Scan Protocol Design of k-Edge Imaging in a Whole-Body Photon-Counting CT

Stefan Sawall[1,2], Laura Klein[1,2], Sabrina Dorn[3], Carlo Amato[1,2], Joscha Maier[1,2], Sebastian Faby[3], Monika Uhrig[1,2], Heinz-Peter Schlemmer[1,2], and Marc Kachelrieß[1,2]

[1]German Cancer Research Center (DKFZ), Heidelberg, Germany

[2]Ruprecht-Karls-University of Heidelberg, Germany

[3]Siemens Healthineers, Forchheim, Germany

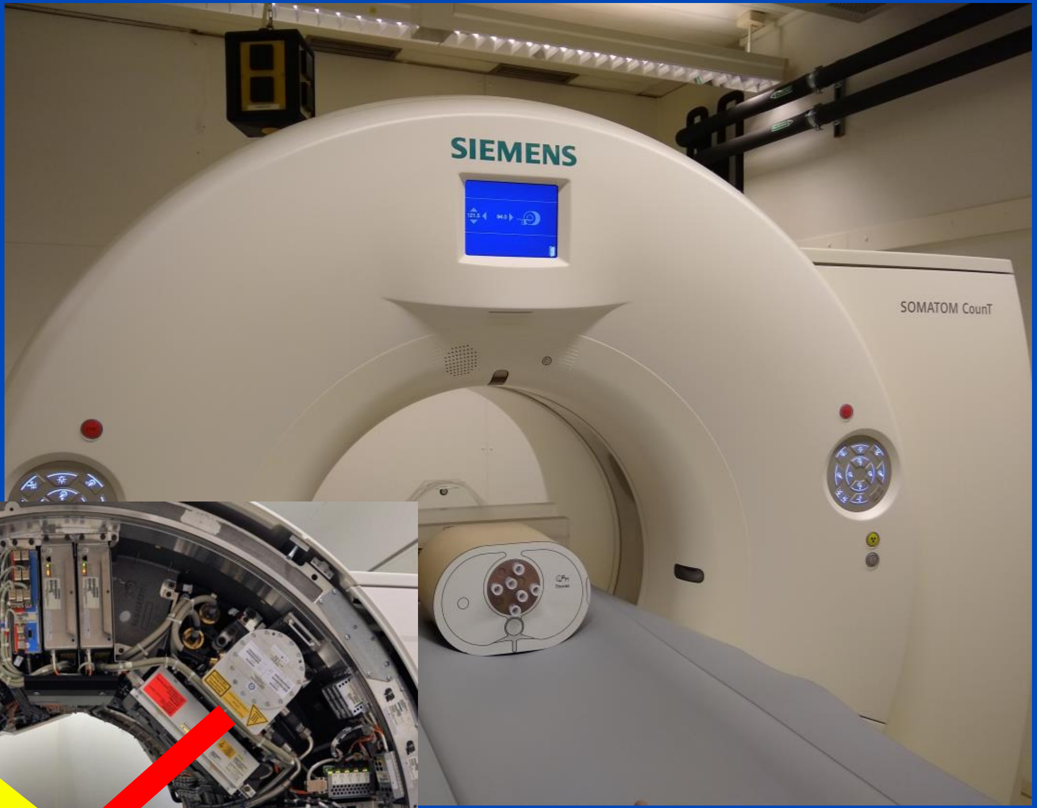
DEUTSCHES KREBSFORSCHUNGSZENTRUM IN DER HELMHOLTZ-GEMEINSCHAFT

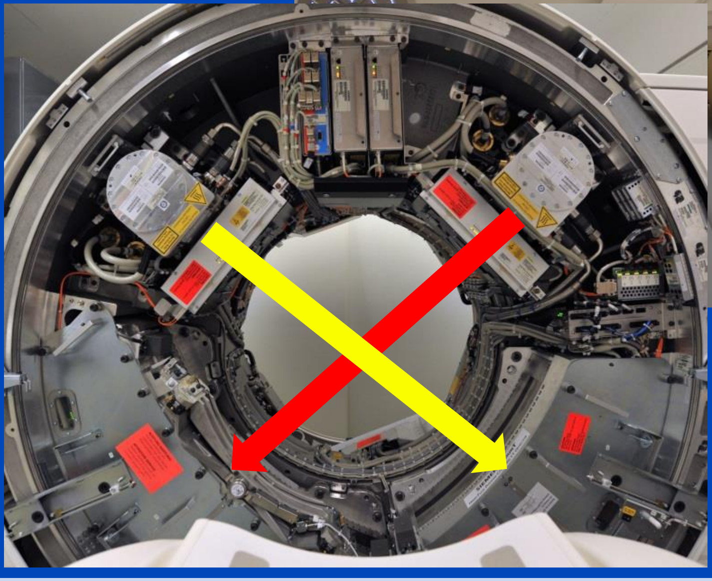
SOMATOM CounT CT @ DKFZ

Gantry from a clinical dual source scanner

A: conventional CT detector (50 cm FOV)

B: Photon counting detector (27.5 cm FOV)

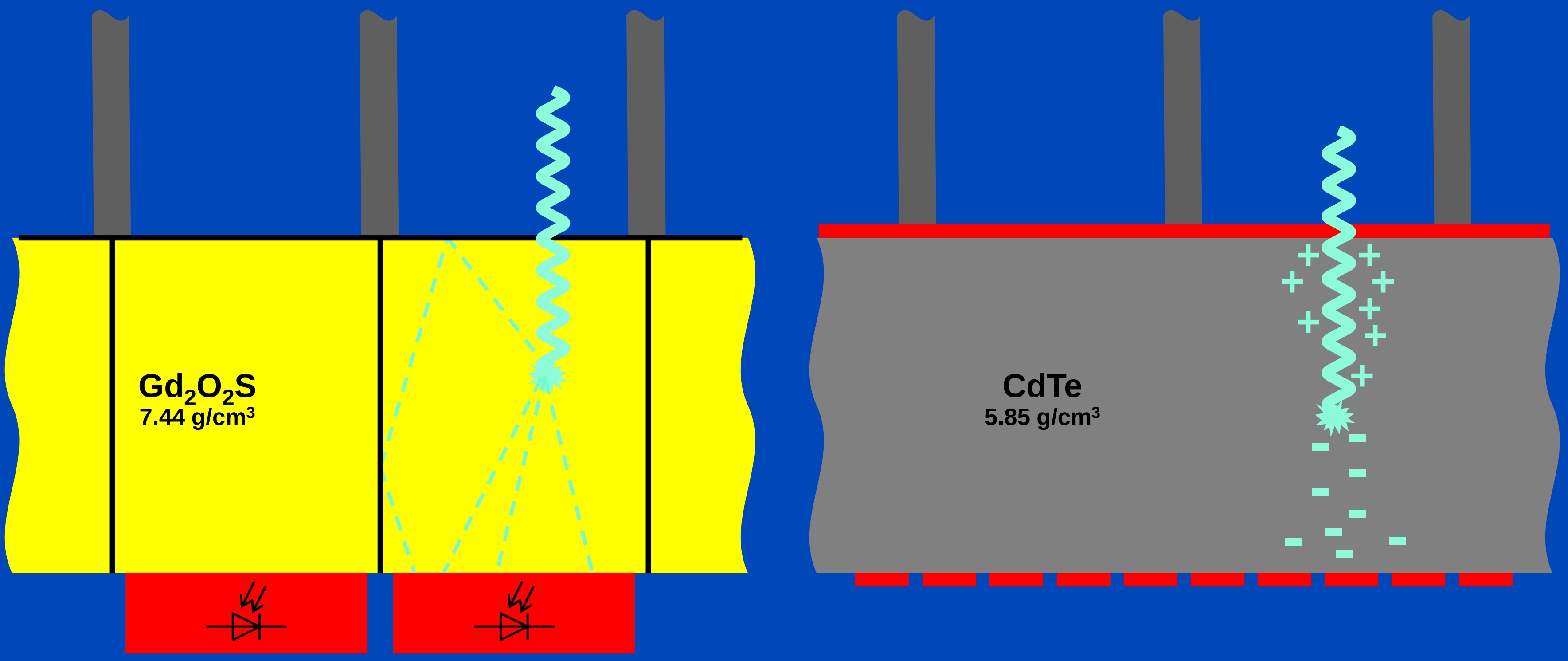




Photon-Counting CT

Counting Single Photons

Energy-Integrating (Today) | **Photon-Counting (Future)**

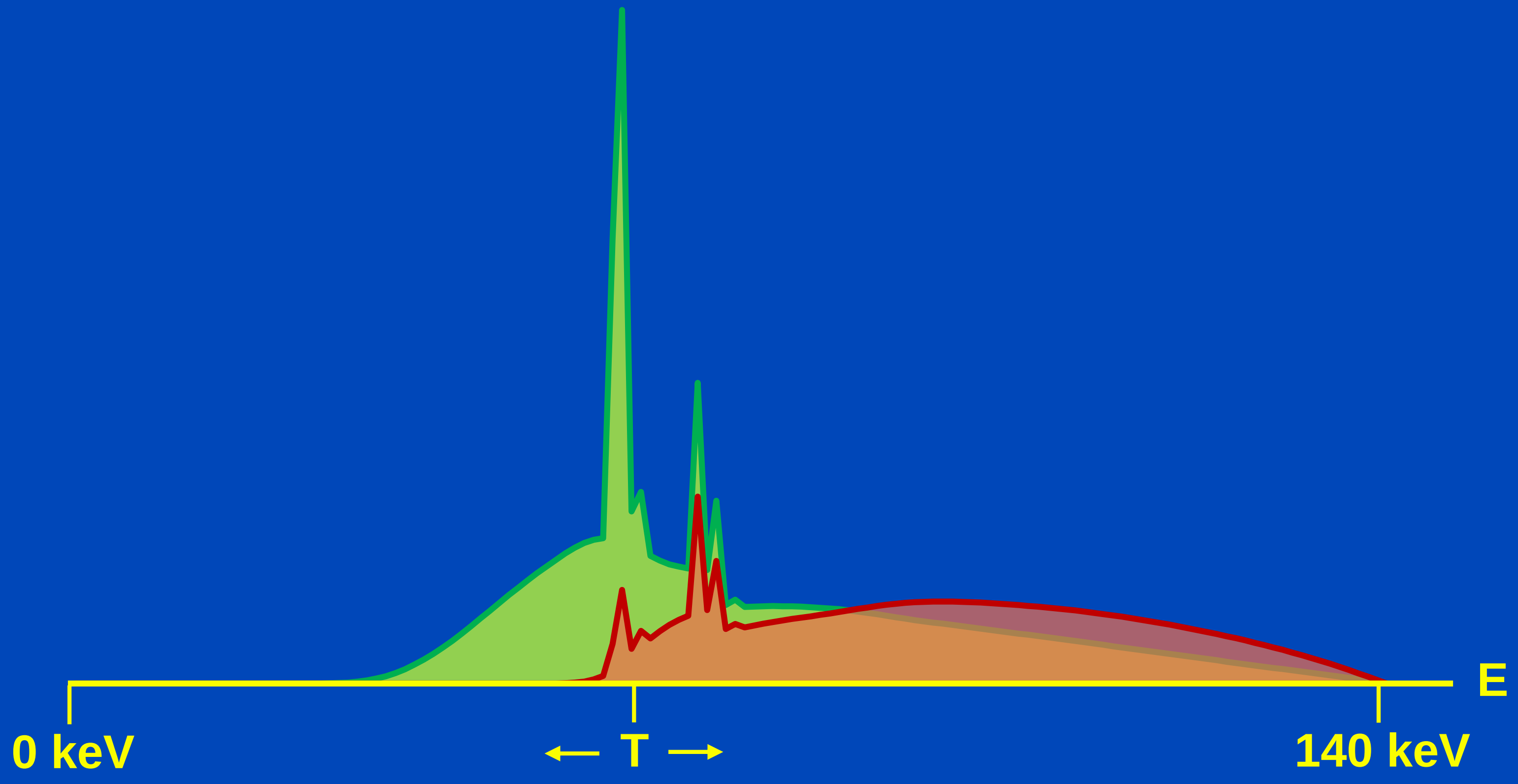


Requirements for CT: up to 10^9 x-ray photon counts per second per mm^2.
Hence, photon counting only achievable for direct converters.

Aim

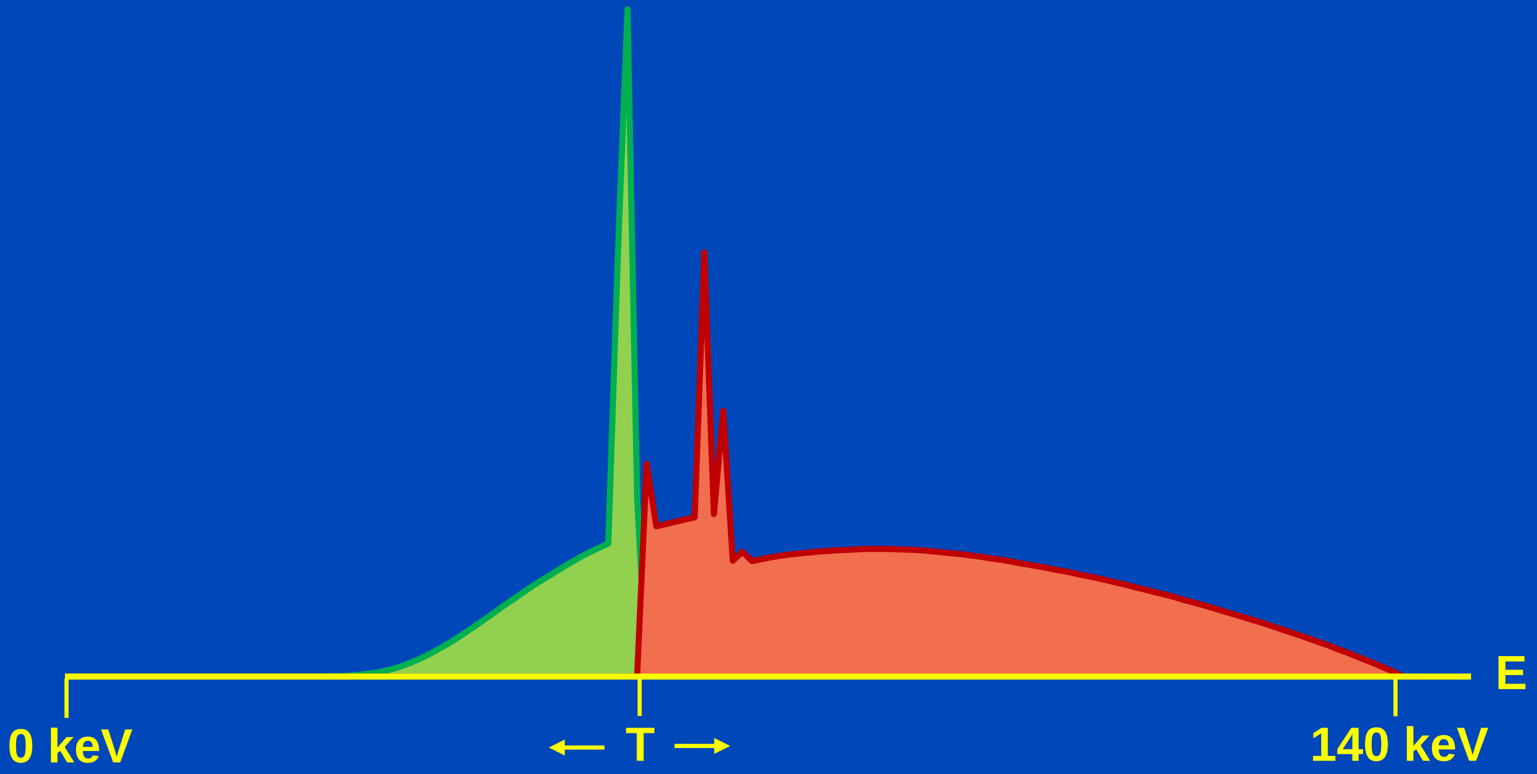
To investigate and validate the spectral properties of high-Z elements and the quality of resulting material images in a photon-counting CT.

2 Bins – Real Case



140 kV spectra as seen after having passed a 30 cm water layer.

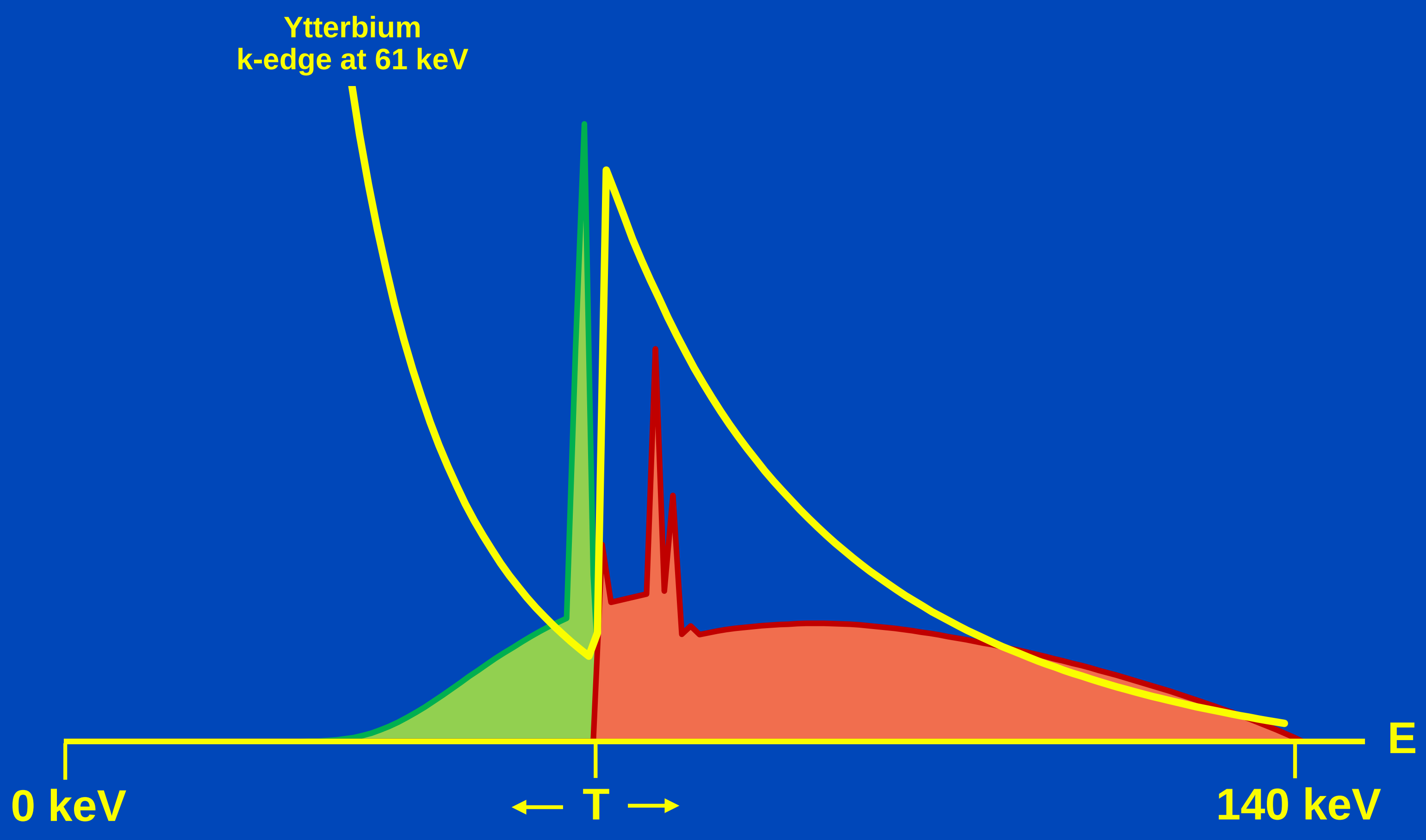
2 Bins – Ideal Case



140 kV spectra as seen after having passed a 30 cm water layer.

dkfz.

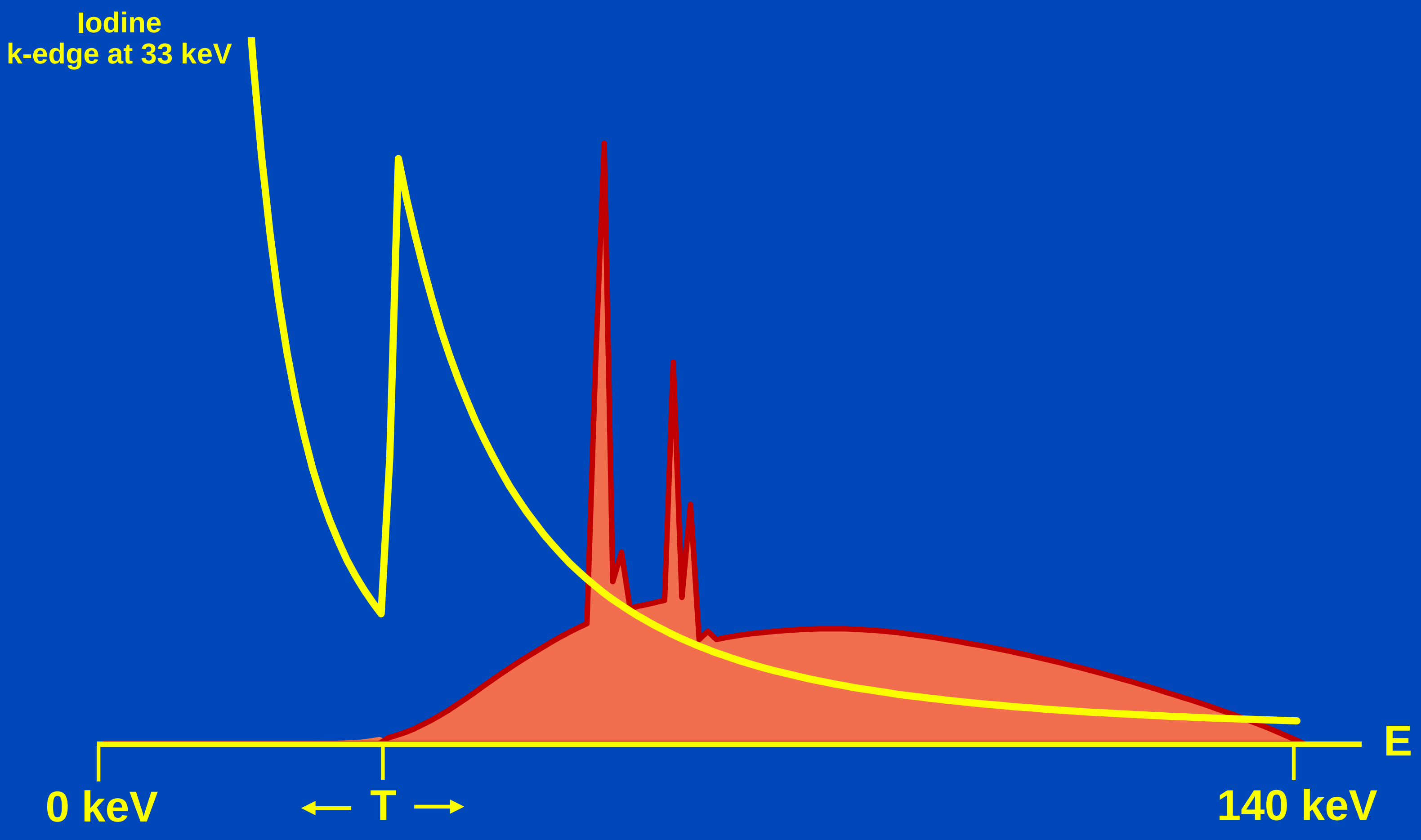
k-Edge Imaging



140 kV spectra as seen after having passed a 30 cm water layer.

dkfz.

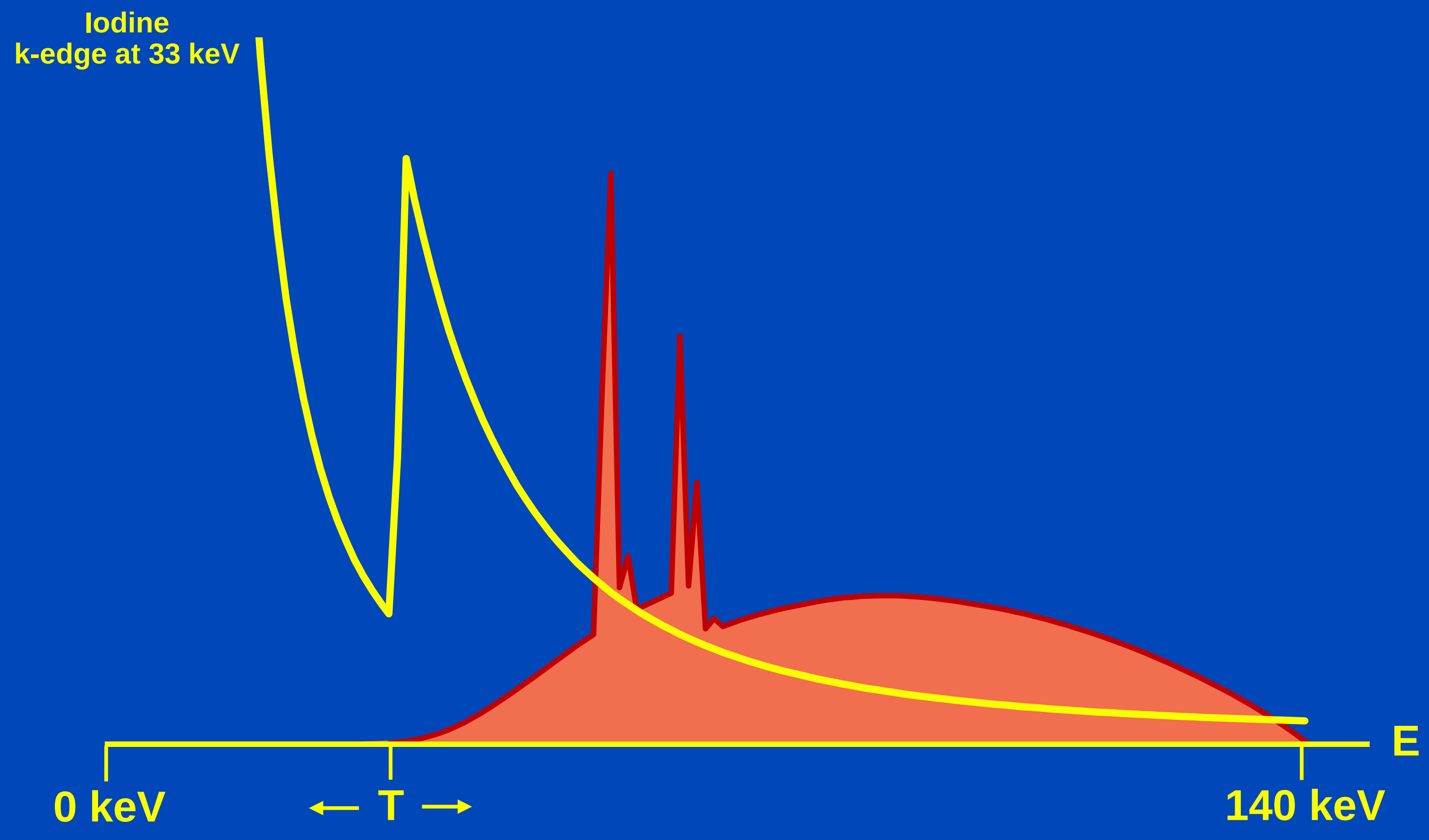
What About Iodine? – 30 cm Patient



140 kV spectra as seen after having passed a 30 cm water layer.

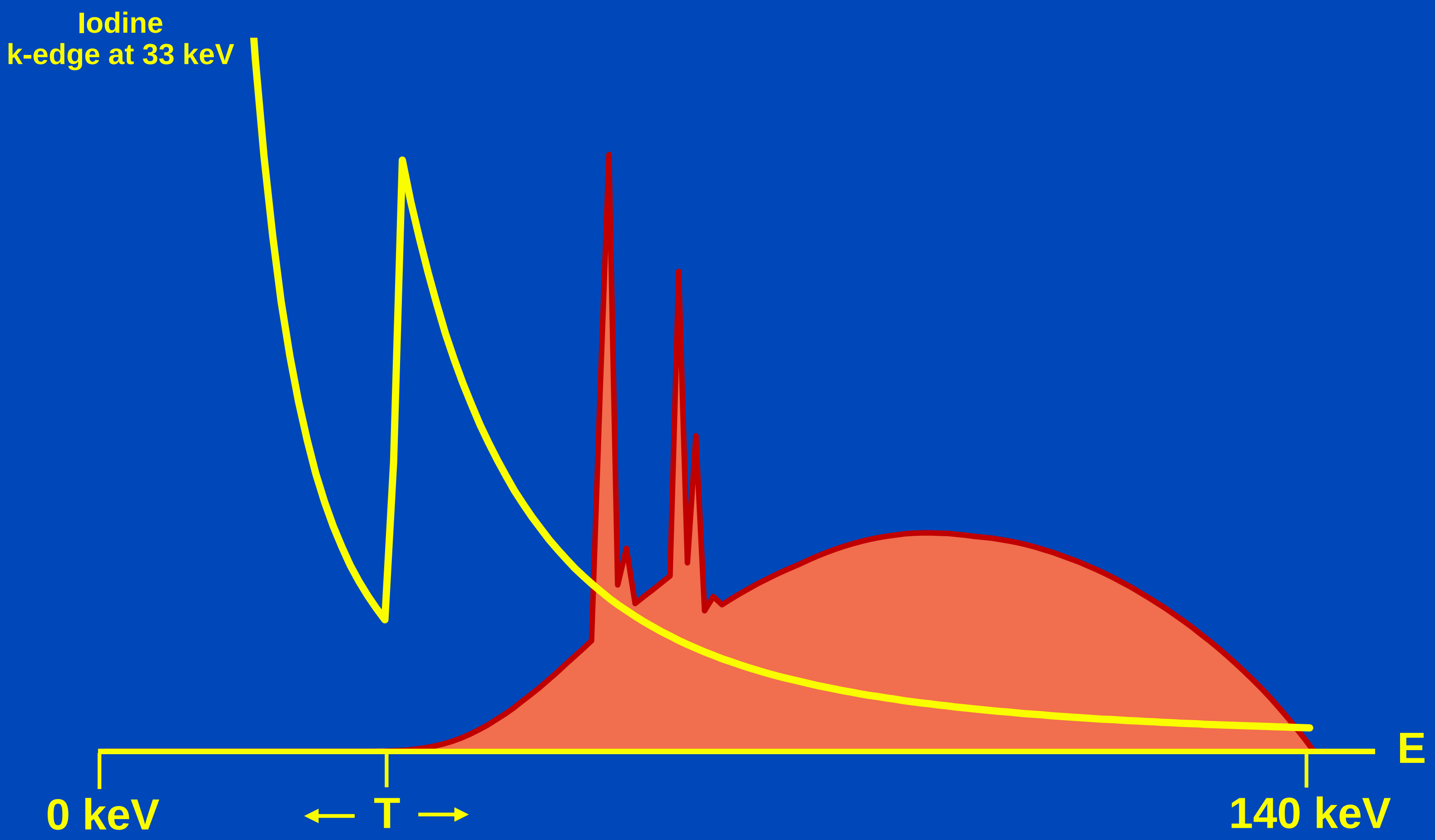
dkfz.

What About Iodine? – 40 cm Patient



140 kV spectra as seen after having passed a 40 cm water layer.

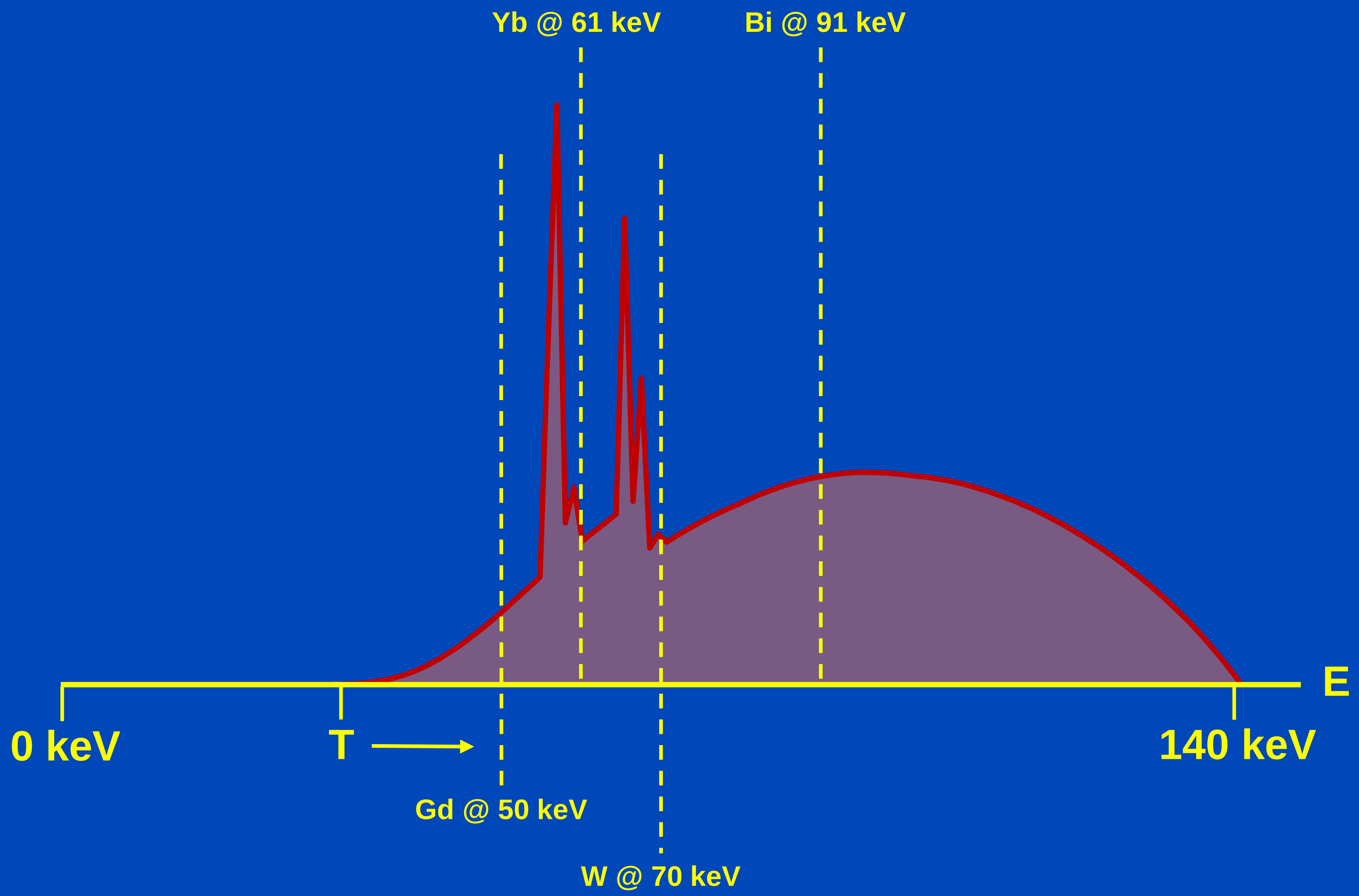
What About Iodine? – 50 cm Patient



140 kV spectra as seen after having passed a 50 cm water layer.

dkfz.

High-Z Materials



140 kV spectra as seen after having passed a 50 cm water layer.

Readout Modes of the Siemens CounT

Macro Mode
0.9 × 1.1 mm focus
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
0.9 × 1.1 mm focus
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
0.9 × 1.1 mm focus
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
0.7 × 0.7 mm focus
4×2 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 photon-counting whole-body CT prototype, 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.

Measurements

- Semianthropomorphic thorax and liver phantoms
- Phantom sizes
  - Small (200 × 300 mm)
  - Medium (250 × 350 mm)
  - Large (300 × 400 mm)
- Voltages 80, 100, 120 and 140 kV
- Macro mode (2 bins) with threshold in 2 keV steps from 50 keV to 90 keV
- Further parameters
  - Tube current time product 200 mAs
  - Collimation 32×0.5 mm
  - Pitch 0.5
  - Rotation 0.5 s

U	$CTDI_{vol32}$
80 kV	4.4 mGy
100 kV	9.2 mGy
120 kV	15 mGy
140 kV	22 mGy

Phantoms and CNRDC

- **Calculation of material-specific images by subtraction**
- **ROI 1 in contrast-enhanced region: Mean M_1, variance V_1**
- **ROI 2 in non-enhanced region: Mean M_2, variance V_2**
- **CNR = contrast-to-noise ratio**

$$\mathrm{CNR} = \frac{M_1 - M_2}{\sqrt{V_1 + V_2}}$$

- **CNRD = CNR at unit dose = CNR/sqrt($CTDI_{vol32}$)**
- **CNRDC = CNRD at unit concentration = CNRD/concentration**

Material-Selective vs. CNR Optimizing

- f_{LH} = low/high energy or bin image, background at 0 HU
- C_{LH} = contrast in low/high energy or bin image
- V_{LH} = variance in low/high energy or bin image
- Material-selective image: $f_\mathrm{L} - f_\mathrm{H}$ (e.g. iodine map)

$$\mathrm{CNR}^2_\mathrm{mat} = \frac{(C_\mathrm{L} - C_\mathrm{H})^2}{V_\mathrm{L} + V_\mathrm{H}}$$

- Optimum CNR image: $(1-\alpha) f_\mathrm{L} + \alpha f_\mathrm{H}$

$$\alpha_\mathrm{opt} = \frac{C_\mathrm{H} V_\mathrm{L}}{C_\mathrm{H} V_\mathrm{L} + C_\mathrm{L} V_\mathrm{H}} \qquad \mathrm{CNR}^2_\mathrm{opt} = \frac{C_\mathrm{L}^2}{V_\mathrm{L}} + \frac{C_\mathrm{H}^2}{V_\mathrm{H}}$$

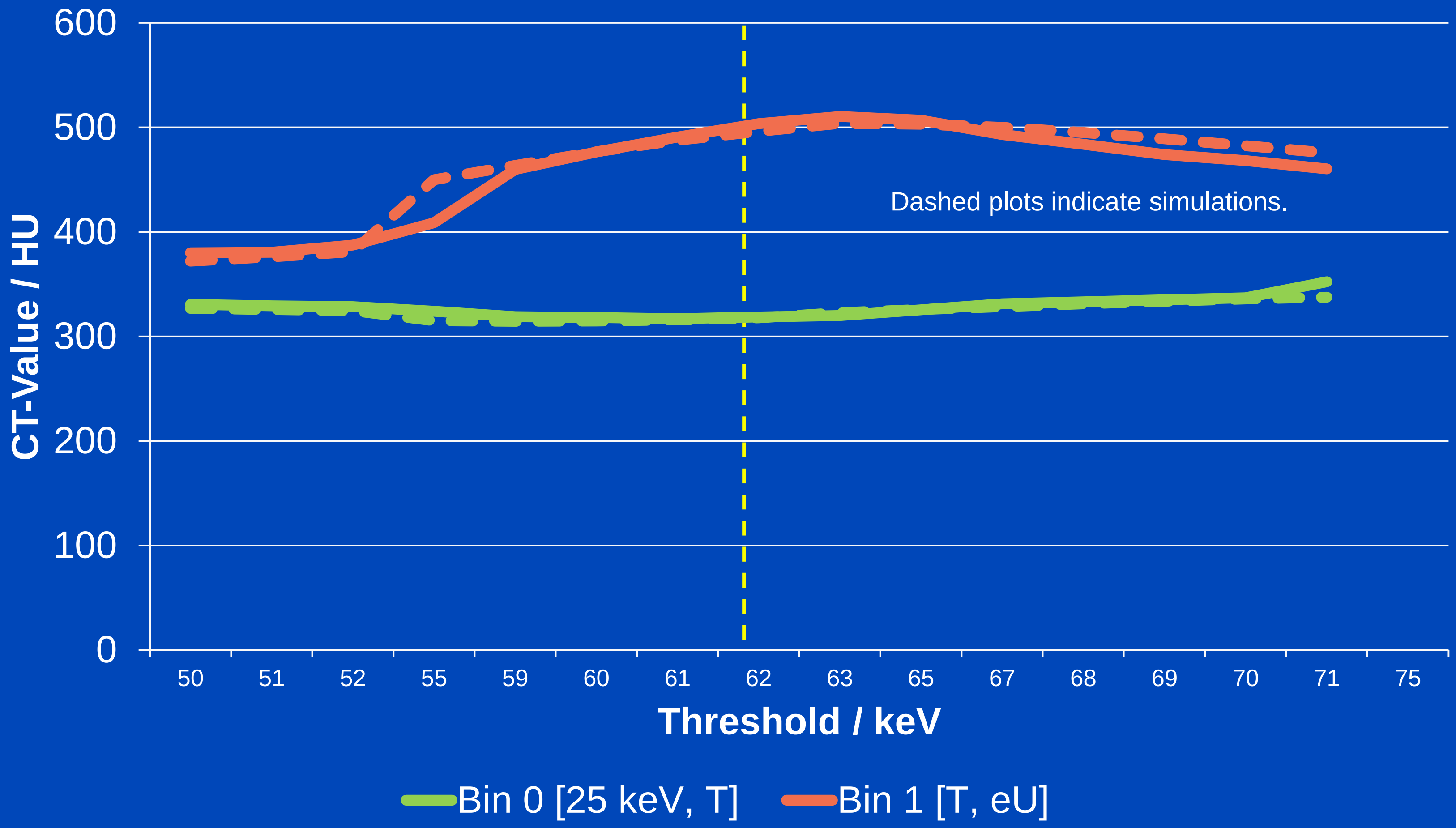
nearly independent of threshold

- Optimum is optimal (needless to say):

$$\mathrm{CNR}^2_\mathrm{opt} = \mathrm{CNR}^2_\mathrm{mat} + \frac{(C_\mathrm{L} V_\mathrm{H} + C_\mathrm{H} V_\mathrm{L})^2}{V_\mathrm{L} V_\mathrm{H} (V_\mathrm{L} + V_\mathrm{H})}$$

CT-Values

Ytterbium



Measurement at 80 kV.

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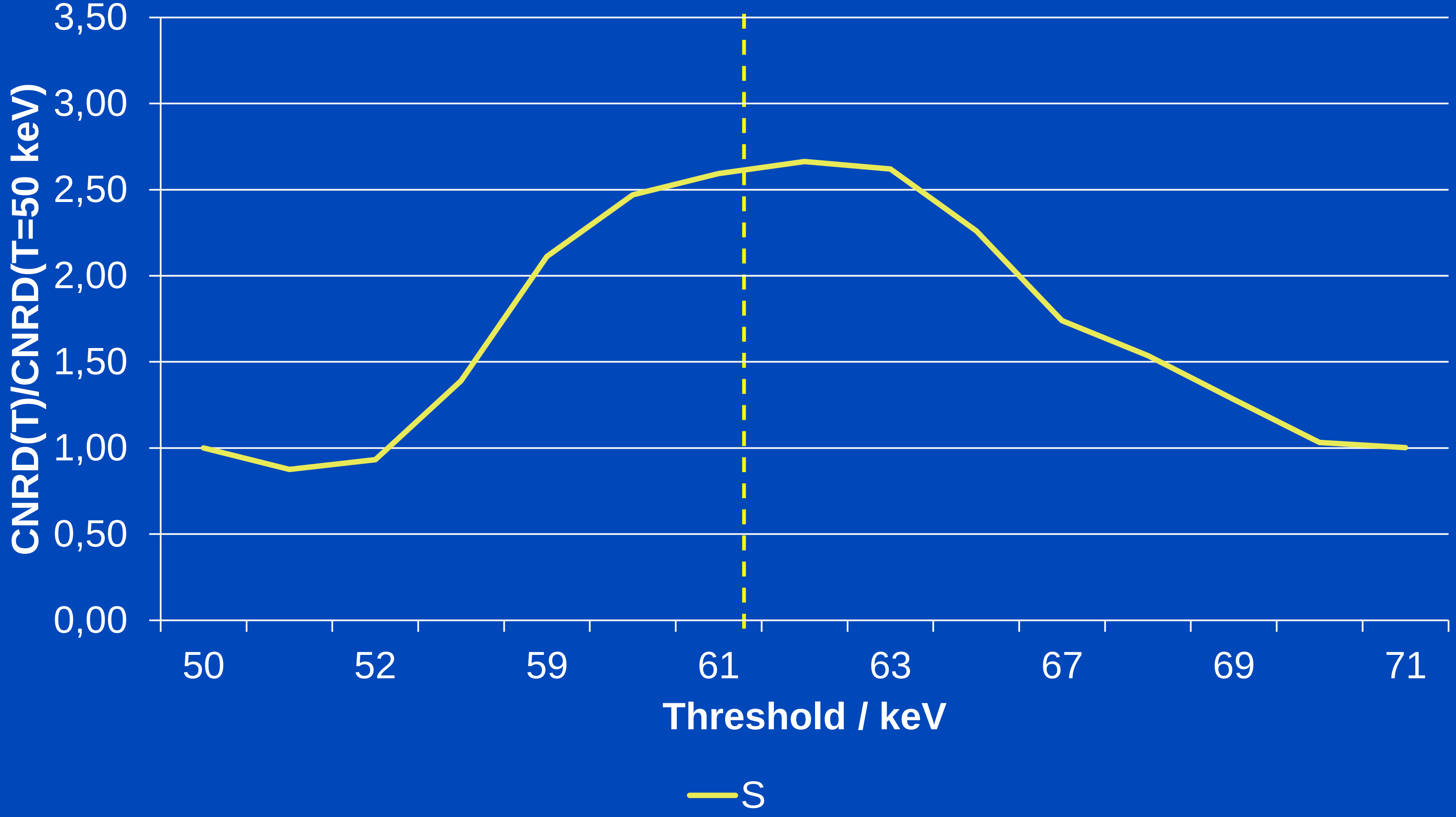
Material Image CNRD

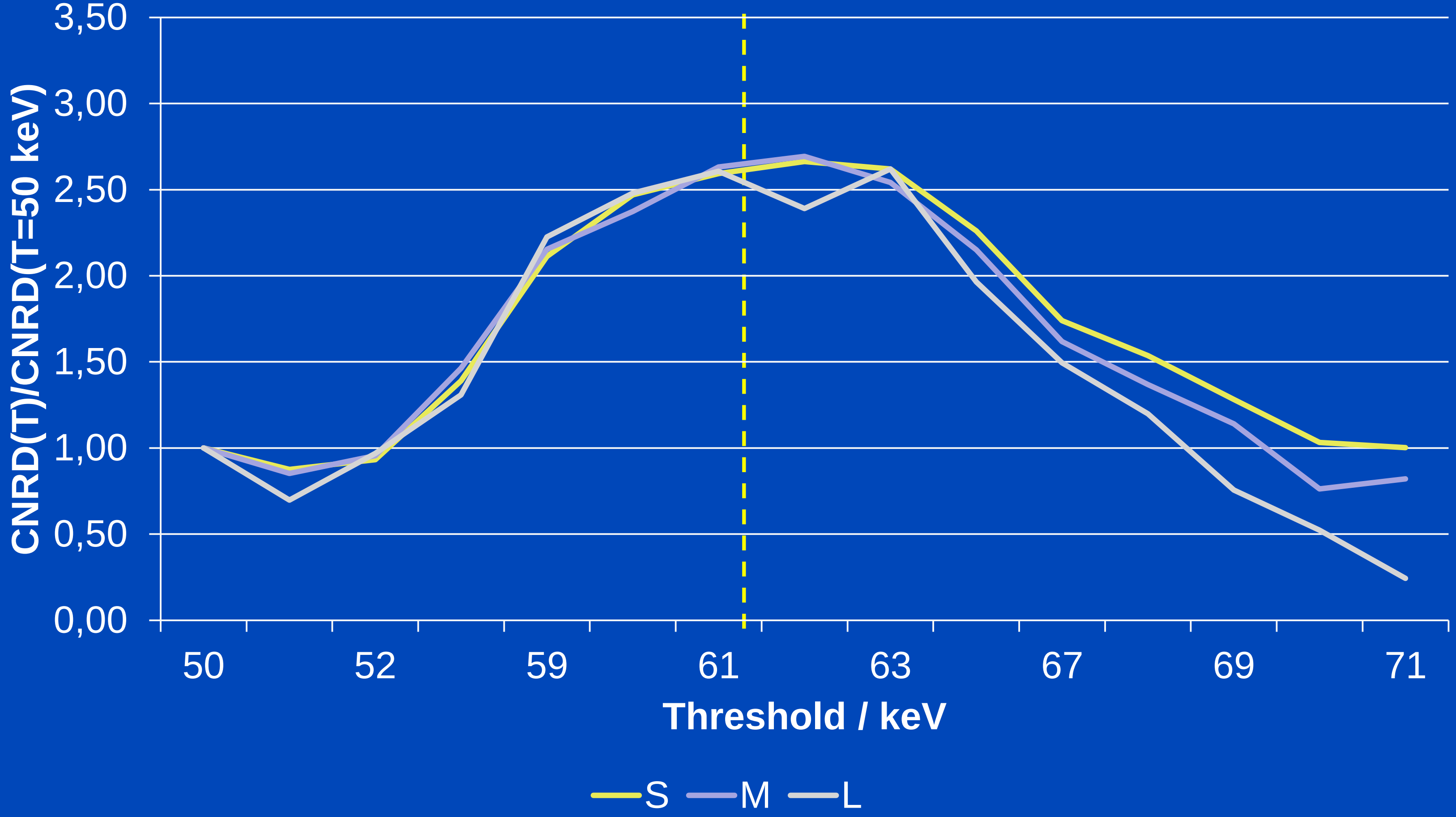
Ytterbium



Measurement at 80 kV.

Material Image CNRD

Ytterbium

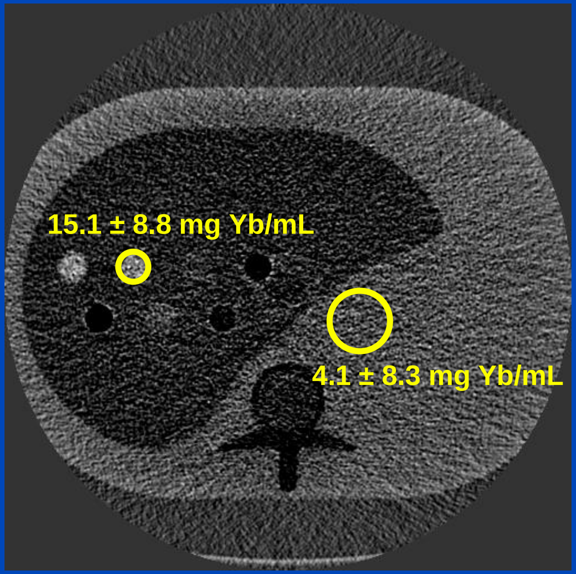


Measurement at 80 kV.

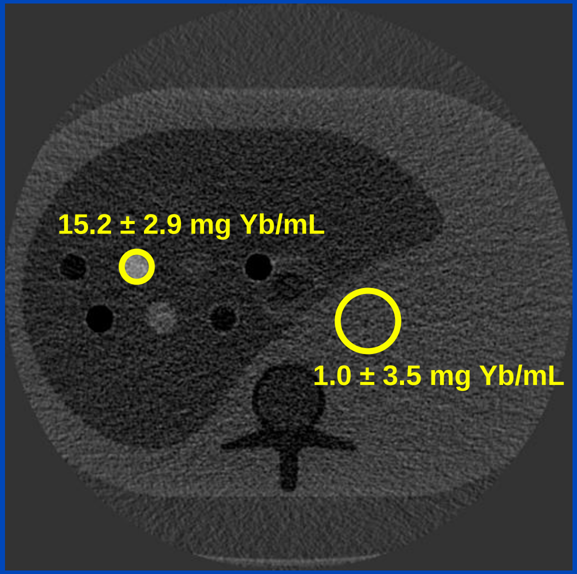
dkfz.

Material Images

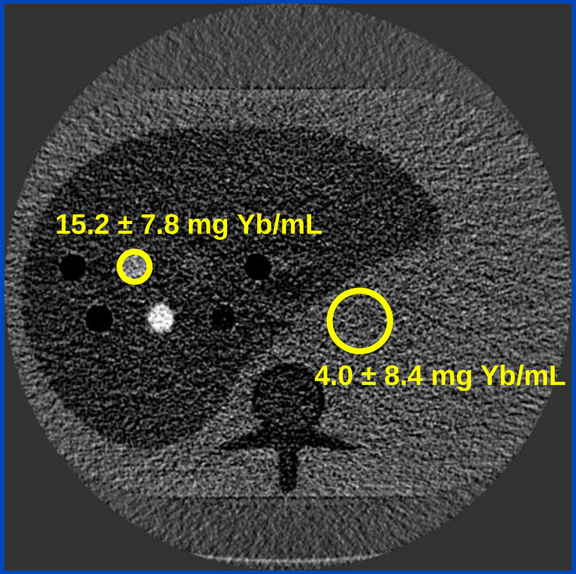
T=50 keV



T=61 keV



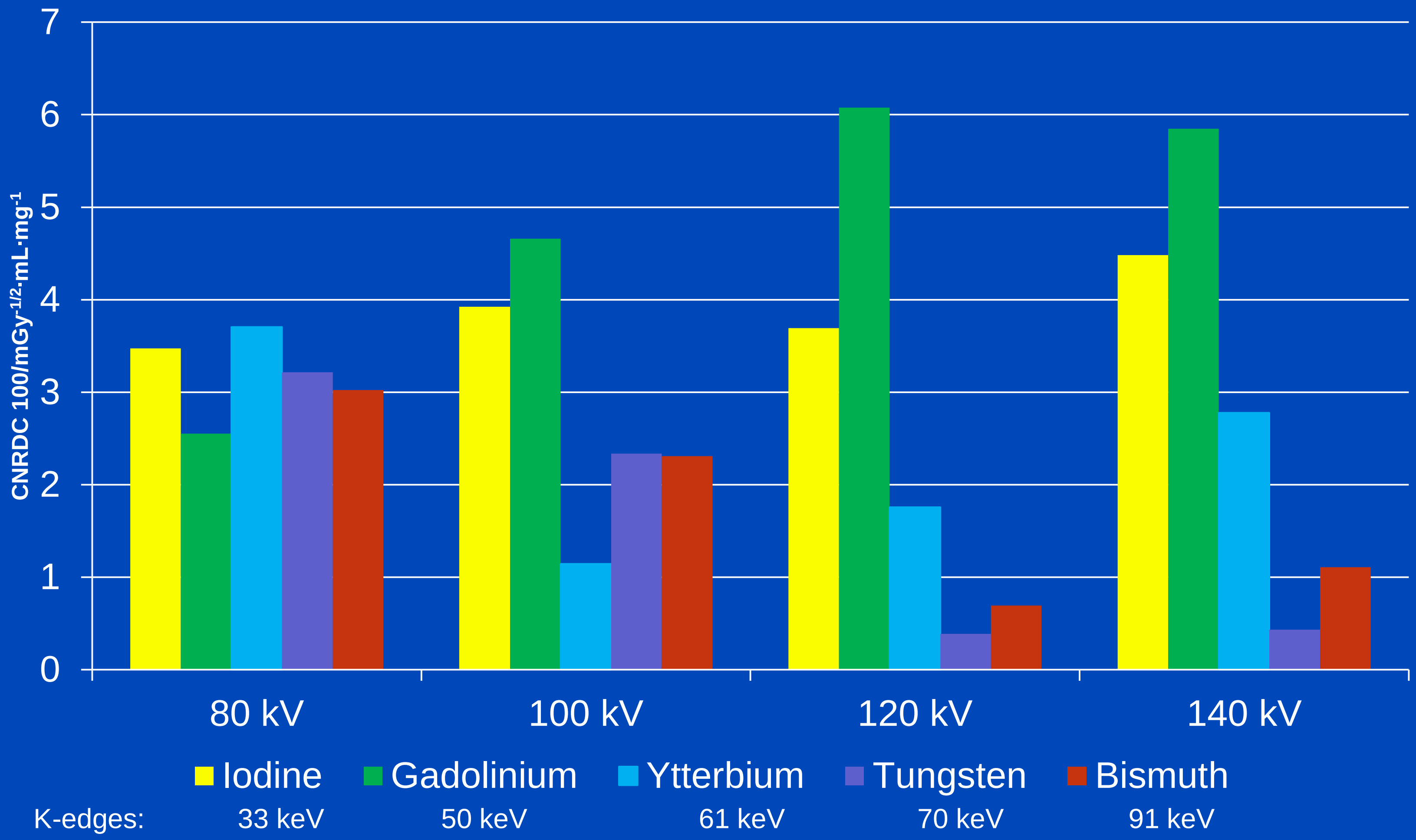
T=71 keV



C = 15 mg Yb/mL, *W* = 50 mg Yb/mL

Peak $CNRDC_{mat}$

S Phantom



Summary & Conclusion

- Clinical k-edge imaging seems reasonable with photon counting CT.
- Thresholds can be set to maximize CNR_{mat} since CNR_{opt} is nearly constant.
- Optimal thresholds increase with
  - higher tube voltage (otherwise too many counts in high bin)
  - and with larger phantom diameters (needs counts in low bin).
- Contrast agents with high-Z materials might
  - help to reduce the required amount of contrast agent
  - or might be combined with iodine.
- Limitations
  - Only one threshold was used.
  - Threshold was optimized only for a single contrast agent.
  - Noise reduction techniques were not applied.
  - High-Z agents do not exist for use in human subjects.

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.