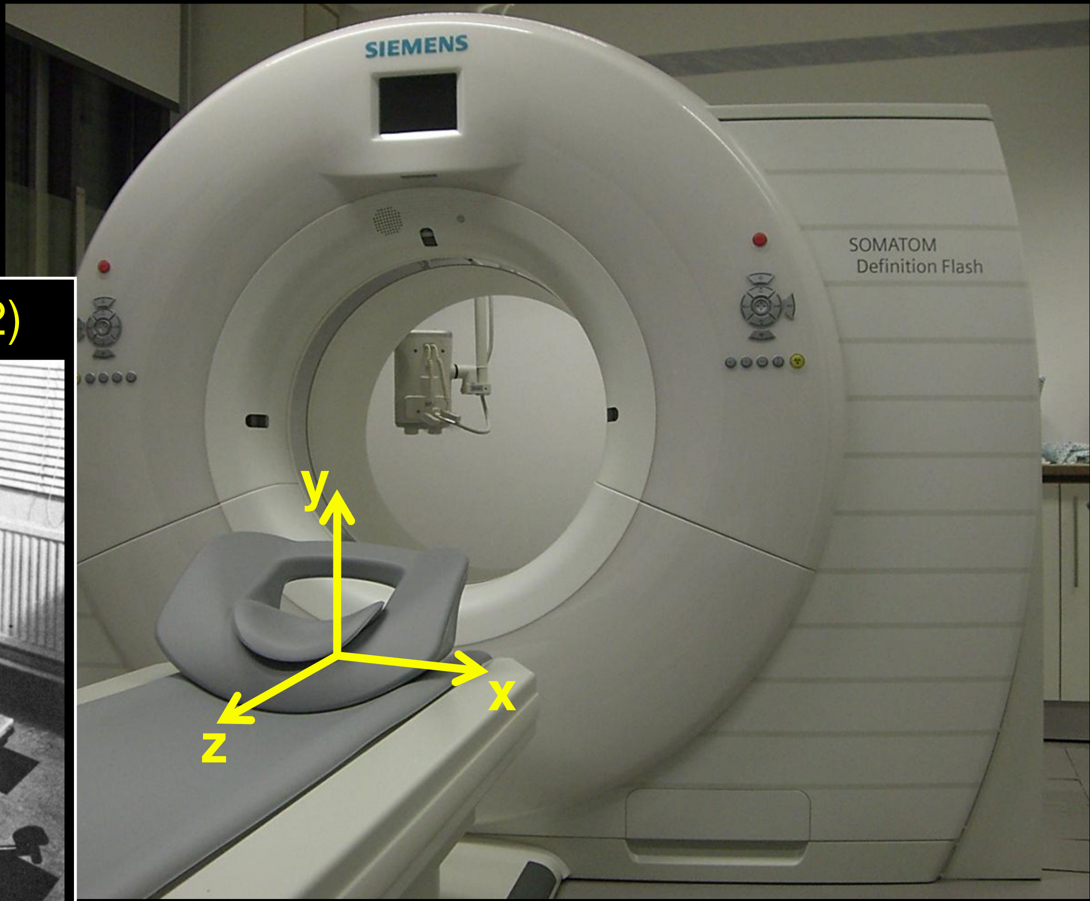
Technical Status and Future Perspectives in Computed Tomography

Prof. Dr. Marc Kachelrieß

German Cancer Research Center (DKFZ)
Heidelberg, Germany
www.dkfz.de/ct

DEUTSCHES KREBSFORSCHUNGSZENTRUM IN DER HELMHOLTZ-GEMEINSCHAFT

Siemens 2·2·64=256-slice
dual source cone-beam spiral **CT**(2008)



1152 views per rotation in 0.28 s

2·64×(736+480) 2-byte channels per view

600 MB/s data transfer rate

5 GB data size typical

EMI parallel beam scanner (1972)

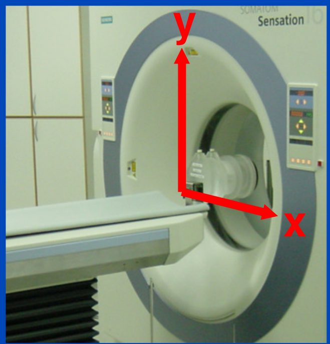
180 views per rotation in 300 s

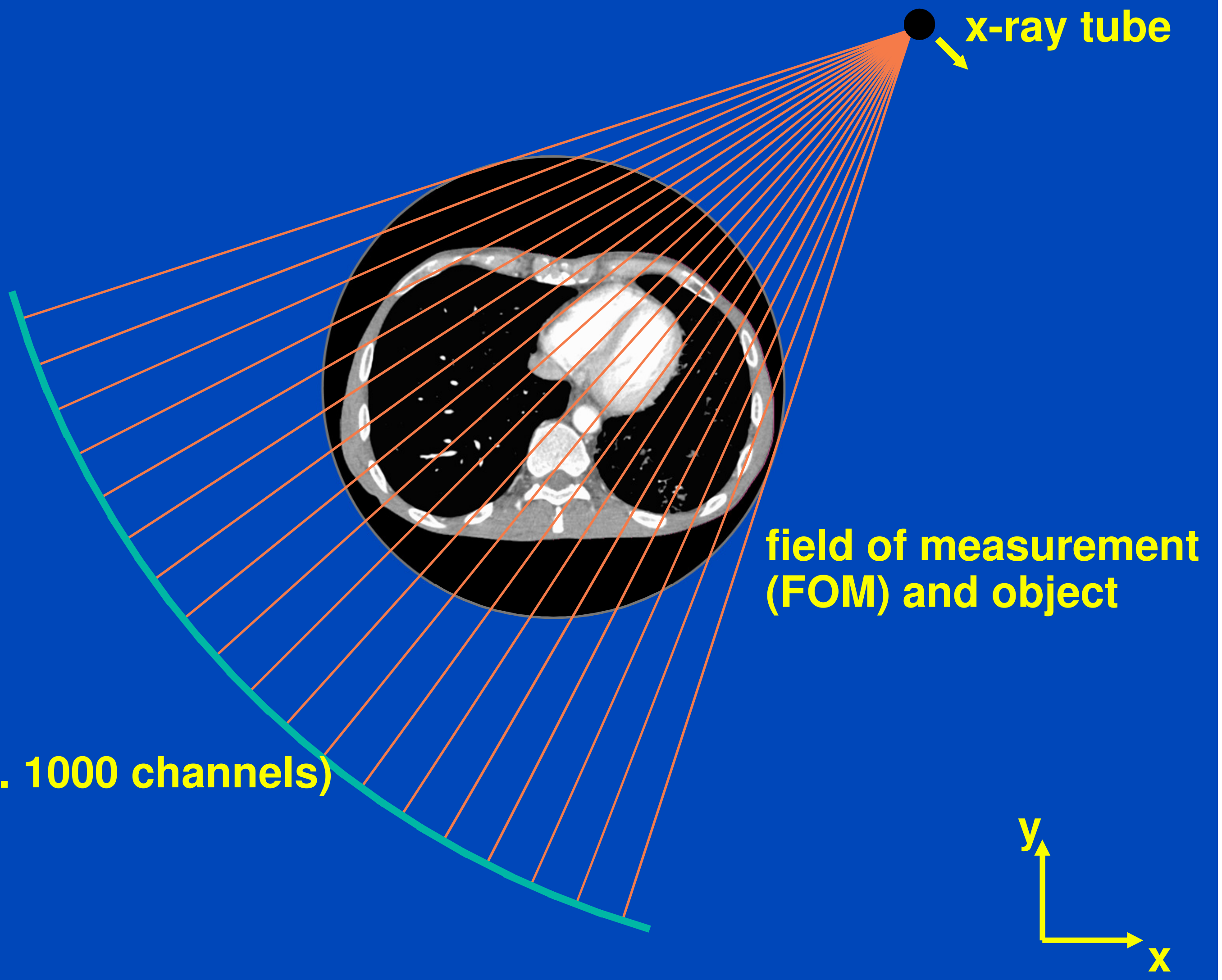
2×160 positions per view

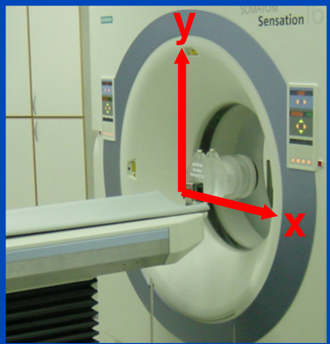
384 B/s data transfer rate

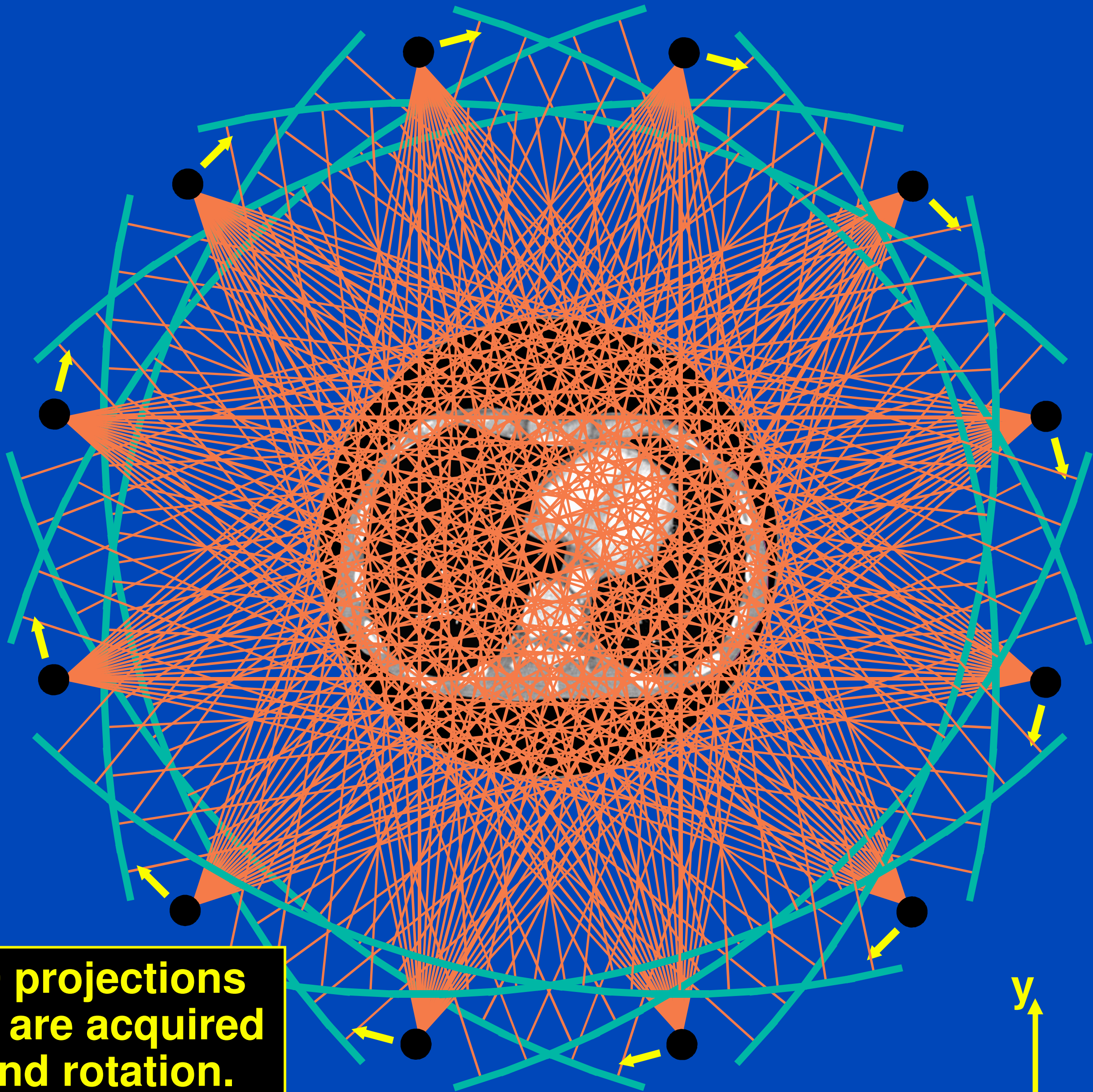
113 kB data size

Fan-Beam Geometry (transaxial / in-plane / x-y-plane)

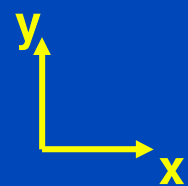






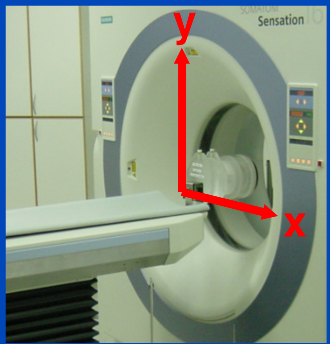


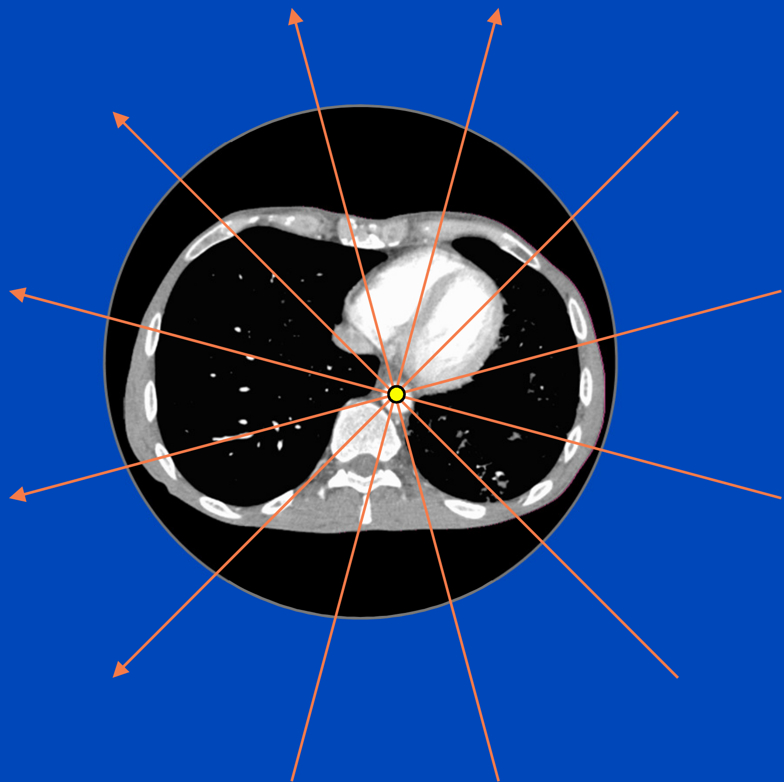
In the order of 1000 projections with 1000 channels are acquired per detector slice and rotation.



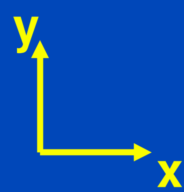
dkfz.

Data Completeness



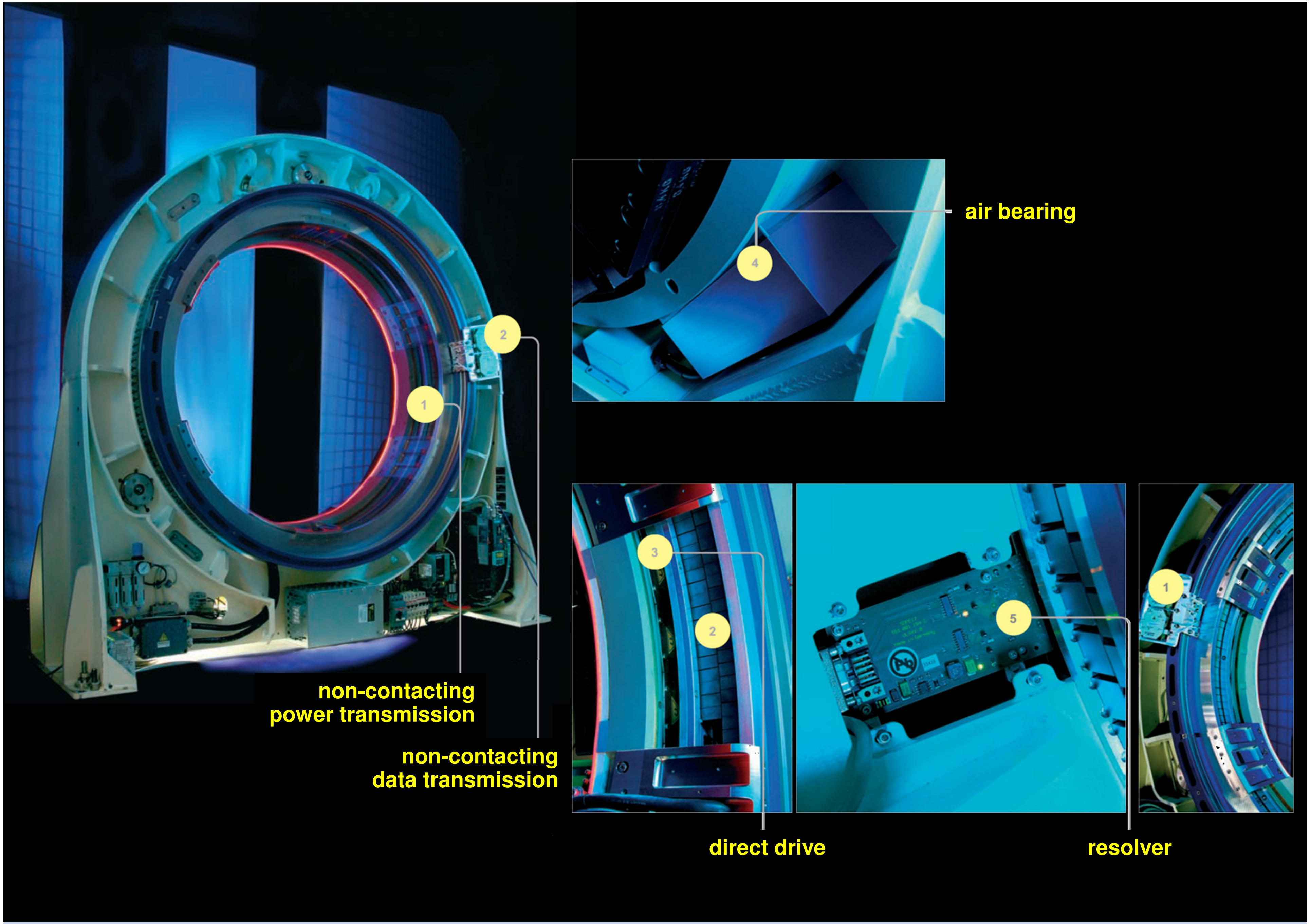


Each object point must be viewed by an angular interval of 180° or more. Otherwise image reconstruction is not possible.



Demands on the Mechanical Design

- Continuous data acquisition (spiral, fluoro, dynamic, …)
- Able to withstand very fast rotation
  - Centrifugal force at 550 mm with 0.5 s: $F = 9\ g$
  - with 0.4 s: $F = 14\ g$
  - with 0.3 s: $F = 25\ g$
  - with 0.2 s: $F = 55\ g$
- Mechanical accuracy better than 0.1 mm
- Compact and robust design
- Short installation times
- Long service intervals
- Low cost



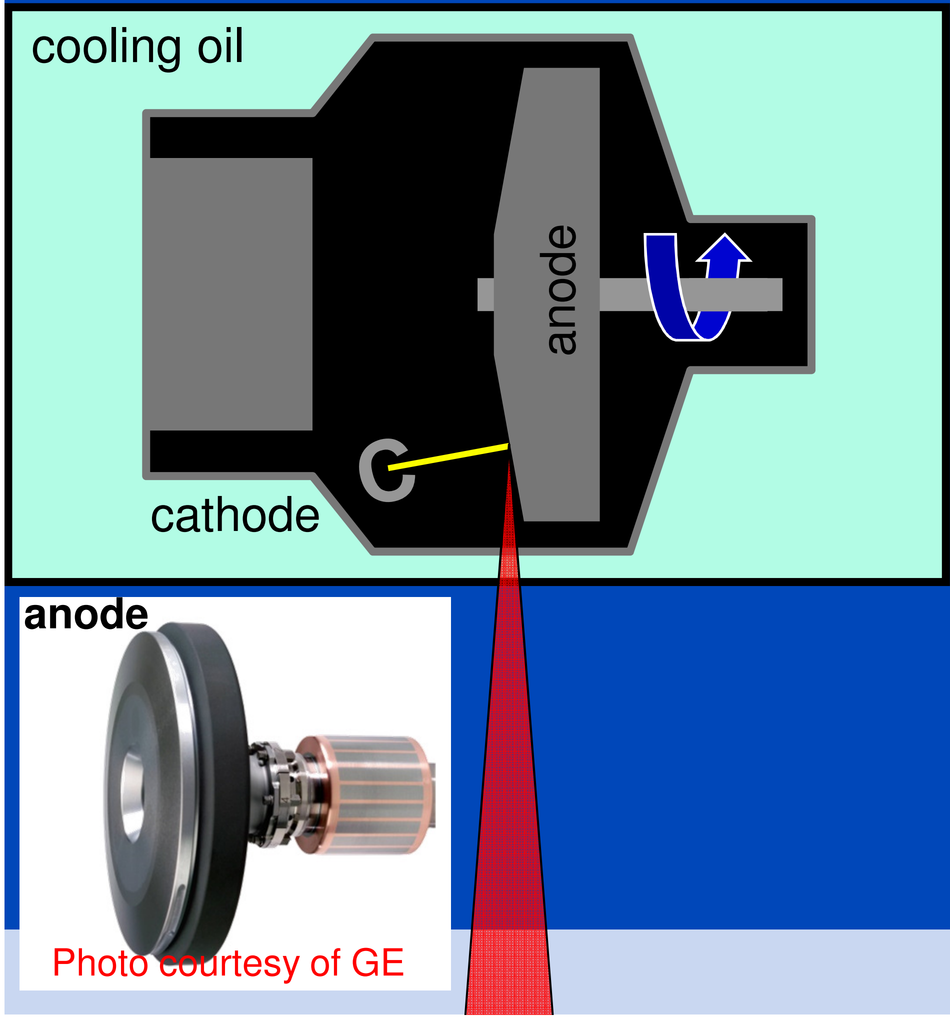
Data courtesy of Schleifring GmbH, Fürstenfeldbruck, Germany
and of rsna2011.rsna.org/exbData/1678/docs/Gantry_Subsystem.pdf

Demands on X-Ray Sources

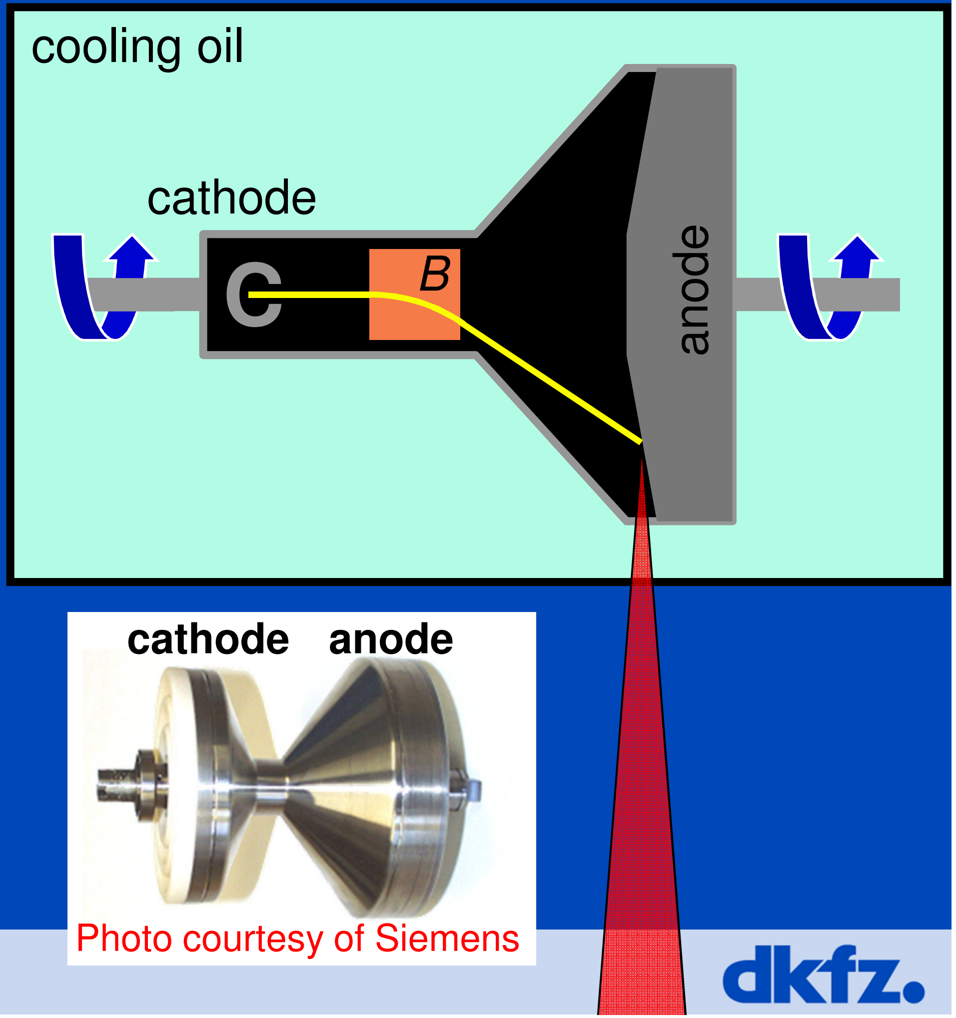
- Tube voltages from 70 to 140 kV
- High instantaneous power levels (typ. 50-100 kW)
- High continuous power levels (typ. >5 kW)
- High cooling rates (typ. >1 MHU/minute)
- High tube current variation (low inertia)
- Must withstand centrifugal forces
- Compact and robust design

Tube Technology

conventional tube
(rotating anode, helical wire emitter)



high performance tube
(rotating cathode, anode + envelope, flat emitter)



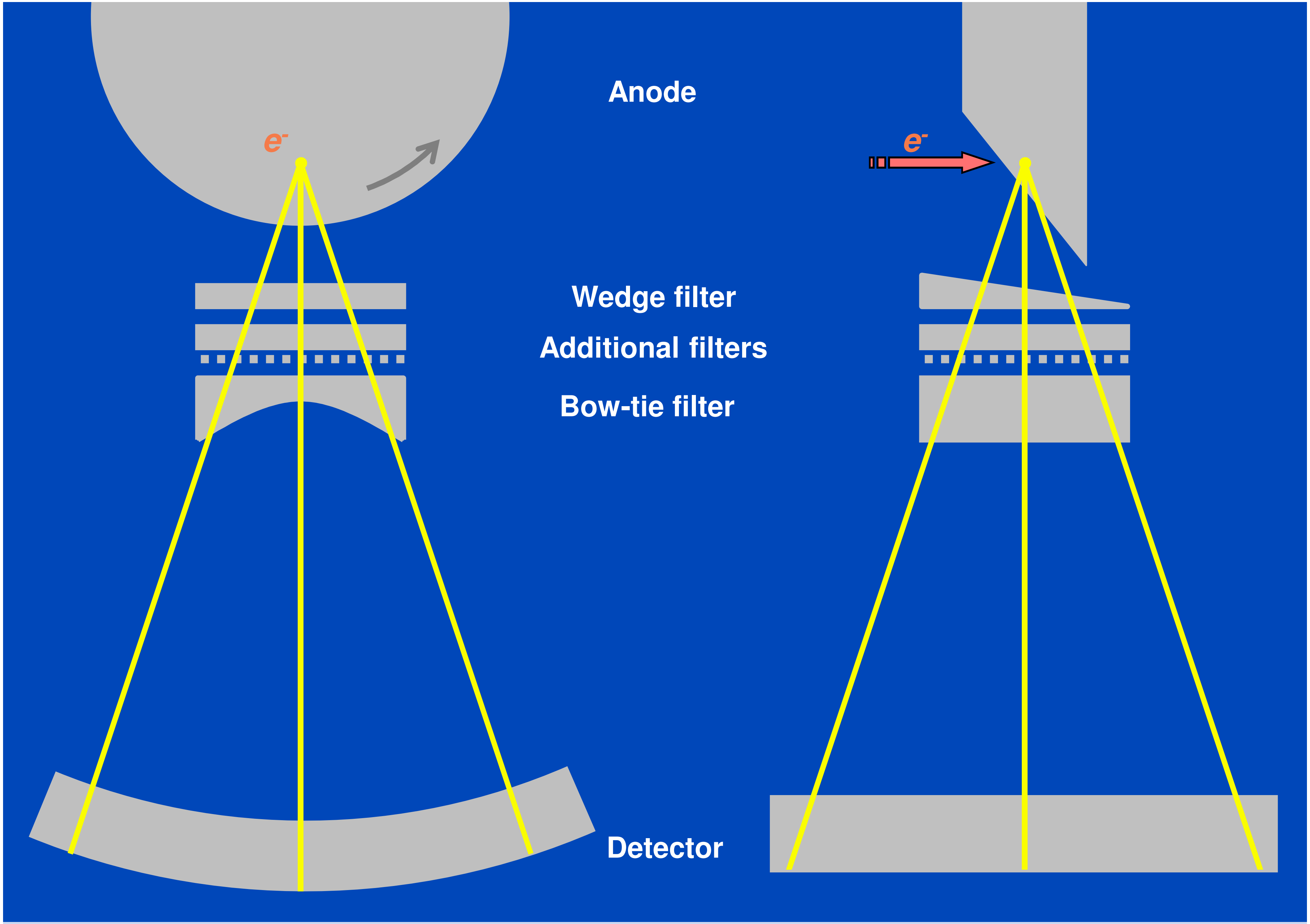


Figure not drawn to scale. Order of prefiltration may differ from scanner to scanner.

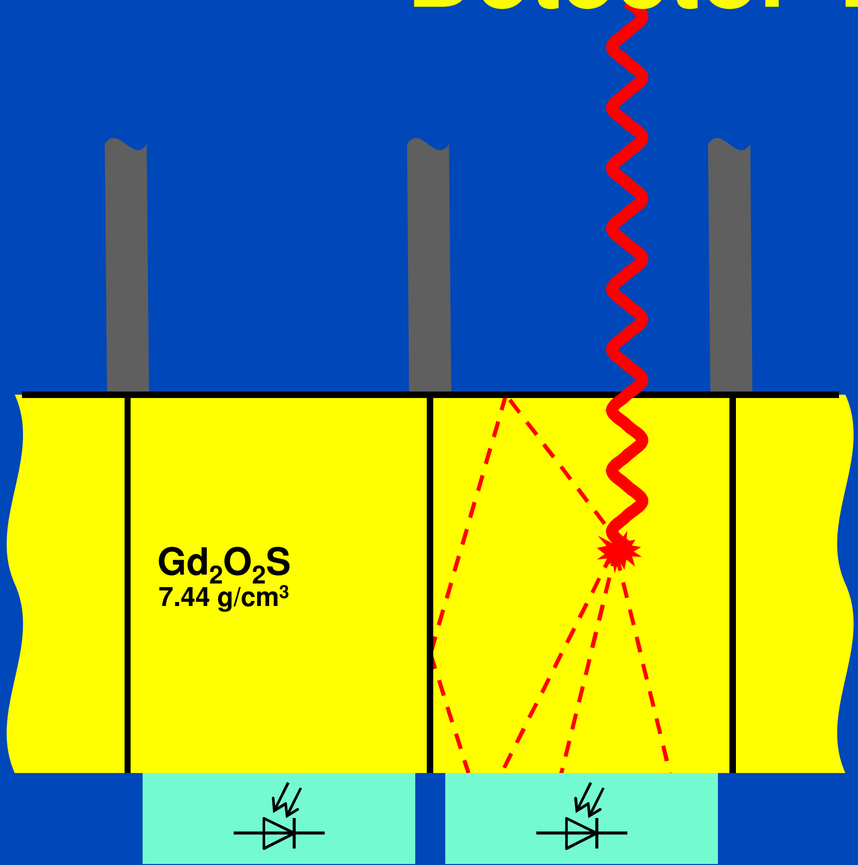
dkfz.

Demands on CT Detector Technology

- **Available as multi-row arrays**
- **Very fast sampling (typ. 300 µs)**
- **Favourable temporal characteristics (decay time < 10 µs)**
- **High absorption efficiency**
- **High geometrical efficiency**
- **High count rate (up to 10^9 cps*)**
- **Adequate dynamic range (at least 20 bit)**

* in the order of 10^5 counts per reading and 10^4 readings per second

Detector Technology



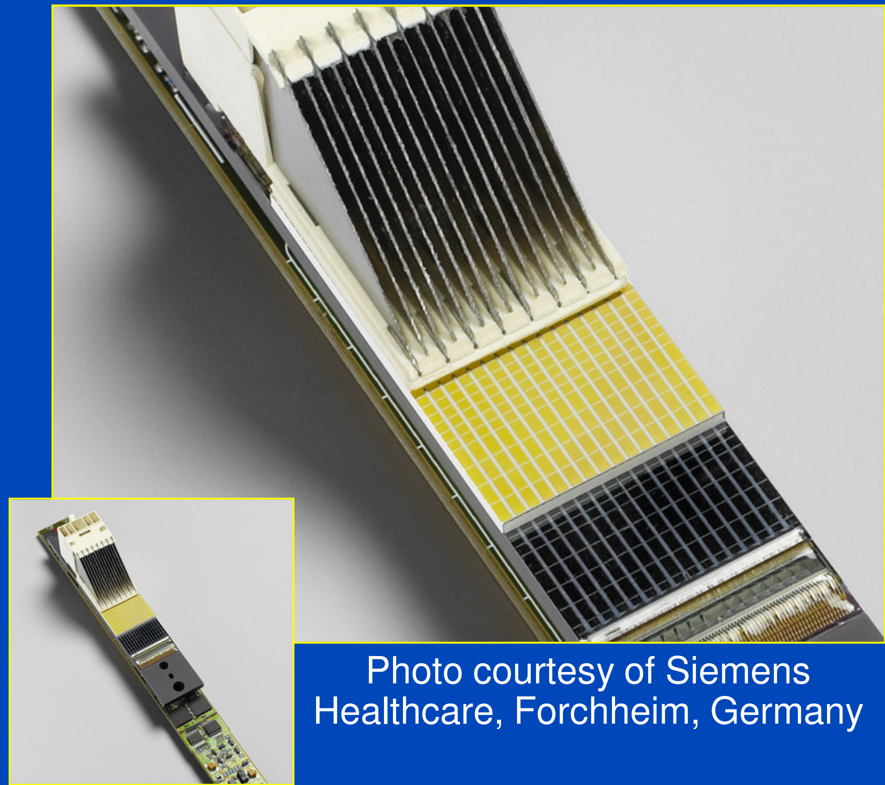
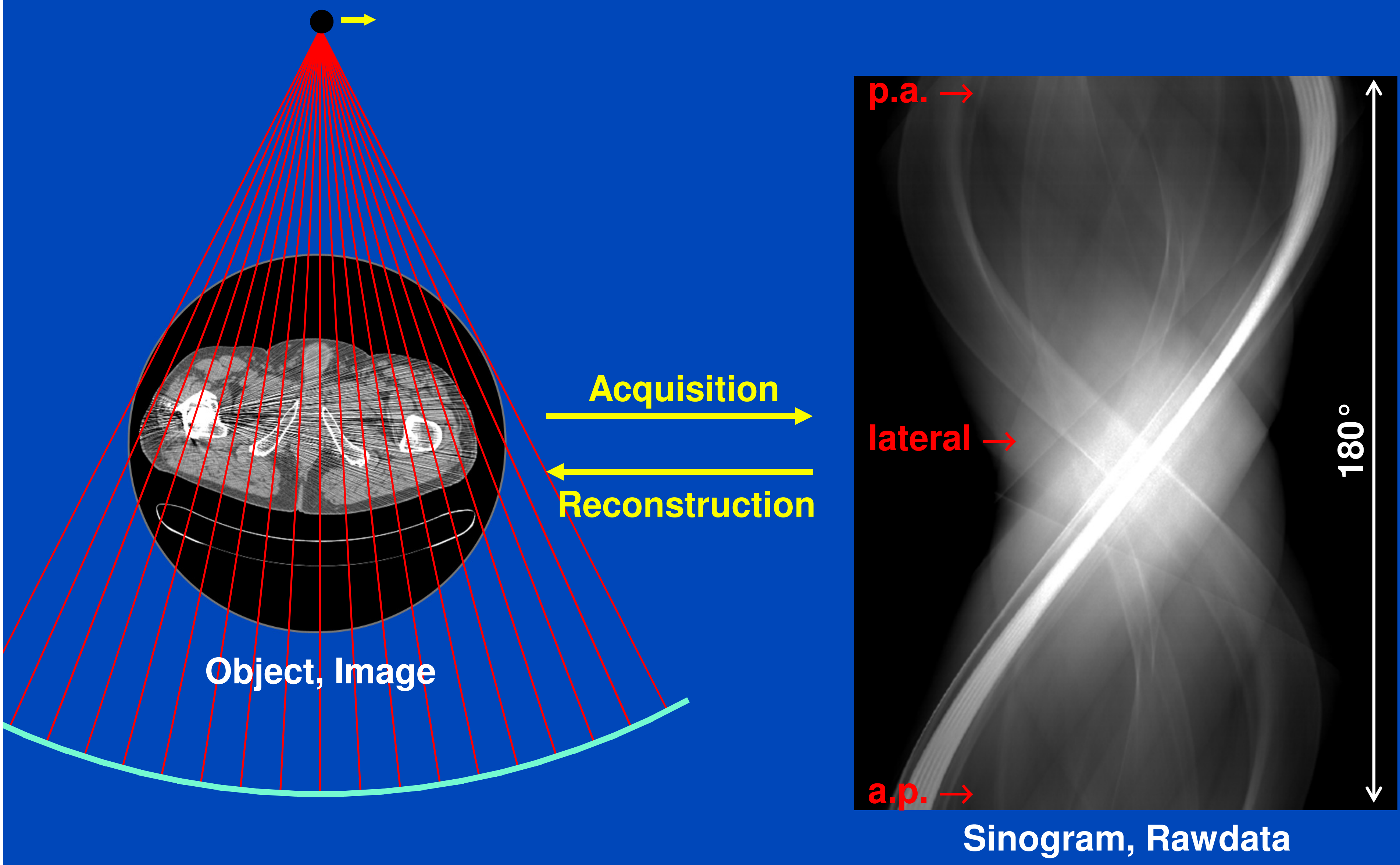


Photo courtesy of Siemens Healthcare, Forchheim, Germany

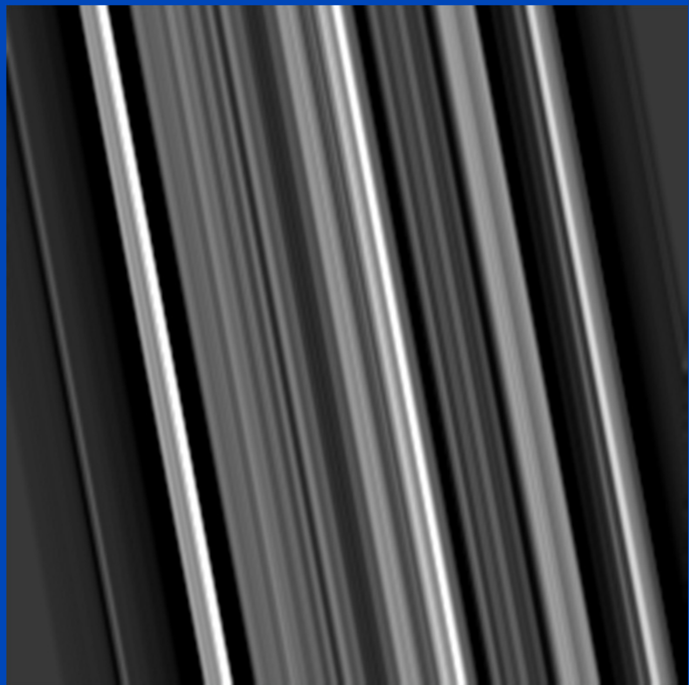


Acquisition
Reconstruction
Object, Image
p.a. →
lateral →
a.p. →
180°
Sinogram, Rawdata

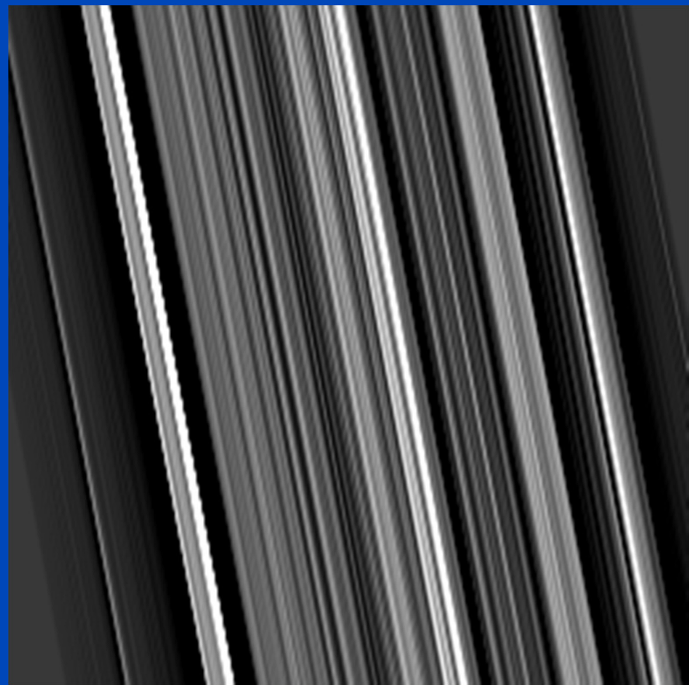
dkfz.

Filtered Backprojection (FBP)

1. Filter projection data with the reconstruction kernel.
2. Backproject the filtered data into the image:



Smooth



Standard

Reconstruction kernels balance between spatial resolution and image noise.

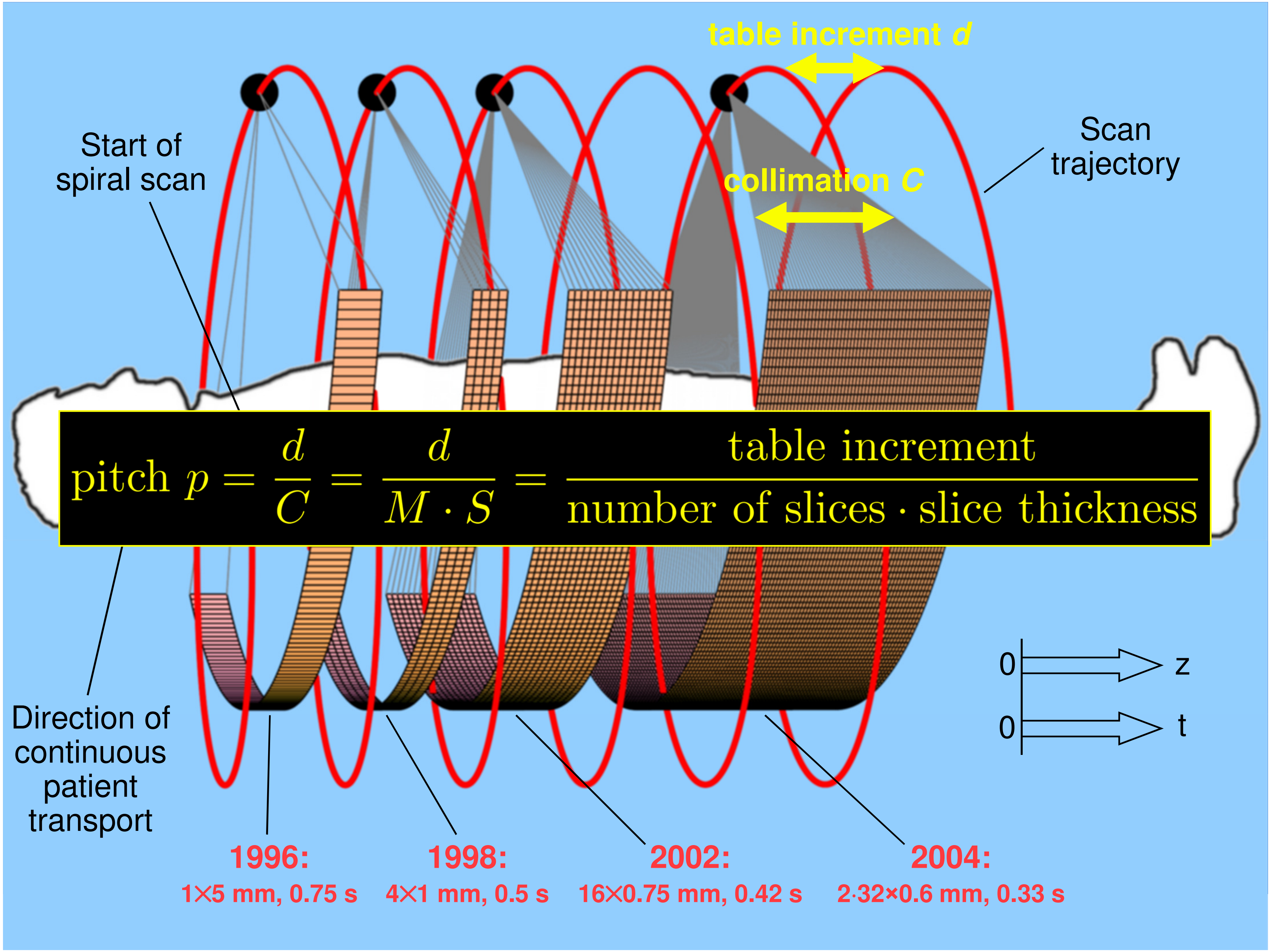
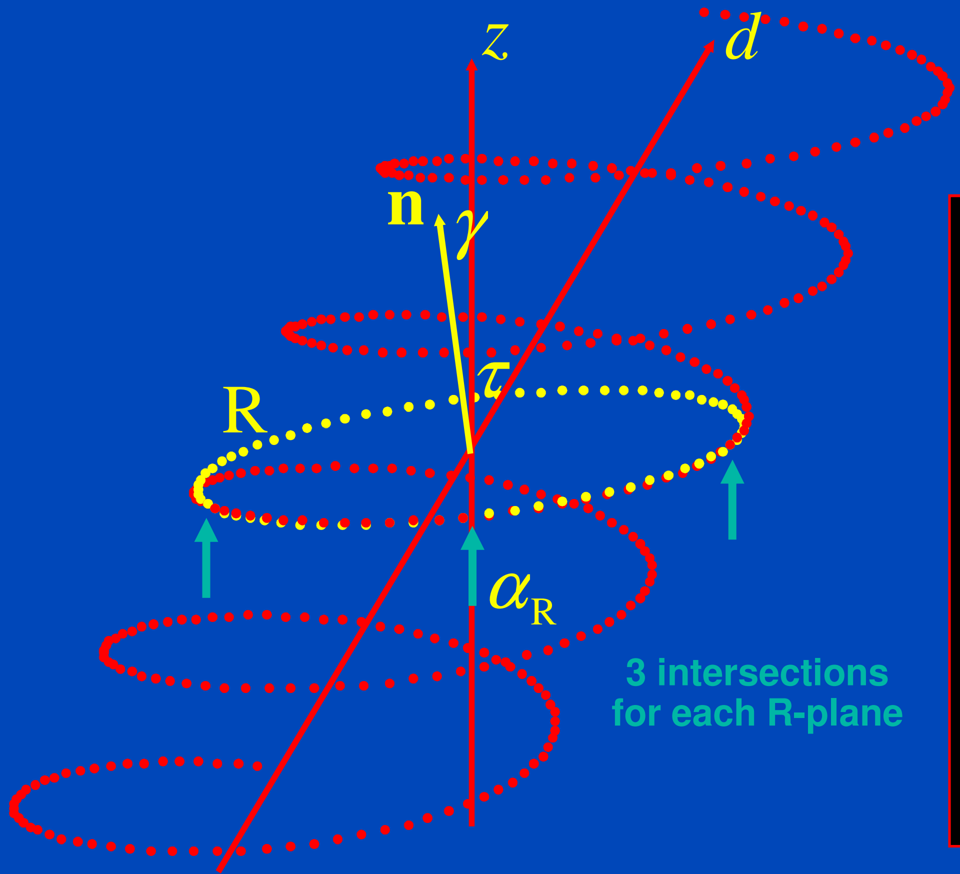
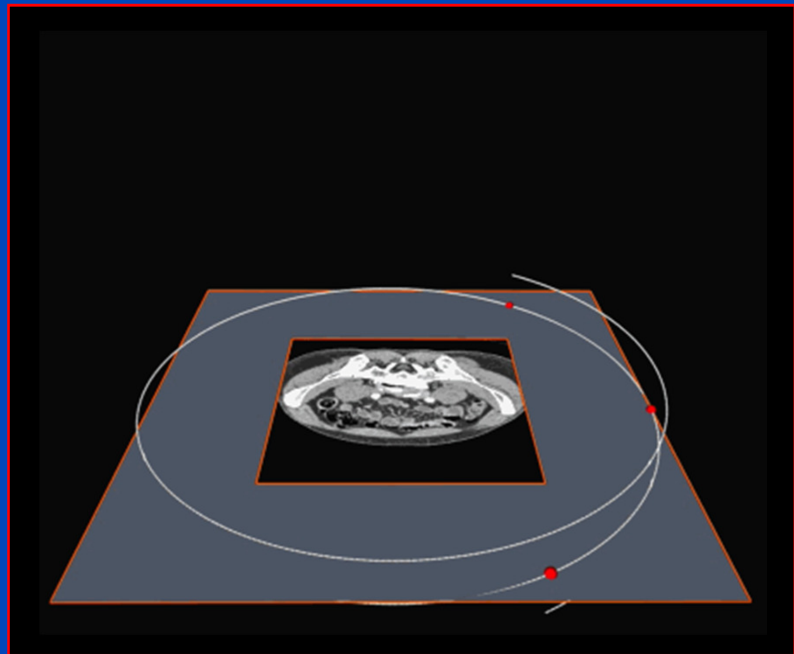


table increment d
Start of spiral scan
Scan trajectory
collimation C
$$\text{pitch } p = \frac{d}{C} = \frac{d}{M \cdot S} = \frac{\text{table increment}}{\text{number of slices} \cdot \text{slice thickness}}$$
0
z
0
t
Direction of continuous patient transport
1996:
1×5 mm, 0.75 s
1998:
4×1 mm, 0.5 s
2002:
16×0.75 mm, 0.42 s
2004:
2·32×0.6 mm, 0.33 s

The ASSR Algorithm



$$p = \frac{d}{MS} \leq 1.5$$

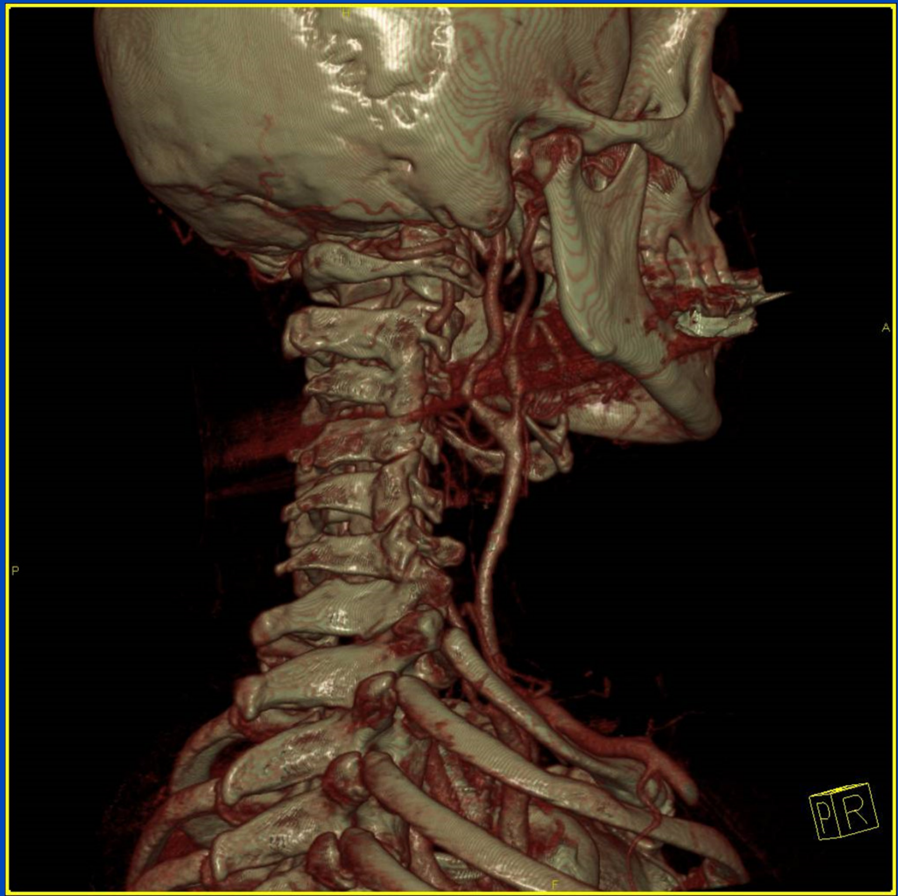


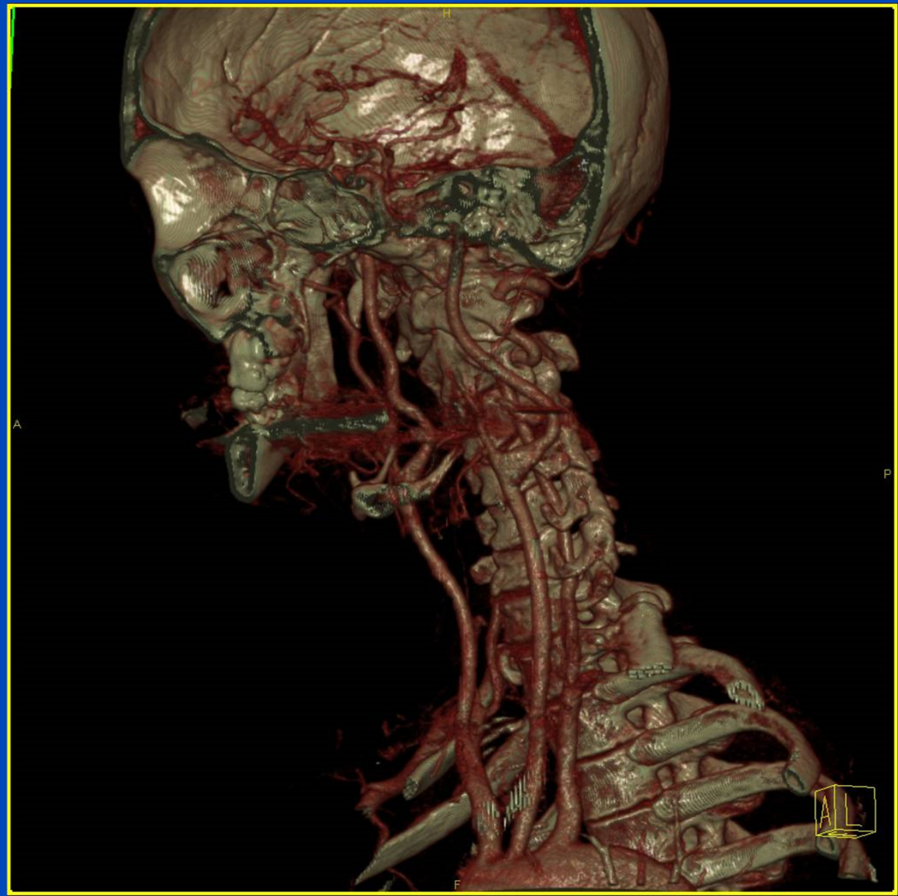
Resulting mean deviation at R_F: $\Delta_\mathrm{mean} \approx 0.014d$

at R_M: $\Delta_\mathrm{mean} \approx 0.007d$

CT-Angiography

Sensation 64 spiral scan with 2·32×0.6 mm and 0.375 s





$$x^2 = y$$

Model

$$x = \sqrt{y}$$

Solution

This is an analytical solution.

Filtered Backprojection[1] (FBP)

Model

Measurement: $p(\vartheta,\xi) = \int dxdy\, f(x,y)\delta(x\cos\vartheta + y\sin\vartheta - \xi)$

Fourier transform:

$$\int d\xi\, p(\vartheta,\xi)e^{-2\pi i\xi u} = \int dxdy\, f(x,y)e^{-2\pi iu(x\cos\vartheta + y\sin\vartheta)}$$

This is the central slice theorem: $P(\vartheta,u) = F(u\cos\vartheta, u\sin\vartheta)$

Inversion: $f(x,y) = \int_0^\pi d\vartheta \int_{-\infty}^{\infty} du\, |u| P(\vartheta,u)e^{2\pi iu(x\cos\vartheta + y\sin\vartheta)}$

$$= \int_0^\pi d\vartheta\, p(\vartheta,\xi) * k(\xi)\Big|_{\xi = x\cos\vartheta + y\sin\vartheta}$$

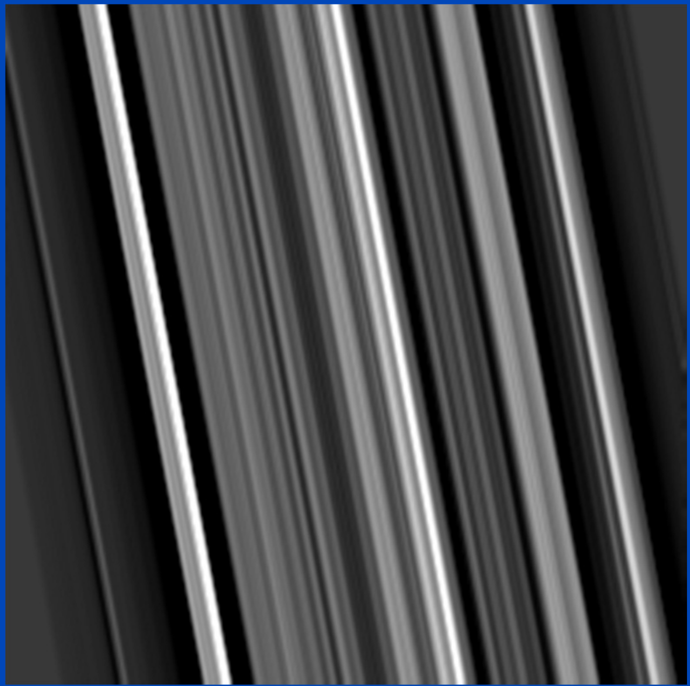
Solution

[1]Ramachandran and Lakshminarayanan. Proc. Nat. Acad. Sci. USA, 1971.

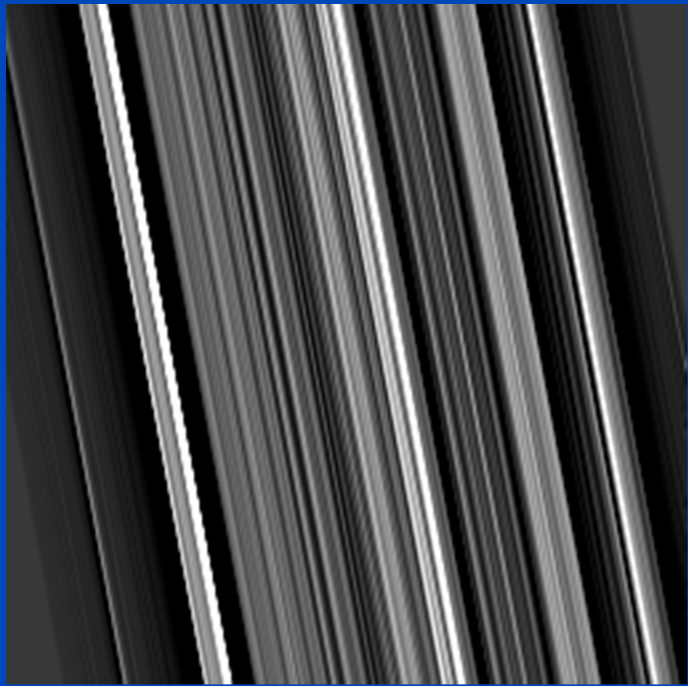
dkfz.

Filtered Backprojection (FBP)

1. Filter projection data with the reconstruction kernel.
2. Backproject the filtered data into the image:



Smooth



Standard

Reconstruction kernels balance between spatial resolution and image noise.

$$x^2 = y$$

Model

$$\cancel{x = \sqrt{y}}$$

$$(x_n + \Delta x_n)^2 = y$$

$$x_n^2 + 2x_n\Delta x_n + \cancel{\Delta x_n^2} = y$$

$$x_n^2 + 2x_n\Delta x_n \approx y$$

$$\Delta x_n = \tfrac{1}{2}(y - x_n^2)/x_n$$

$$x_{n+1} = x_n + \Delta x_n$$

Update equation

This is an iterative solution.

dkfz.

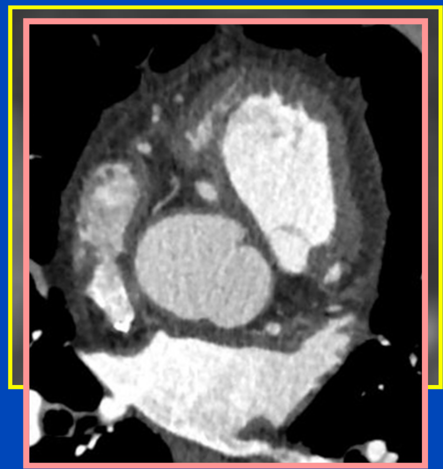
Influence of Update Equation and Model

$0.5\,(3 - x_n^2)/x_n$	$0.4\,(3 - x_n^2)/x_n$	$0.5\,(3 - x_n^{2.1})/x_n$
$x_0 = 1.$	$x_0 = 1.$	$x_0 = 1.$
$x_1 = 2.$	$x_1 = 1.8$	$x_1 = 2.$
$x_2 = 1.75$	$x_2 = 1.74667$	$x_2 = 1.67823$
$x_3 = 1.73214$	$x_3 = 1.73502$	$x_3 = 1.68833$
$x_4 = 1.73205$	$x_4 = 1.73265$	$x_4 = 1.68723$
$x_5 = 1.73205$	$x_5 = 1.73217$	$x_5 = 1.68734$
$x_6 = 1.73205$	$x_6 = 1.73207$	$x_6 = 1.68733$
$x_7 = 1.73205$	$x_7 = 1.73206$	$x_7 = 1.68733$
$x_8 = 1.73205$	$x_8 = 1.73205$	$x_8 = 1.68733$

$$x^2 = 3, \quad x_0 = 1, \quad x_{n+1} = x_n + \Delta x_n$$

Iterative Reconstruction

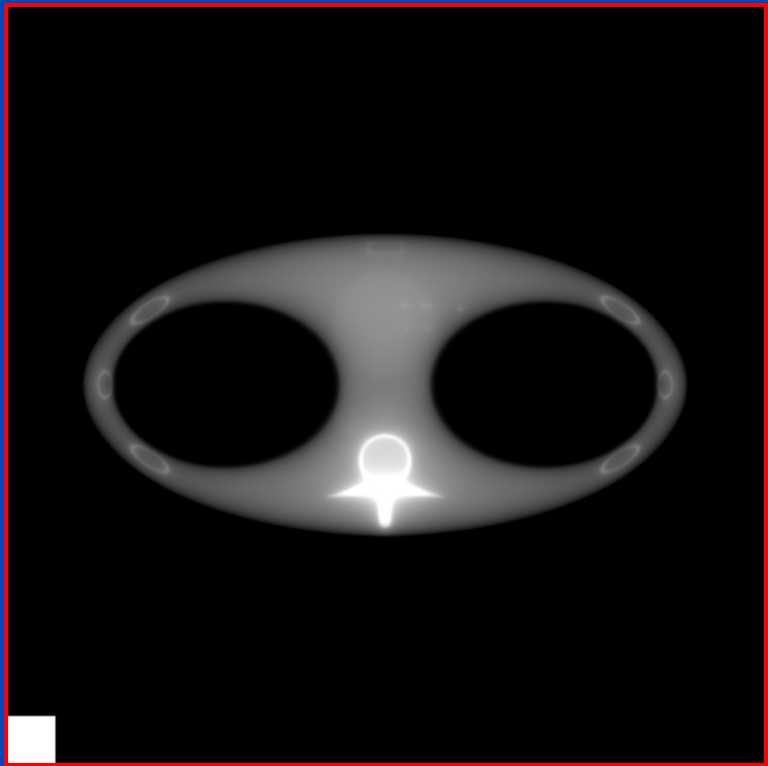
- Aim: less artifacts, lower noise, lower dose
- Iterative reconstruction
  – Reconstruct an image.
  – Regularize the image.
  – Does the image correspond to the rawdata?
  – If not, reconstruct a correction image and continue.
- SPECT + PET are iterative for a long time!
- Until recently, the computational demand prohibited to use iterative recon in CT.
- First CT product implementations
  – AIDR (adaptive iterative dose recuction, Toshiba)
  – ASIR (adaptive statistical iterative reconstruction, GE)
  – iDose (Philips)
  – IRIS (image reconstruction in image space, Siemens)
  – VEO, MBIR (model-based iterative reconstruction, GE)
  – SAFIRE (sinogram-affirmed iterative reconstruction, Siemens)

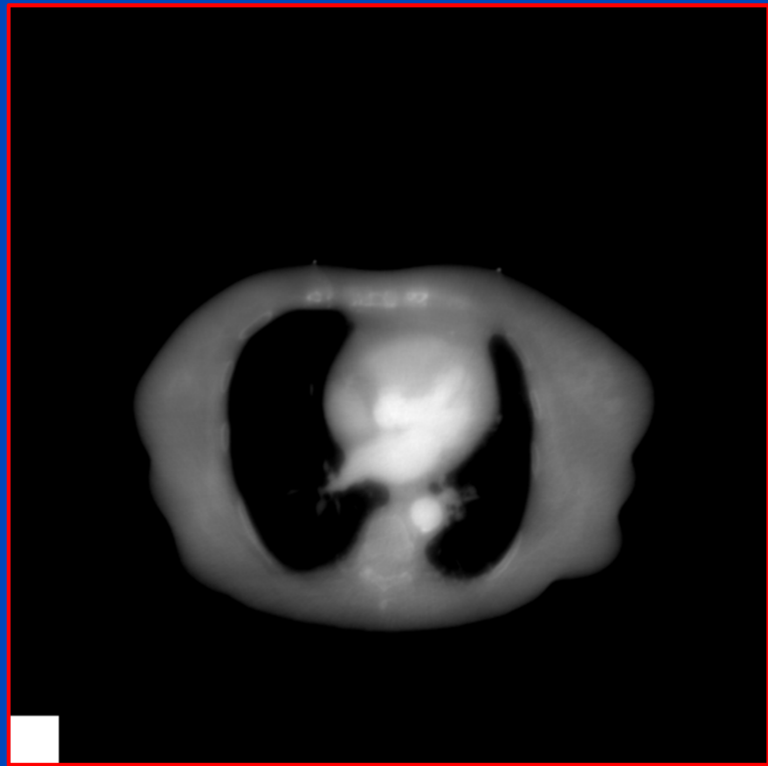


Flavours of Iterative Reconstruction

- **ART** $$\boldsymbol{f}_{\nu+1} = \boldsymbol{f}_\nu + \boldsymbol{R}^{\mathrm{T}} \cdot \frac{\boldsymbol{p} - \boldsymbol{R}\cdot\boldsymbol{f}_\nu}{\boldsymbol{R}^2 \cdot \boldsymbol{1}}$$
- **SART** $$\boldsymbol{f}_{\nu+1} = \boldsymbol{f}_\nu + \frac{1}{\boldsymbol{R}^{\mathrm{T}} \cdot \boldsymbol{1}} \boldsymbol{R}^{\mathrm{T}} \cdot \frac{\boldsymbol{p} - \boldsymbol{R}\cdot\boldsymbol{f}_\nu}{\boldsymbol{R} \cdot \boldsymbol{1}}$$
- **MLEM** $$\boldsymbol{f}_{\nu+1} = \boldsymbol{f}_\nu \frac{\boldsymbol{R}^{\mathrm{T}} \cdot \left(e^{-\boldsymbol{R}\cdot\boldsymbol{f}_\nu}\right)}{\boldsymbol{R}^{\mathrm{T}} \cdot \left(e^{-\boldsymbol{p}}\right)}$$
- **OSC** $$\boldsymbol{f}_{\nu+1} = \boldsymbol{f}_\nu + \boldsymbol{f}_\nu \frac{\boldsymbol{R}^{\mathrm{T}} \cdot \left(e^{-\boldsymbol{R}\cdot\boldsymbol{f}_\nu} - e^{-\boldsymbol{p}}\right)}{\boldsymbol{R}^{\mathrm{T}} \cdot \left(e^{-\boldsymbol{R}\cdot\boldsymbol{f}_\nu}\boldsymbol{R}\cdot\boldsymbol{f}_\nu\right)}$$
- **and hundreds more …**

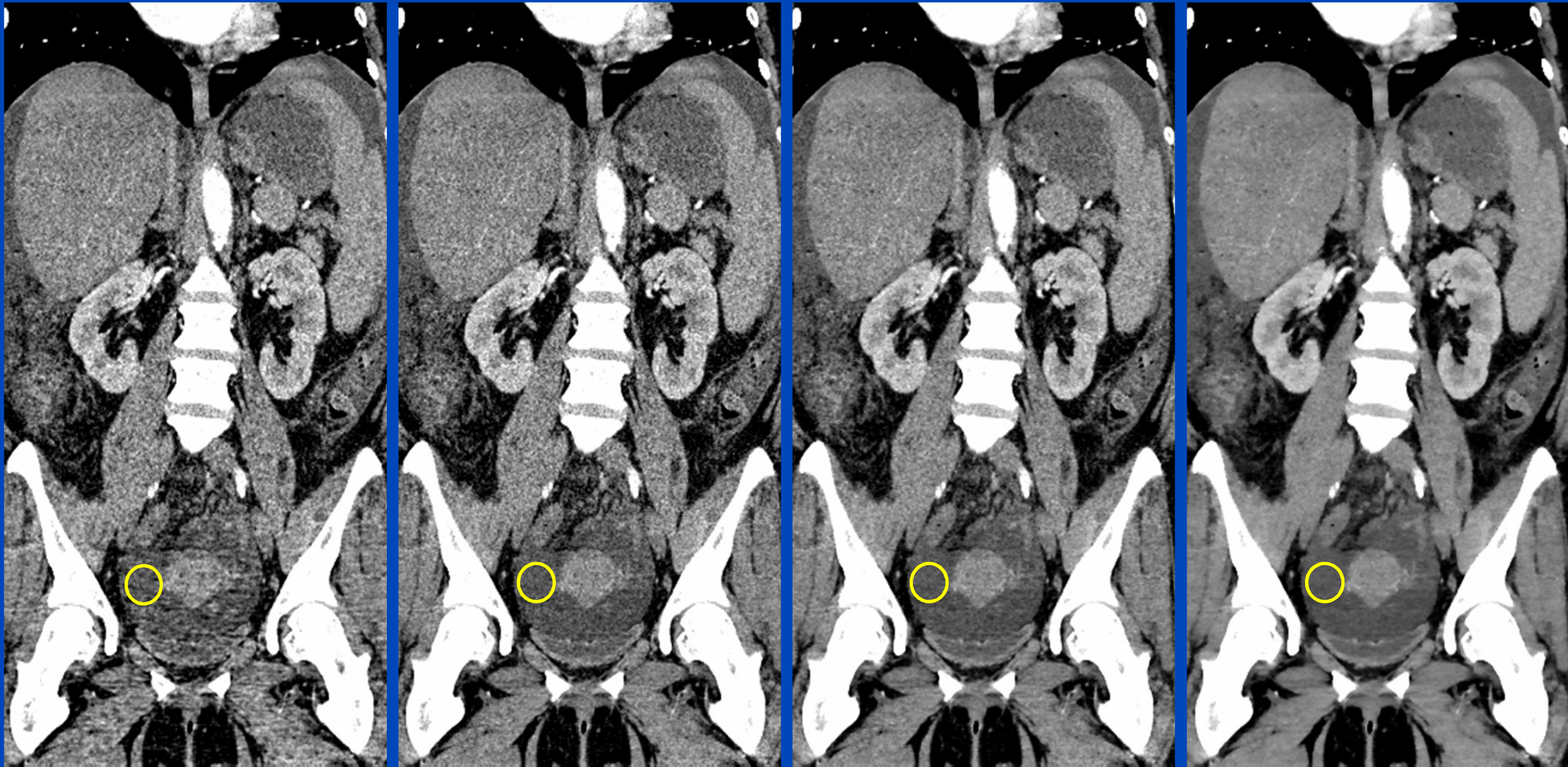
Iterative Reconstruction



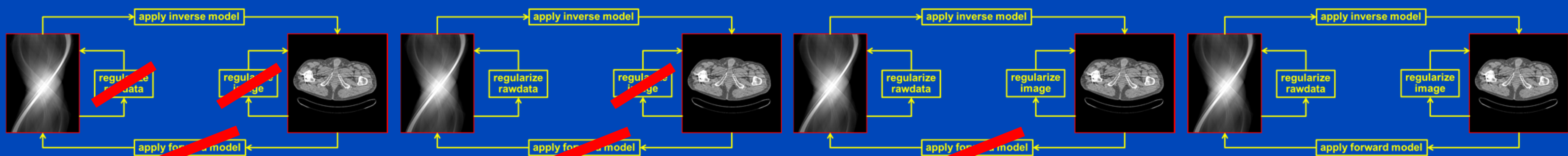


16 ordered subsets iterations

Plain FBP | **Siemens Standard** | **IRIS VA34** | **SAFIRE VA40**



σ = 26.8 HU | σ = 17.6 HU | σ = 12.3 HU | σ = 7.8 HU

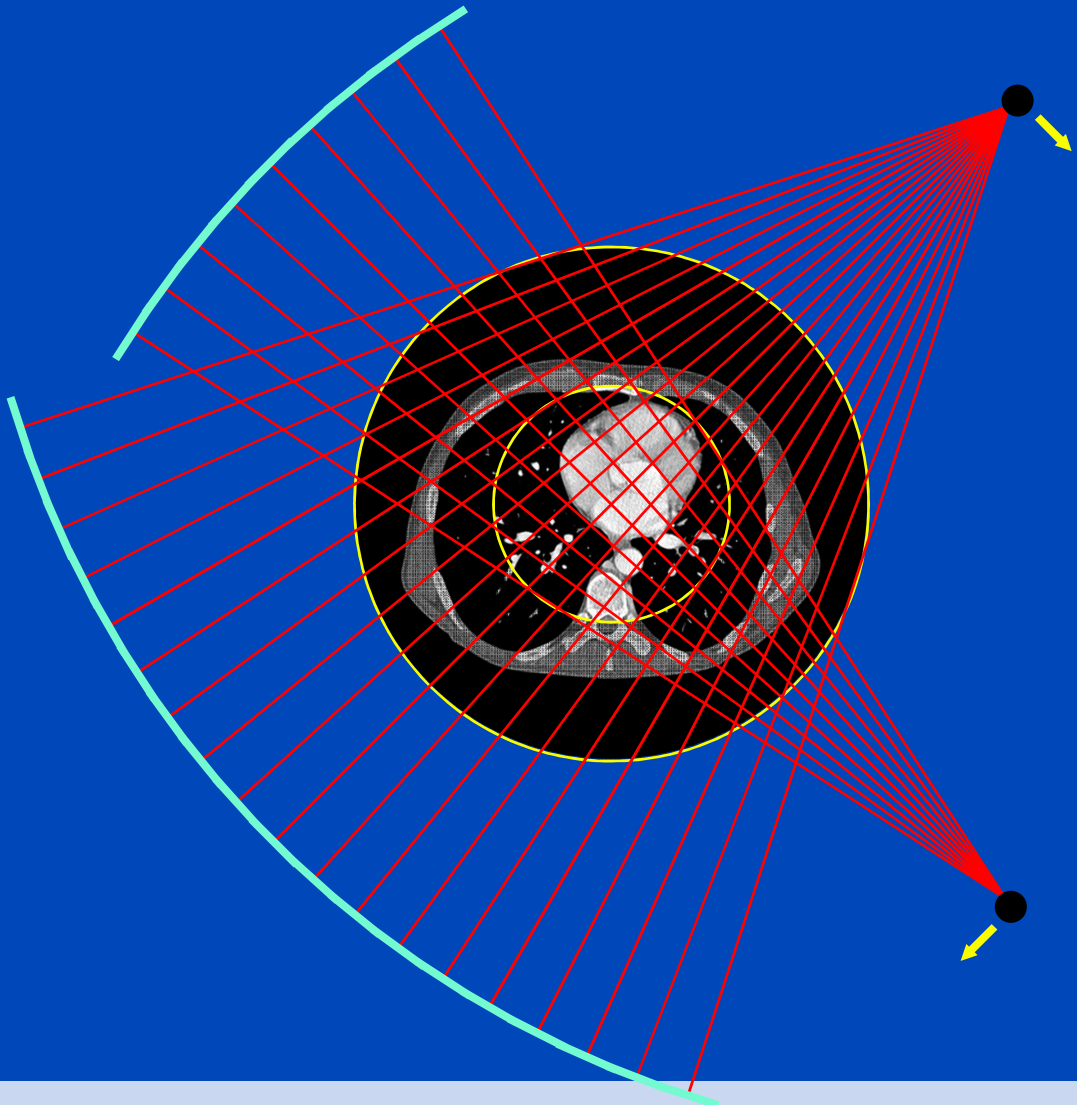


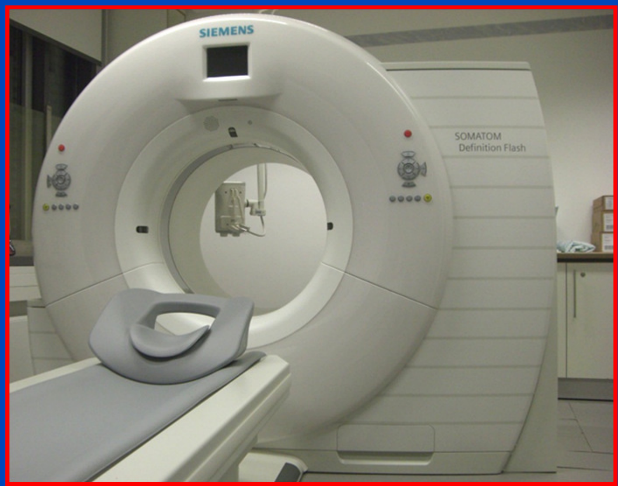
CT images provided by Siemens Healthcare, Forchheim, Germany

dkfz.

Dual Energy CT

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



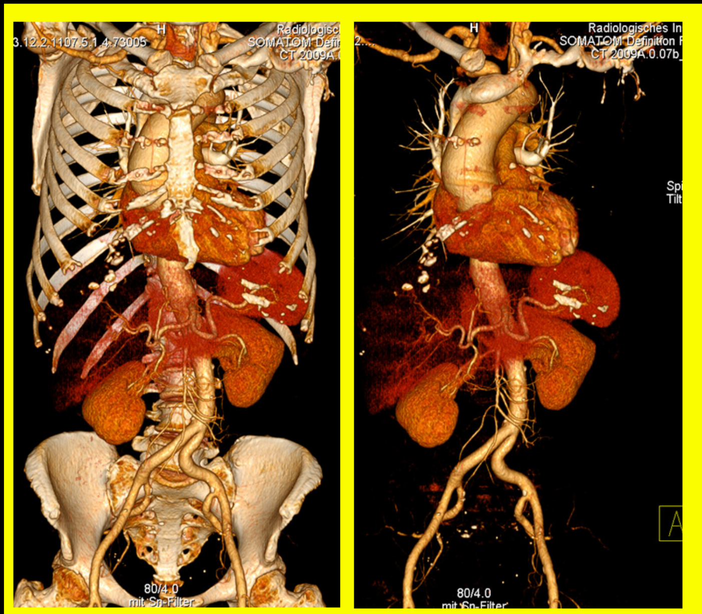


Siemens SOMATOM Definition Flash
dual source cone-beam spiral CT scanner

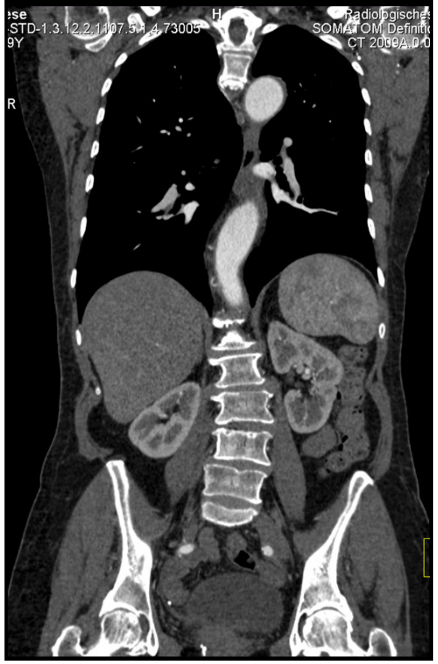
Examples

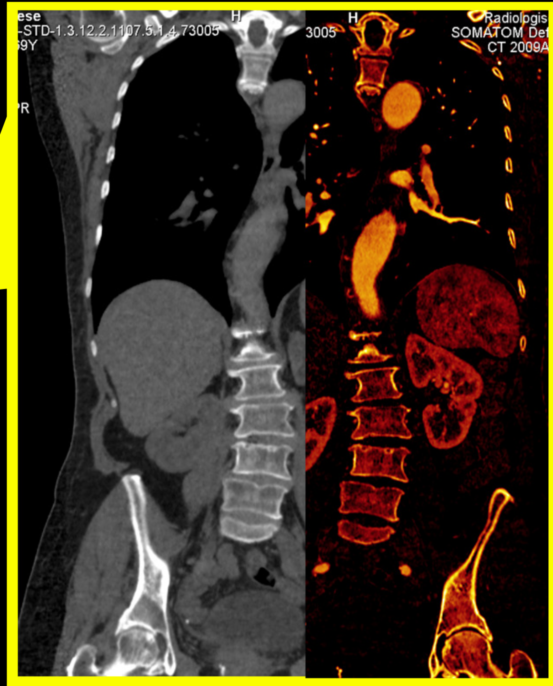
(Slide Courtesy of Siemens Healthcare)

DE bone removal



Single DECT Scan





Virtual non-contrast and Iodine image

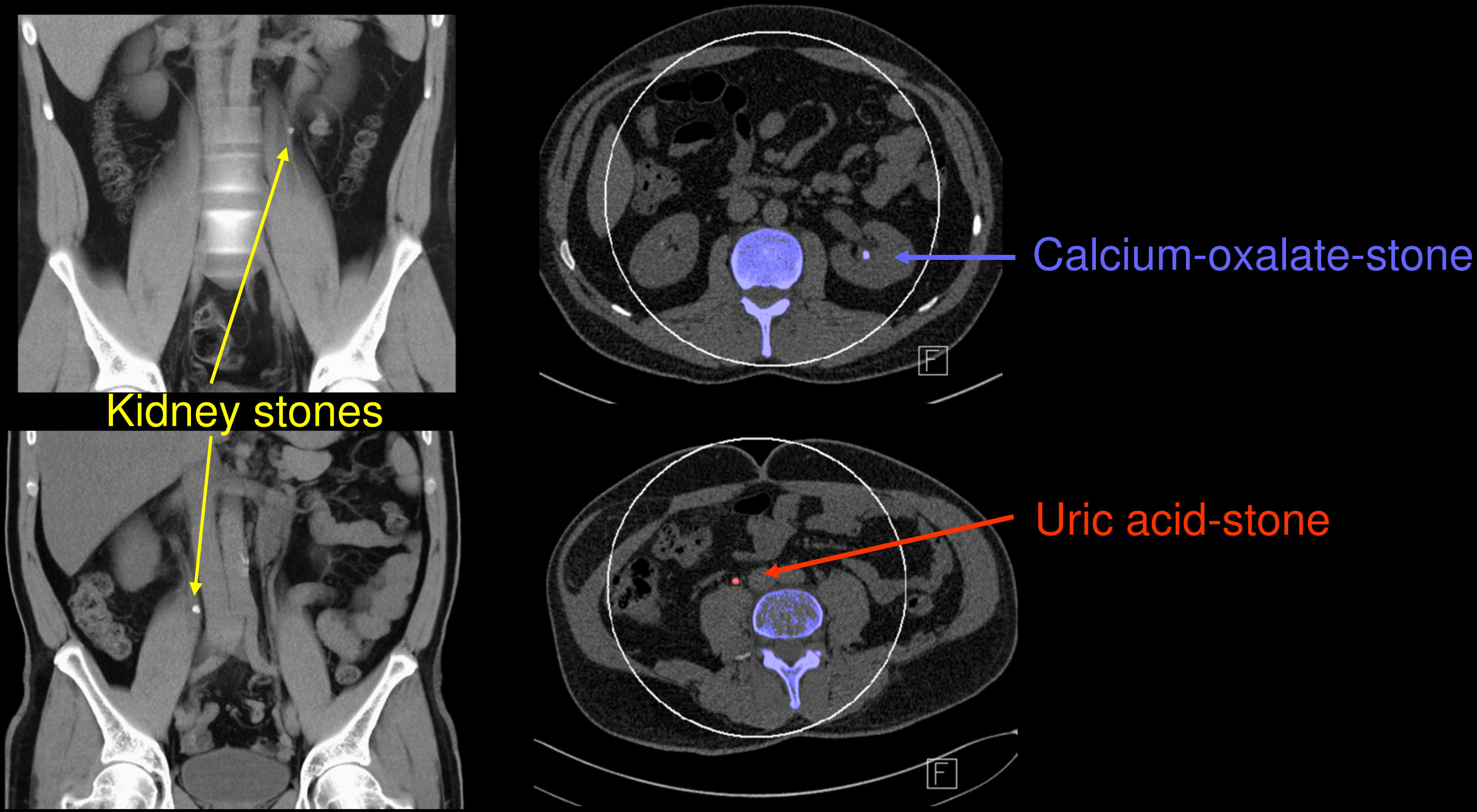
Dual Energy whole body CTA: 100/140 Sn kV @ 0.6mm

Courtesy of Friedrich-Alexander-University Erlangen-Nürnberg

DECT Today: Widely Available via DSCT

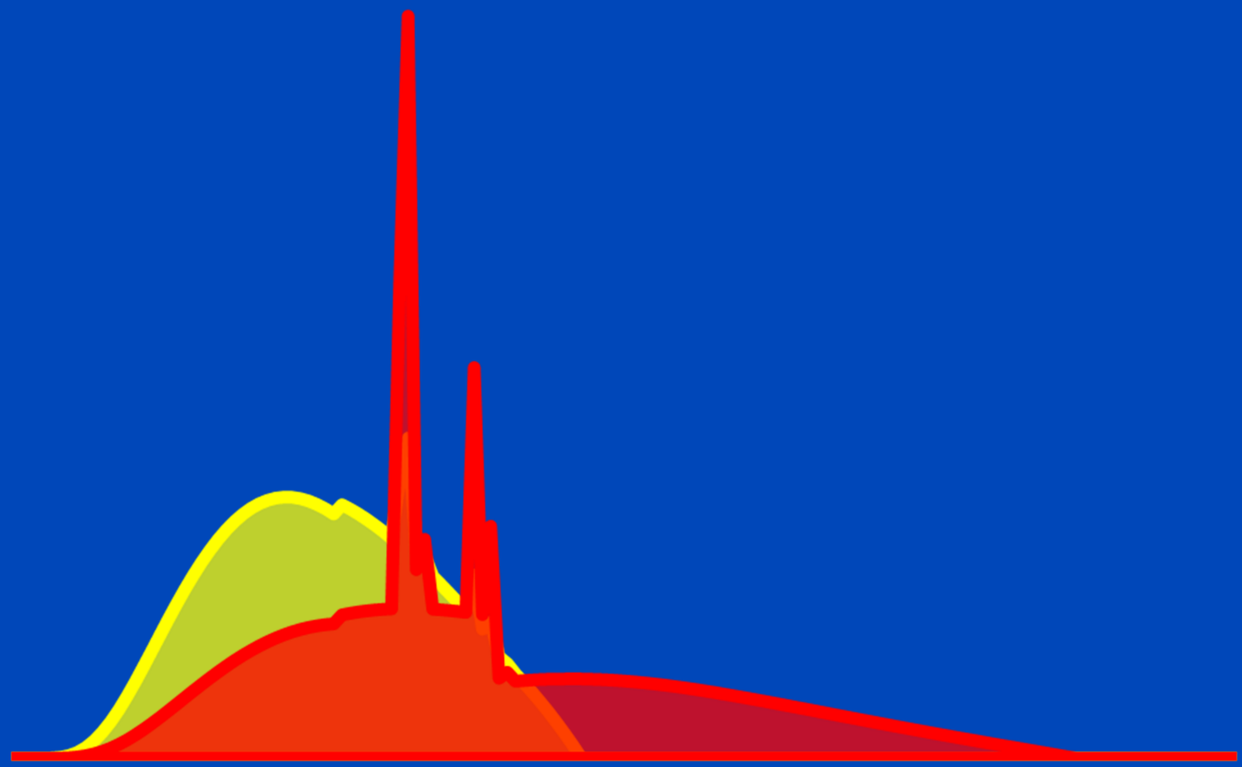
(Slide Courtesy of Siemens Healthcare)

- **“Spectroscopy“: more specific tissue characterization**
  → Detection and visualization of calcium, iron, uric acid, …..

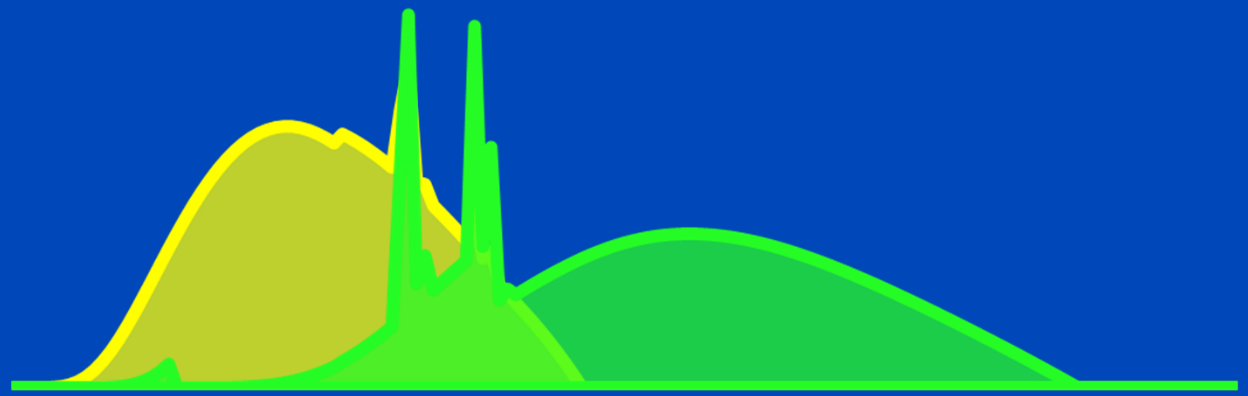


Courtesy of Klinikum Großhadern, LMU München

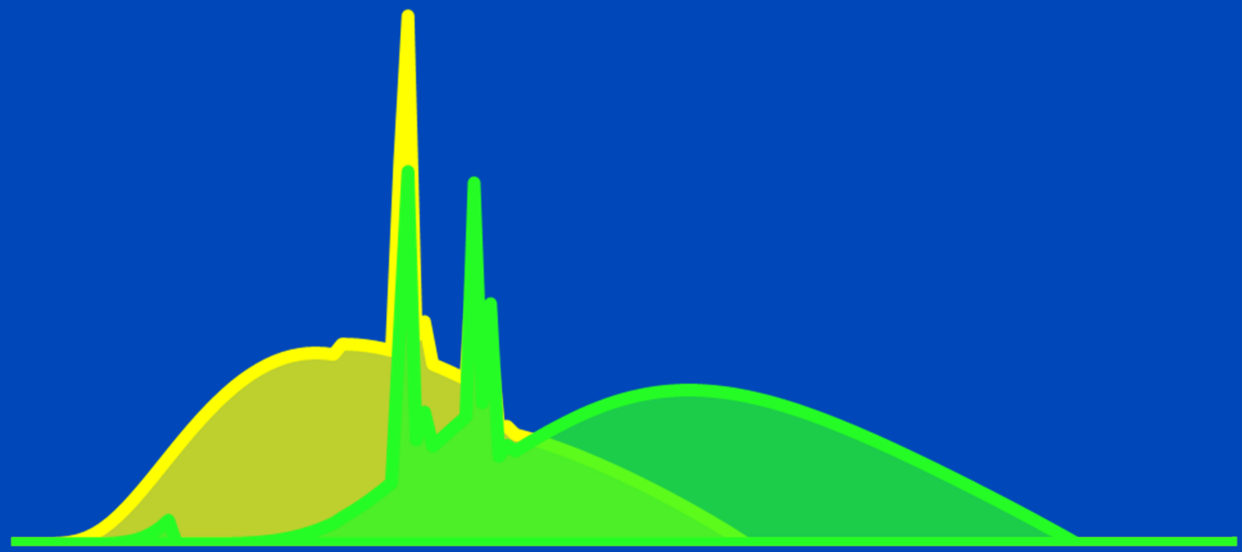
80 kV / 140 kV



80 kV / 140 kV Sn

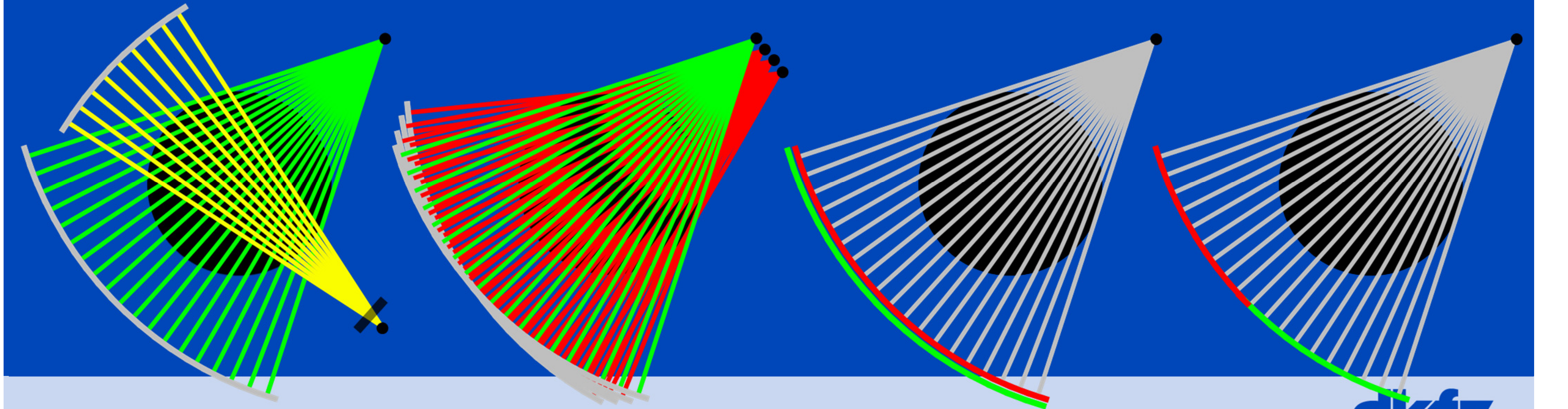


100 kV / 140 kV Sn



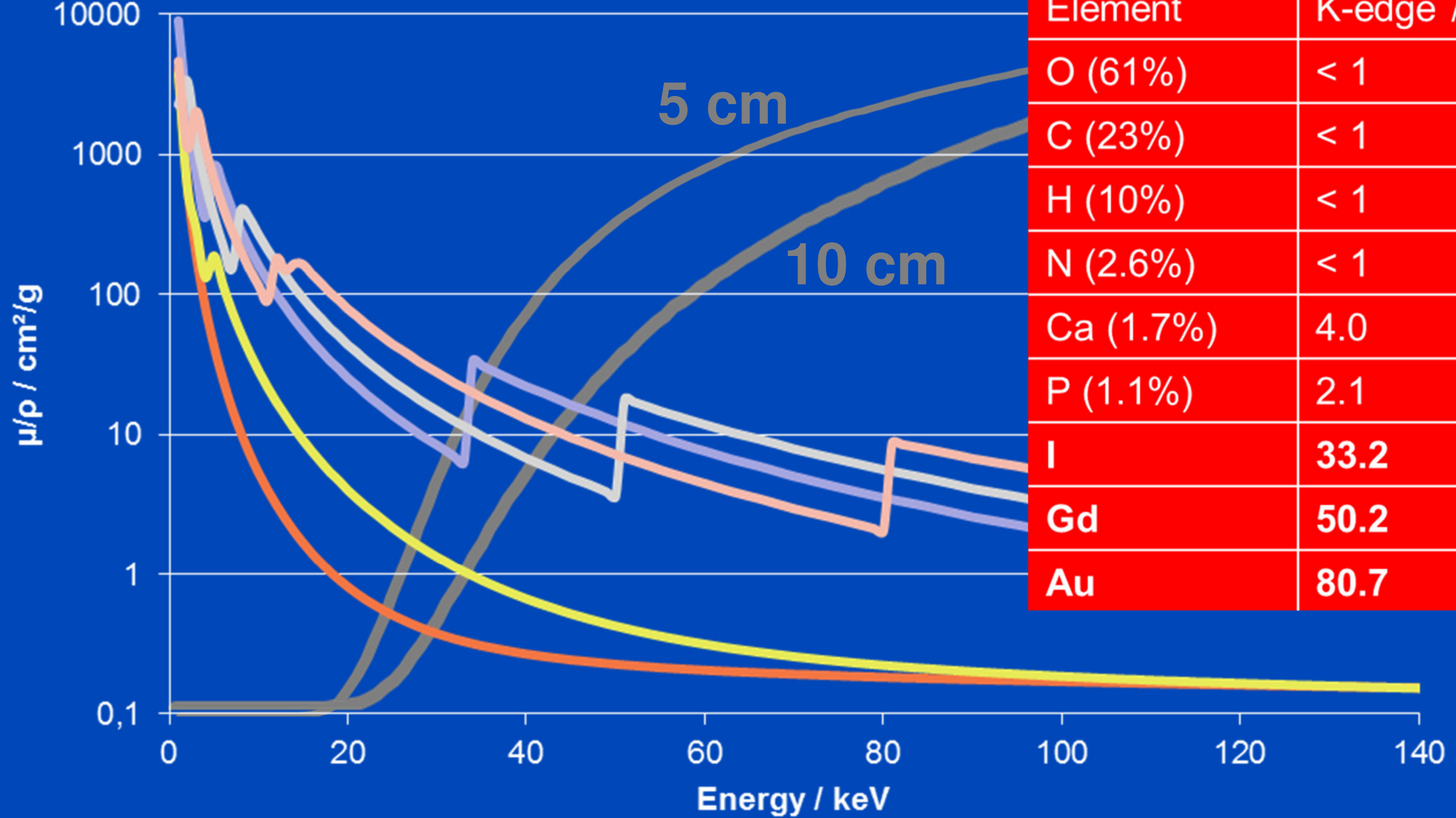
Technology Approaches

- Multiple scans at different spectra
- Dual source CT
- Fast tube voltage switching
- Slow tube voltage modulation
- Dual layer detectors (sandwich detectors)
- Split detector (different prefiltration)
- Photon counting detectors (two or more energy bins)



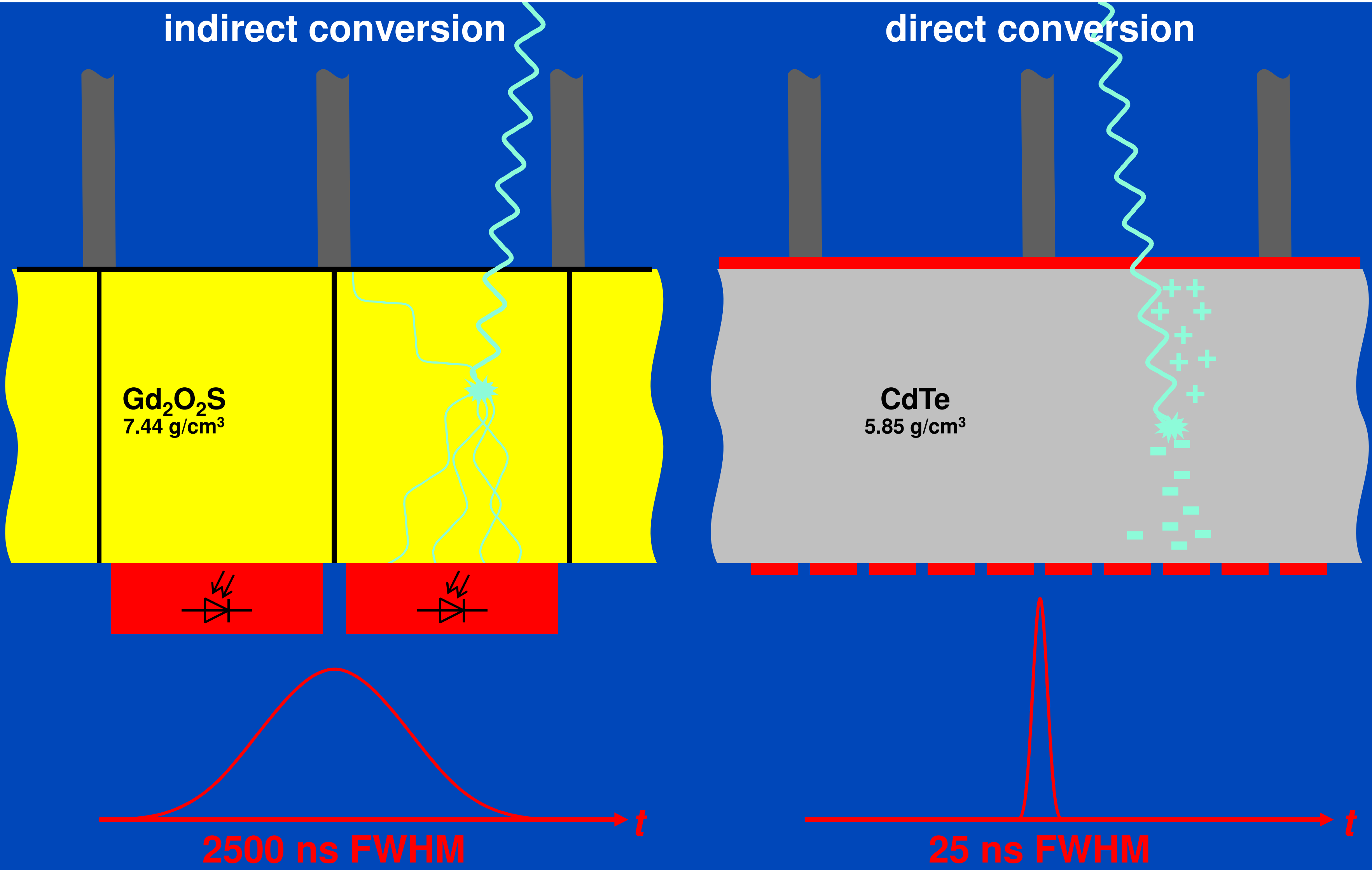
More than Dual Energy CT?

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



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

120 kV water transmission curves (gray) given in relative units on a non-logarithmic ordinate.



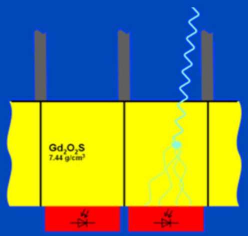
i.e. max O(40·10³) cps

i.e. max O(40·10⁶) cps

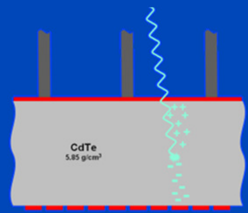
Requirements for CT: up to 10^9 x-ray photon counts per second per mm^2.
Hence, photon counting only achievable for direct converters.

dkfz.

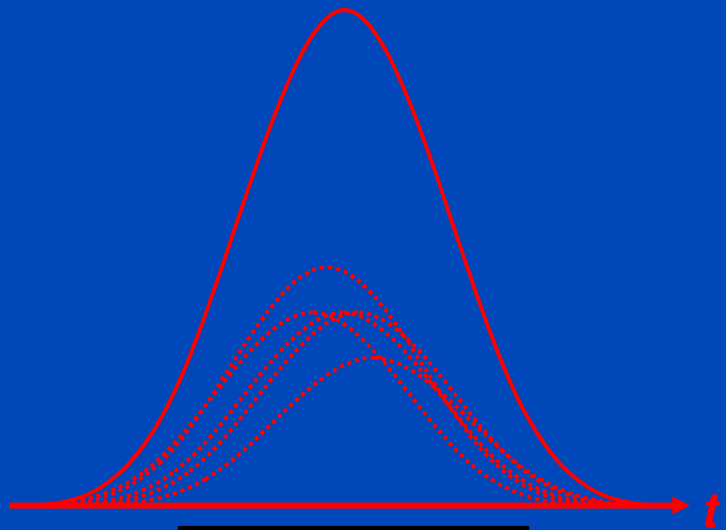
Integrating vs. Photon Counting Detector Technology

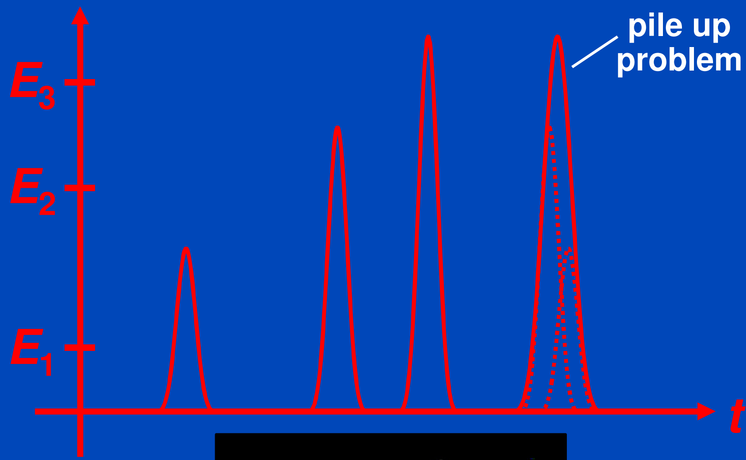


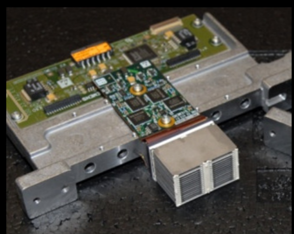
indirect conversion
= pile up only
= energy integrating



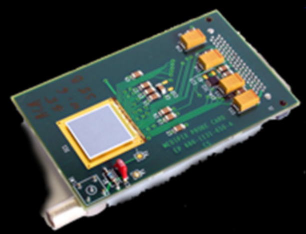
direct conversion
= photon counting
= energy-selective







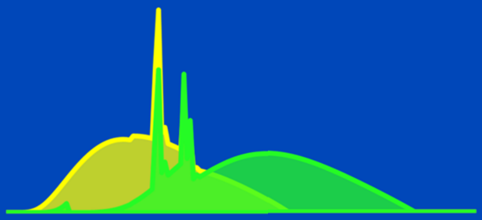
CT detector

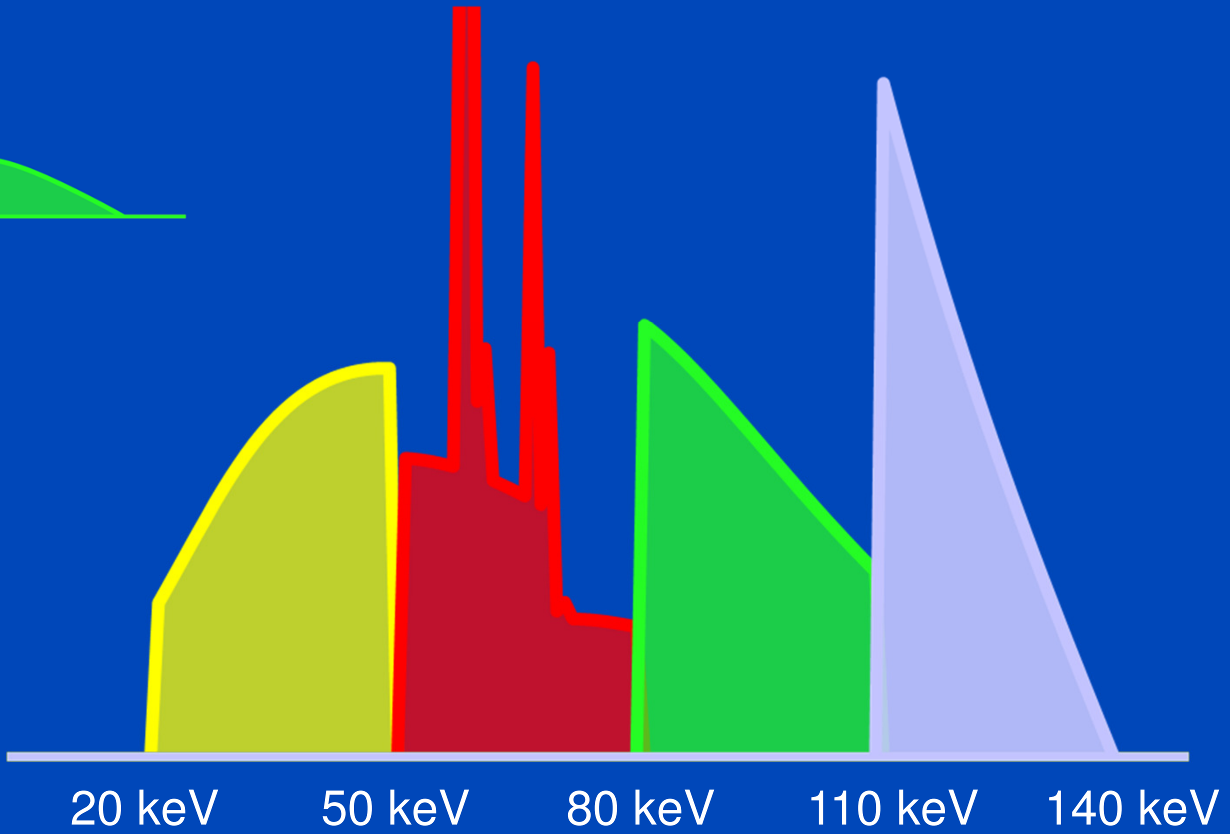


MedipPix detector

Energy-Selective Detectors: Lower Dose? Improved Spectroscopy?

Conventional dual energy CT





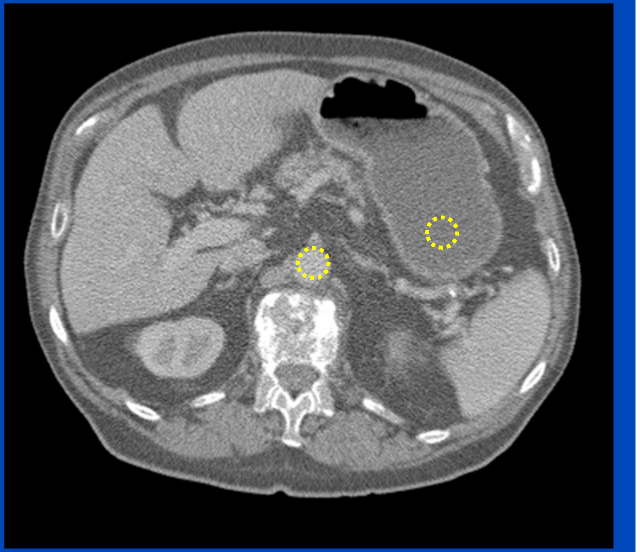
Photon Counting used to Maximize CNR

- To optimize the CNR the optimal bin weighting factor is given by (weighting after log):

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

- The resulting CNR is

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



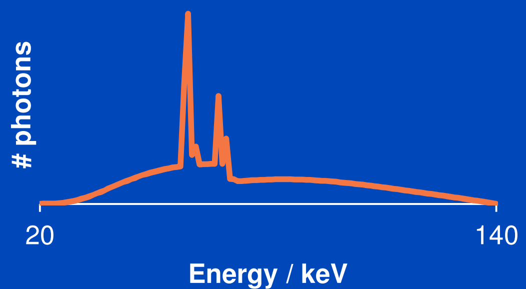
- At the optimum this evaluates to

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

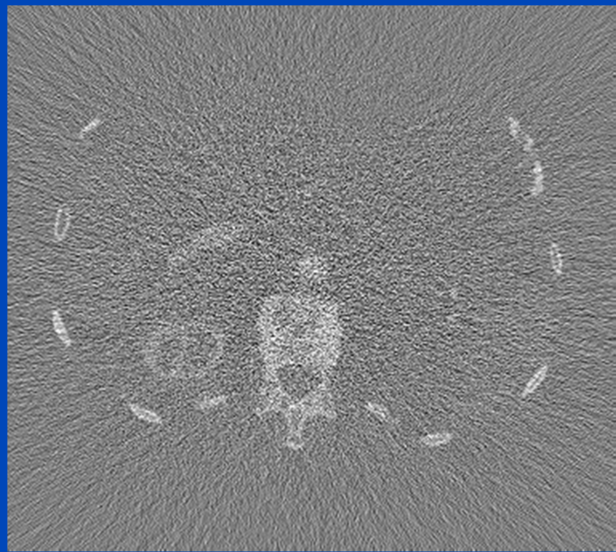
Energy Integrating vs. Photon Counting
with counting starting at 20 keV

Energy Integrating

CNR = 2.11

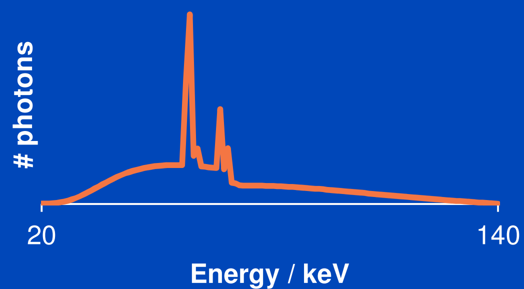


PC minus EI



Photon Counting

CNR = 2.95



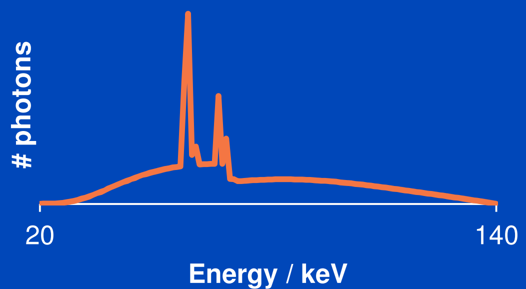
40% CNR improvement or
49% dose reduction achievable.

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

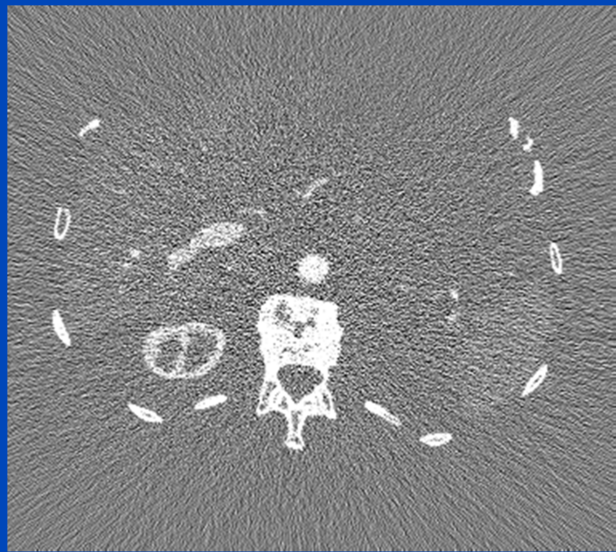
Energy Integrating vs. Photon Counting
with 4×30 keV wide Gaussian bins

Energy Integrating

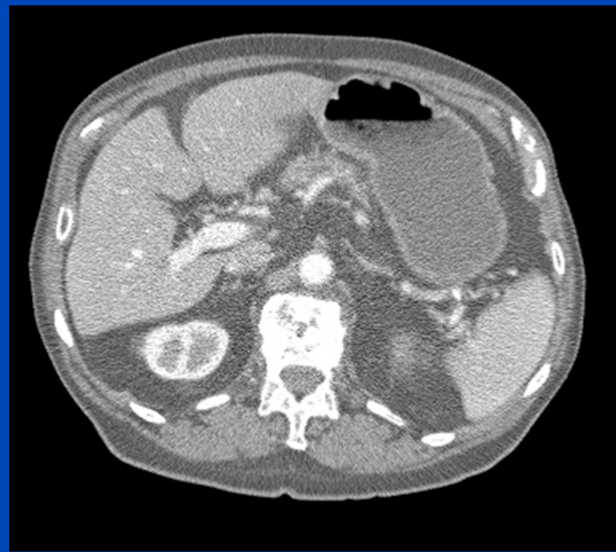
CNR = 2.11



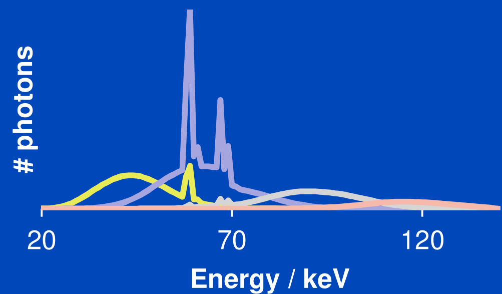
PC minus EI



Photon Counting



CNR = 4.19



99% CNR improvement or 75% dose reduction achievable.

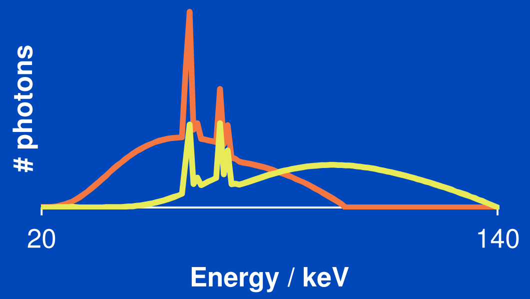
Images: C = 0 HU, W = 700 HU, difference image: C = 0 HU W = 350 HU, bins start at 20 keV

Photon Counting used for Spectral Imaging

- DECT scan with 100 kV / 140 kV Sn
- Photon counting acquisition at 140 kV
- Same patient dose in both cases

100 kV

140 kV Sn



Images: *C* = 0 HU, *W* = 700 HU

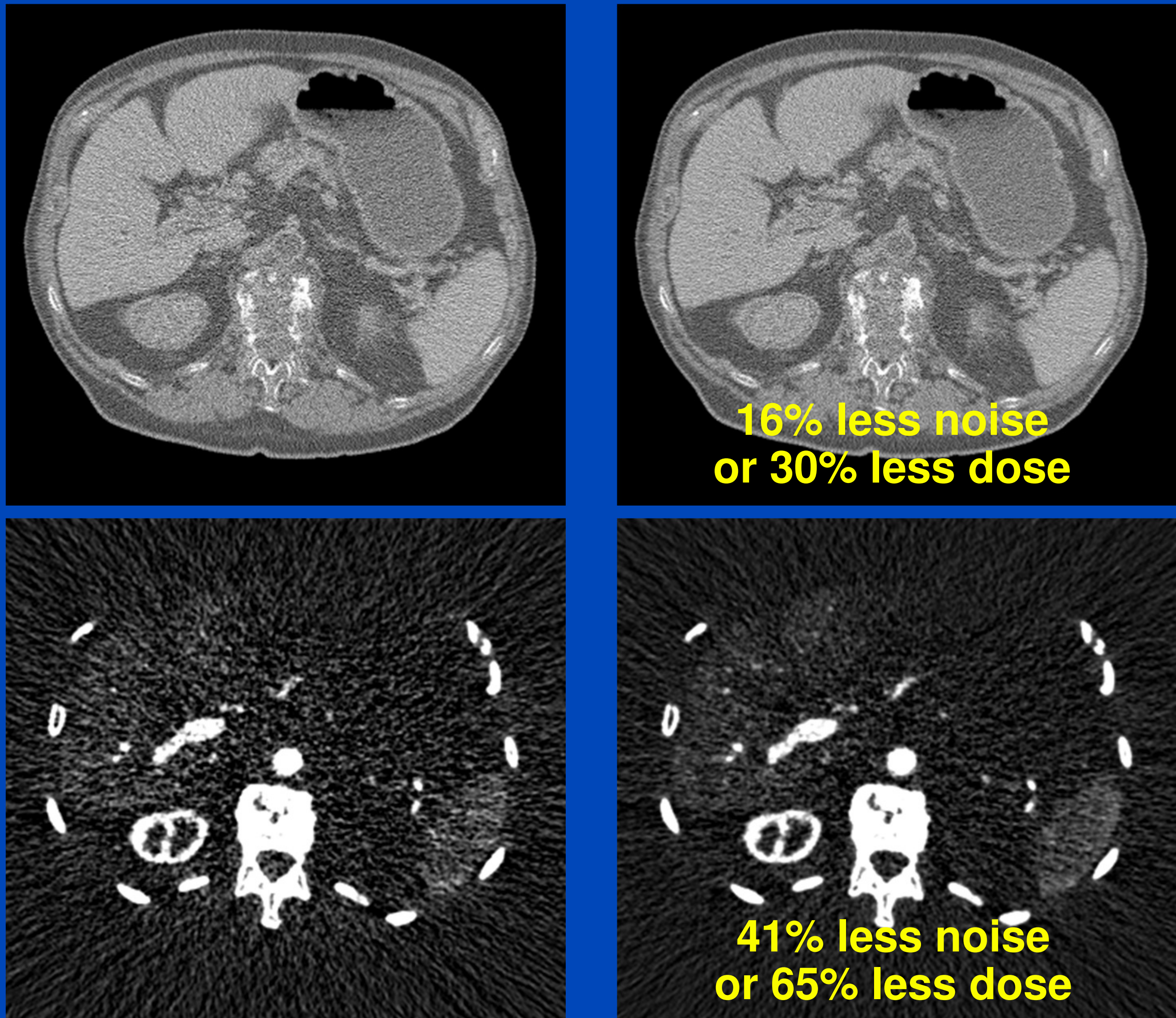
Energy Integrating vs. Photon Counting

Energy Integrating
100 kV / 140 kV Sn

Photon Counting 140 kV
4×30 keV Gaussian bins

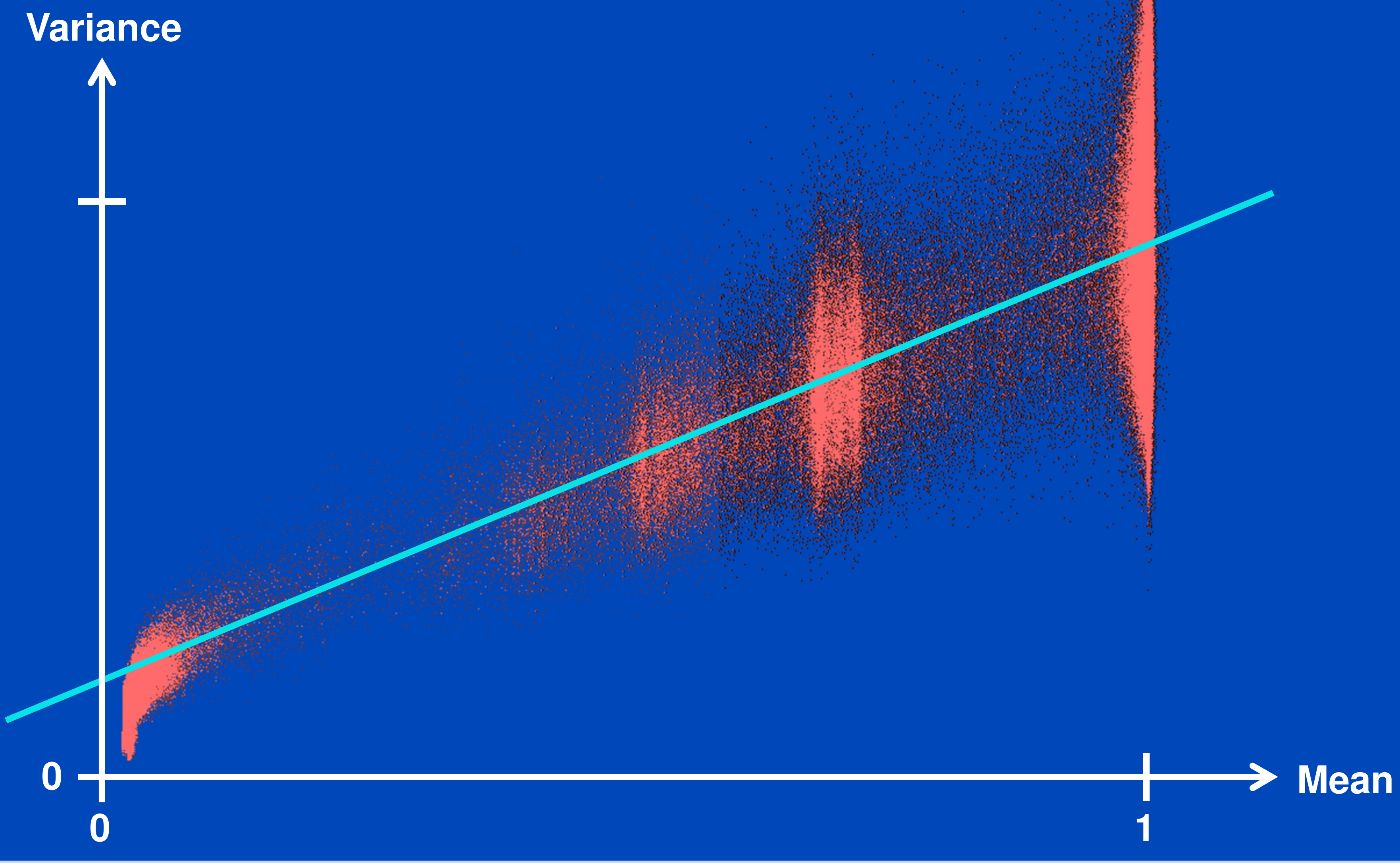
Water

Iodine



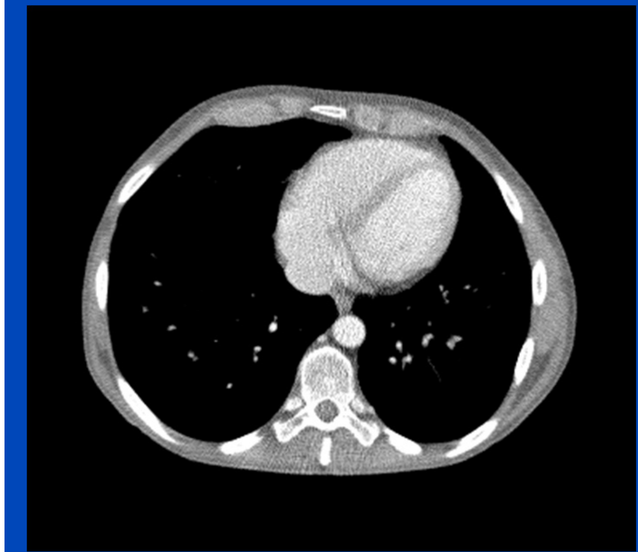
Water image: *C* = 0 HU, *W* = 700 HU, iodine image: *C* = 0 HU, *W* = 2000 HU, bins start at 20 keV

Mean vs. Variance Plot



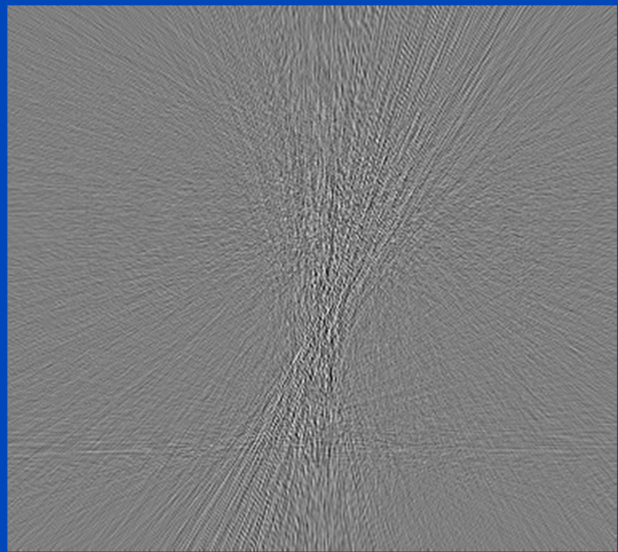
CT with and without Electronic Noise

Without electronic noise

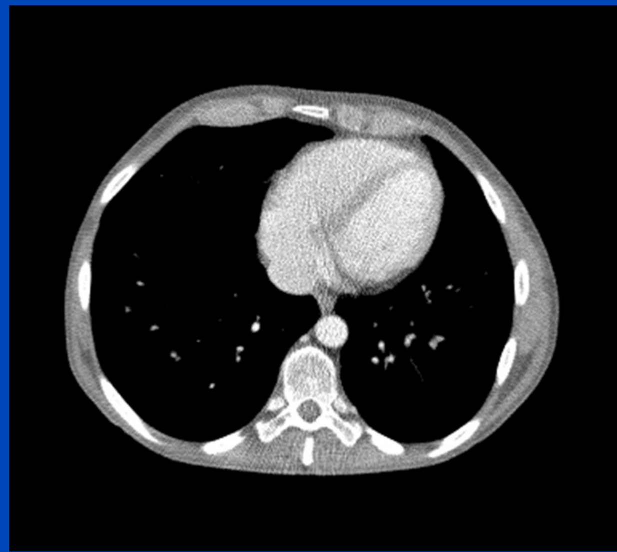


Noise: 82 HU

Difference



With electronic noise



Noise 85 HU

Images: *C* = 0 HU, *W* = 700 HU, difference image: *C* = 0 HU *W* = 100 HU


Thank You!
marc.kachelriess@dkfz.de