From X-Rays to CT Images

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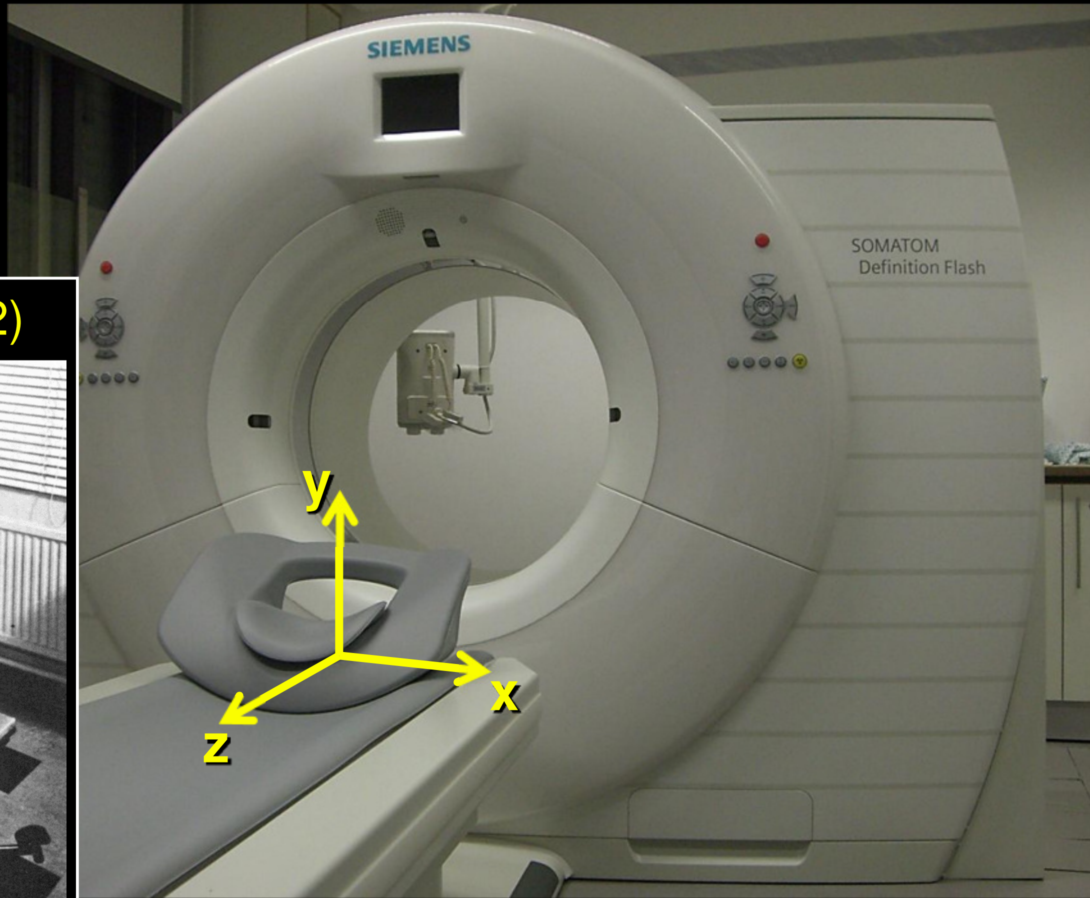
DEUTSCHES KREBSFORSCHUNGSZENTRUM IN DER HELMHOLTZ-GEMEINSCHAFT

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

dkfz.

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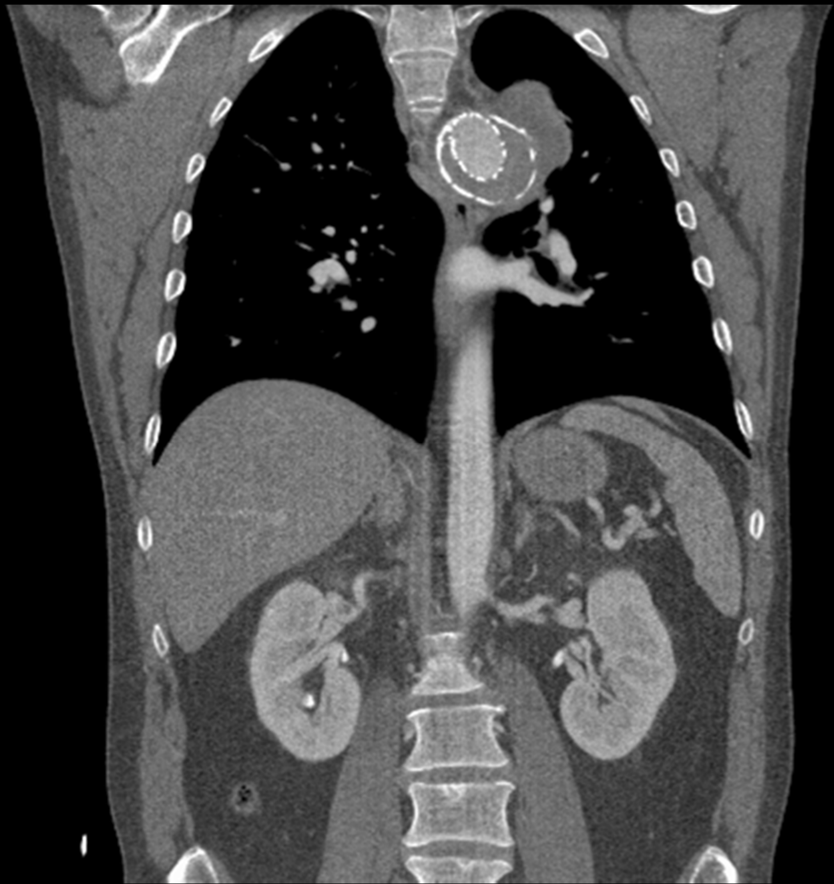
dual source cone-beam spiral **CT**(2008)



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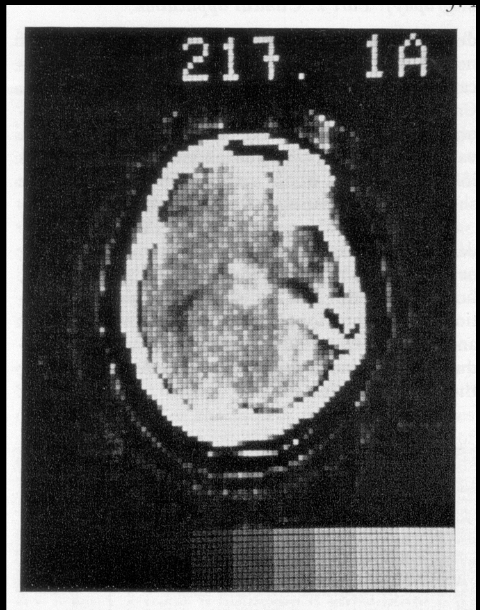

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2×160 positions per view

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113 kB data size

GE Discovery

Toshiba Aquilion ONE Vision

Philips Brilliance iCT

Siemens Definition Flash

Wieviele Schichten akquiriert der von Ihnen genutzte CT-Scanner?

1. Bis zu 4 Schichten
2. Bis zu 16 Schichten
3. Bis zu 64 Schichten
4. Bis zu 128 Schichten
5. Mehr als 128 Schichten

TED

What does CT Measure?

- **Polychromatic Radon transform**

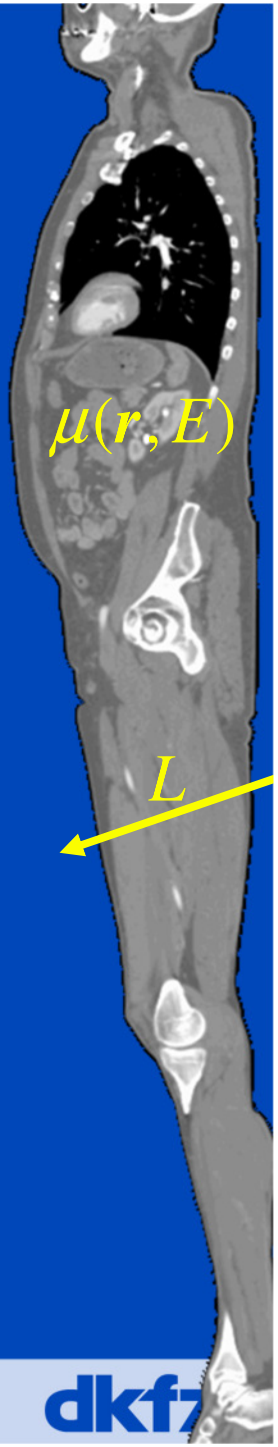
$$p(L) = -\ln \int dE\, w(E)\, e^{-\int dL\, \mu(\boldsymbol{r}, E)}$$

with normalized detected spectrum: $1 = \int dE\, w(E)$

- **Widely used monochromatic approximation:**

$$p(L) \approx \int dL\, \mu(\boldsymbol{r}, E_{\text{eff}})$$

with the effective energy being around 70 keV



Tube Technology

conventional tube
(rotating anode, helical wire emitter)

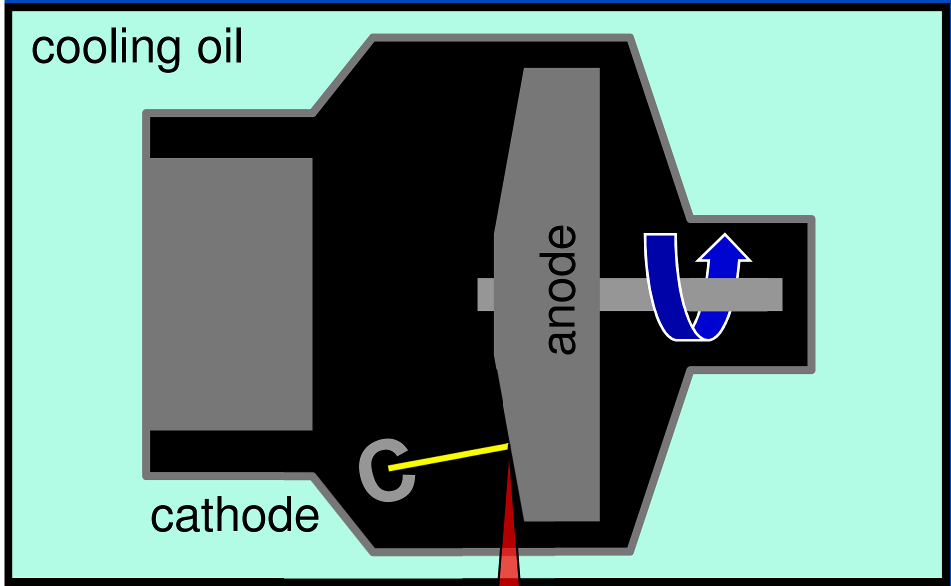


Photo courtesy of GE

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

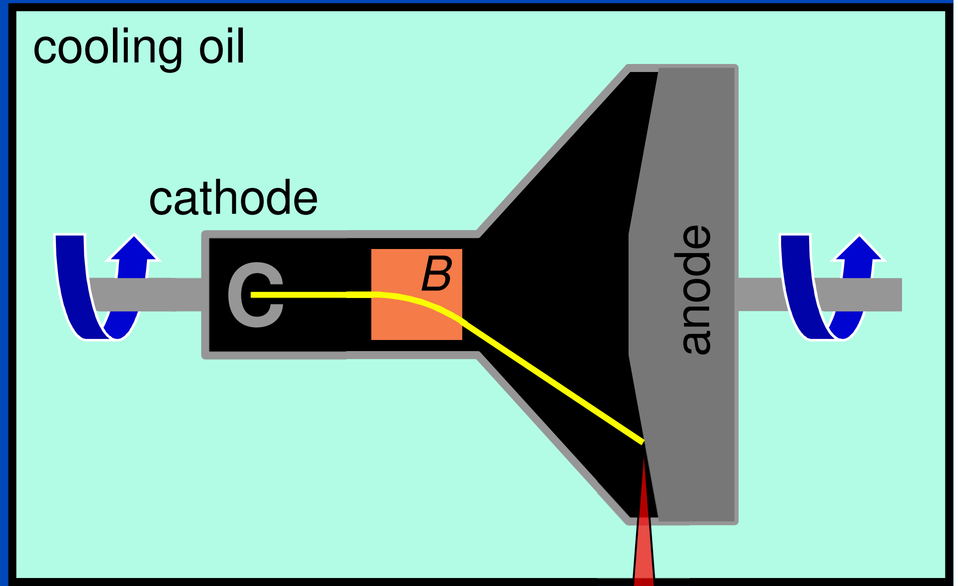


Photo courtesy of Siemens

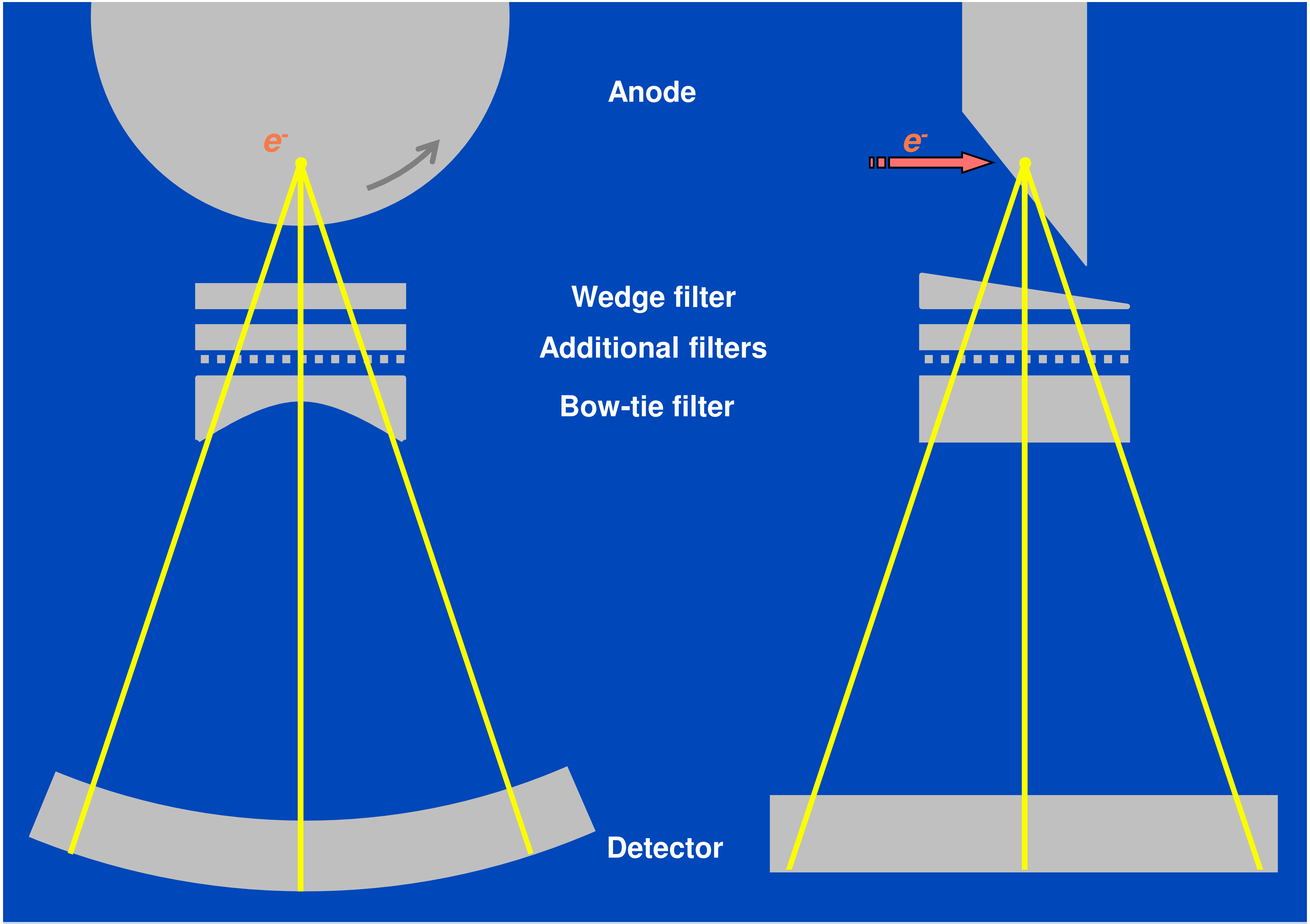
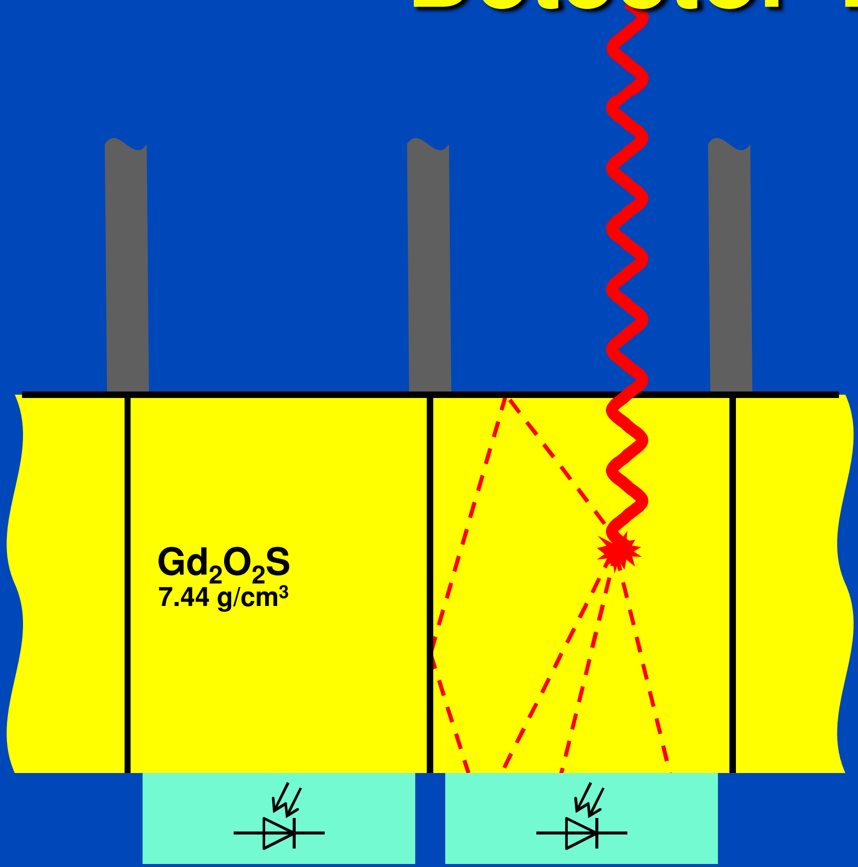


Figure not drawn to scale. Order of prefiltration may differ from scanner to scanner.

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

Detector Technology



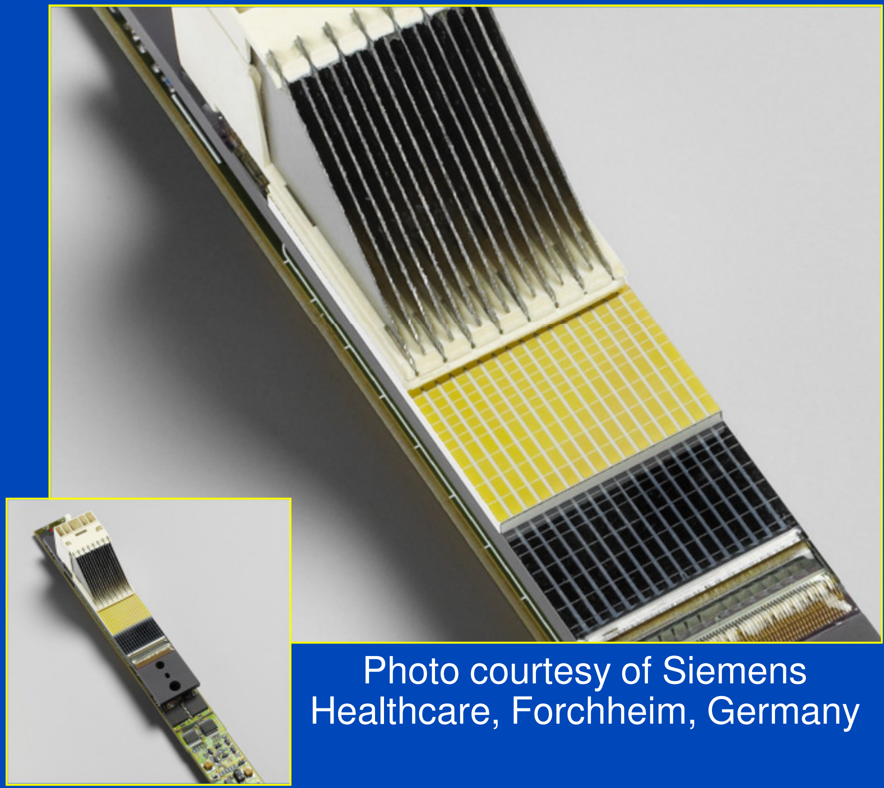


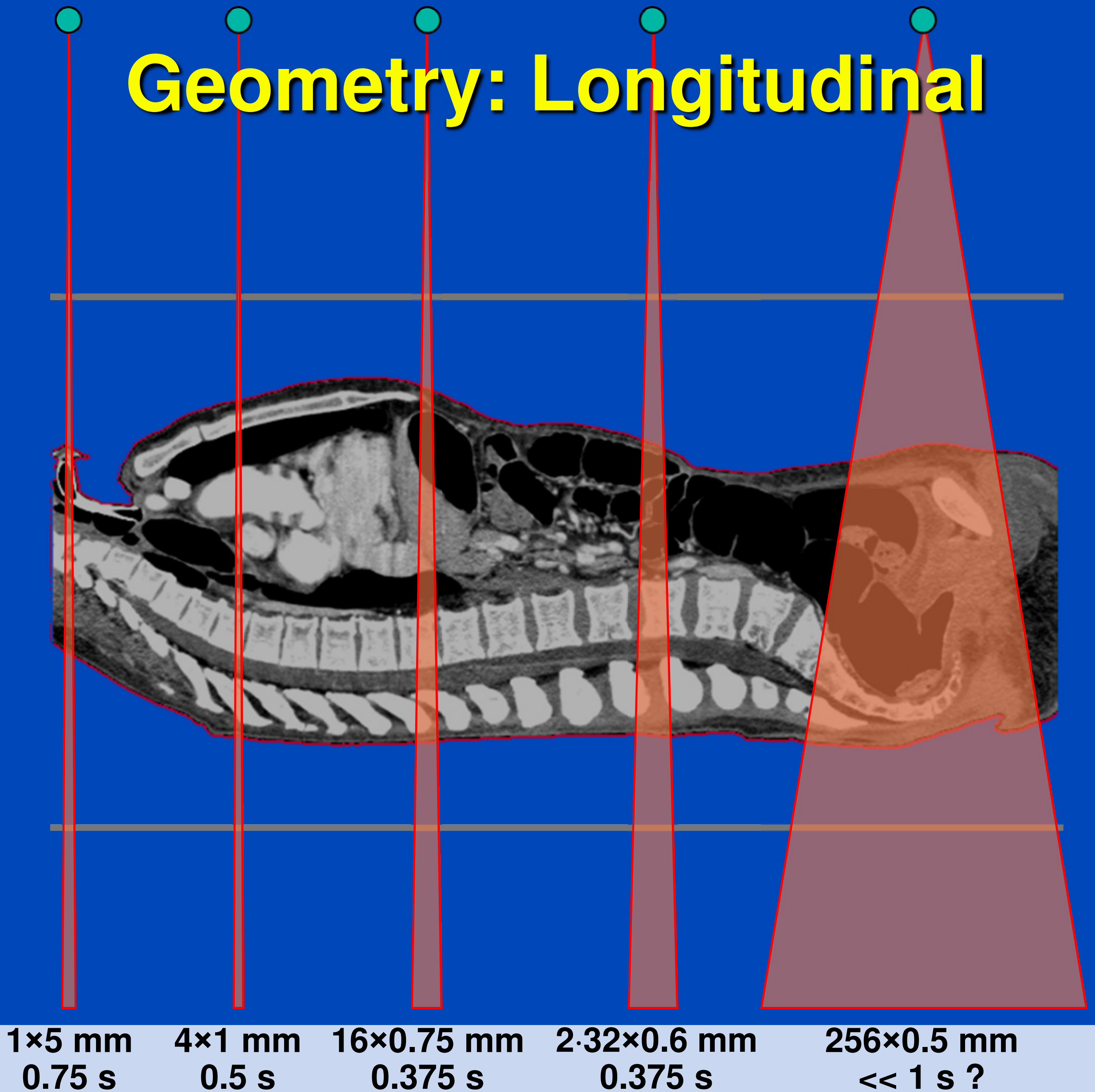
Photo courtesy of Siemens Healthcare, Forchheim, Germany

Demands on CT Detector Technology

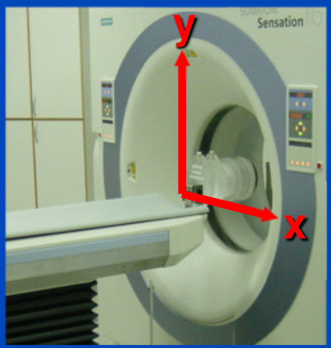
- **Available as multi-row arrays**
- **Very fast sampling (typ. 300 µs)**
- **Favourable temporal characteristics (decay time < 10 µs)**
- **High absorption efficiency**
- **High geometrical efficiency**
- **High count rate (up to 10^9 cps*)**
- **Adequate dynamic range (at least 20 bit)**

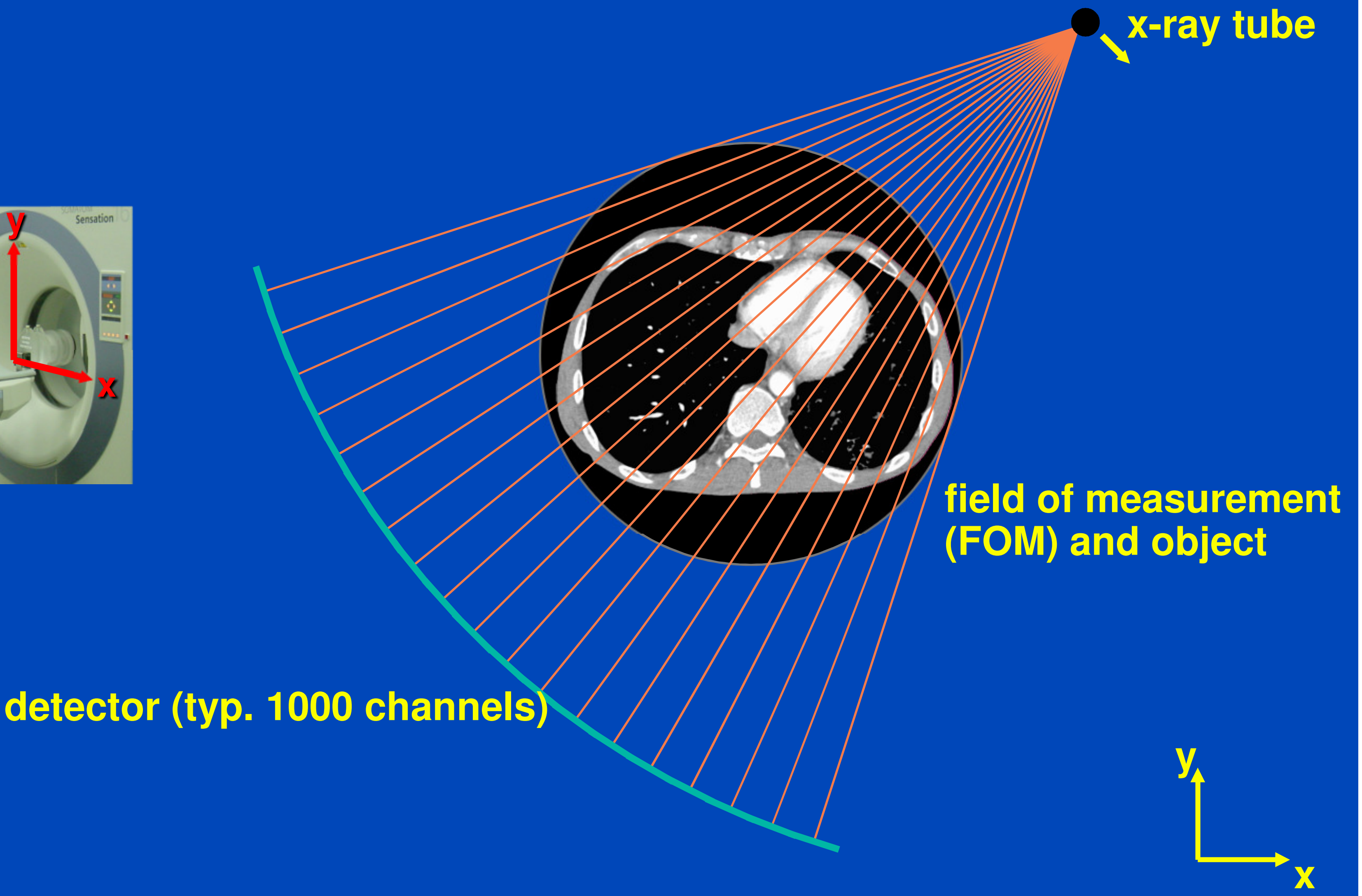
* in the order of 10^5 counts per reading and 10^4 readings per second

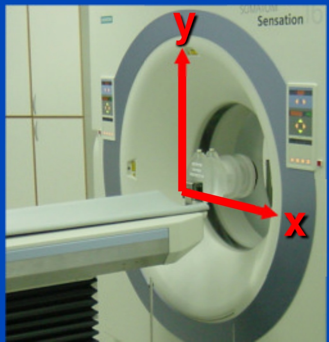
Geometry: Longitudinal

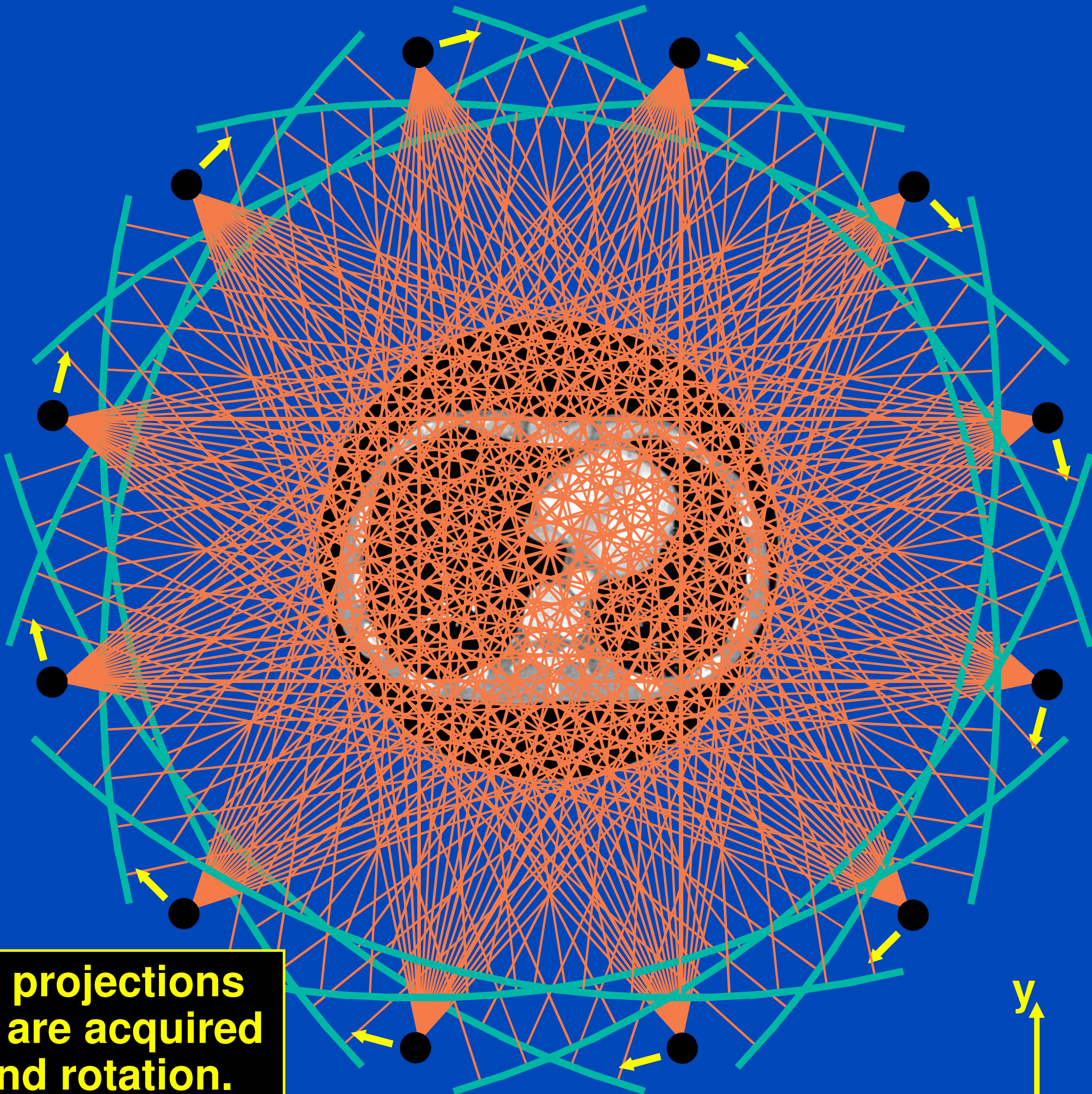


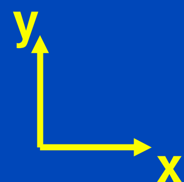
Geometry: Lateral



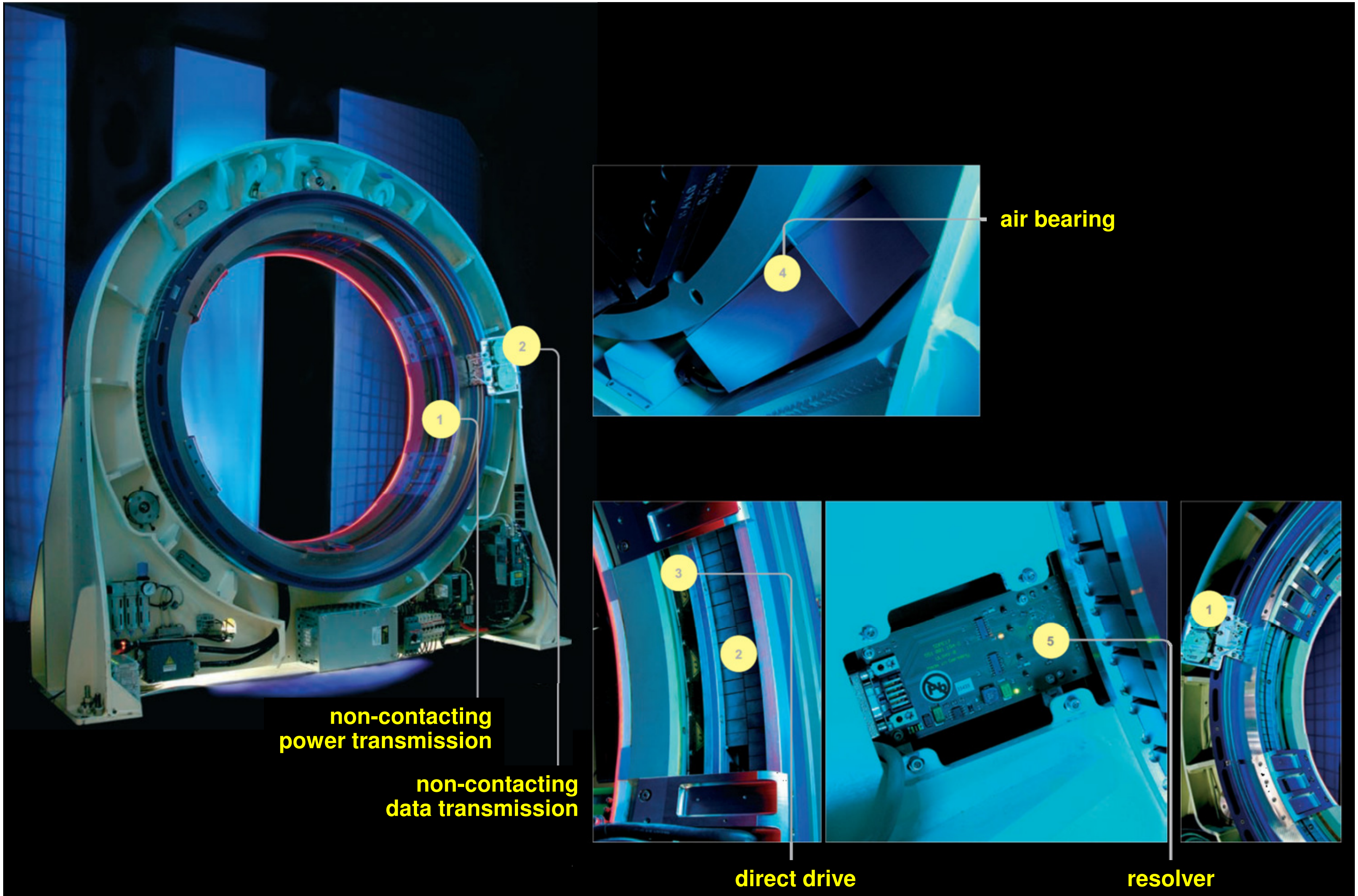








In the order of 1000 projections with 1000 channels are acquired per detector slice and rotation.



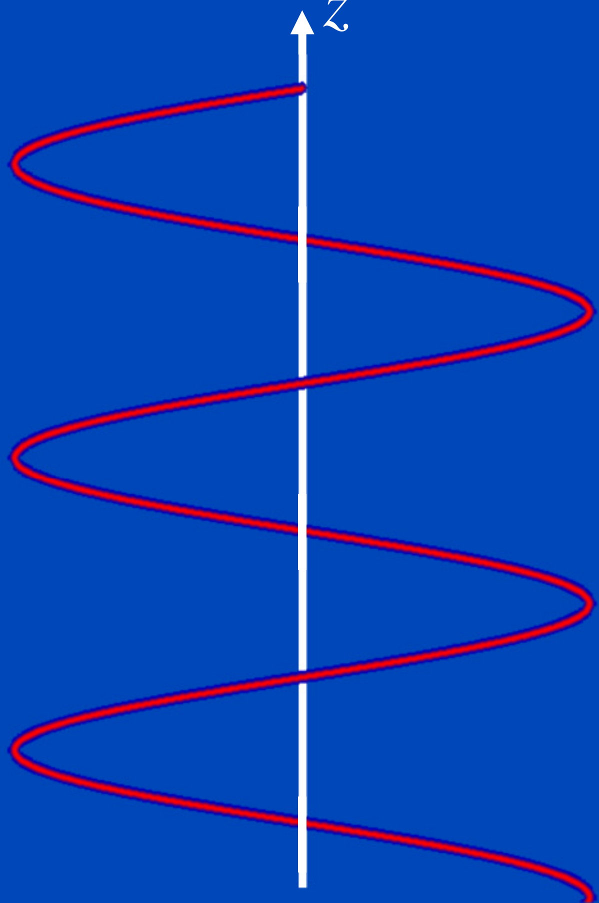
Data courtesy of Schleifring GmbH, Fürstenfeldbruck, Germany
and of rsna2011.rsna.org/exbData/1678/docs/Gantry_Subsystem.pdf

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

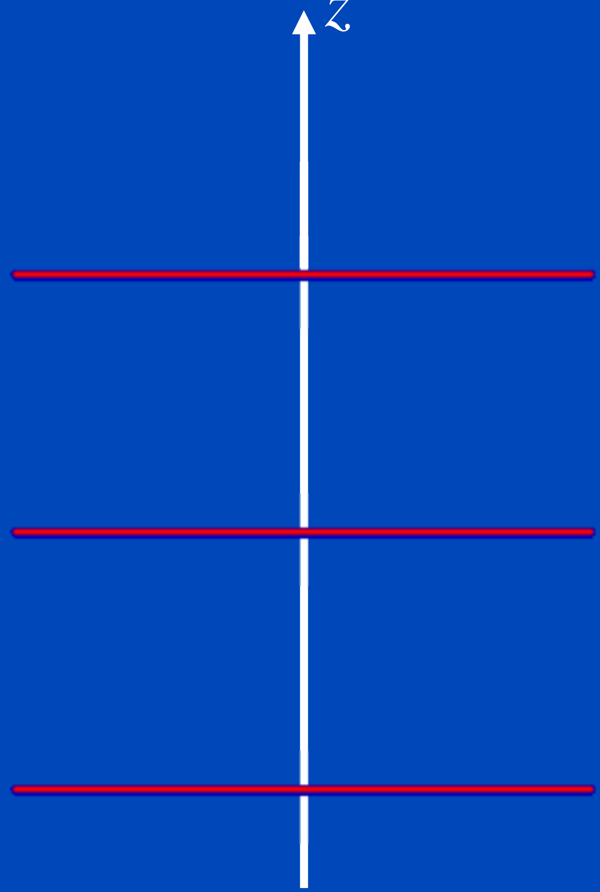
Scan Trajectories

Spiral



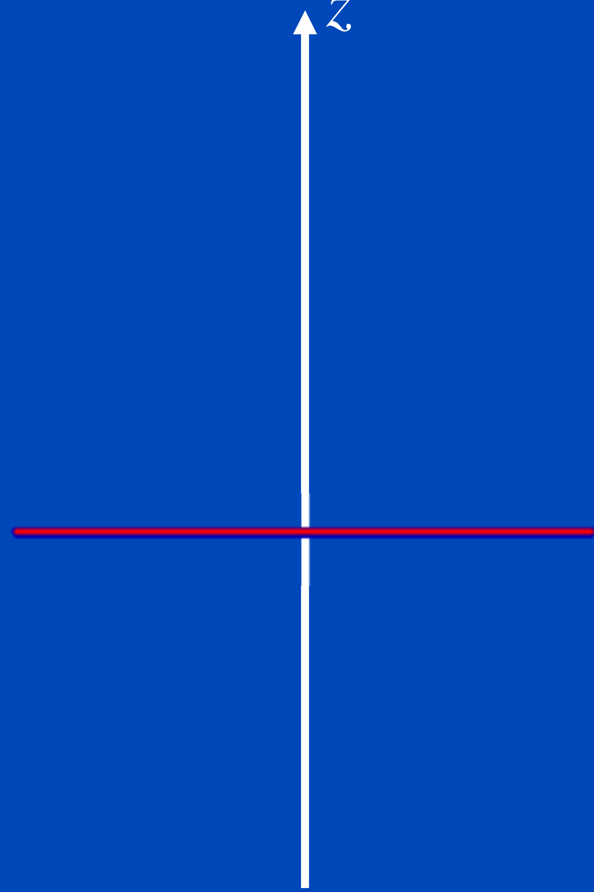
$$p = \frac{d}{M \cdot S} \leq 1.5$$

Sequence



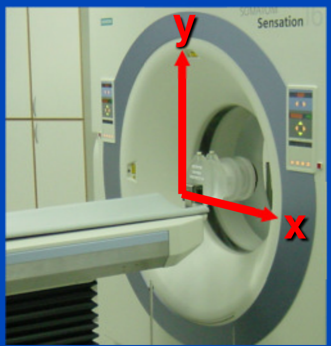
$$p = \frac{d}{M \cdot S} \leq 0.9$$

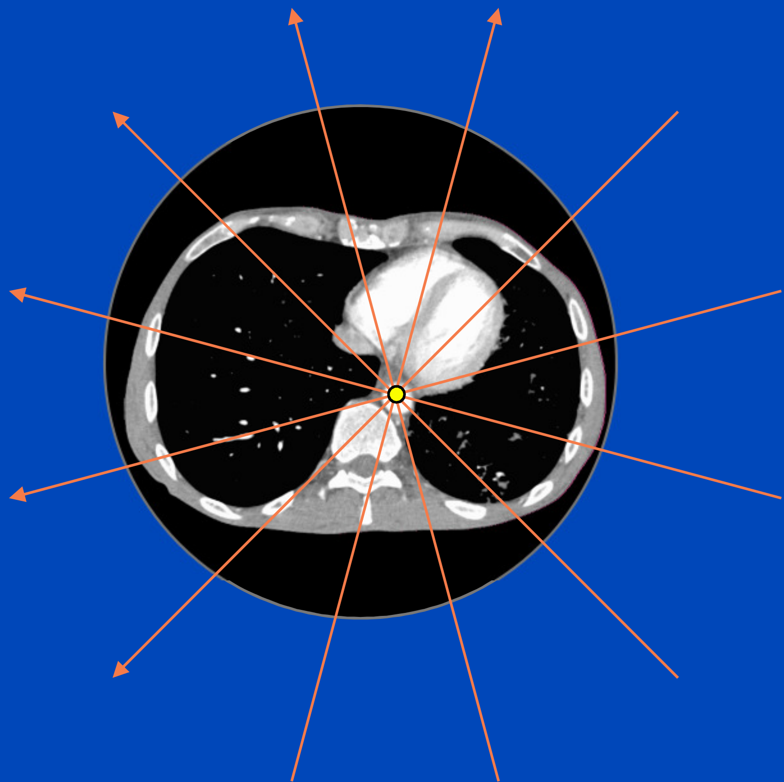
Circle



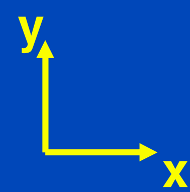
$$p = \frac{1}{N_{\text{rot}}}$$

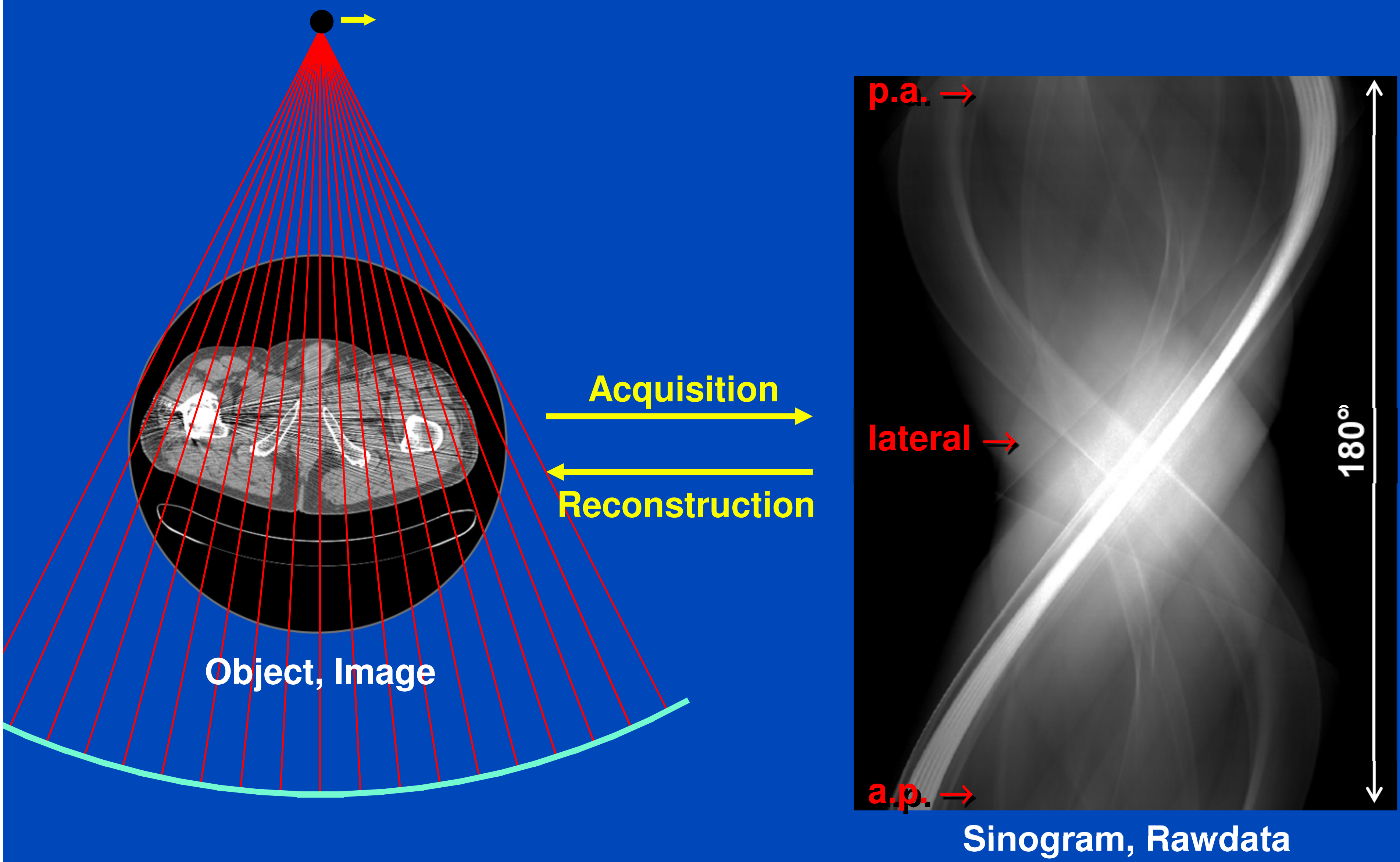
Data Completeness





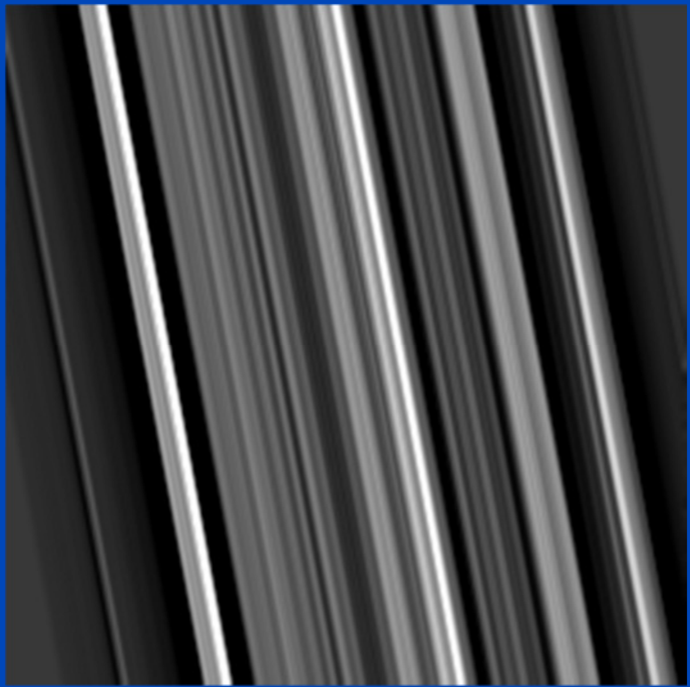
Each object point must be viewed by an angular interval of 180° or more. Otherwise image reconstruction is not possible.



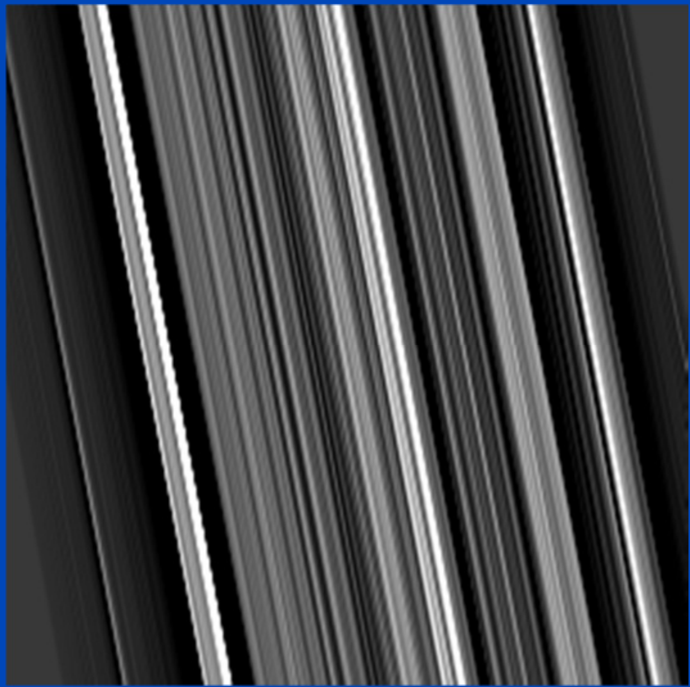

Acquisition
Reconstruction
Object, Image
p.a.
lateral
a.p.
180°
Sinogram, Rawdata

Filtered Backprojection (FBP)

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

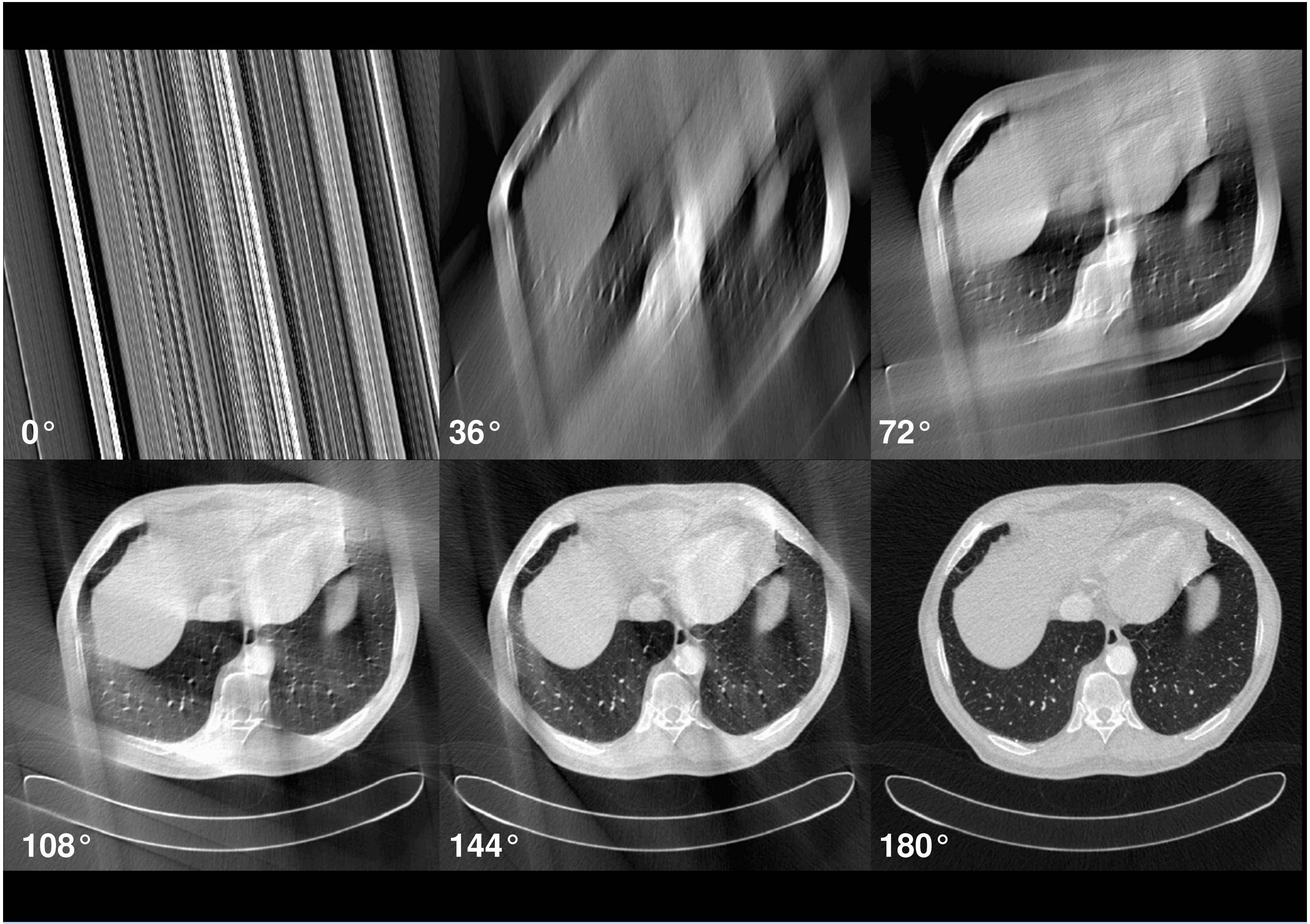


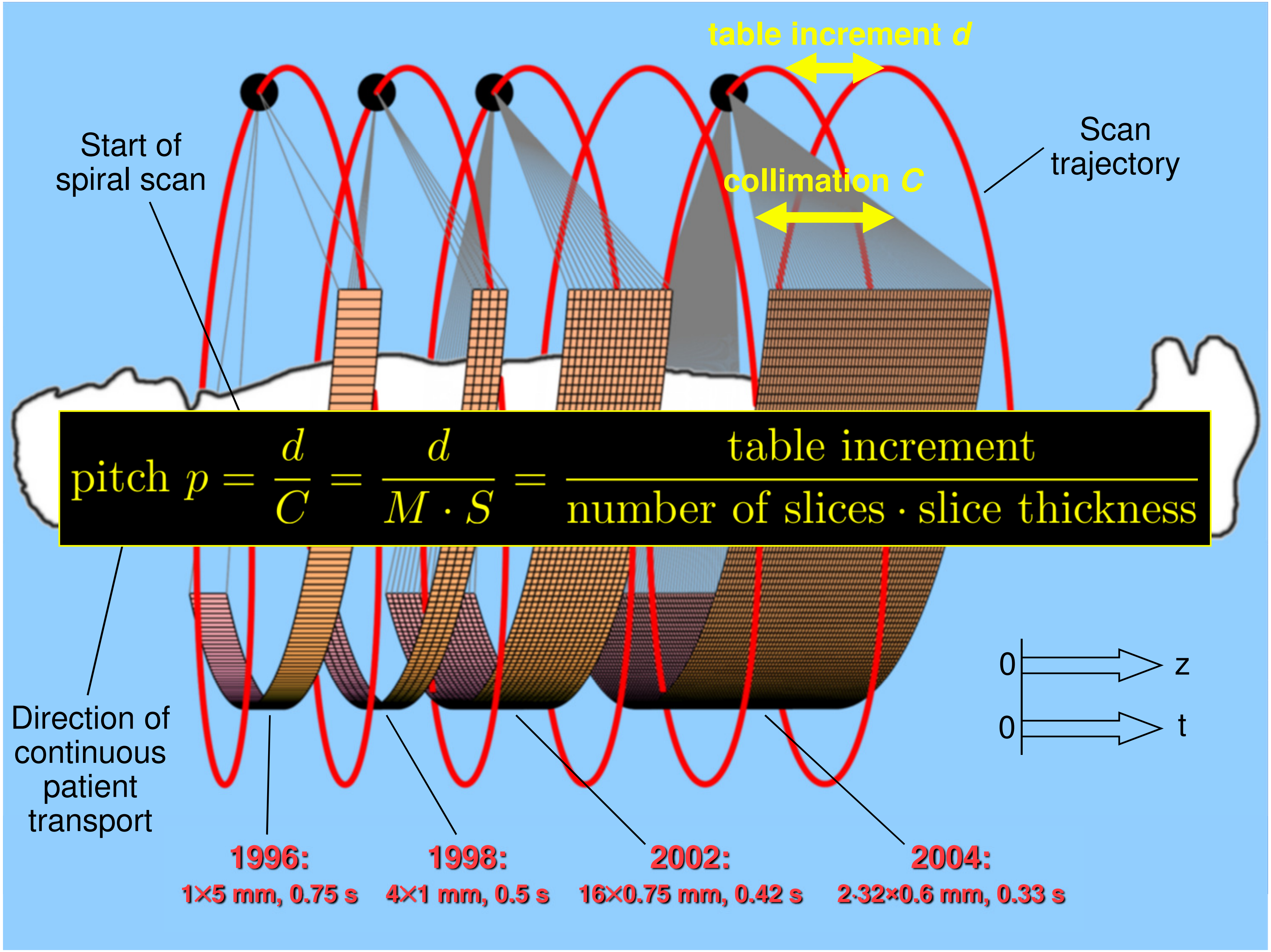
Smooth



Standard

Reconstruction kernels balance between spatial resolution and image noise.

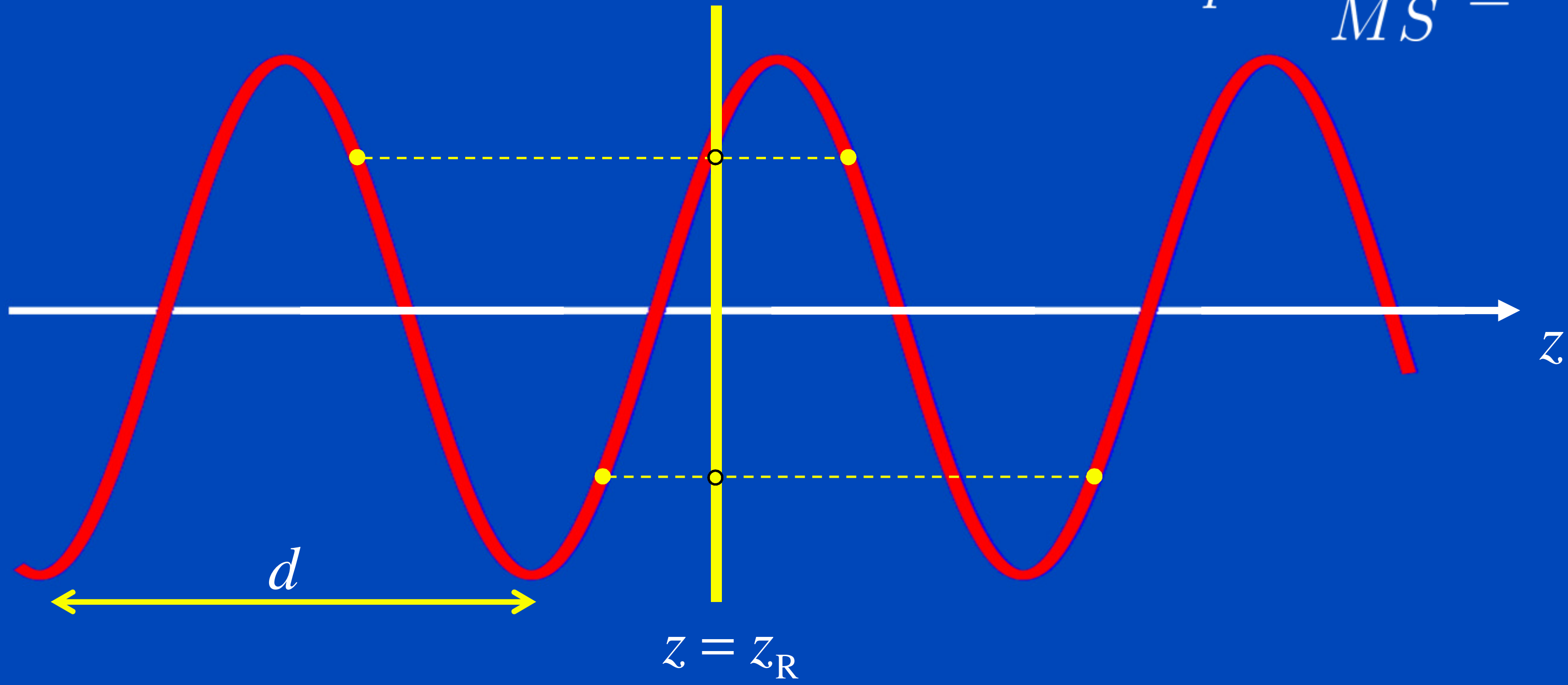

0°
36°
72°
108°
144°
180°



$$\text{pitch } p = \frac{d}{C} = \frac{d}{M \cdot S} = \frac{\text{table increment}}{\text{number of slices} \cdot \text{slice thickness}}$$

360° LI Spiral z-Interpolation for Single-Slice CT (*M*=1)

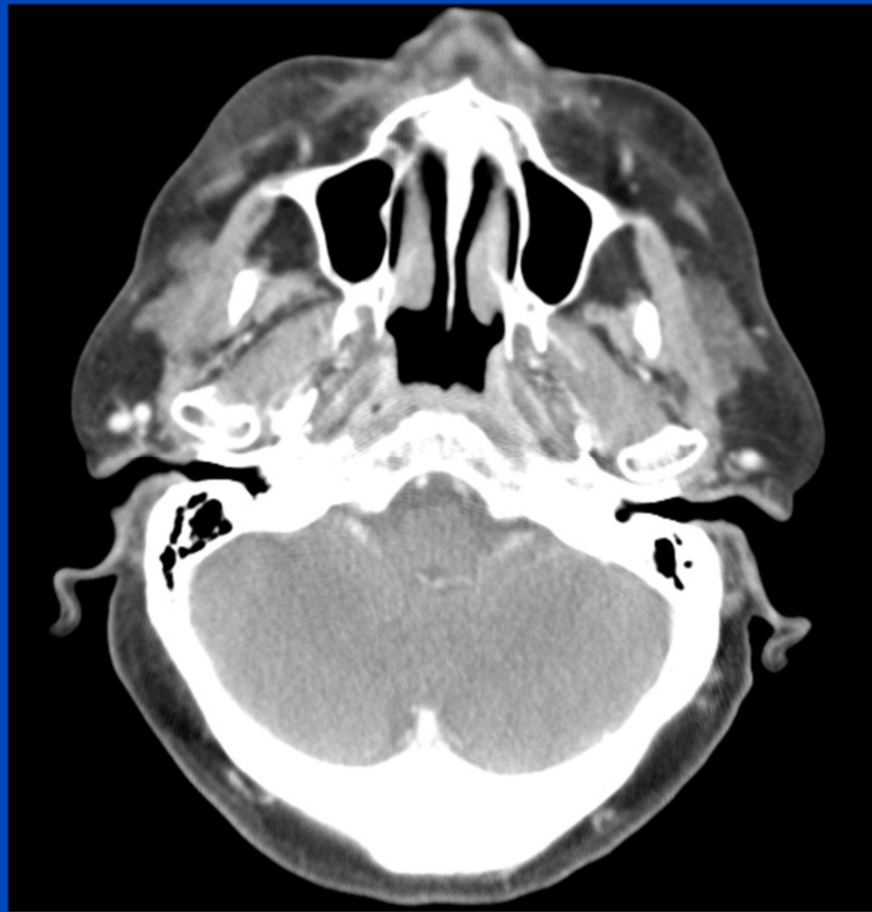
$$p = \frac{d}{MS} \leq 2$$



Spiral z-interpolation is typically a linear interpolation between points adjacent to the reconstruction position to obtain circular scan data.

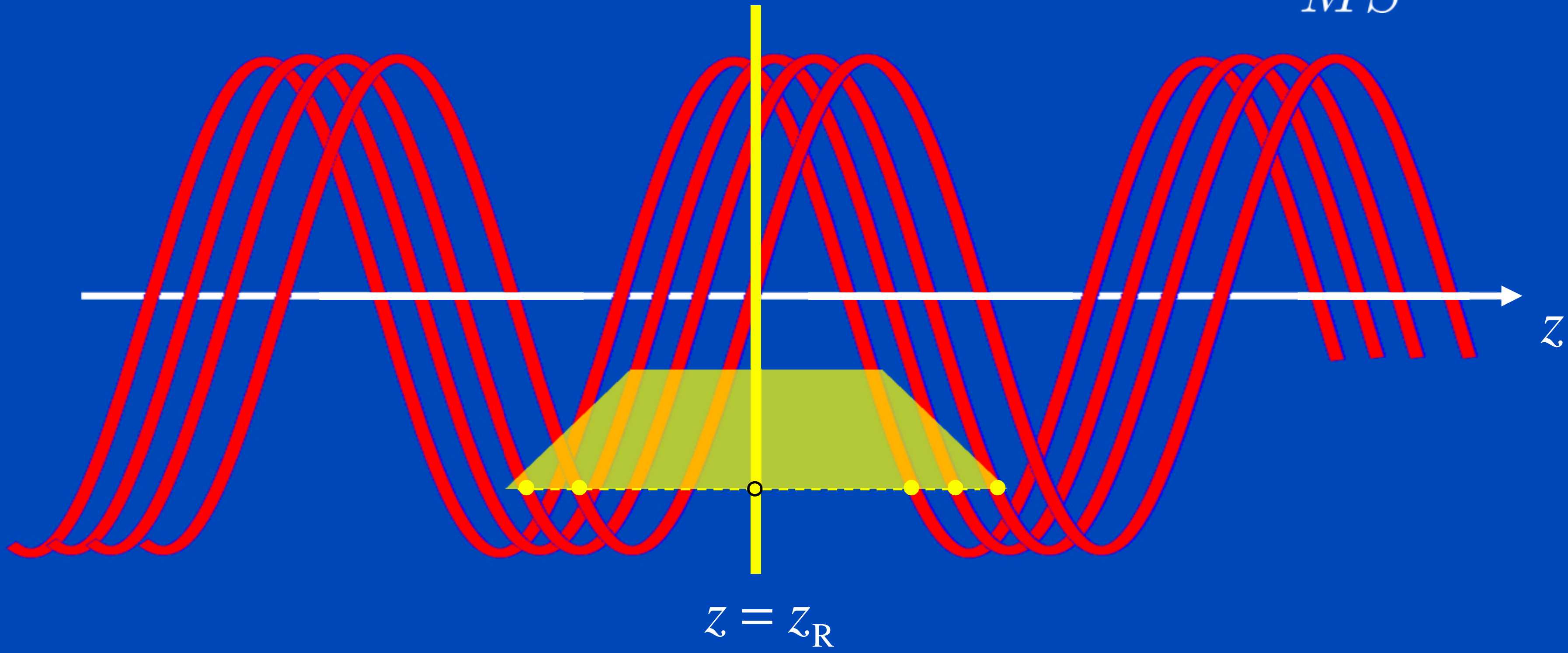
without *z*-interpolation

with *z*-interpolation



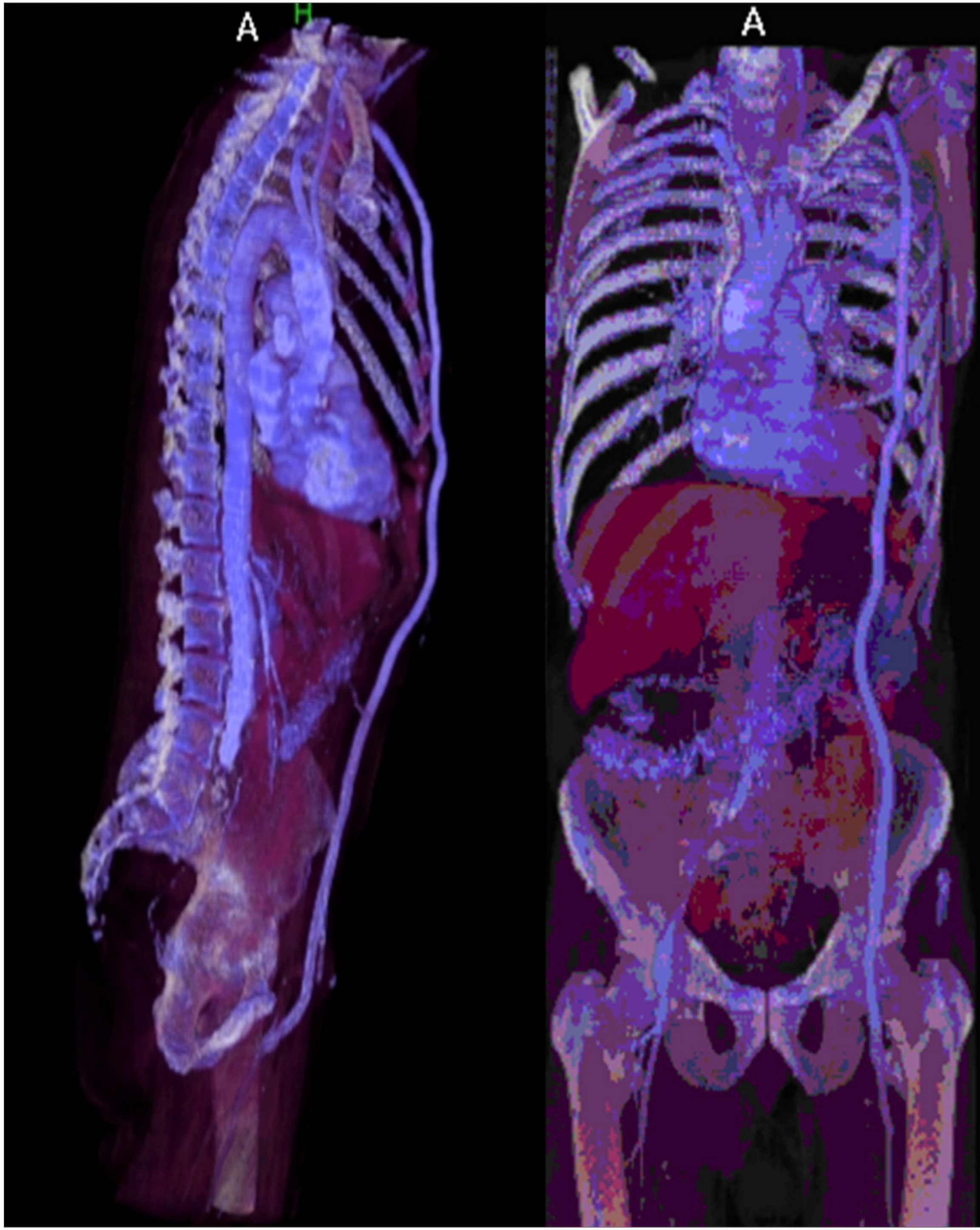
Spiral z-Filtering for Multi-Slice CT
M=2, …, 6

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



Spiral z-filtering is collecting data points weighted with a triangular or trapezoidal distance weight to obtain circular scan data.

dkfz.



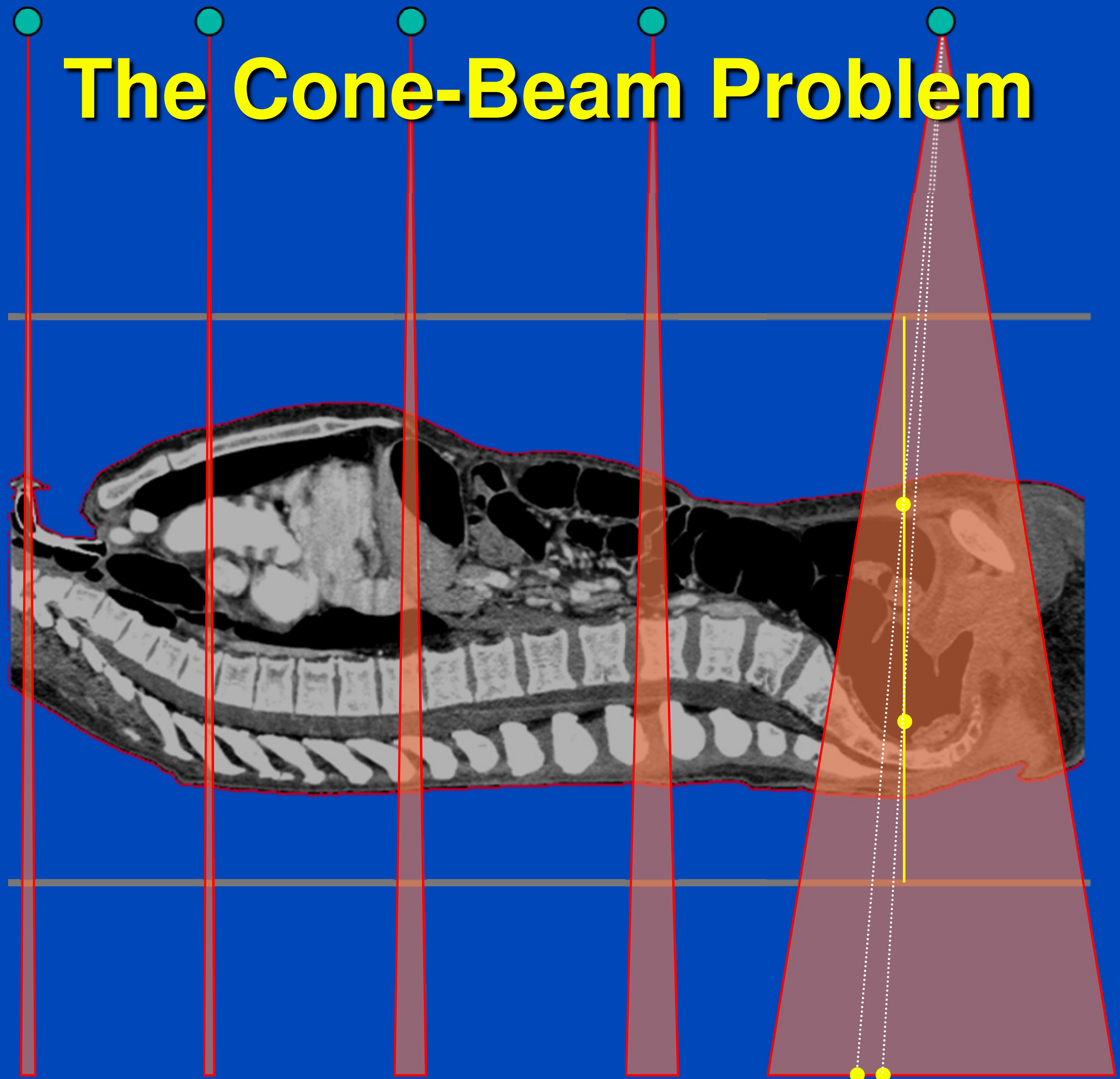
CT Angiography: Axillo-femoral bypass

M = 4

120 cm in 40 s

0.5 s per rotation
4×2.5 mm collimation
pitch 1.5

The Cone-Beam Problem

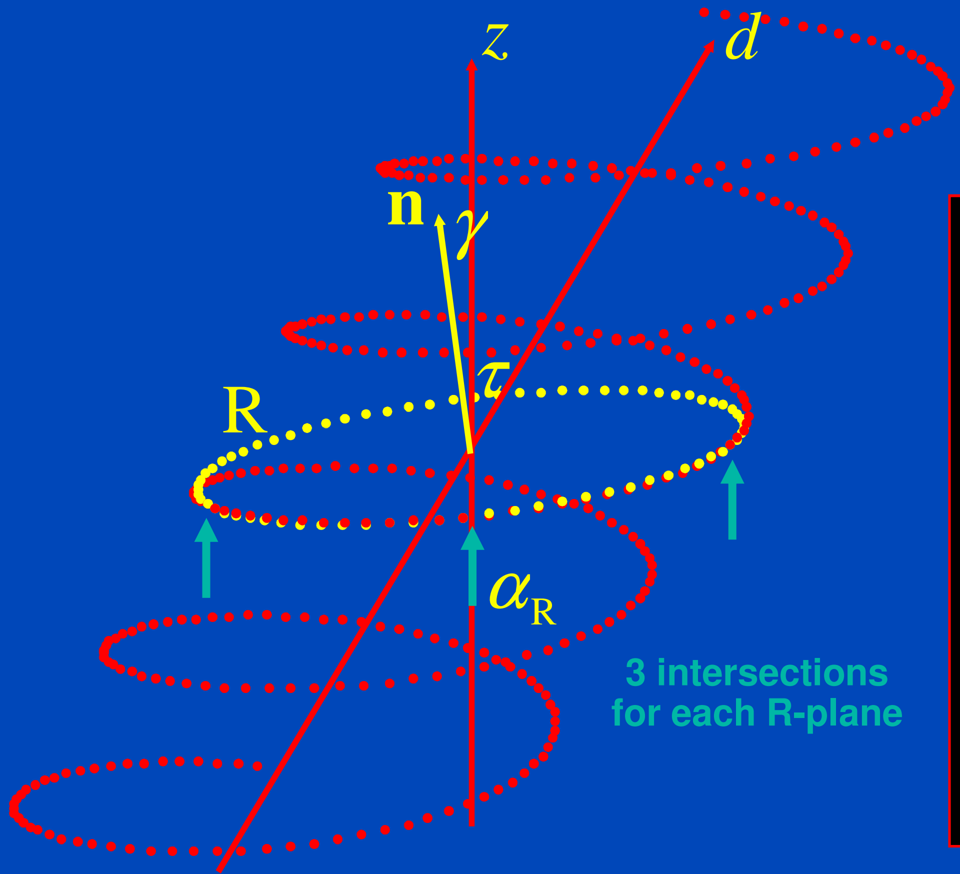


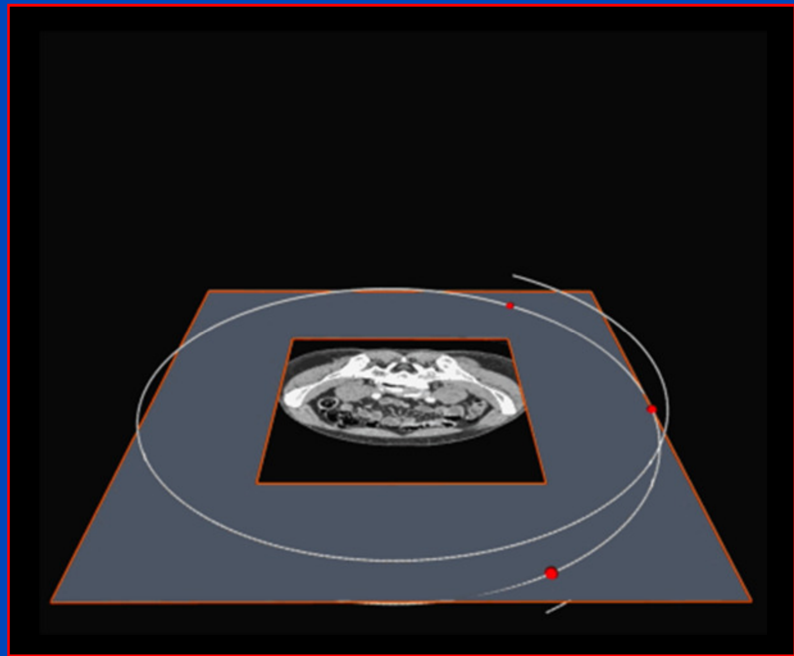
1×5 mm	4×1 mm	16×0.75 mm	2·32×0.6 mm	256×0.5 mm
0.75 s	0.5 s	0.375 s	0.375 s	<< 1 s ?

dkfz.

The ASSR Algorithm

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



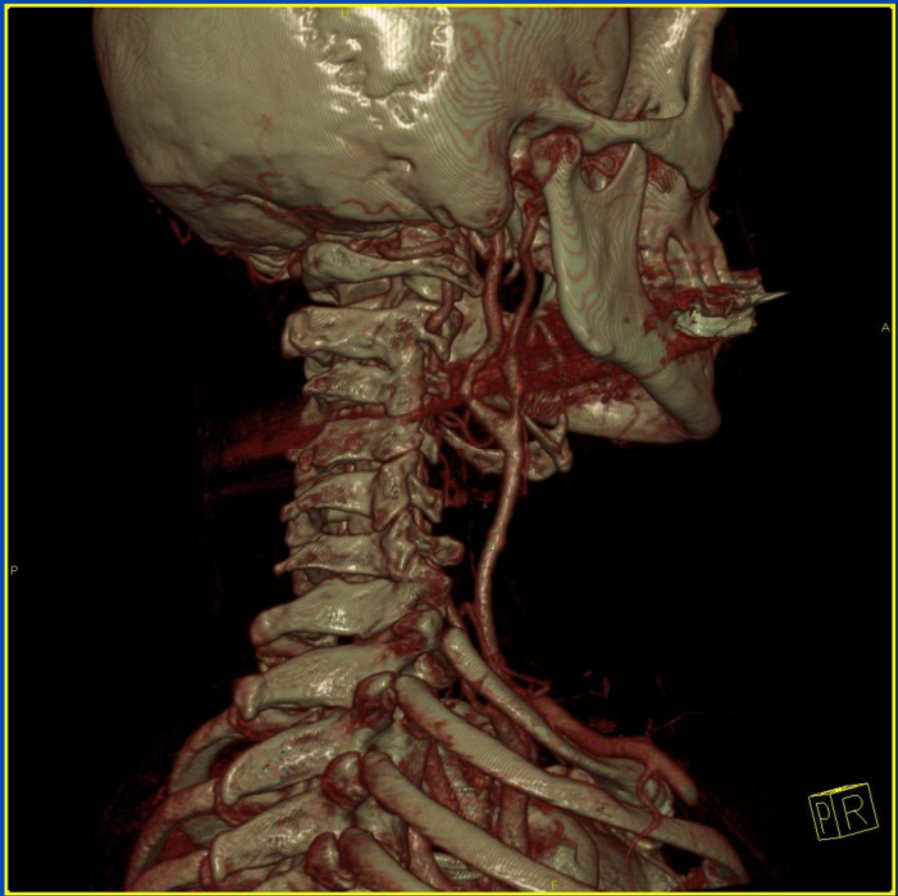


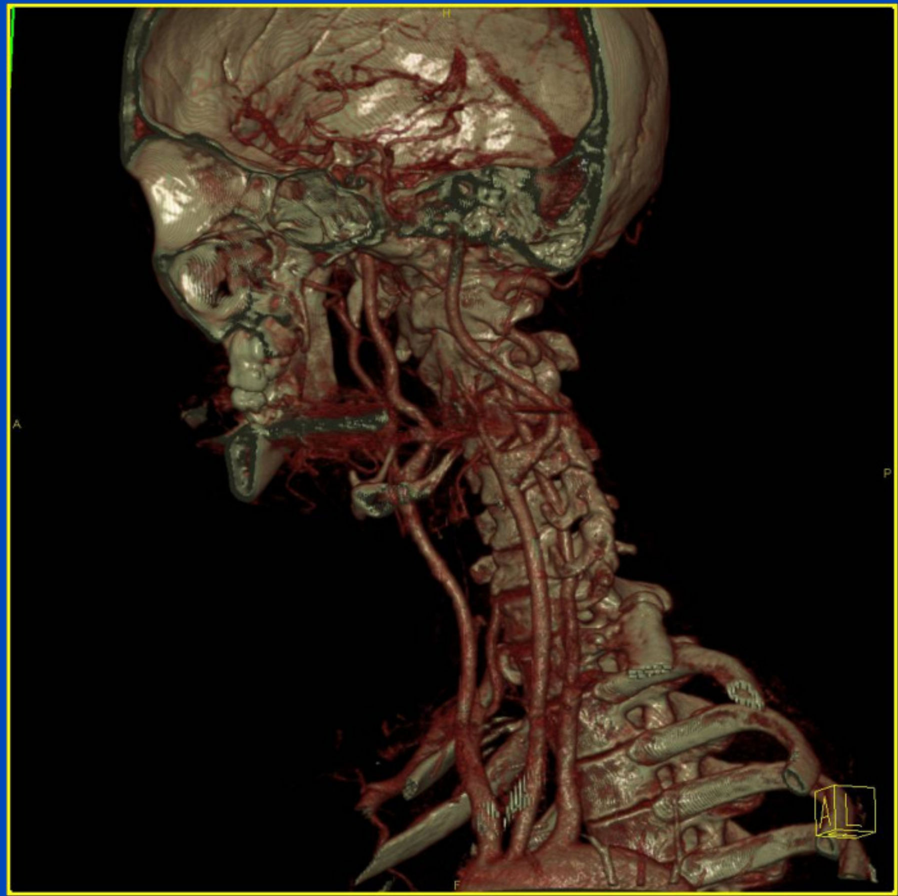
Resulting mean deviation at R_F: $\Delta_\mathrm{mean} \approx 0.014d$

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





Wie muss gemessen werden, um Bilder rekonstruieren zu können?

1. Es muss jeder Voxel aus einem genau 180° langen Winkelintervall gemessen werden.
2. Man kann Bilder rekonstruieren, sofern mindestens 90° an Projektionsdaten für jeden Pixel zur Verfügung stehen.
3. Für jedes CT-Bild ist ein Vollumlauf an Daten nötig.
4. Wenn mindestens 180° an Daten zur Verfügung stehen, können CT-Bilder errechnet werden.

TED

dkfz.

Wie soll es weitergehen?

Iterative Bildrekonstruktion

oder lieber

Cardio CT?

$$x^2 = y$$

Model

$$x = \sqrt{y}$$

Solution

This is an analytical solution.

Filtered Backprojection[1] (FBP)

Model

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

Fourier transform:

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

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

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

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

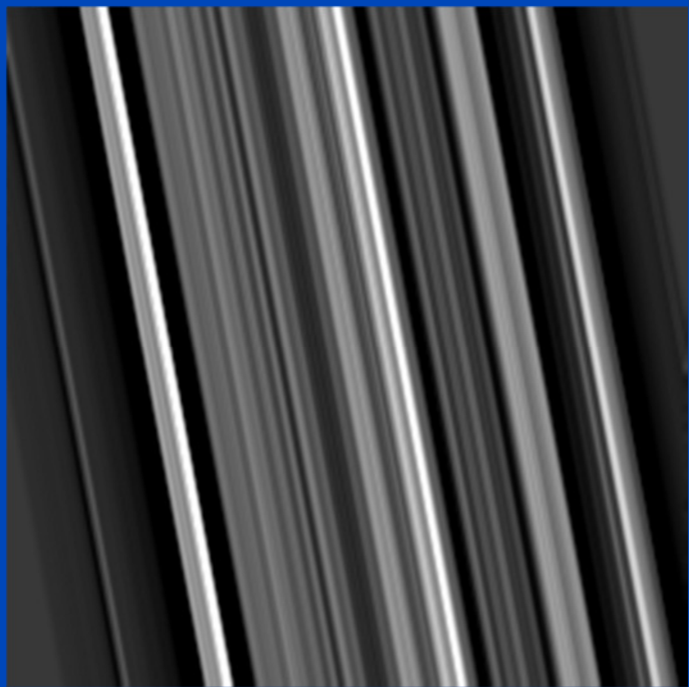
Solution

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

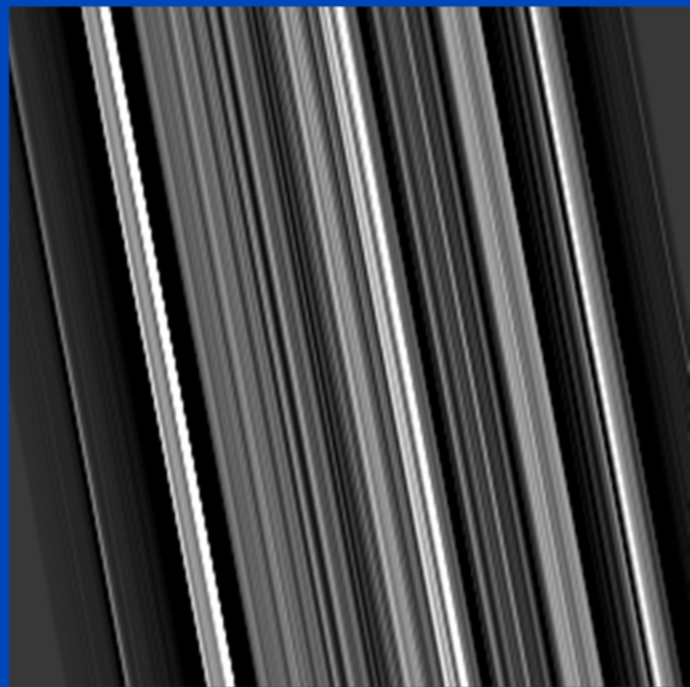
dkfz.

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

$$x = \sqrt{y}$$

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

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

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

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

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

Update equation

This is an iterative solution.

Influence of Update Equation and Model

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

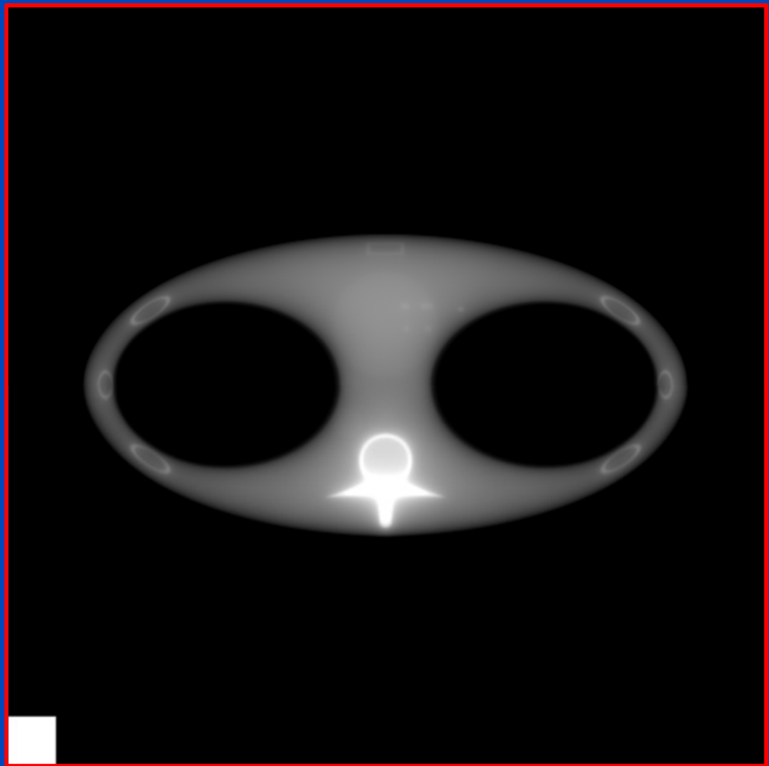
$$x^2 = 3, \quad x_0 = 1, \quad x_{n+1} = x_n + \Delta x_n$$

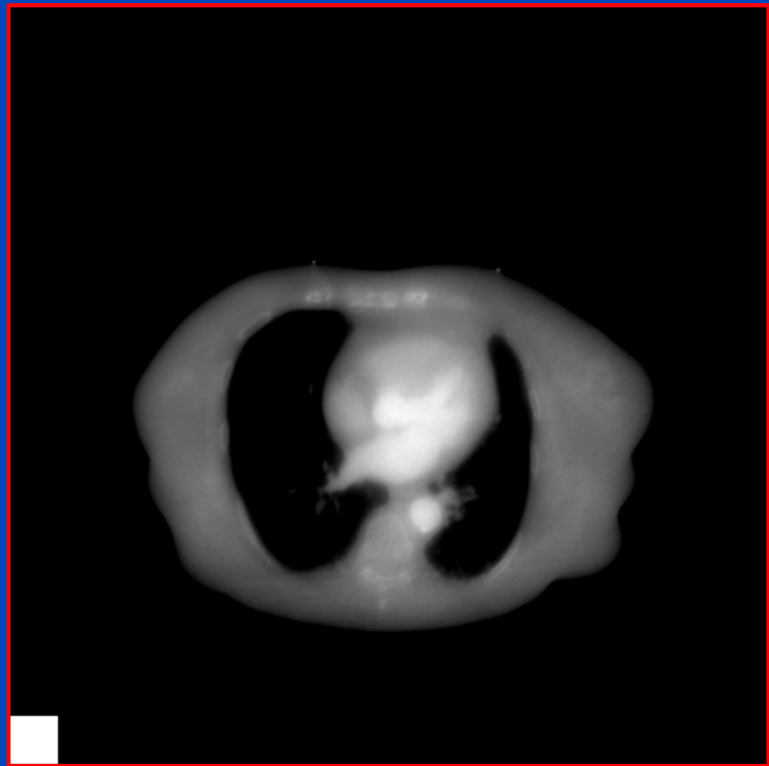
dkfz.

Flavours of Iterative Reconstruction

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

Iterative Reconstruction

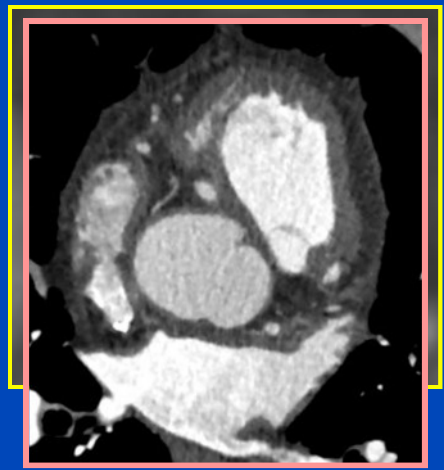




16 ordered subsets iterations

Iterative Reconstruction

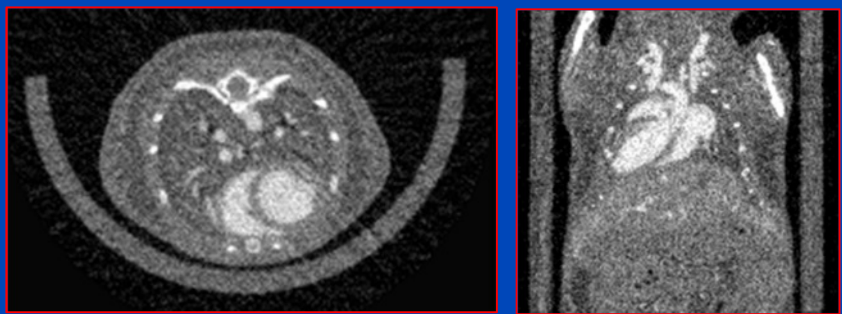
- Aim: less artifacts, lower noise, lower dose
- Iterative reconstruction
  - Reconstruct an image.
  - Regularize the image.
  - Does the image correspond to the rawdata?
  - If not, reconstruct a correction image and continue.
- SPECT + PET are iterative for a long time!
- Until recently, the computational 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)



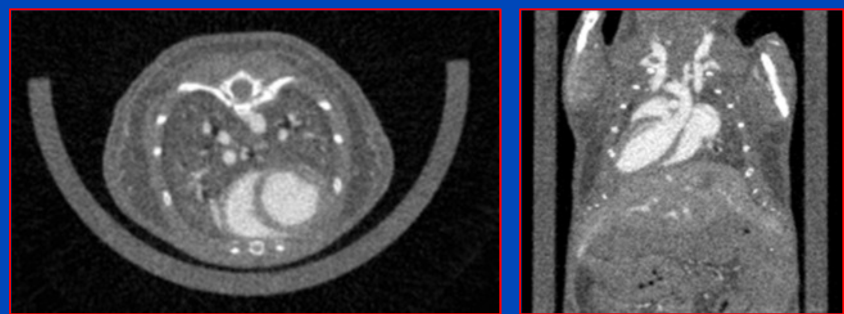
What Makes Iterative Recon Attractive?

- No need to come find an analytical solution
- Works for all geometries with only small adaptations
- Allows to model any effect
- Allows to incorporate prior knowledge
  - noise properties (quantum noise, electronic noise, noise texture, …)
  - prior scans (e.g. planning CT, full scan data, …)
  - image properties such as smoothness, edges (e.g. minimum TV)
  - …
- Handles missing data implicitly (but not necessarily better)

Phase-correlated Feldkamp



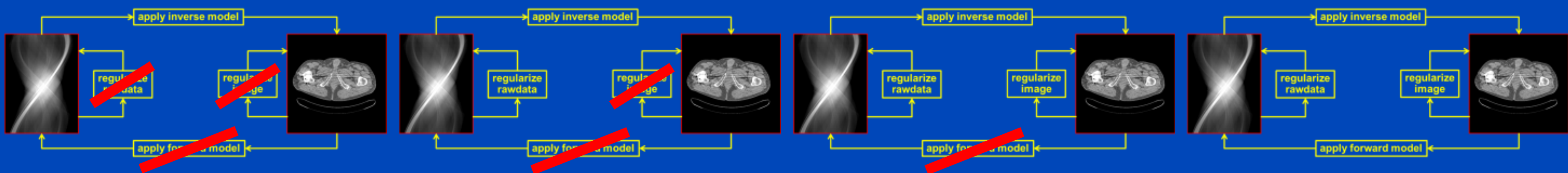
High dimensional TV minimization[1]



[1]L. Ritschl, S. Sawall, M. Knaup, A. Hess, and M. Kachelrieß, Phys. Med. Biol. 57, Jan. 2012

Plain FBP | Siemens Standard | IRIS VA34 | SAFIRE VA40

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

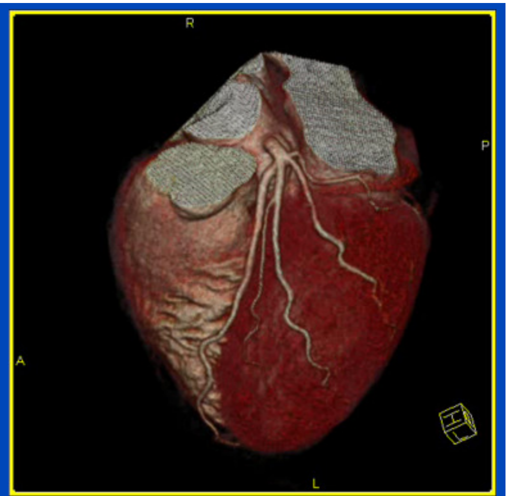


CT images provided by Siemens Healthcare, Forchheim, Germany

dkfz.

Imaging the Heart with CT

(Cardiac-CT = phase-correlated CT)



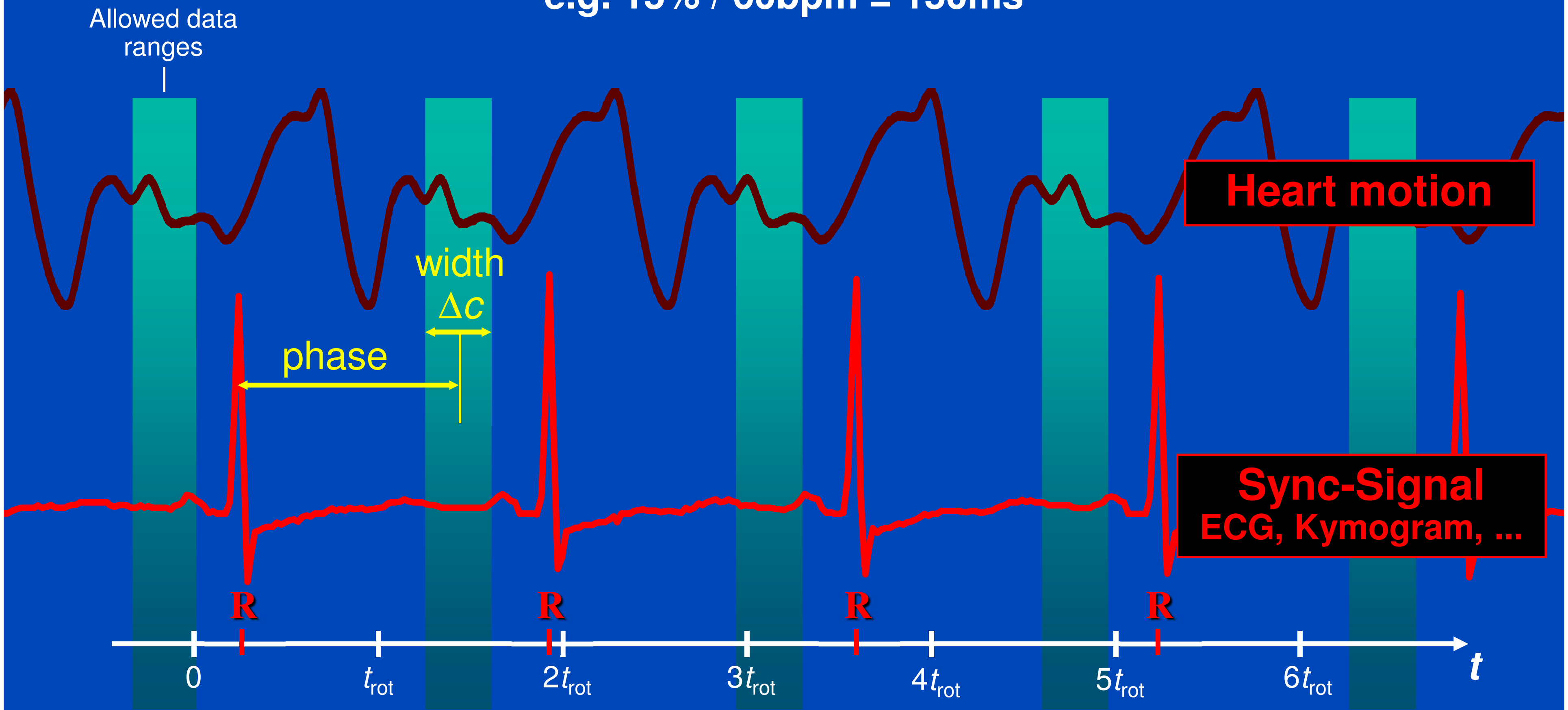
- **Periodic motion**
- **Synchronisation (ECG, Kymogram, …)**
- **Phase-correlated scanning = Prospective Gating**
  - Used in the 80s and 90s with little success.
  - Comes into use again due to large cone-angles.
- **Phase-correlated reconstruction = Retrospective Gating**
  - Single-phase (partial scan) approaches, e.g. 180°MCD
  - Bi-phase approaches, e.g. ACV (Flohr et al.)
  - Multi-phase Cardio Interpolation methods, e.g. 180°MCI (gold-standard)
  - Generations
    - » Single-slice spiral CT: 180°CD, 180°CI (introduced 1996[1])
    - » Multi-slice spiral CT: 180°MCD, 180°MCI (introduced 1998[2])
    - » Cone-beam spiral CT: ASSR CD, ASSR CI (introduced 2000[3])
    - » Wide cone-beam CT: EPBP (introduced 2002[4])
    - » Multi-source CBCT: EPBP (introduced 2005[5])

[1]Med. Phys. 25(12):2417-2431 (1998), [2]Med. Phys. 27(8):1881-1902 (2000), [3]Proc. Fully 3D-2001:179-182 (2001),
[4]Med. Phys. 31(6): 1623-1641 (2004), [5]Med. Phys. 33(7): 2435-2447 (2006)

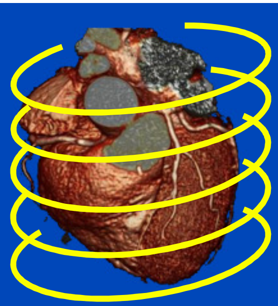
Synchronization with the Heart Phase

t_{eff} = width / heart rate

e.g. 15% / 60bpm = 150ms



Width, and thus t_{eff}, corresponds to the FWTM of the phase contribution profile.



Retrospective Gating

=

Standard scan + ECG-correlated recon

Standard spiral scan with low pitch value ($p \le f_H \cdot t_{rot}$)

Phase-correlated reconstruction

$p \cdot T_{rot} / 2 \le$ Temp. resolution $\le T_{rot} / 2$

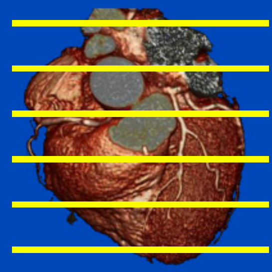
Works also at high heart rates

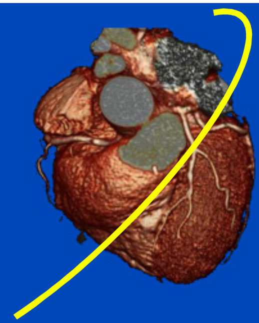
Dose management: ECG-based TCM

Full phase selectivity

Highly robust (also with arrhythmia)

Good dose usage





Prospective Gating

=

ECG-triggered scan + standard recon

ECG-triggered sequence- or spiral scan with high pitch value

Standard image reconstruction

Temporal resolution = $T_{rot} / 2$

Good at low heart rates

Dose management: inherent

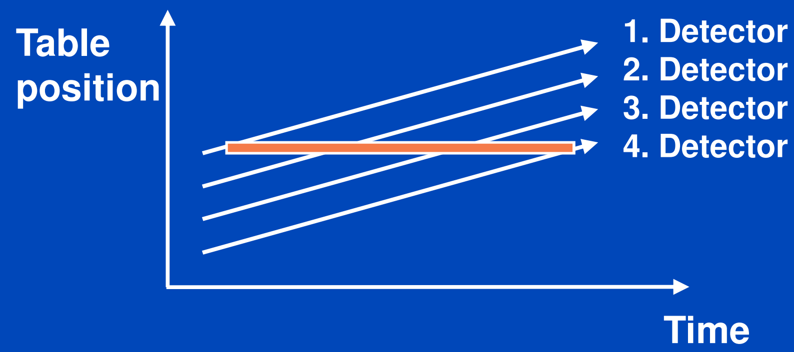
No phase selectivity

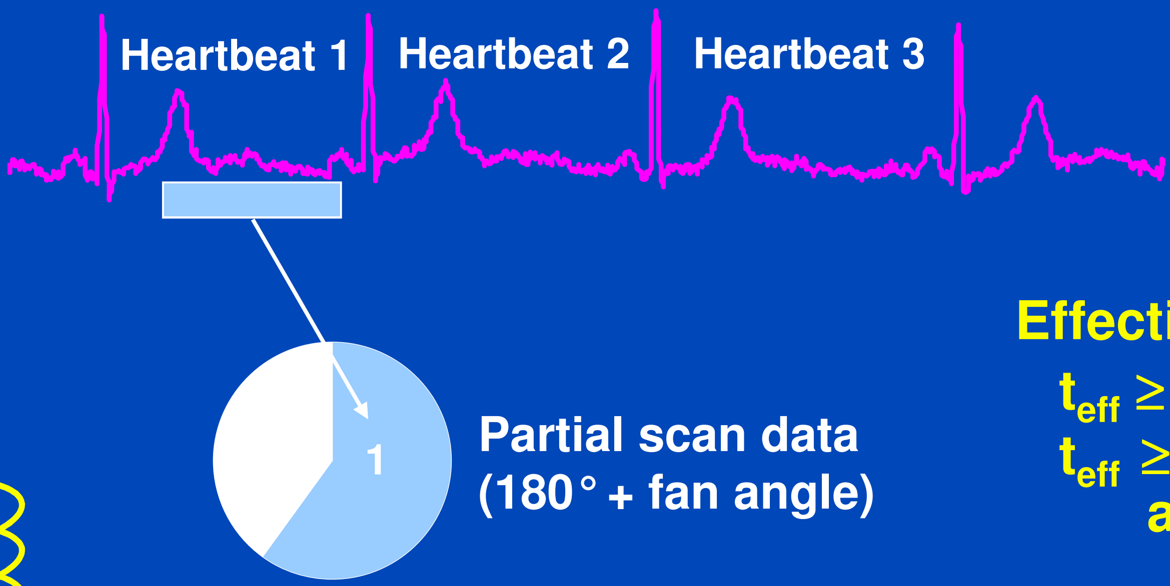
Sufficiently robust (not with arrythmia)

Very good dose usage

Partial Scan Reconstruction

Use one segment of 180°+δ data of phase-coherent data for a selected heart phase

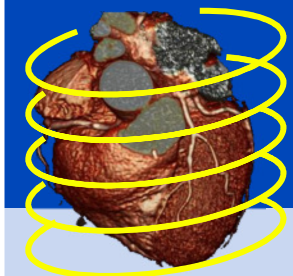




Effective scan time

$t_{eff} \geq t_{rot}/2$

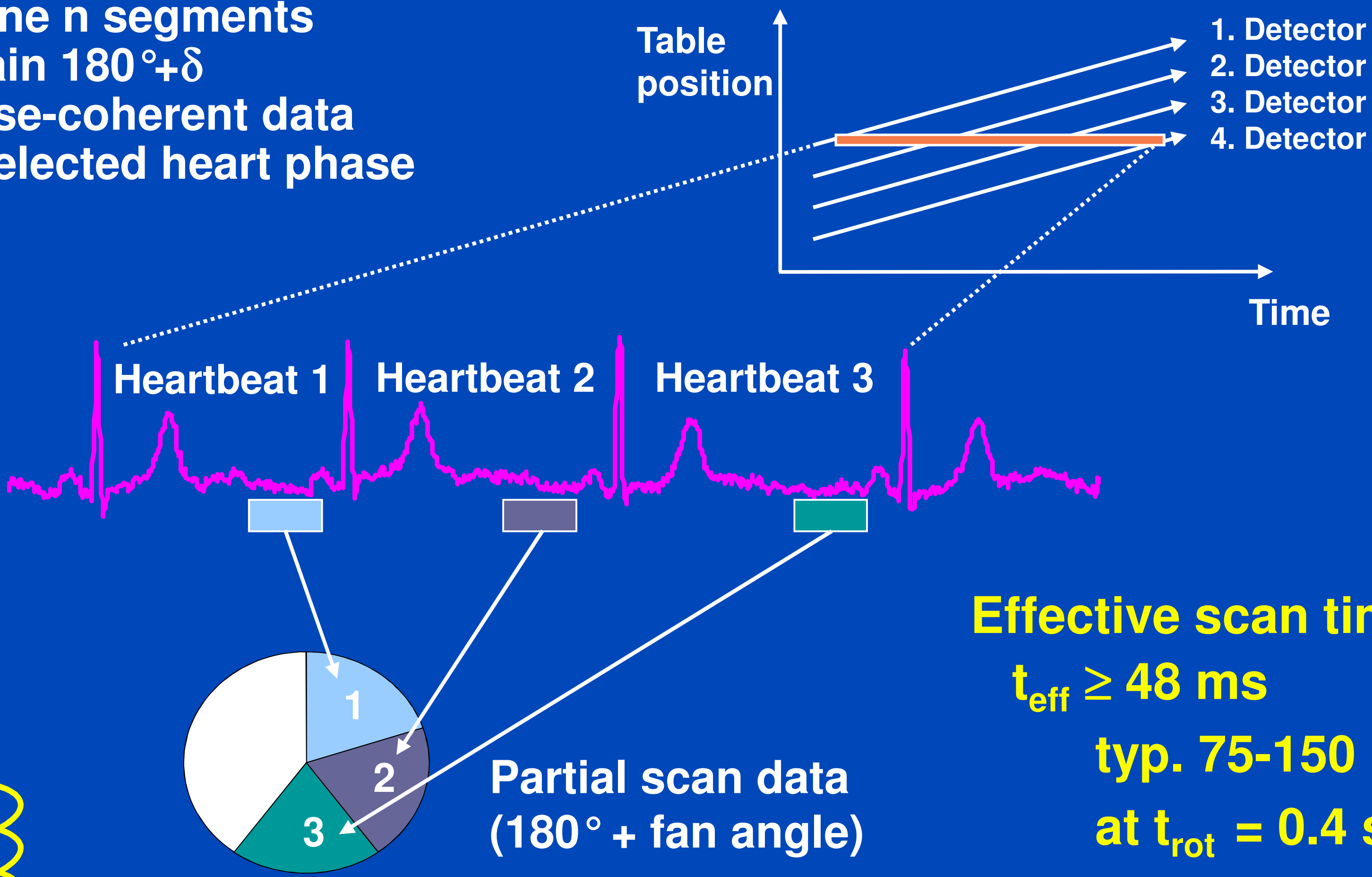
$t_{eff} \geq 200$ ms at $t_{rot} = 0.4$ s



Kachelrieß, Ulzheimer, Kalender, Med. Phys. 27(8):1881-1902 (2000)

Multi-Segment Reconstruction

Combine n segments
to obtain 180°+δ
of phase-coherent data
for a selected heart phase

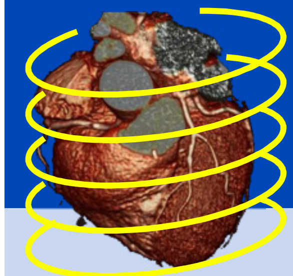


Effective scan time

$t_{eff} \geq 48$ ms

typ. 75-150 ms

at $t_{rot} = 0.4$ s



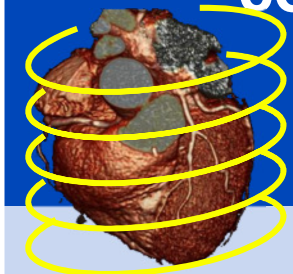
Kachelrieß, Ulzheimer, Kalender, Med. Phys. 27(8):1881-1902 (2000)

Pitch Value and Full Phase Selectivity

- Each voxel must be illuminated by the x-rays at least as long as one motion cycle of the heart takes
- The table increment per motion cycle must not be larger than the collimation of the scanner

$$p \leq f_{\mathrm{H}} t_{\mathrm{rot}}$$

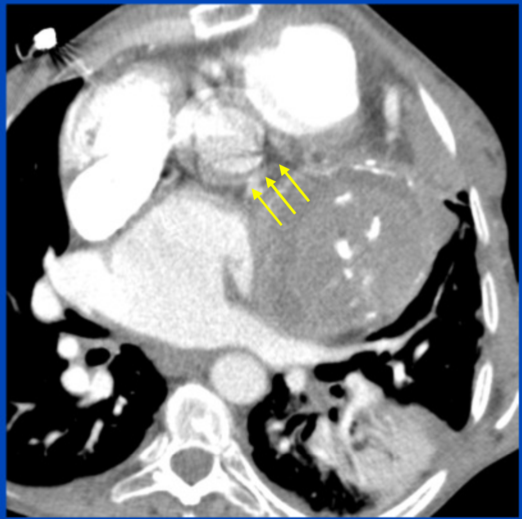
- For example $t_{rot} = 0.5$ s and $f_H = 60$ bpm imply that a pitch value of $p < 0.5$ must be chosen.
- The lower the pitch value the more segments can be combined in multi-segment image reconstruction.



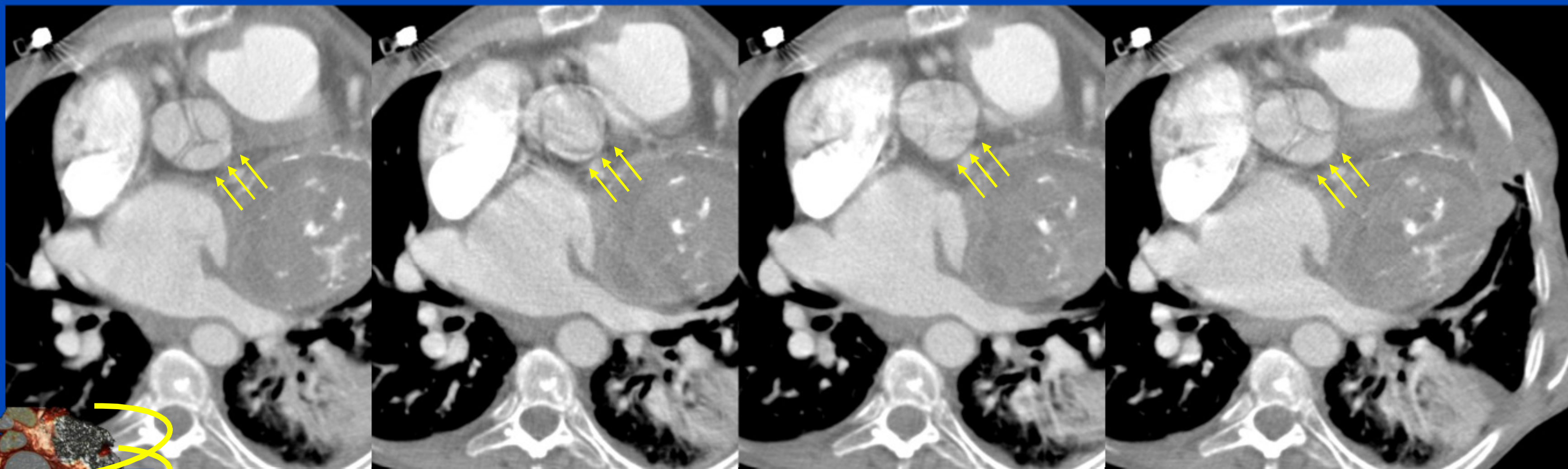
dkfz.

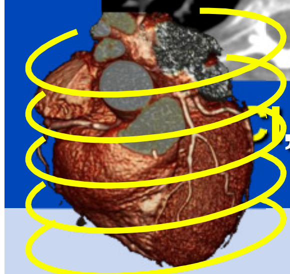
Cardiac CT is Phase-Selective

Volume Zoom, 4 × 2.5 mm, 0.5 s, 1998, f_H = 90 bpm



180°MLI



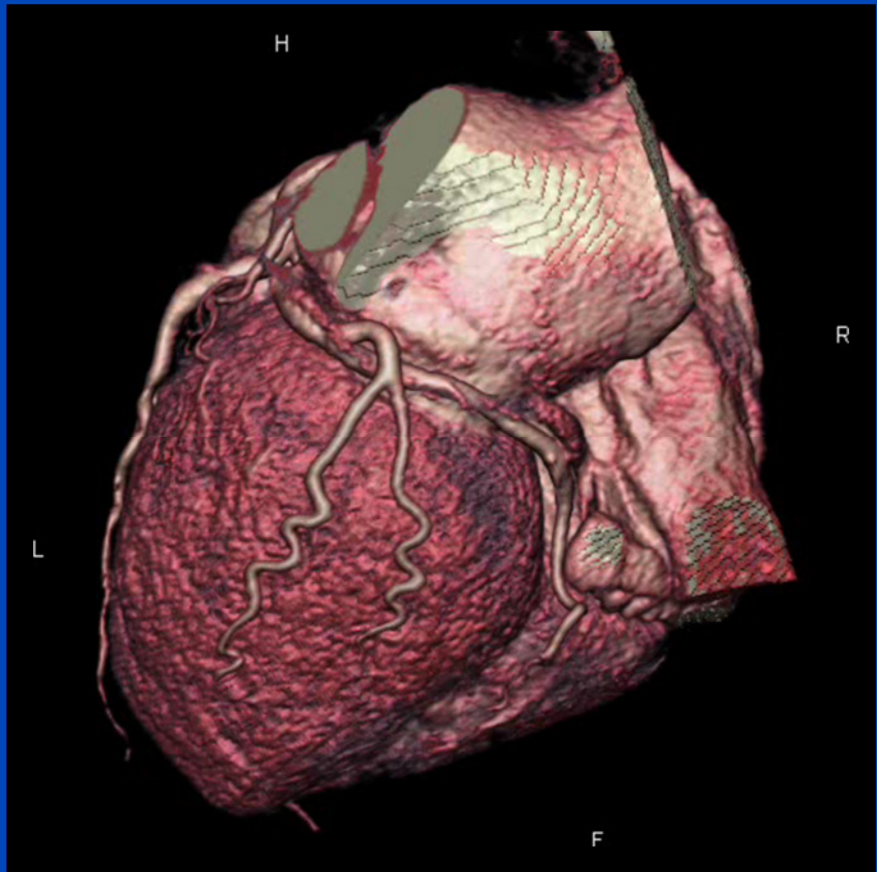


0%R-R | 25%R-R | 50%R-R | 75%R-R

Kachelrieß, Ulzheimer, Kalender, Med. Phys. 27(8):1881-1902 (2000)

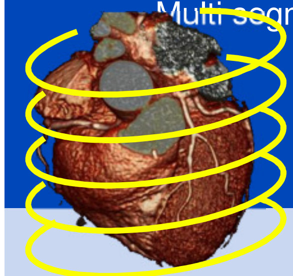
Volume Zoom, 4 × 2.5 mm, 0.5 s, 1998

Multi-segment reconstruction 180°MCI, 90 bpm

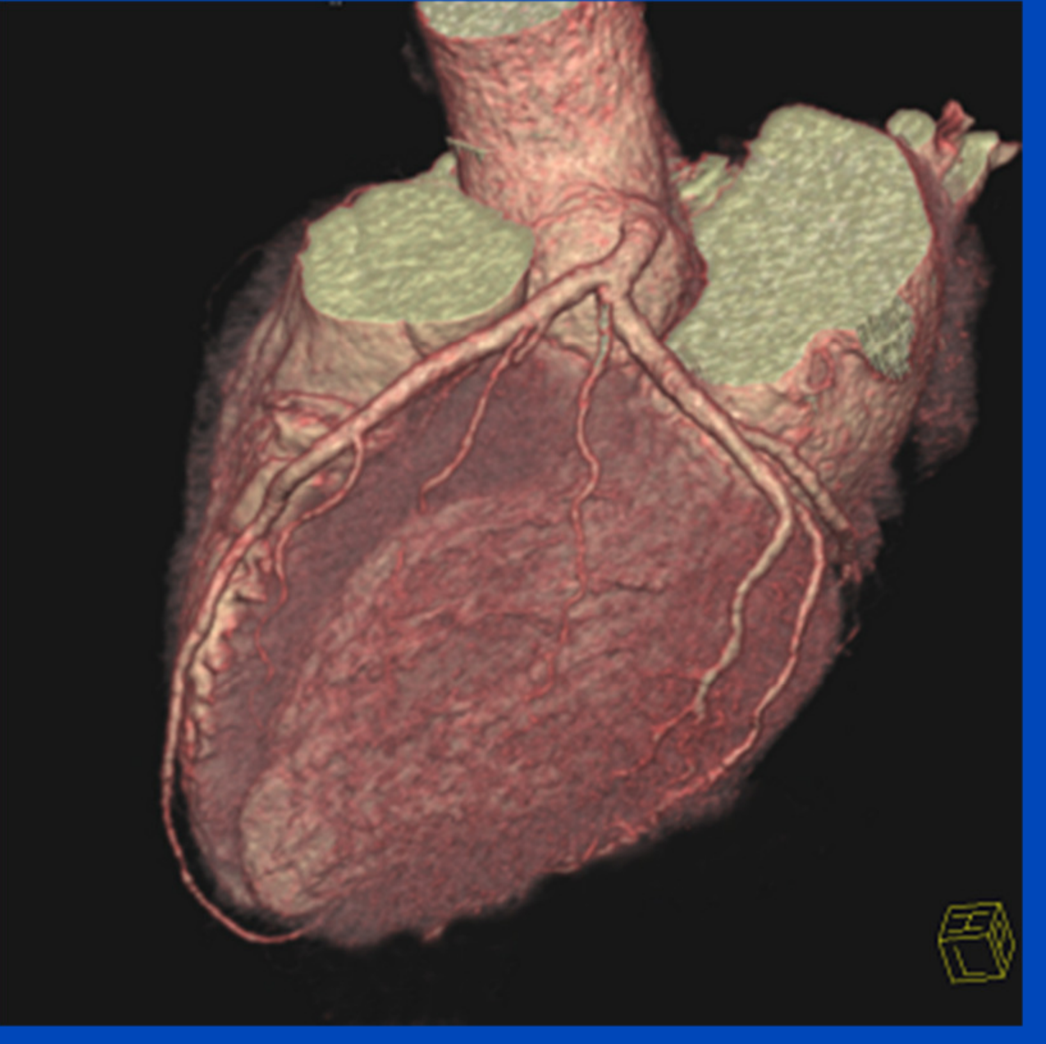


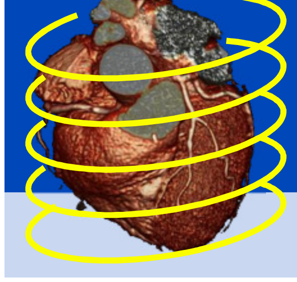
Sensation 64, 2·32 × 0.6 mm, 0.33 s, 2004

Data courtesy of Dr. Stephan Achenbach



Dual-source-CT, 330 ms rotation, partial scan reconstruction, 83 ms temporal resolution

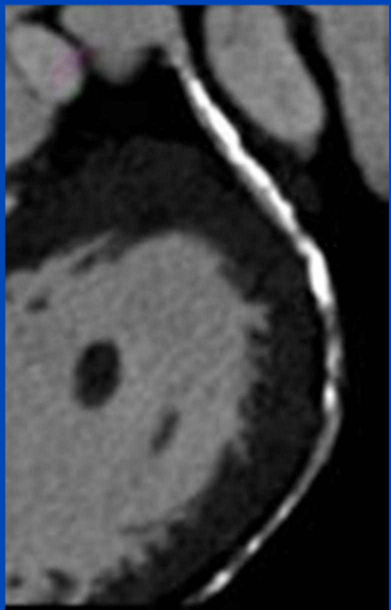


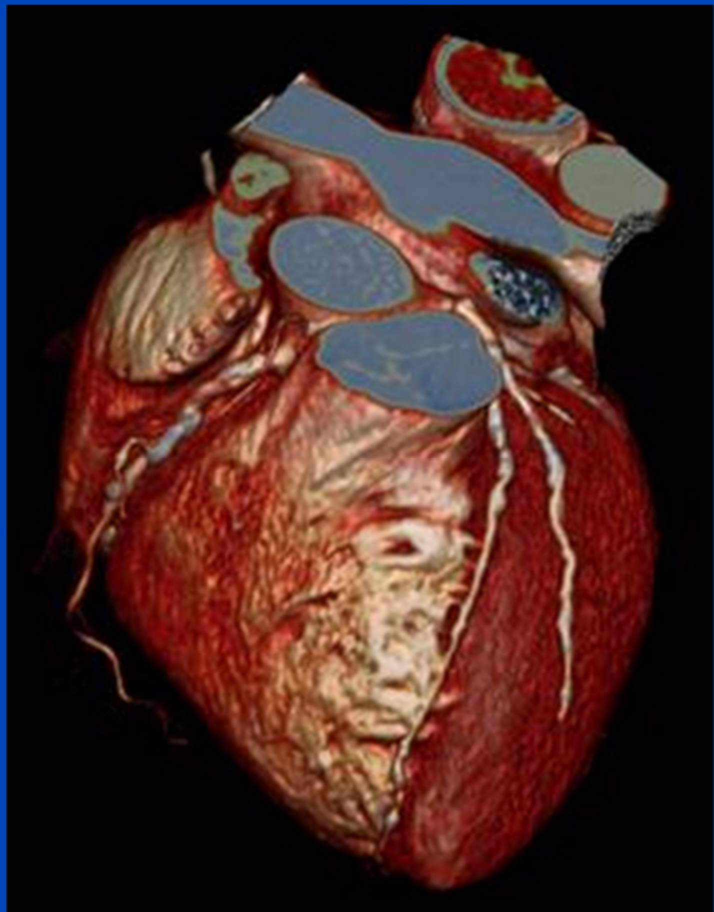


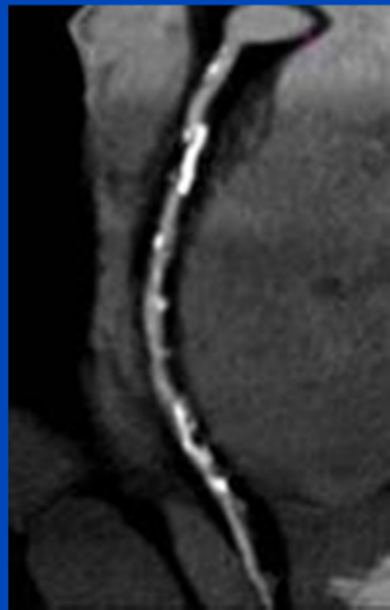
Data courtesy of Dr. Stephan Achenbach, Erlangen, Germany

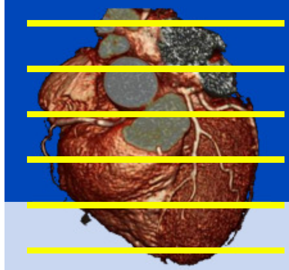
dkfz.

2·64×0.6 mm, 300 ms rotation, partial scan recon, 150 ms temporal resolution



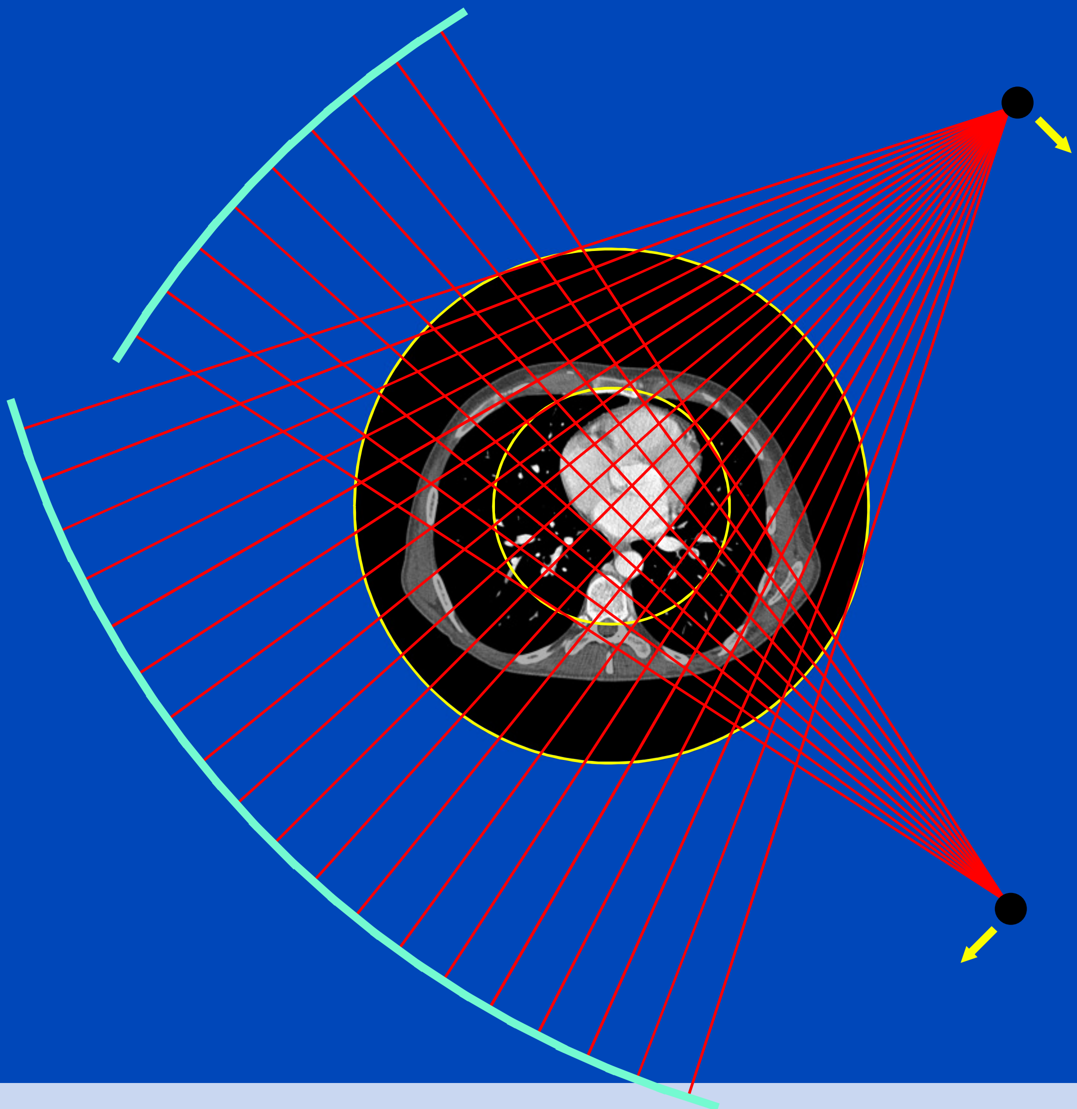


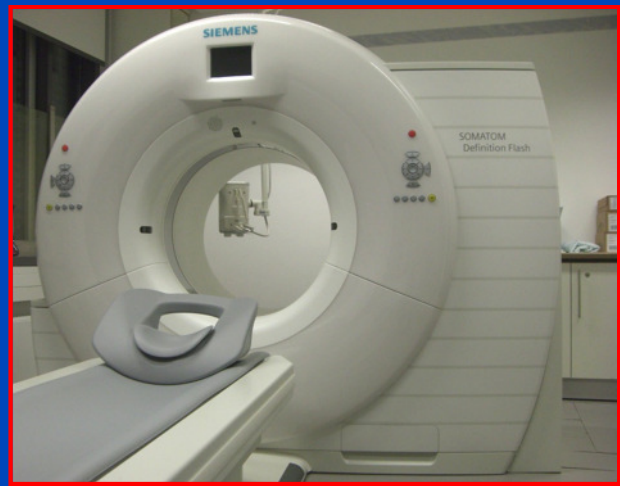




Data courtesy of Dr. Michael Lell, Erlangen, Germany

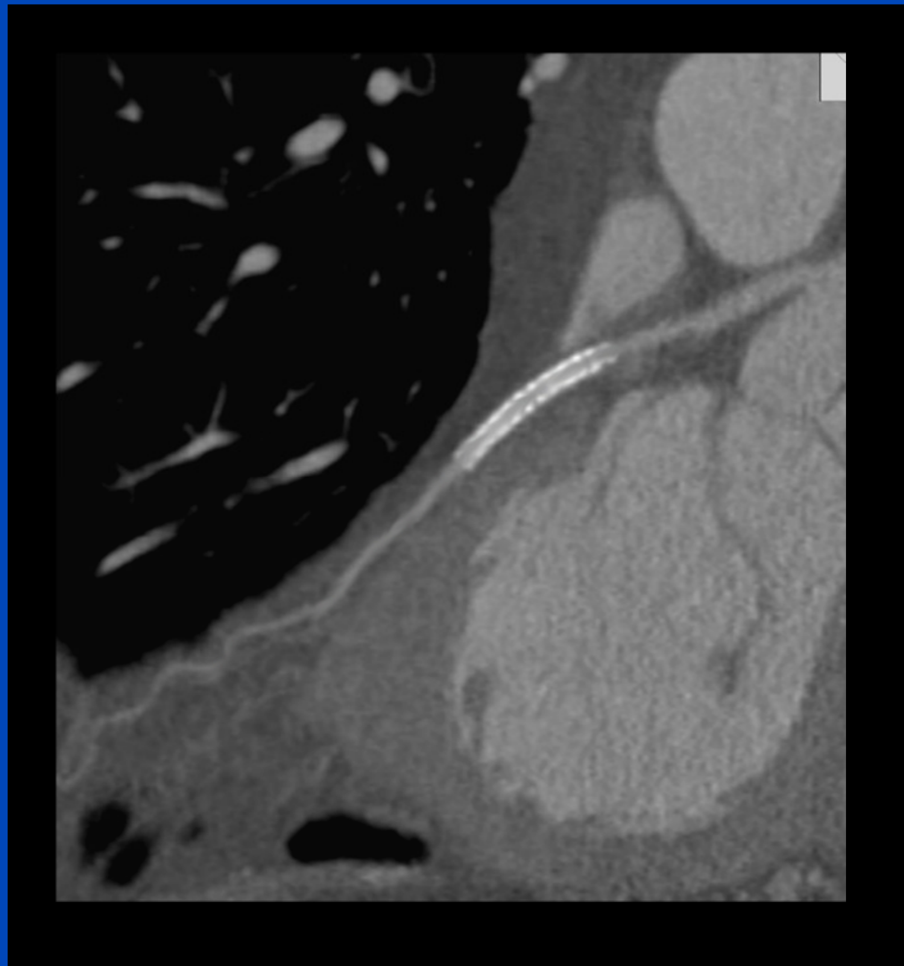
Multi-Threaded CT Scanners and Dual-Source-CT

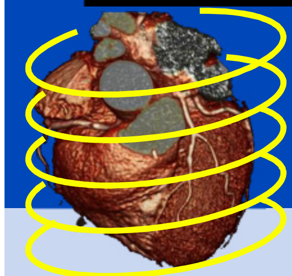




Siemens SOMATOM Definition Flash
dual source cone-beam spiral CT scanner

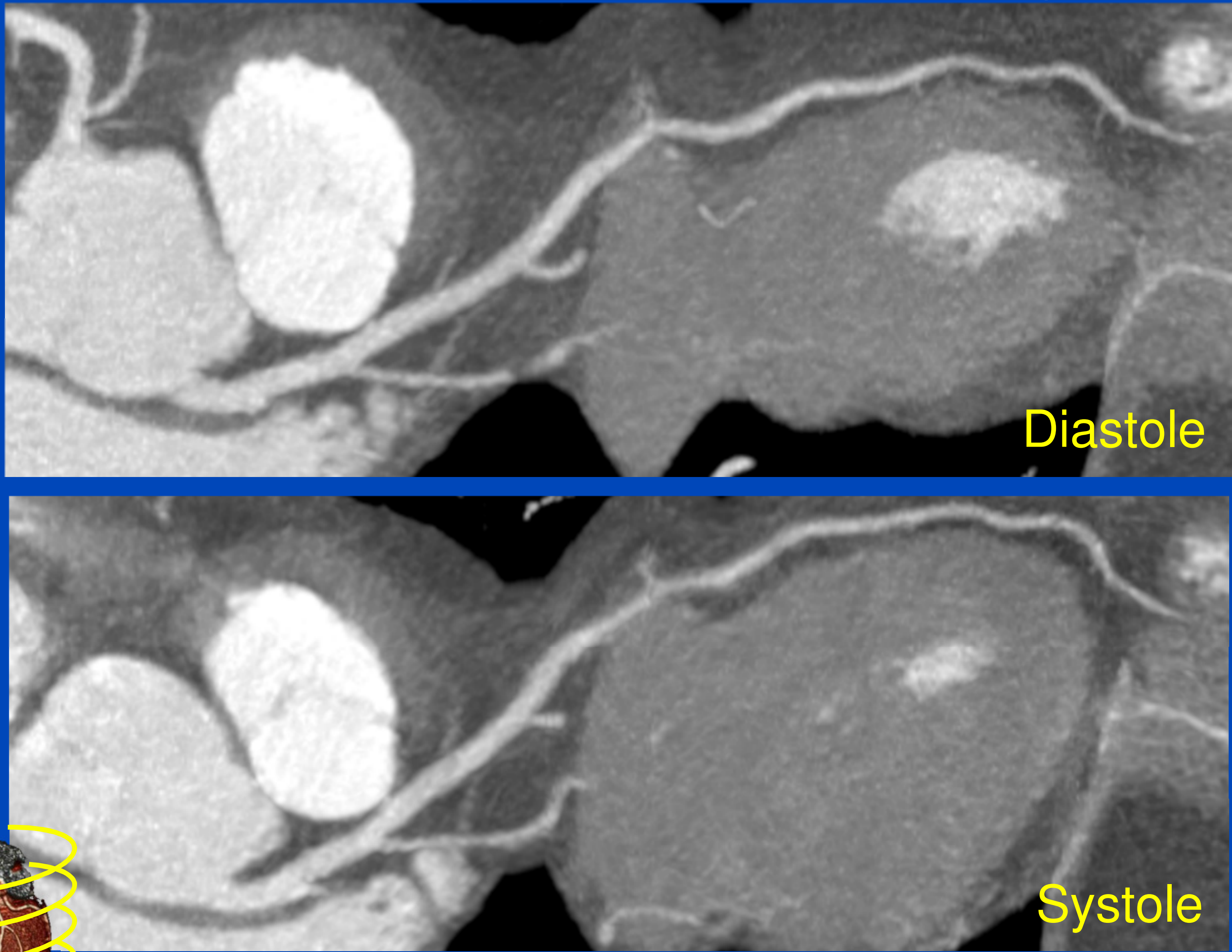
Dual-source-CT, 330 ms rotation, partial scan reconstruction, 83 ms temporal resolution

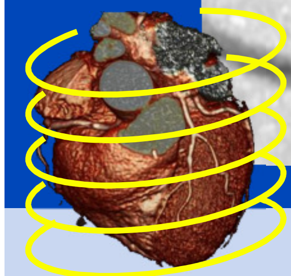




Data courtesy of Dr. Stephan Achenbach, Erlangen, Germany

Dual-source-CT, 330 ms rotation, partial scan reconstruction, 83 ms temporal resolution





Data courtesy of Dr. Stephan Achenbach, Erlangen, Germany

dkfz.

Dual-Source-CT

Flash Mode

280 ms Rotation

Partial scan reconstruction

70 ms temporal resolution

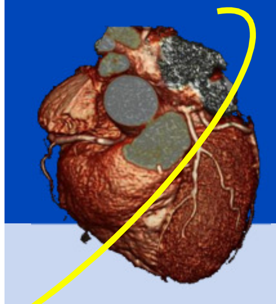
Pitch = 3.2 (43 cm/s)

320 mAs, 100 kV

10.6 cm scan range

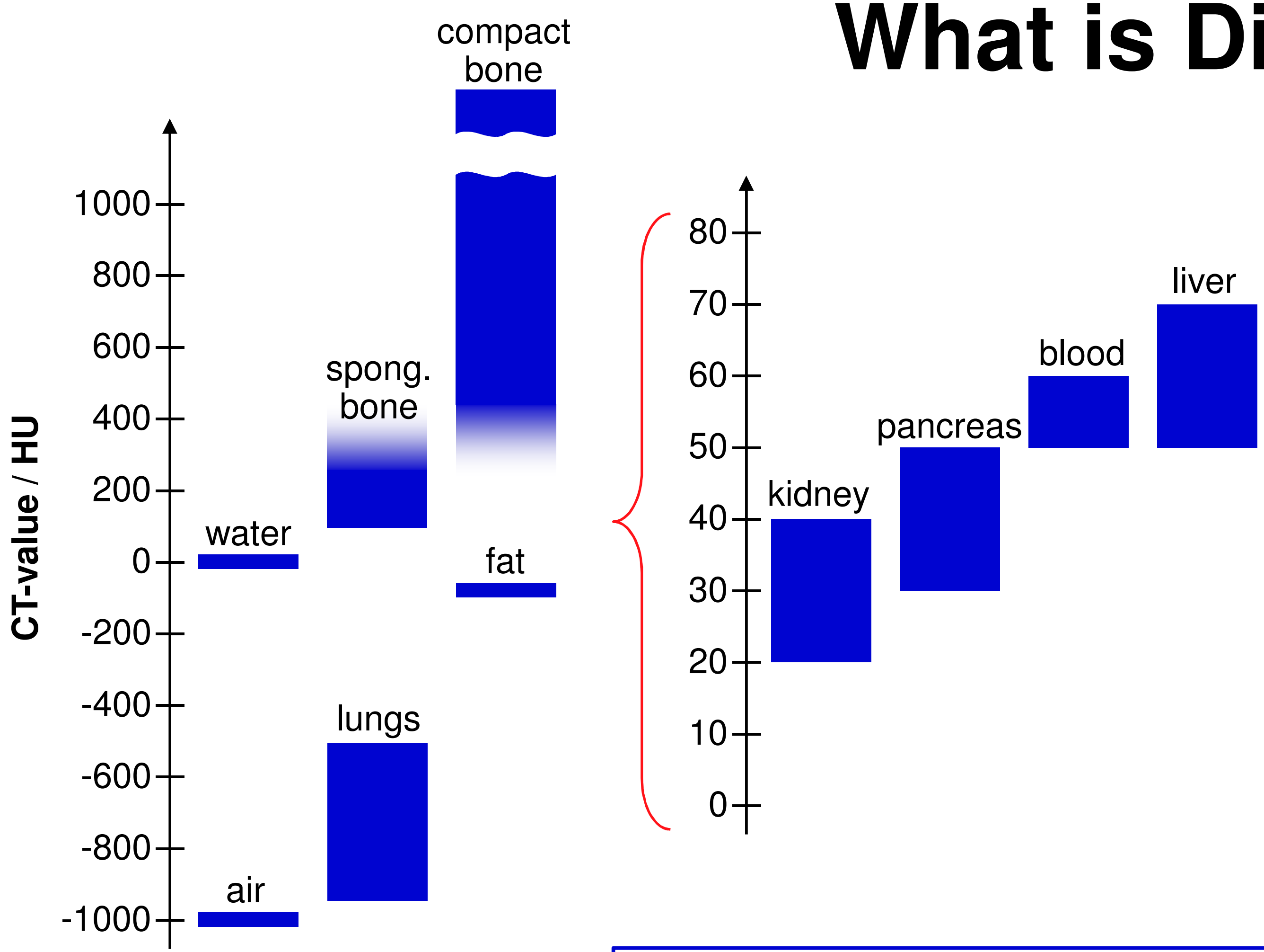
DLP = 64 mGy·cm

D_{eff} = 0.89 mSv

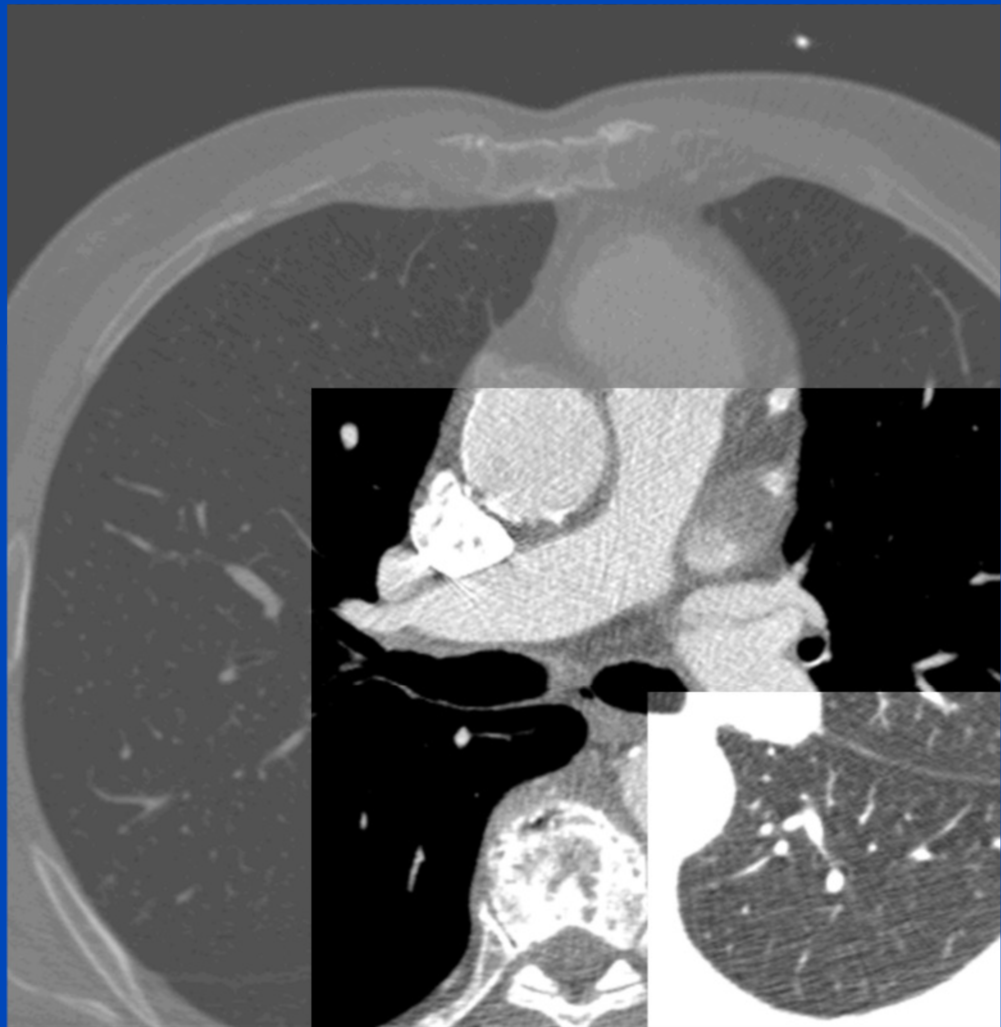


Data courtesy of Dr. Stephan Achenbach, Erlangen, Germany

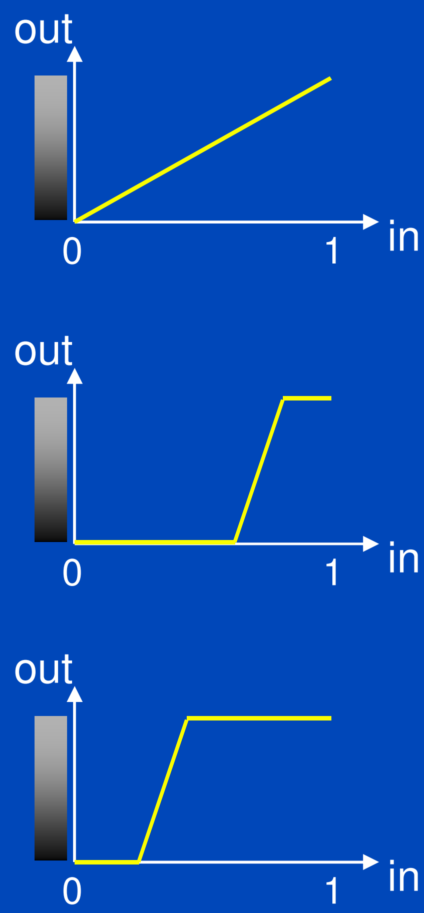
What is Displayed?



$$CT(\boldsymbol{r}) = \frac{\mu(\boldsymbol{r}) - \mu_{\text{Water}}}{\mu_{\text{Water}}} \cdot 1000\ \text{HU}$$

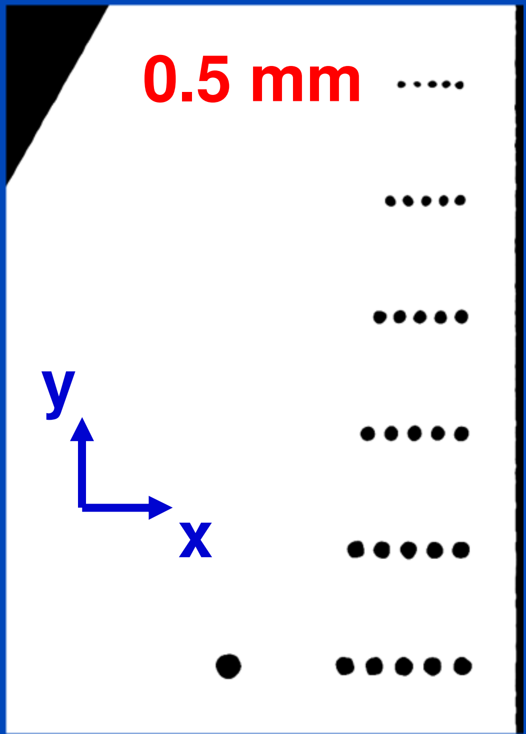


(0, 5000) (0, 1000) (-750, 1000)

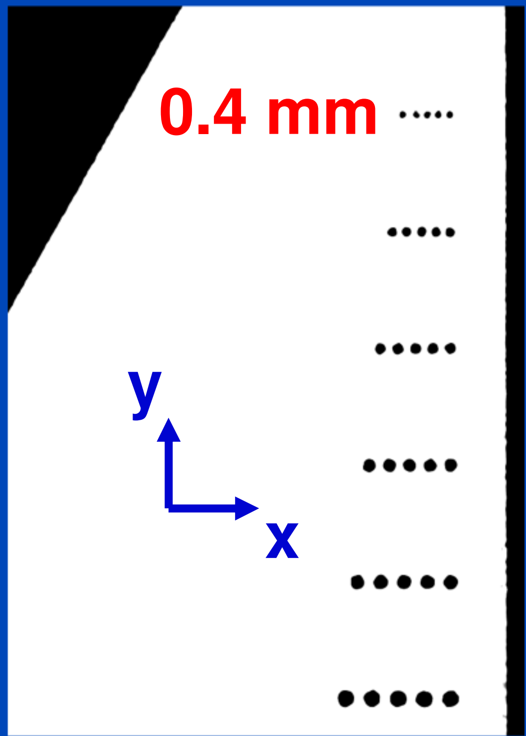


Spatial Resolution 1

In-plane resolution

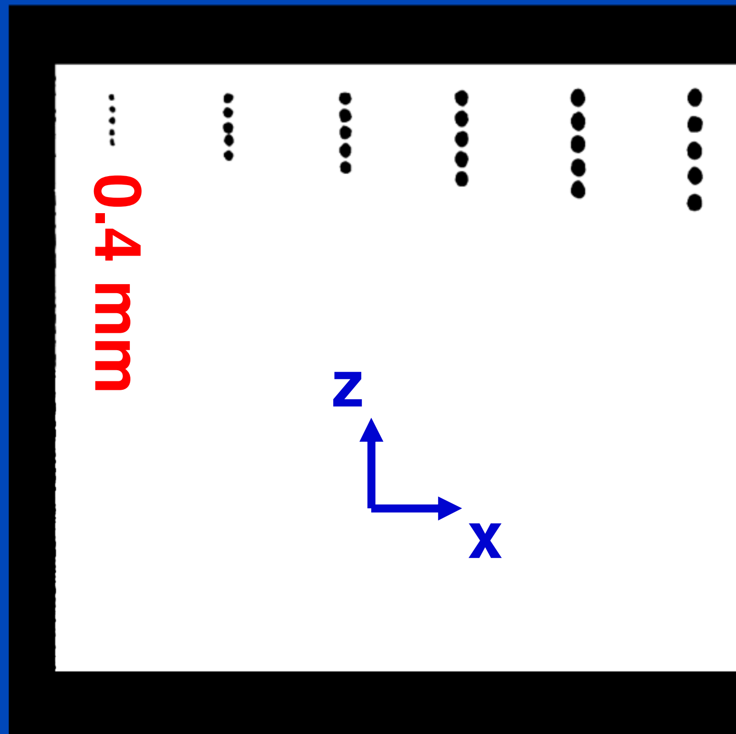


Std. scan, x/y



UHR scan, x/y

z-resolution



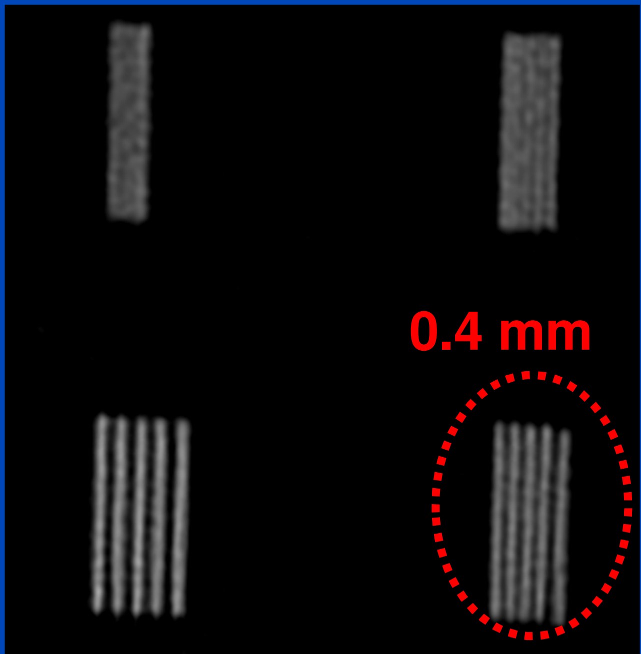
Std. or UHR scan, x/z

Sensation 64, collimation: $2 \cdot 32 \times 0.6$ mm

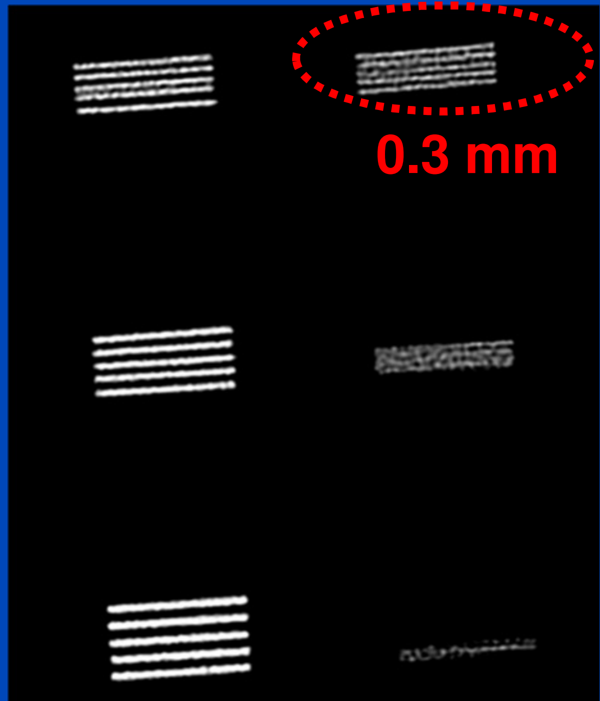
Spatial Resolution 2

In-plane resolution

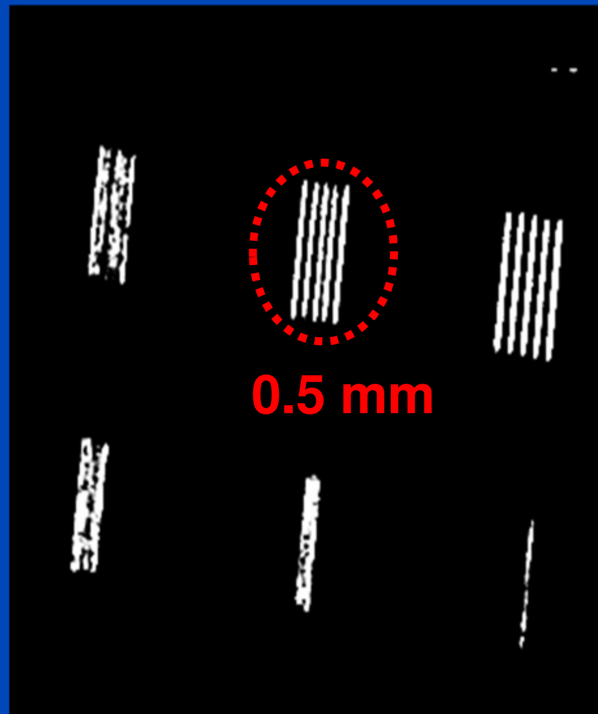
z-resolution



Std. scan, x/y



UHR scan, x/y

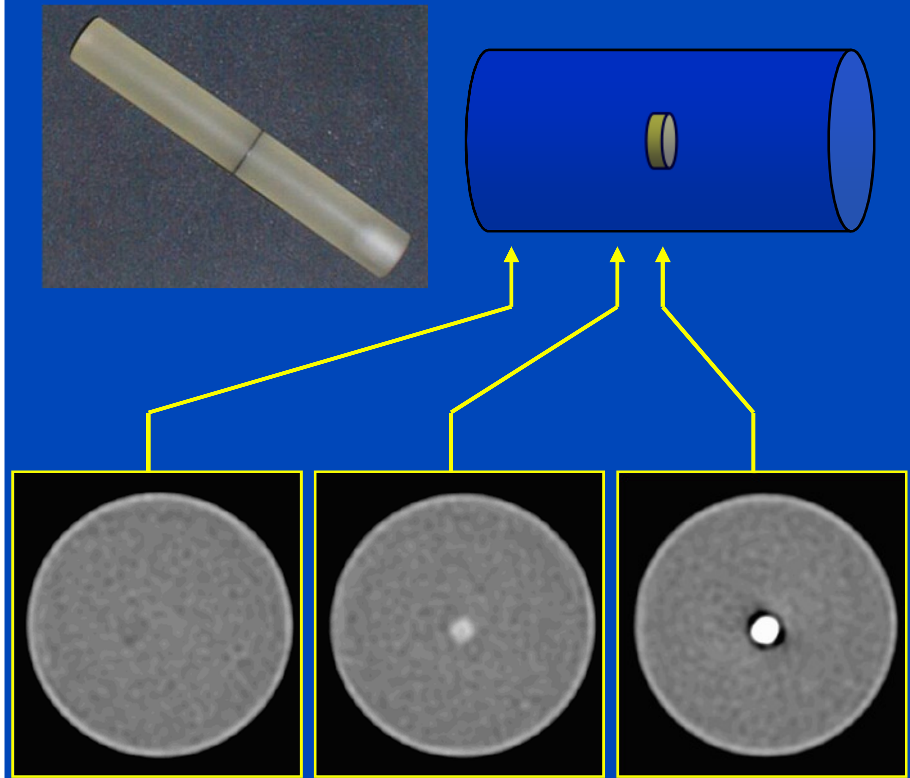


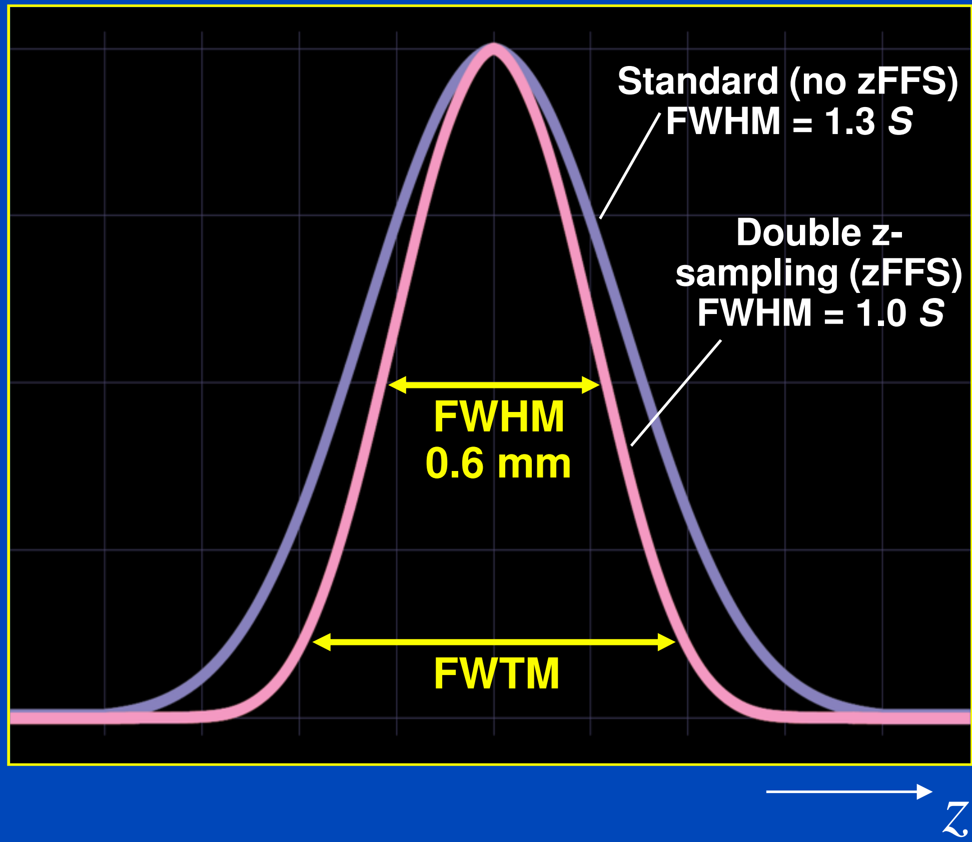
Std. or UHR scan, x/z

Sensation 64, collimation: $2 \cdot 32 \times 0.6$ mm

Spatial Resolution 3

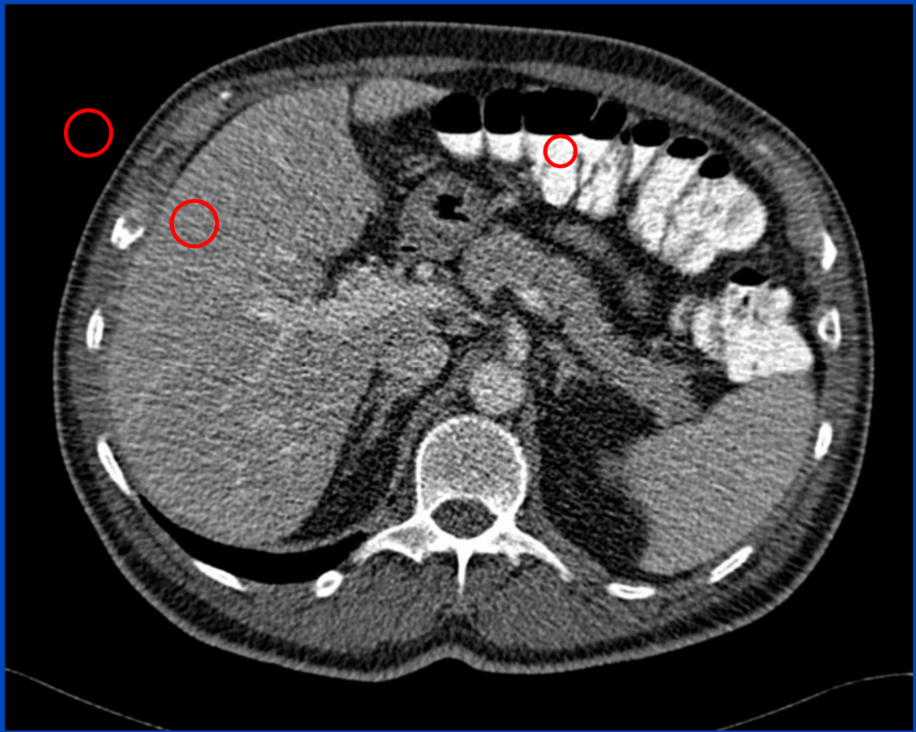
Point Spread Function (PSF), Slice Sensitivity Profile (SSP)



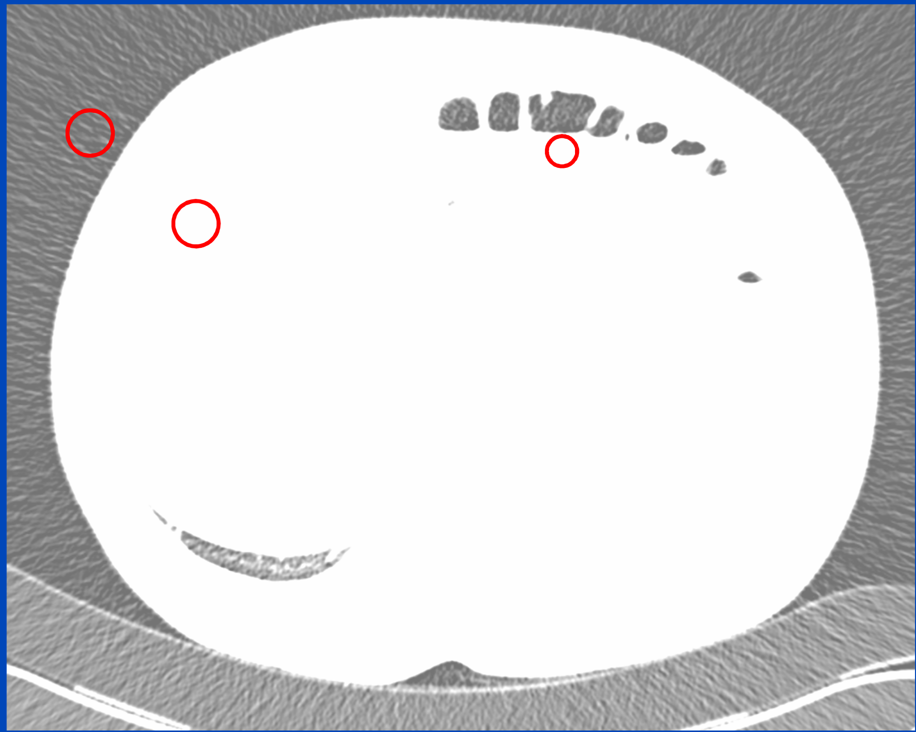


FWHM = S_{eff} = effective slice thickness = freely selectable parameter during image recon.

Image Noise



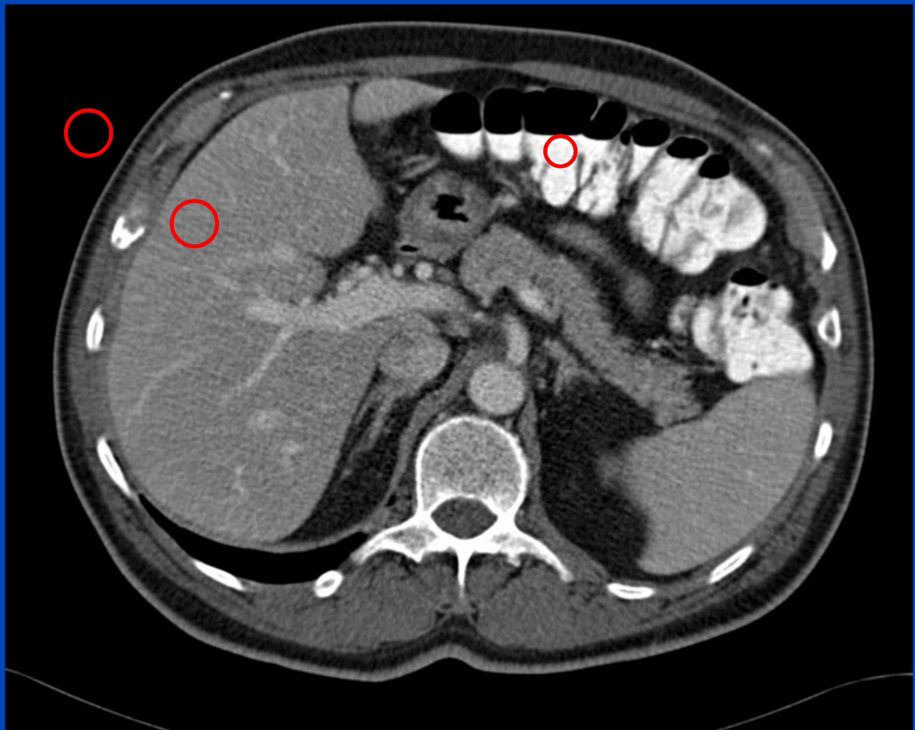
150 HU / 600 HU



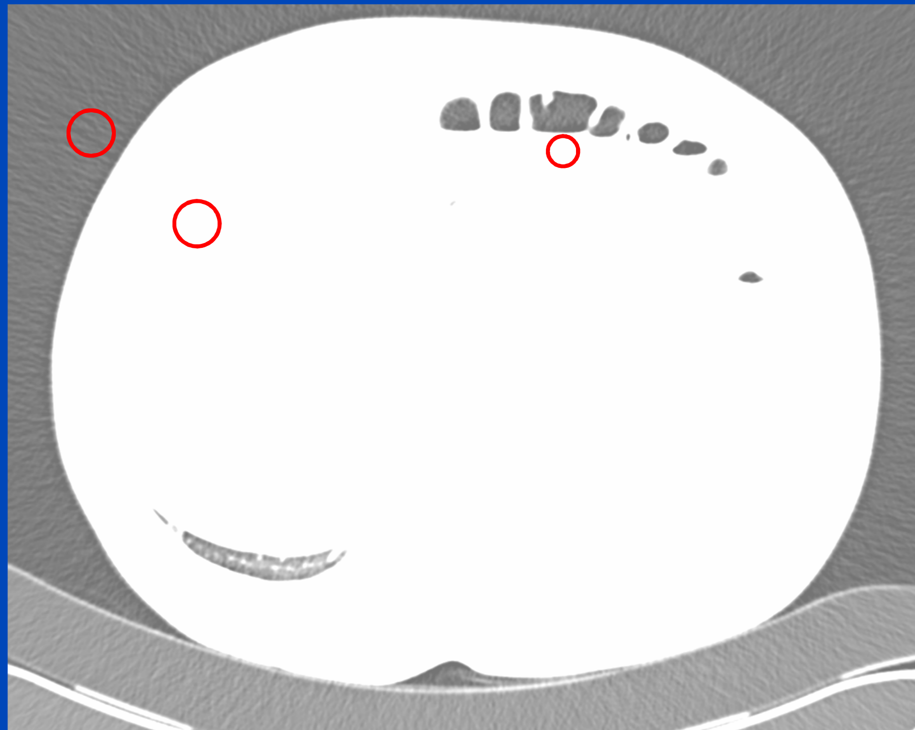
-1000 HU / 600 HU

Air ROI: μ = -995 HU, σ = 31 HU
Soft tissue ROI: μ = 148 HU, σ = 59 HU
Iodine ROI: μ = 423 HU, σ = 62 HU

Image Noise



150 HU / 600 HU



-1000 HU / 600 HU

Air ROI: μ = -998 HU, σ = 15 HU
Soft tissue ROI: μ = 149 HU, σ = 29 HU
Iodine ROI: μ = 425 HU, σ = 32 HU

Four-fold mAs reduces image noise by a factor of two.

Dependencies of IQ and Dose

- Image quality is determined by spatial resolution and contrast resolution (image noise)
- Image noise σ decreases with the square-root of dose

$$\sigma^2 = \text{Noise}^2 \propto \frac{1}{\text{Dose}} \propto \frac{1}{\text{mAs}_{\text{eff}}}$$

- Dose increases with the fourth power of the spatial resolution for a given object and image noise

$$\underbrace{\left(\frac{\sigma}{\mu}\right)^2}_{\text{Noise relative to the background (= 1/SNR)}} \propto \frac{e^{\mu 2R} + 1}{\mu^2 \underbrace{\Delta x^4}_{\text{Fourth power of the spatial resolution}}}$$

Noise relative to the background (= 1/SNR)

Fourth power of the spatial resolution

The Effective mAs-Value

A Measure of Dose

The effective current-time-product mAs_{eff} is a measure of the number of quanta contributing to a given axial slice $z = z_R$. It accounts for the degree of scan overlap by normalization with the pitch value:

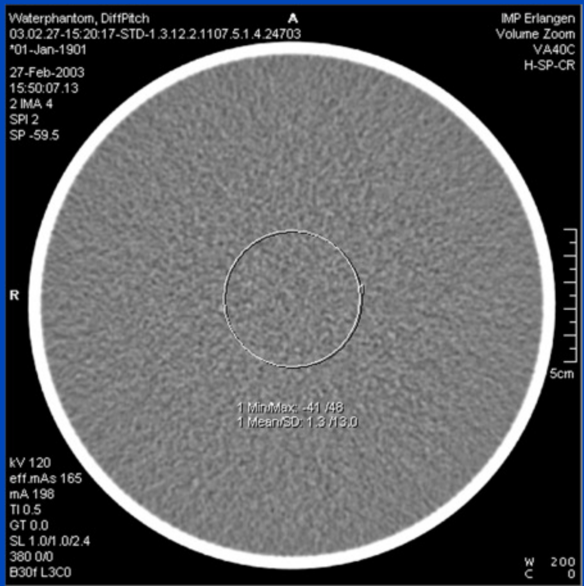
$$\mathrm{mA} = \frac{\mathrm{mAs_{eff}} \cdot \mathrm{Pitch}}{\mathrm{RotationTime}}$$

$$\mathrm{Dose} \propto \mathrm{mAs_{eff}}$$

The concept of selecting the mAs_{eff} guarantees pitch independent image noise and pitch independent patient dose. In addition, MSCT provides pitch independent axial resolution.

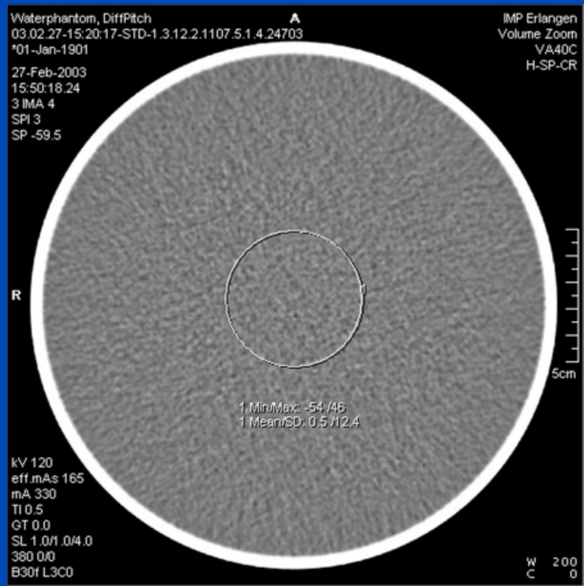
Water Phantom at 165 mAs_{eff}

Scan 1
Pitch 0.6



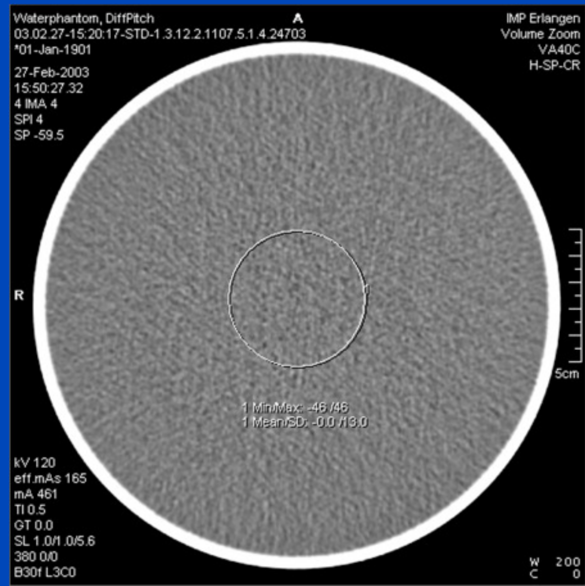
165 mAs_{eff} yields:
198 mA, σ=13.0 HU

Scan 2
Pitch 1.0



165 mAs_{eff} yields:
330 mA, σ= 12.4 HU

Scan 3
Pitch 1.4



165 mAs_{eff} yields:
461 mA, σ= 13.0 HU

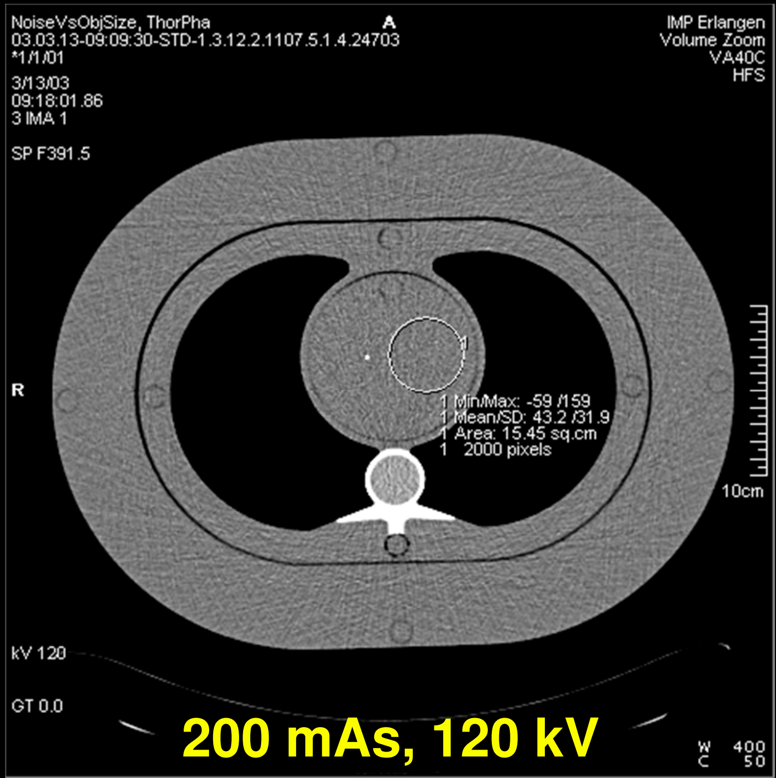
⇒ Same noise, same image quality and same dose with MSCT regardless of the pitch value!

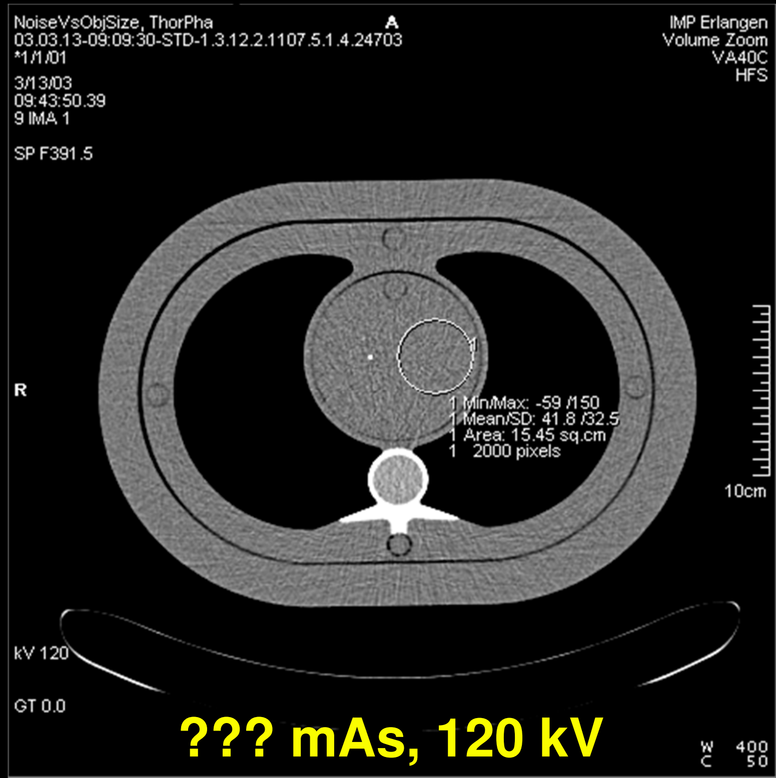
Wie muss der mAs-Wert geändert werden, um das Bildpunktrauschen zu halbieren?

1. Man muss den mAs-Wert halbieren.
2. Der mAs-Wert muss mit $\sqrt{2} \approx 1{,}4$ multipliziert werden.
3. Der mAs-Wert sollte verdoppelt werden.
4. Der vierfache mAs-Wert muss gewählt werden.
5. Das Rauschen ist vom mAs-Wert unabhängig.

TED

Patient Size vs. mAs at Constant Noise

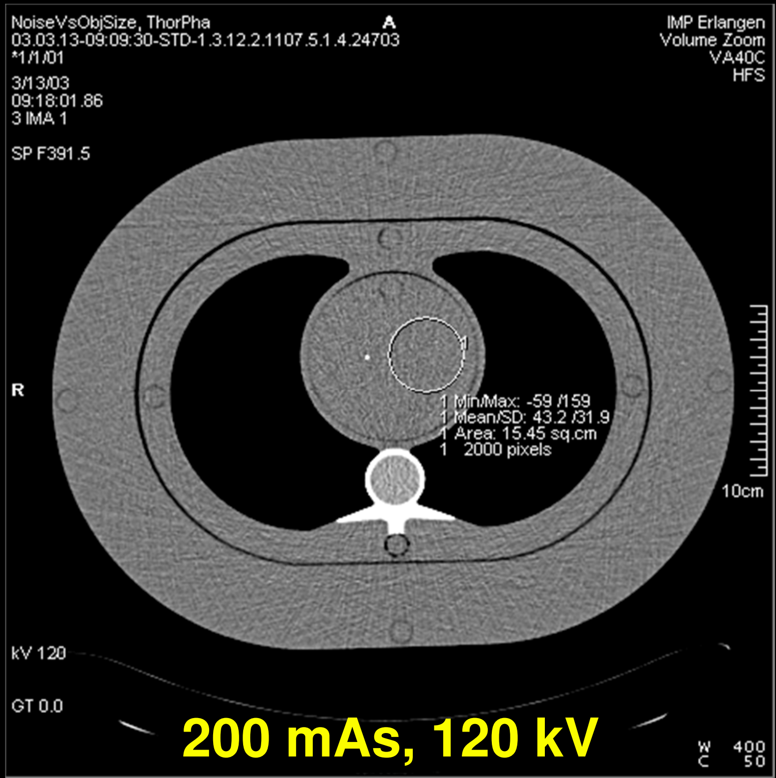


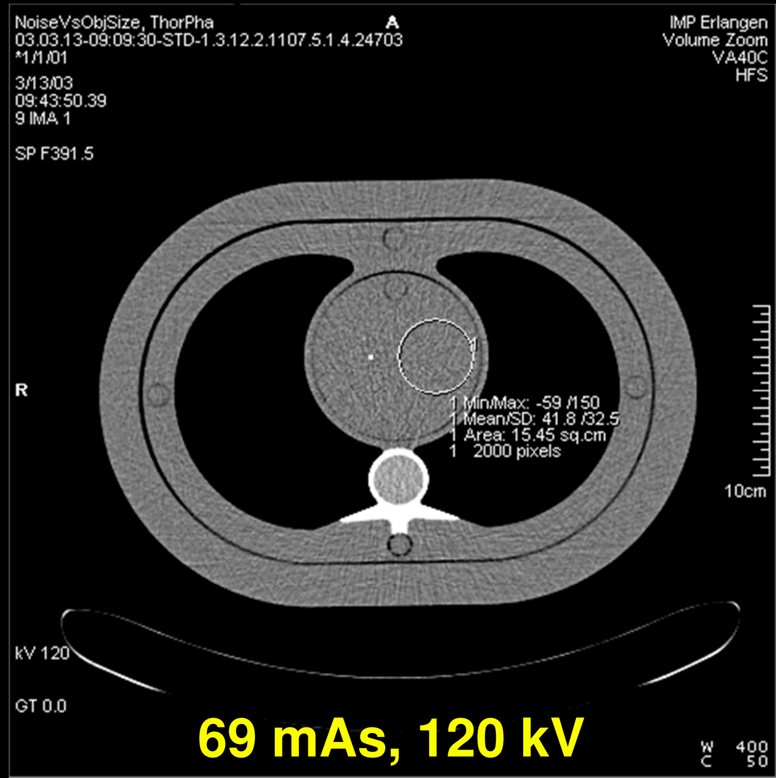


How much would you reduce the tube current when the patient size is reduced by 5 cm?

a) to 130 mAs b) to 100 mAs c) to 70 mAs

Patient Size vs. mAs at Constant Noise





	40 cm × 30 cm	35 cm × 25 cm	HVL_{eff}
80 kV	957 mAs	287 mAs	2.9 cm
120 kV	200 mAs	69 mAs	3.3 cm
140 kV	142 mAs	51 mAs	3.4 cm

HVL_{eff} is the effective (wrt the phantom) half value layer


Thank You!
marc.kachelriess@dkfz.de