Kernel Considerations for High Resolution Photon-Counting CT: Dose Reduction vs. Spatial Resolution

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Aim

To evaluate the dependency of noise reduction in high resolution whole-body photon-counting (PC) CT compared to conventional energy-integrating (EI) CT as a function of the reconstruction kernel.

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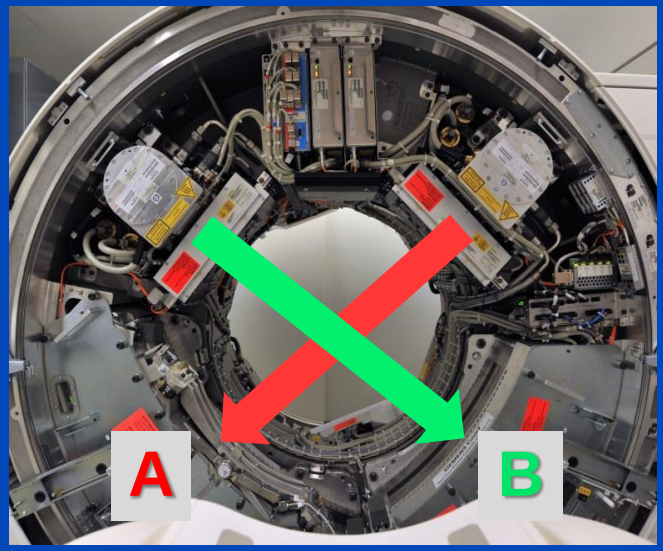
CounT CT System at the DKFZ

Gantry from a clinical dual source scanner

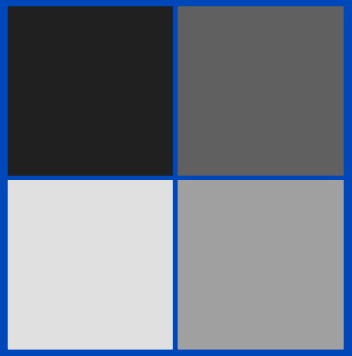
A: conventional CT detector (50.0 cm FOV)

B: Photon counting detector (27.5 cm FOV)

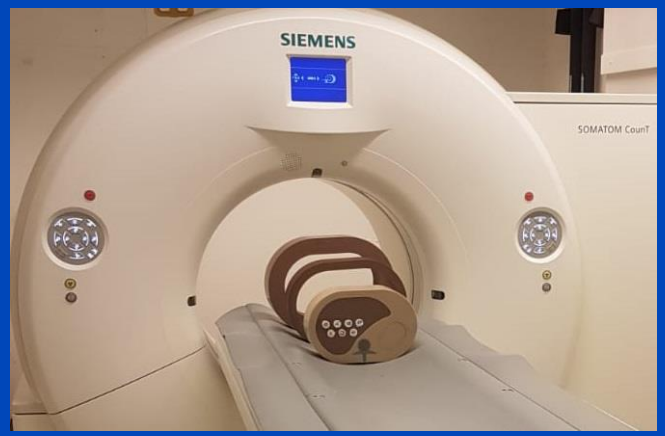


Readout Modes of the CounT

PC-UHR Mode
0.25 mm pixel size
20 lp/cm



EI detector
0.60 mm pixel size
9 lp/cm



Experimental CT, not commercially available.

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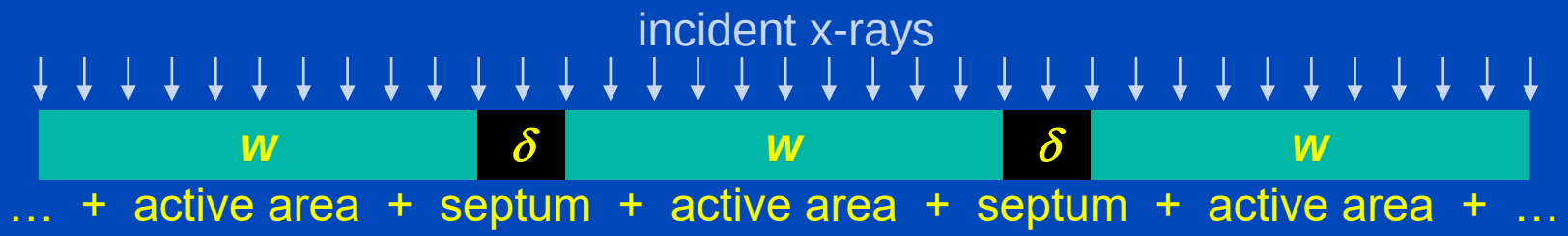
System Model

- **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

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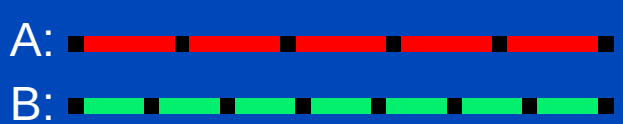
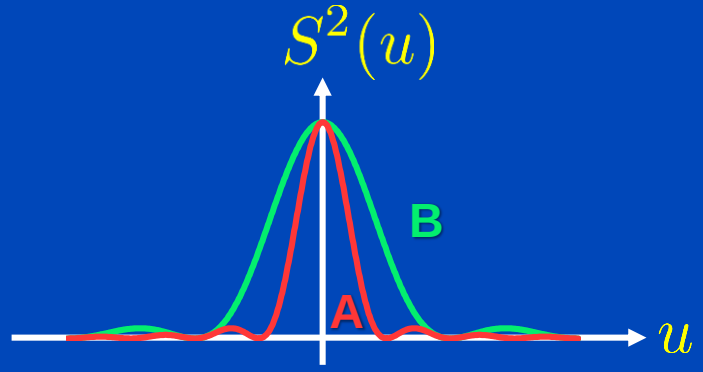
To Bin or not to Bin?

- 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

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

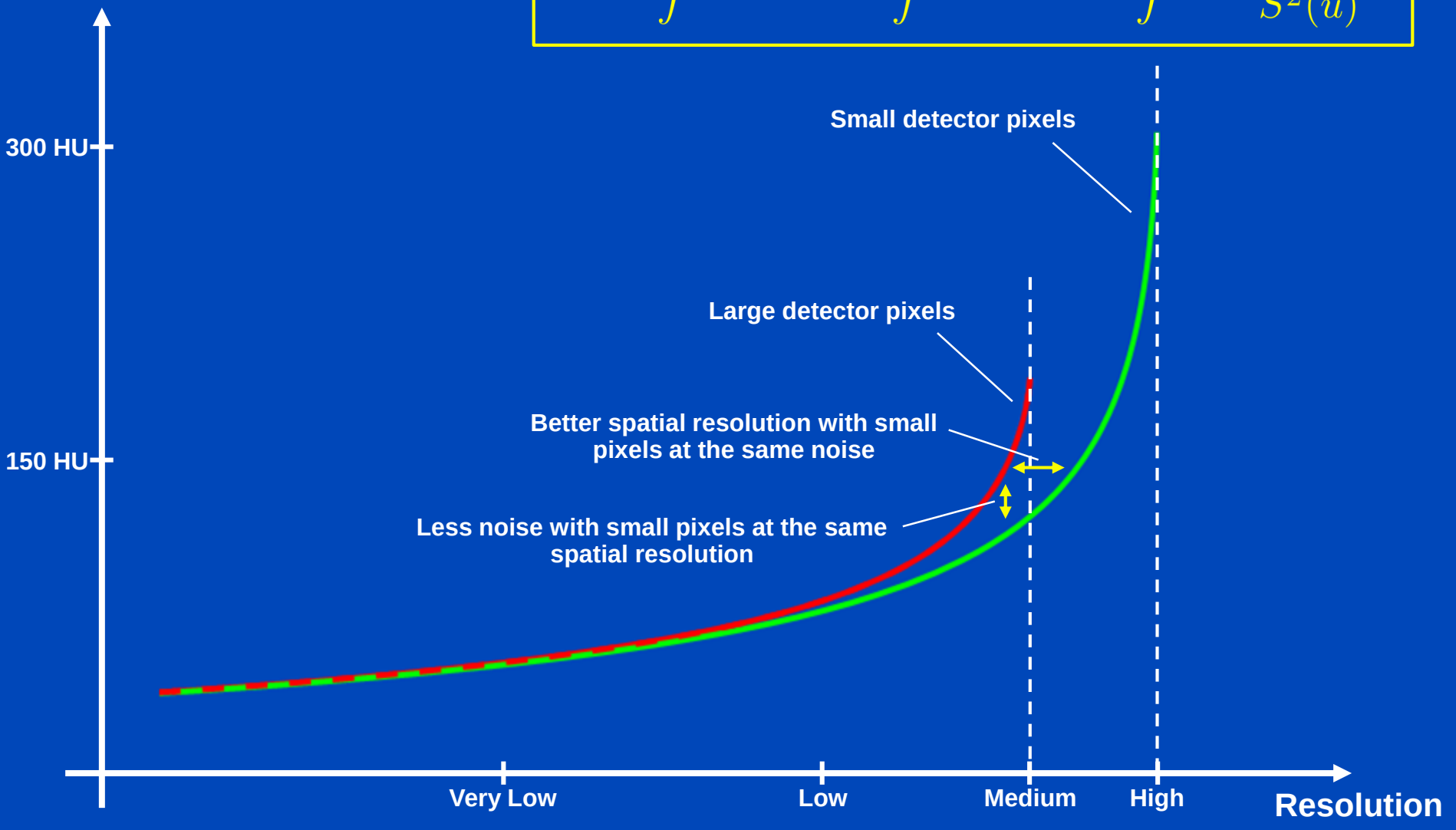
- Compare large (A) with small (B) detector pixels:





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

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



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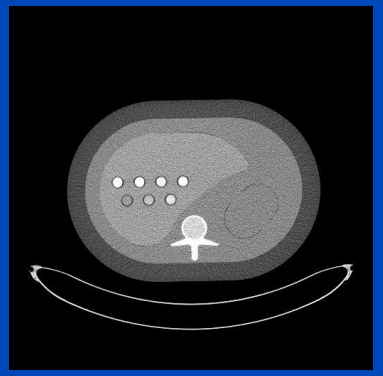
Materials and Methods

Measurements

- Abdomen phantom with iodine inserts of different concentrations
  - Tube current of 200 mAs
  - Tube voltage of 120 kV
- Cadaver measurements
  - Tube current of 300 mAs
  - Tube voltage of 120 kV
- With
  - EI detector, 0.60 mm pixel size
  - PC detector, 0.25 mm pixel size

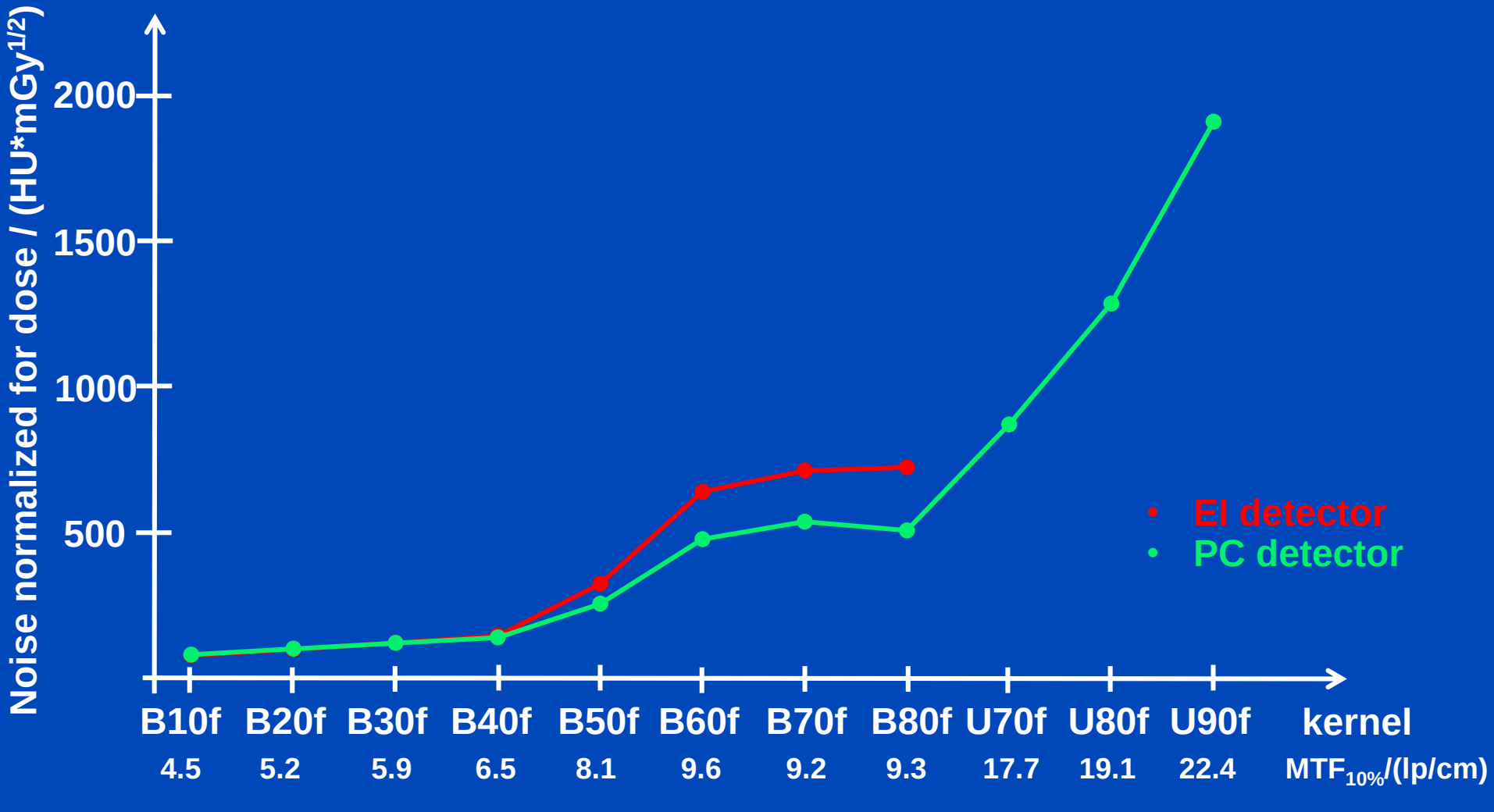


Reconstructions

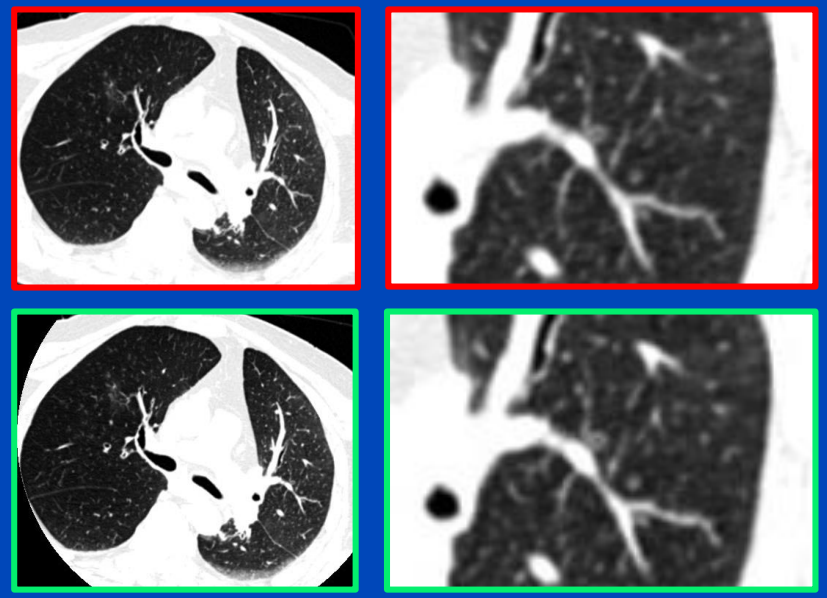
- For different resolution levels by using a broad range of reconstruction kernels: B10f – B80f, U70f, U80f, U90f

C = 20 HU, *W* = 600 HU

Noise and Spatial Resolution



Noise and Spatial Resolution



Noise normalized for dose / (HU*mGy$^{1/2}$)

2000
1500
1000
500

EI detector
PC detector

B10f	B20f	B30f	B40f	B50f	B60f	B70f	B80f	U70f	U80f	U90f	kernel
4.5	5.2	5.9	6.5	8.1	9.6	9.2	9.3	17.7	19.1	22.4	MTF$_{10\%}$/(lp/cm)

C = -500 HU, *W* = 1000 HU

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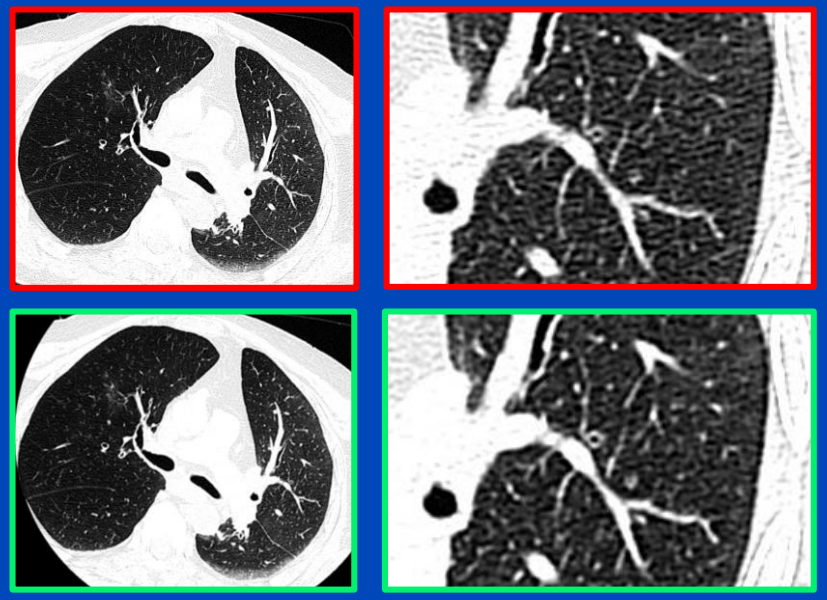
Noise and Spatial Resolution



Noise normalized for dose / (HU*mGy$^{1/2}$)

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• EI detector
• PC detector

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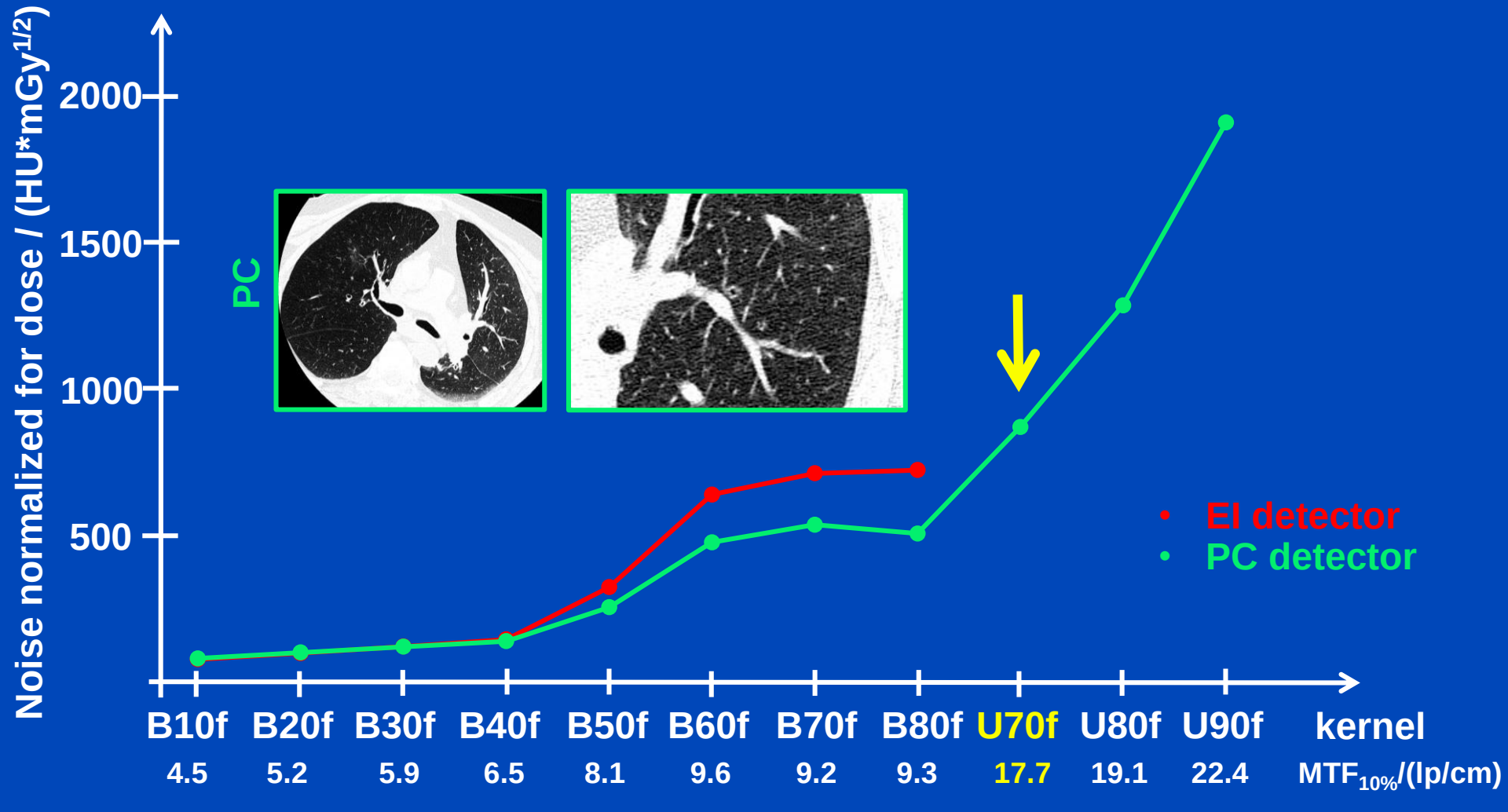
Noise and Spatial Resolution



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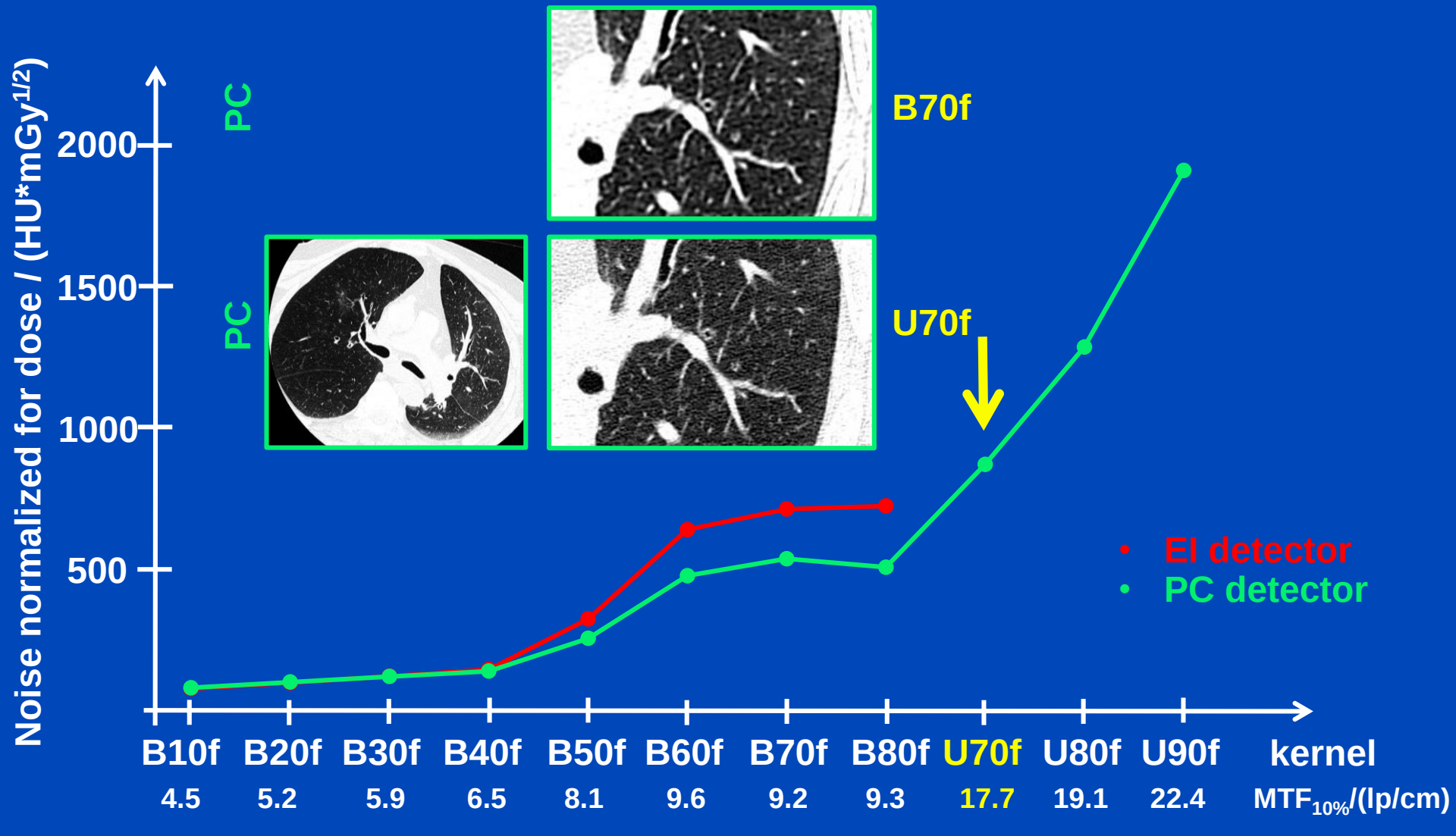
Noise and Spatial Resolution



C = -500 HU, *W* = 1000 HU

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Conclusions

- Clinical applications requiring medium or high resolution, e.g. visualization of small nodules and fungal hyphae, will improve by the usage of PC-UHR and the consequent noise reduction of up to 30%.
- PC-UHR will not improve situations requiring low resolution, e.g. mediastinal and hilary lymph nodes. However, PC detectors also provide higher iodine contrast.
- Additionally, PC-UHR provides a significantly higher spatial resolution compared to EI detectors.

Thank You!

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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.