 120 kV, and 140 kV
- Effective tube current of 200 mAs
- Collimation:
  - UHR: Acq. 64 × 0.25 mm
  - Macro: Acq. 32 × 0.50 mm
  - EID: Acq. 32 × 0.60 mm

C = 20 HU, *W* = 600 HU

Contrast-to-Noise Ratio (CNR)

- By selecting two ROIs, the CNR can be calculated using

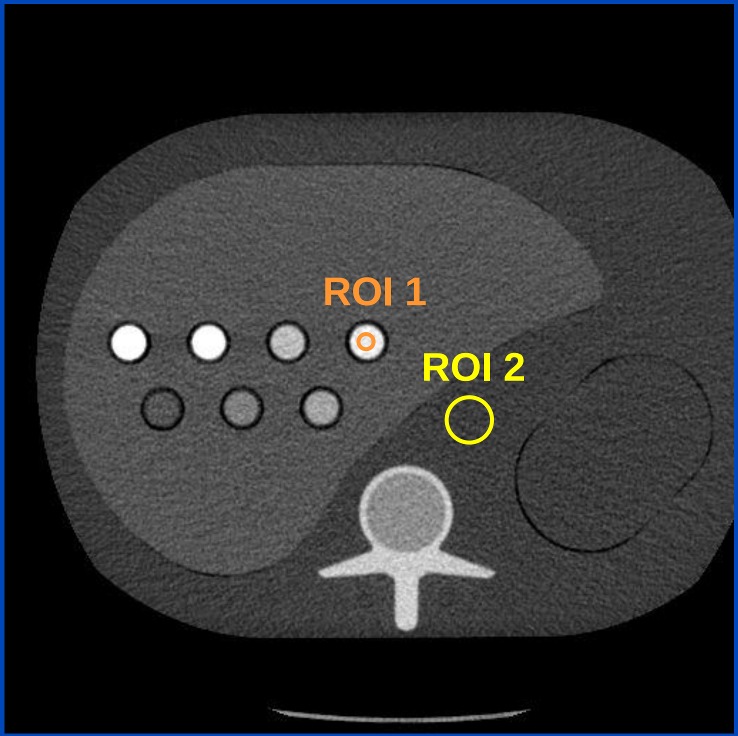
$$\mathrm{CNR} = \frac{|\mu_1 - \mu_2|}{\sqrt{\sigma_1^2 + \sigma_2^2}}$$

- Normalization to dose D:

$$\mathrm{CNRD} = \frac{\mathrm{CNR}}{\sqrt{\mathrm{D}}}$$

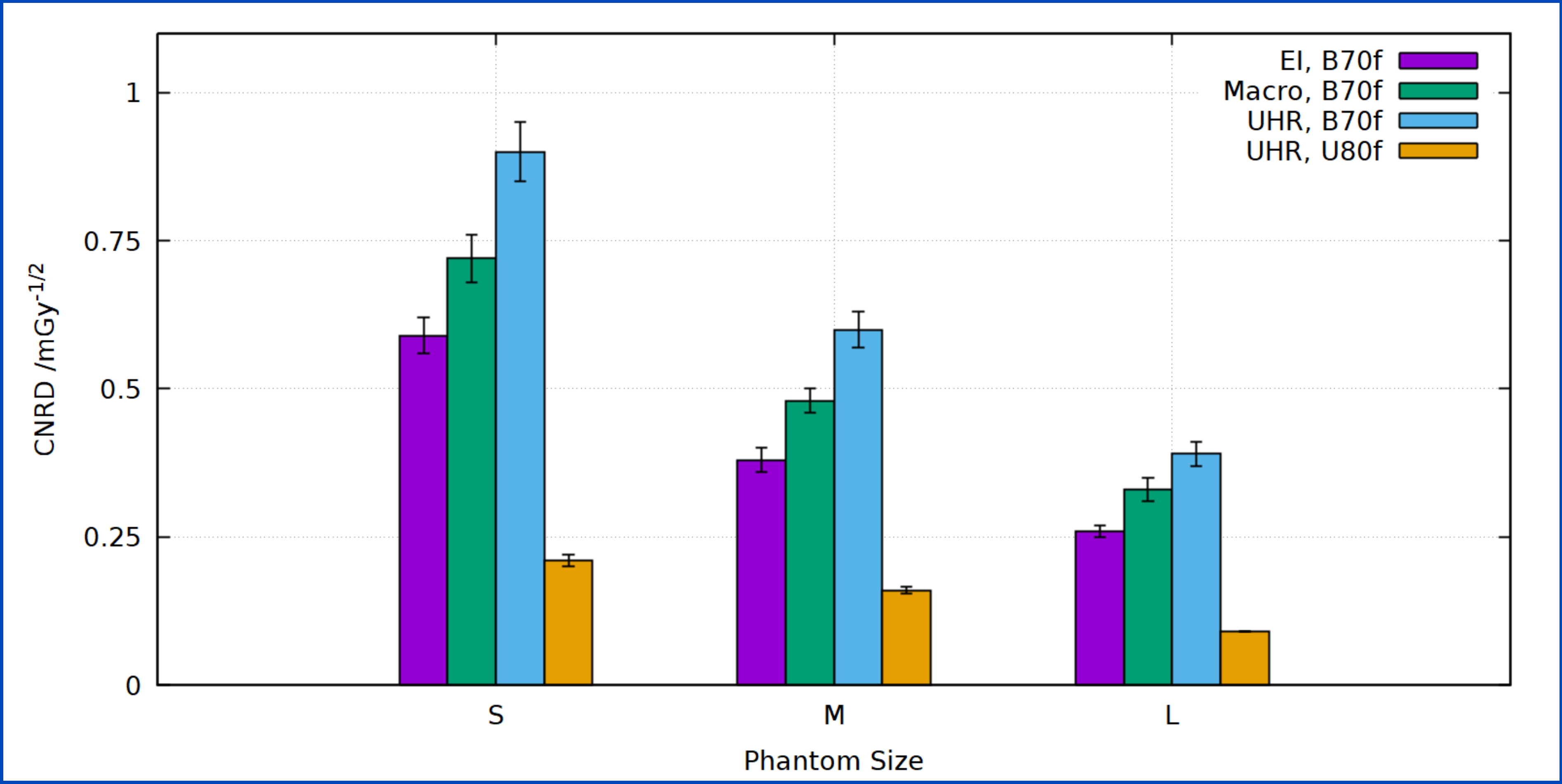
- The potential x-ray dose reduction can be calculated by

$$\text{Dose Reduction} = 1 - \frac{\mathrm{CNRD}^2_{\mathrm{Ref}}}{\mathrm{CNRD}^2_{\mathrm{PC}}}$$



The iodine concentration in ROI 1 is 25 mg/mL. The CT value is about 520 HU at 120 kV.

Results at 120 kV

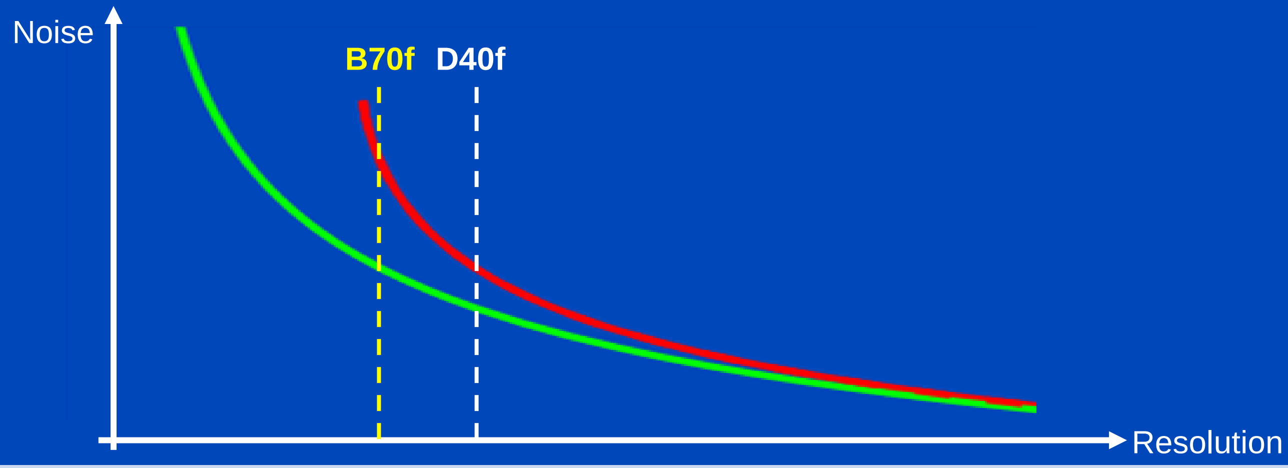


Error bars indicate the errors when analyzing 15 different slices of the same contrast.

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%

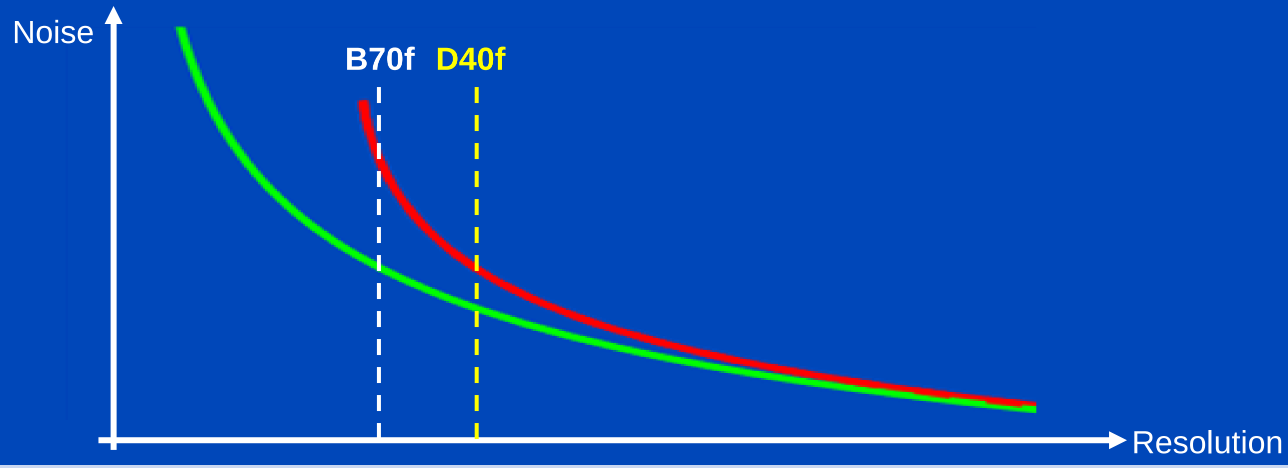
UHR vs. EI	80 kV	100 kV	120 kV	140 kV
S	33% ± 9%	52% ± 5%	57% ± 7%	57% ± 6%
M	41% ± 8%	47% ± 7%	60% ± 6%	62% ± 4%
L	48% ± 8%	43% ± 10%	54% ± 6%	63% ± 5%

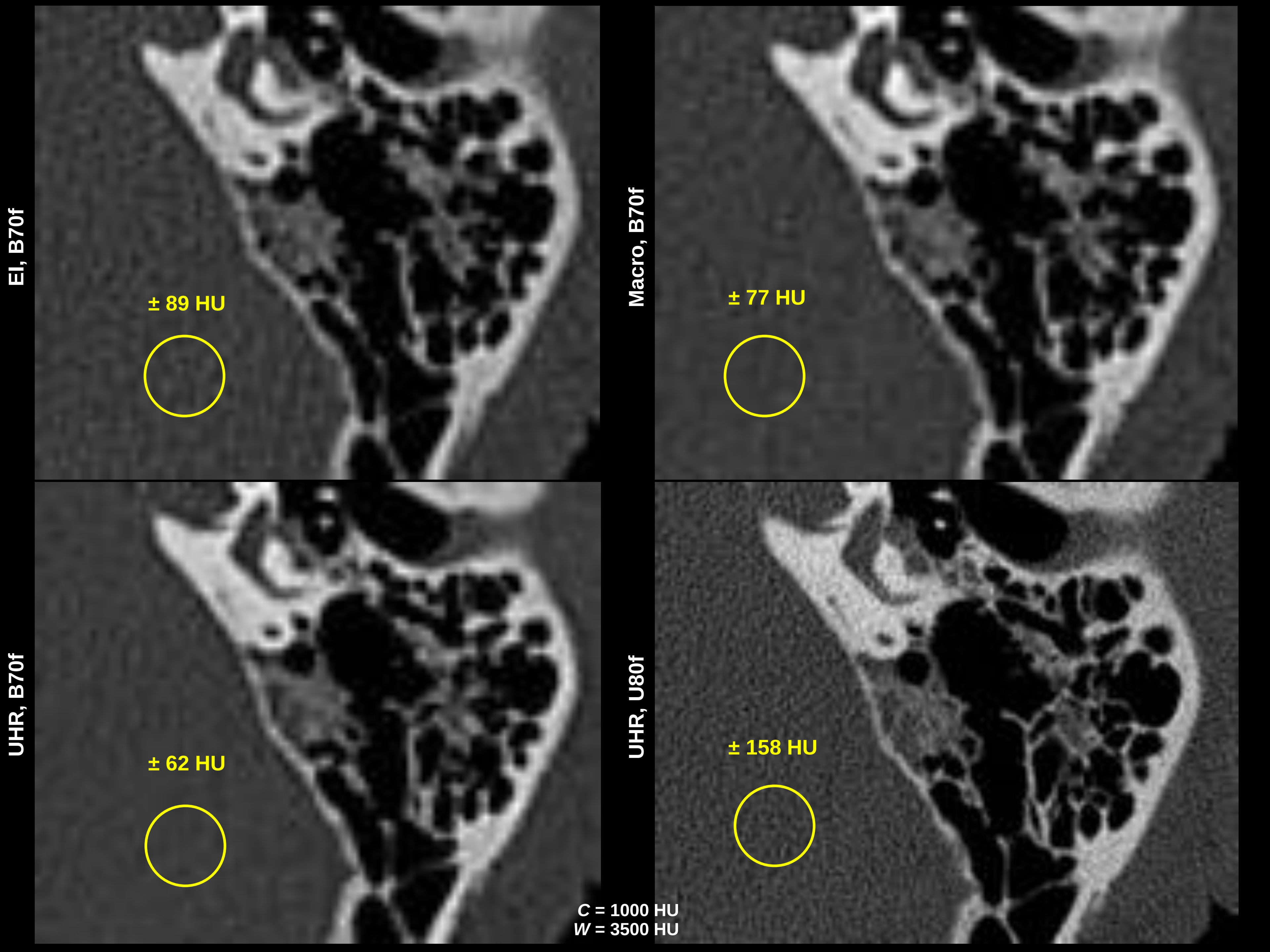


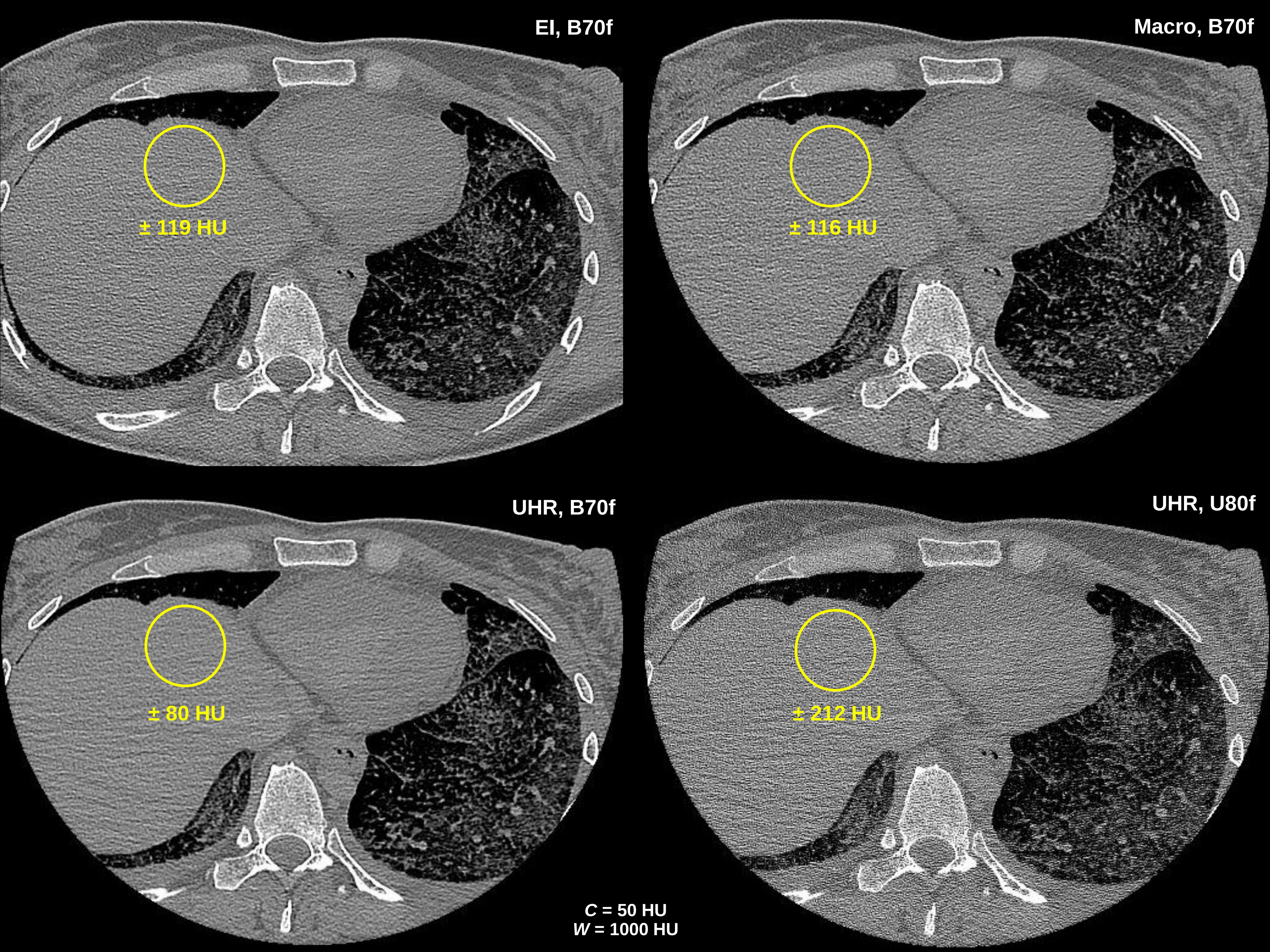
X-Ray Dose Reduction of D40f

UHR vs. Macro	80 kV	100 kV	120 kV	140 kV
S	5% ± 16%	12% ± 17%	17% ± 17%	9% ± 15%
M	11% ± 14%	9% ± 12%	16% ± 16%	13% ± 13%
L	11% ± 14%	6% ± 17%	6% ± 17%	4% ± 17%

UHR vs. EI	80 kV	100 kV	120 kV	140 kV
S	10% ± 11%	28% ± 11%	36% ± 12%	38% ± 12%
M	15% ± 12%	23% ± 12%	40% ± 10%	43% ± 9%
L	24% ± 14%	17% ± 11%	33% ± 12%	43% ± 9%




EI, B70f
± 89 HU
Macro, B70f
± 77 HU
UHR, B70f
± 62 HU
UHR, U80f
± 158 HU
C = 1000 HU
W = 3500 HU


EI, B70f
± 119 HU
Macro, B70f
± 116 HU
UHR, B70f
± 80 HU
UHR, U80f
± 212 HU
C = 50 HU
W = 1000 HU

Conclusions

- This is the first systematic study[1] quantifying the effects of detector sampling on noise reduction in a clinical whole-body photon counting CT.
- The results illustrate that it is favorable to measure with smaller pixels even if the high spatial resolution is not of interest.
- A significant clinical dose reduction can be achieved, depending on the chosen resolution.

[1]Further evaluation and the discussion of other studies can be found in Klein et al. Invest. Radiol. 55(2), Feb 2020, in press.

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Thank You!

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