Understanding Image Quality and Radiation Dose in MSCT and CBCT

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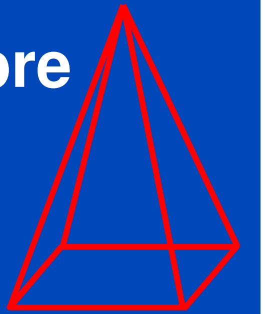
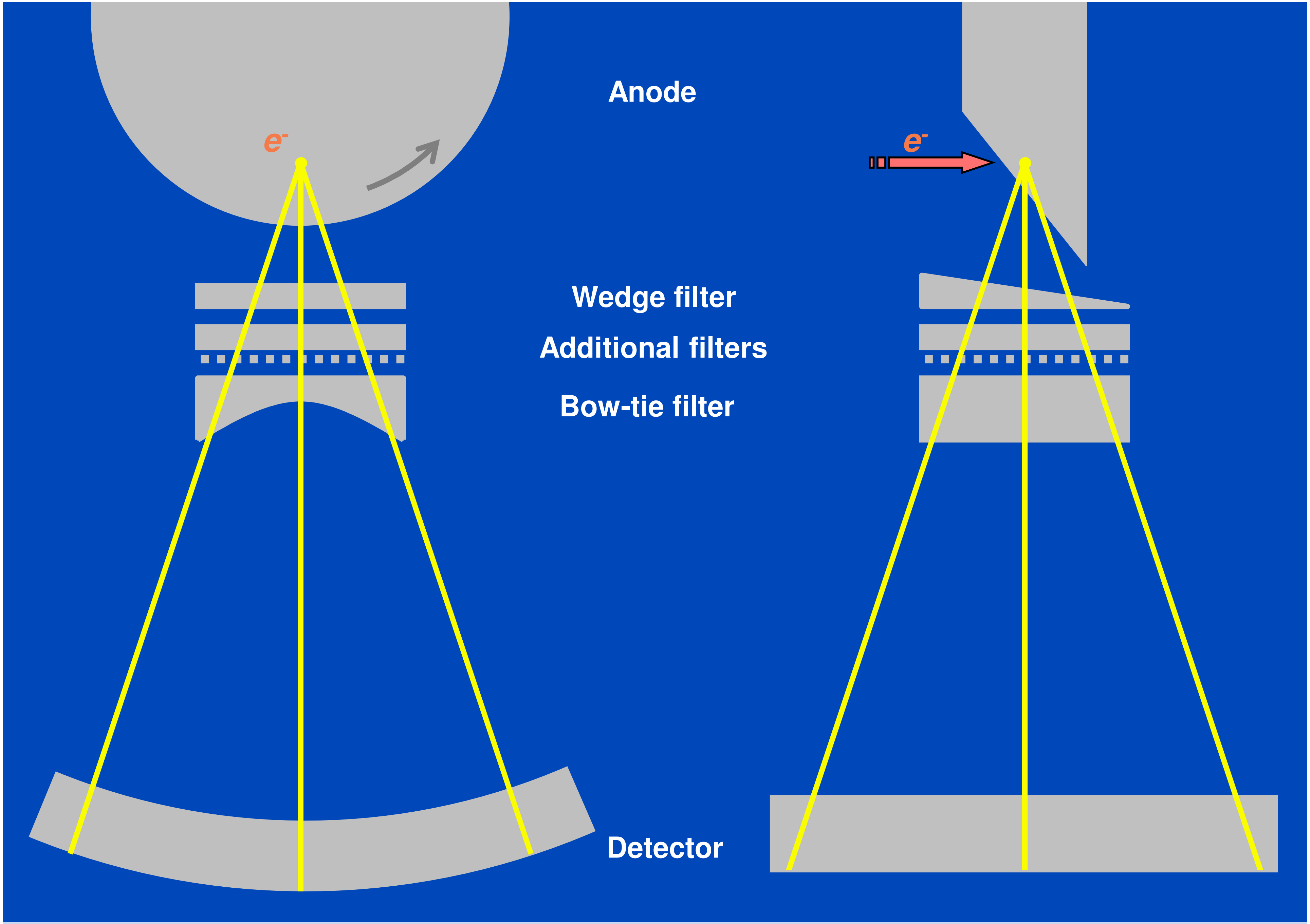
DEUTSCHES
KREBSFORSCHUNGSZENTRUM
IN DER HELMHOLTZ-GEMEINSCHAFT

Terminology Cone-Beam CT



- The shape of the x-ray ensemble depends on the pre patient collimation and can be approximated by
  - a cone if the detector is a circle (e.g. an image intensifier)
  - a pyramid if the detector is a rectangle (e.g. a flat detector)
  - a distorted pyramid if the detector is an arc (e.g. a clinical CT detector)
- Cone-beam CT =
  - a CT with many detector rows?
  - a CT equipped with flat detectors?
  - a CT that requires a volumetric reconstruction!
- Flat detector =
  - indirect converting, based on TFT (amorph. Si) or CMOS (cryst. Si)
  - flat surface, low aspect ratio (number of columns ≈ number of rows)
- Often used synonymously:
  - CT = clinical CT = diagnostic CT = multi slice CT (MSCT)
    = multi detector <u>row</u> CT (MDCT)
  - CBCT = cone-beam CT = flat detector CT (FDCT)



Anode

Wedge filter

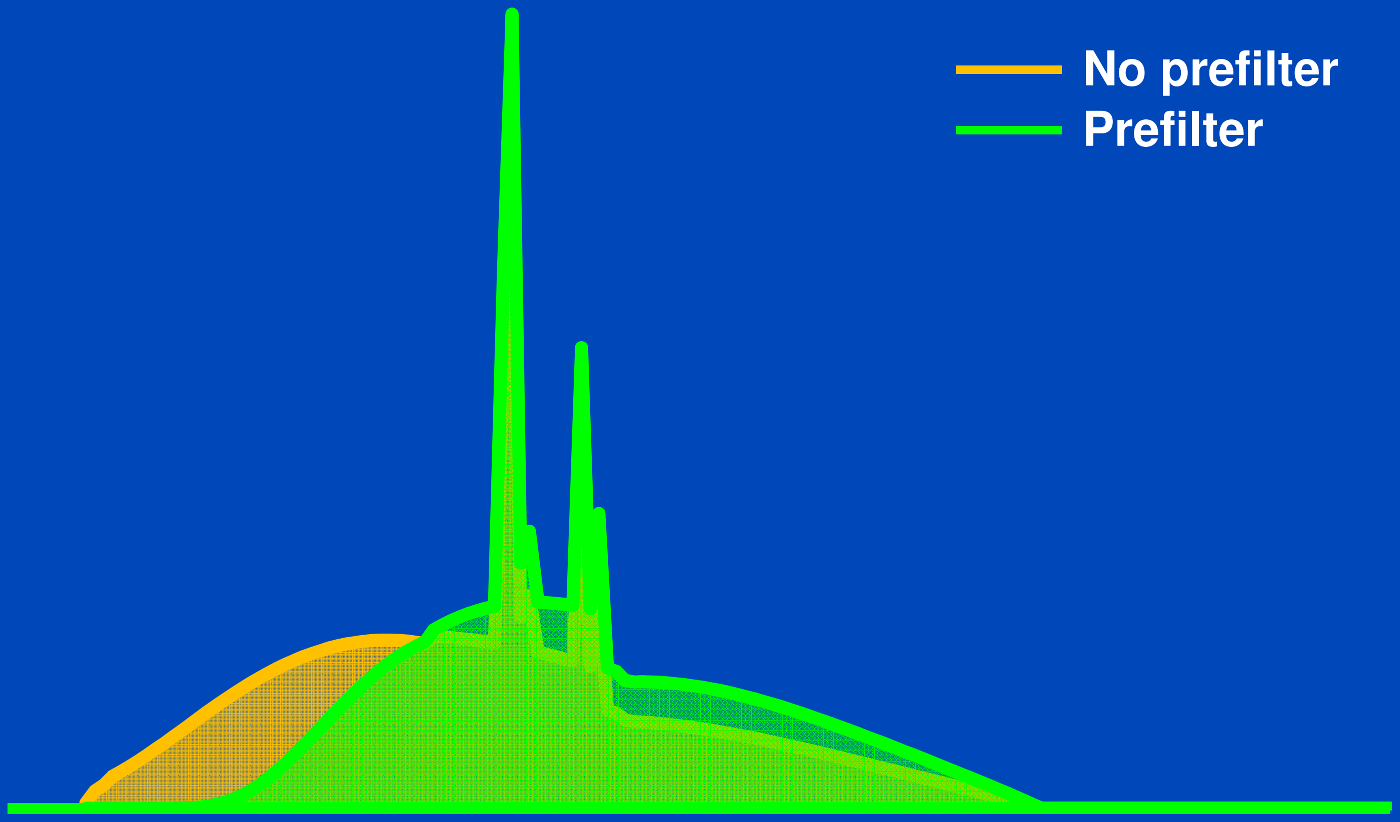
Additional filters

Bow-tie filter

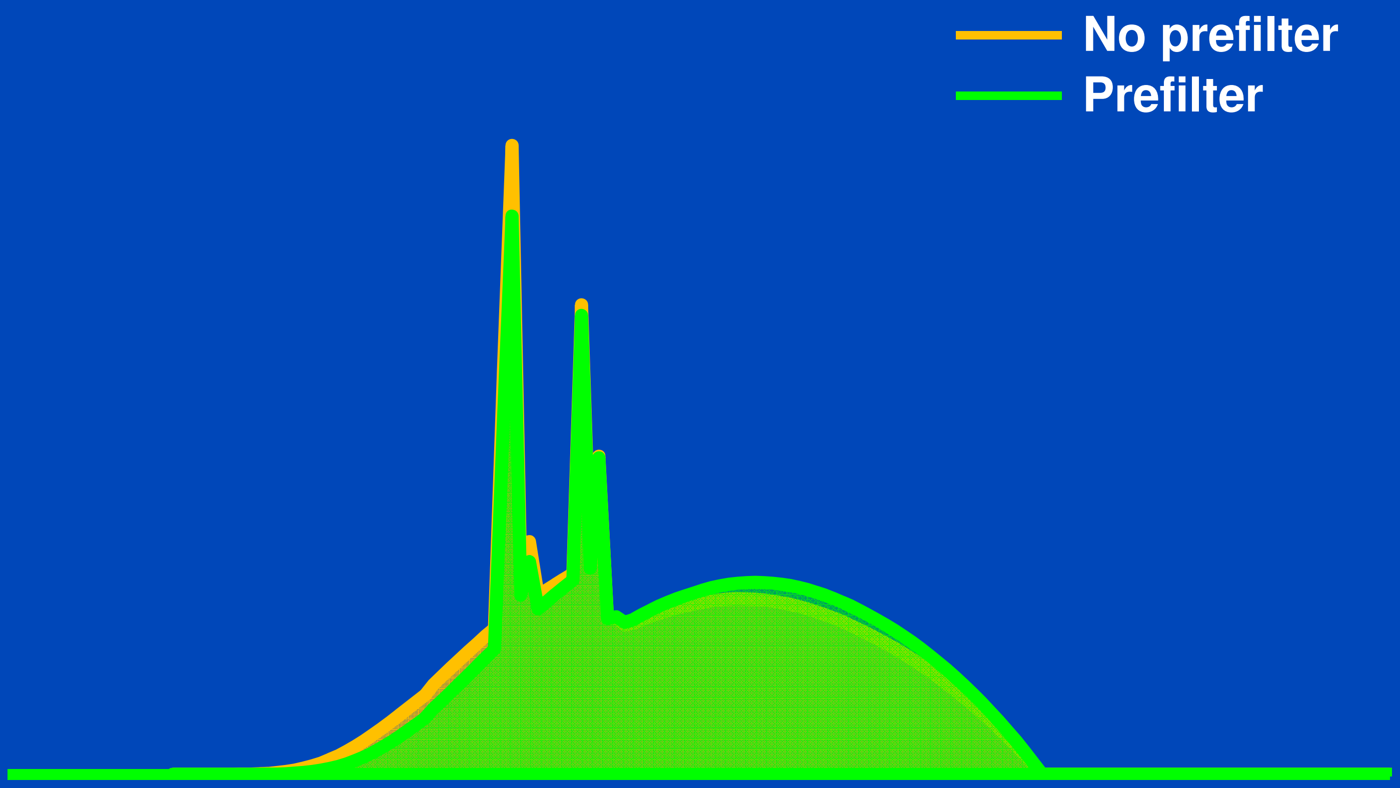
Detector

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

120 kV + 0 mm water with and without prefilter



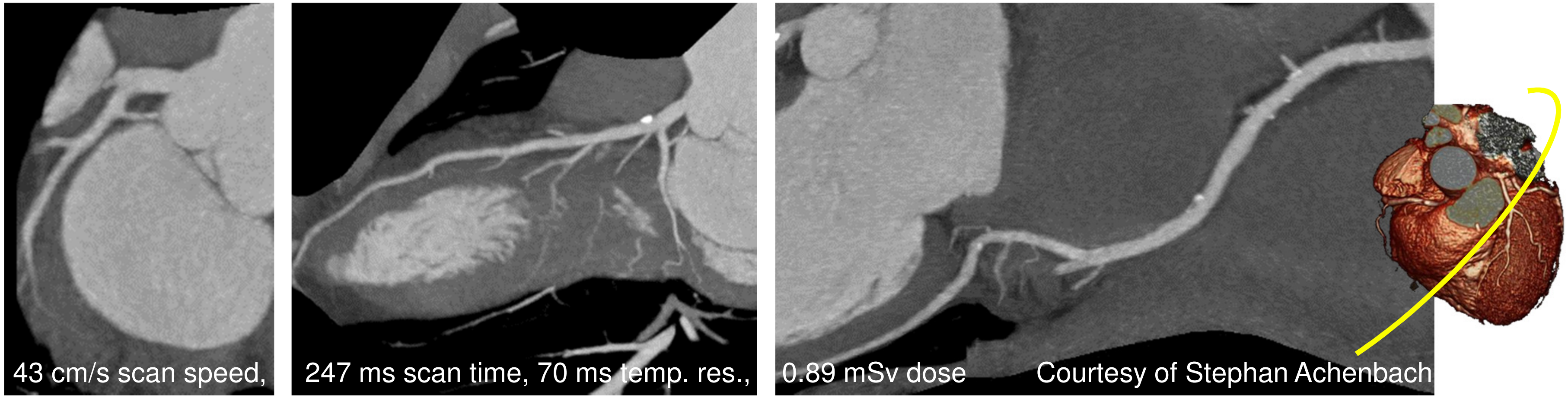
120 kV + 320 mm water with and without prefilter



Clinical CT

Clinical CT detector inside

e.g. Definition Flash dual source spiral cone-beam CT scanner,
Siemens Healthcare, Forchheim, Germany.



43 cm/s scan speed, 247 ms scan time, 70 ms temp. res., 0.89 mSv dose

Courtesy of Stephan Achenbach

Image courtesy by Siemens Healthcare

Fixed C-Arm CT

Flat detector inside

e.g. floor-mounted Artis Zeego or ceiling-mounted Artis Zee, Siemens Healthcare, Forchheim, Germany

Image courtesy by Siemens Healthcare

Mobile C-Arm CT

Flat detector inside

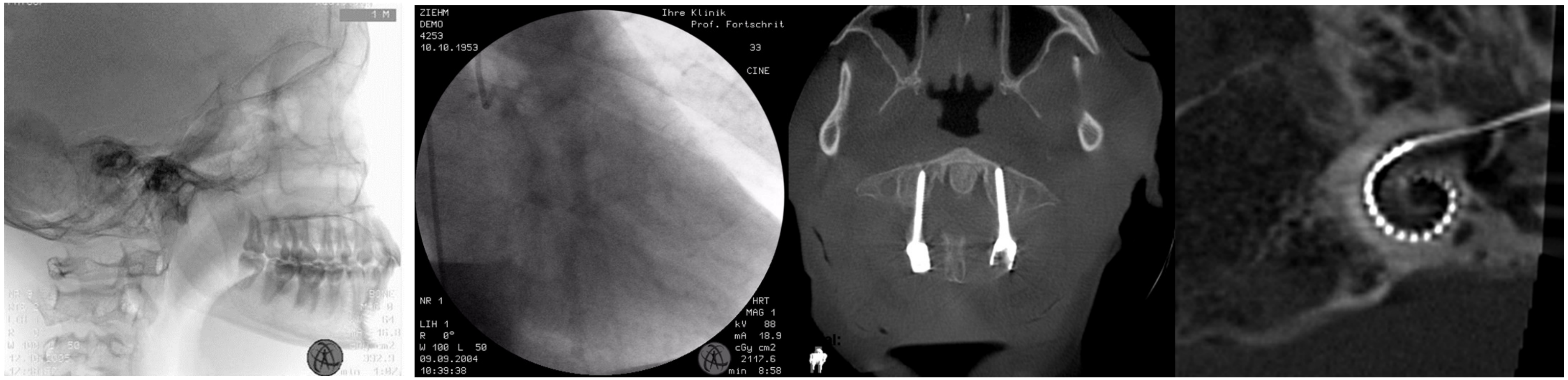
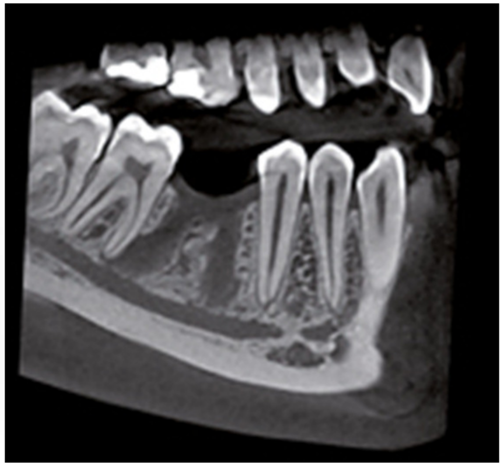


Image courtesy by Ziehm Imaging

Dental Volume Tomography (DVT)

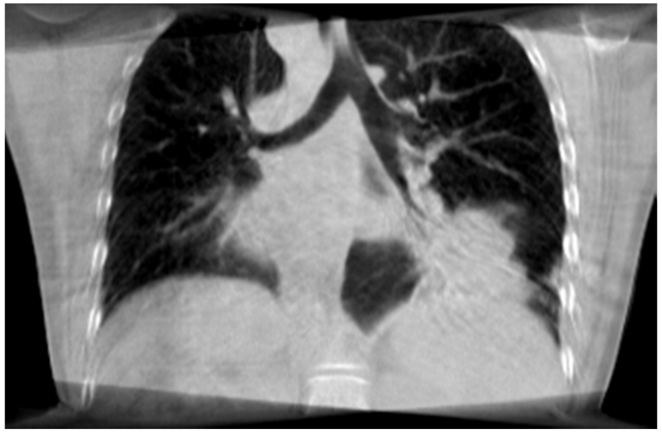
Flat detector inside



e.g. Orthophos XG 3D, Sirona Dental Systems GmbH, Bensheim, Germany

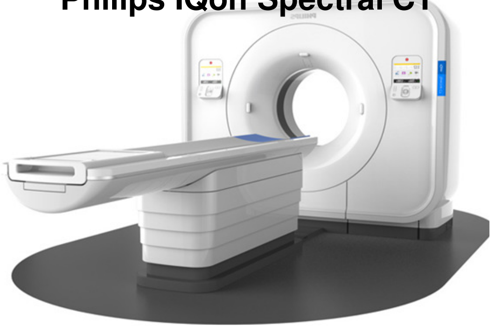
Image courtesy by Sirona Dental

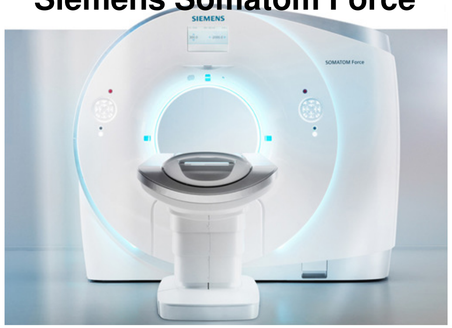
CBCT Guidance for Radiation Therapy



e.g. TrueBeam, Varian Medical Systems, Palo Alto, CA, USA


GE Revolution CT


Philips IQon Spectral CT


Siemens Somatom Force


Toshiba Aquilion ONE Vision

In-plane resolution: 0.4 … 0.7 mm

Nominal slice thickness: S = 0.5 … 1.5 mm

Tube (max. values): 120 kW, 150 kV, 1300 mA

Effective tube current: mAs_{eff} = 10 mAs … 1000 mAs

Rotation time: T_{rot} = 0.25 … 0.5 s

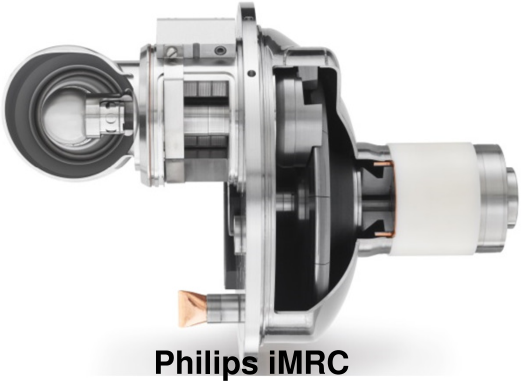
Simultaneously acquired slices: M = 16 … 320

Table increment per rotation: d = 1 … 183 mm

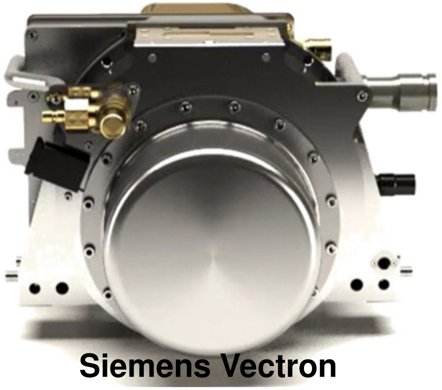
Scan speed: up to 73 cm/s

Temporal resolution: 50 … 250 ms

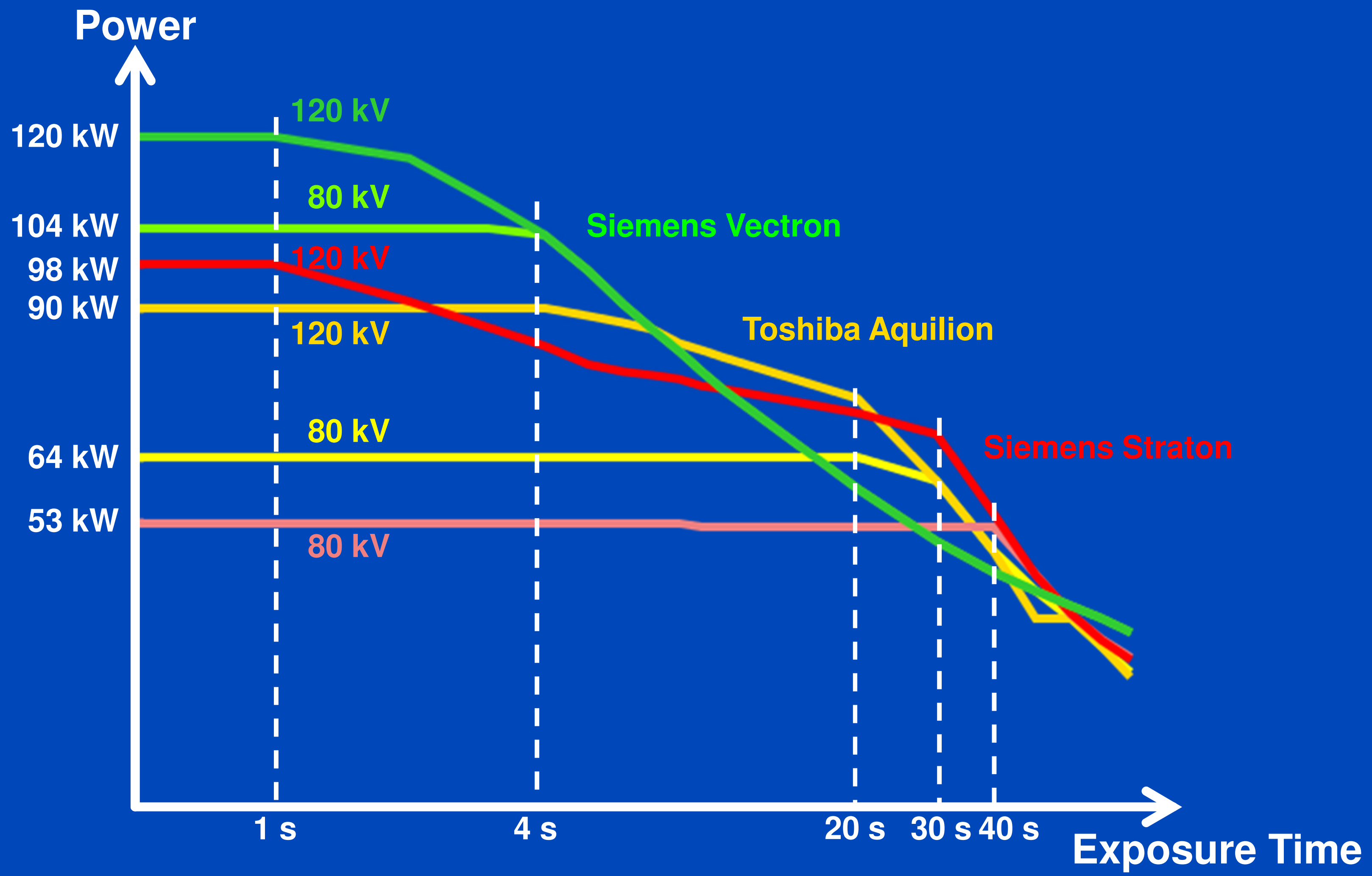

GE Performix HDw


Philips iMRC


Siemens Straton


Siemens Vectron

Flash, Force, Aquilion



With more Powerful X-Ray Tubes

a) faster scans are possible.

b) we can go to lower tube voltages, implying a contrast increase and a decrease in patient dose.

c) an increase in patient dose is likely.

d) stronger prefiltration can be used, resulting in a decrease in patient dose.

e) the detector thickness can be decreased and thereby the spatial resolution will increase.

E-Vote

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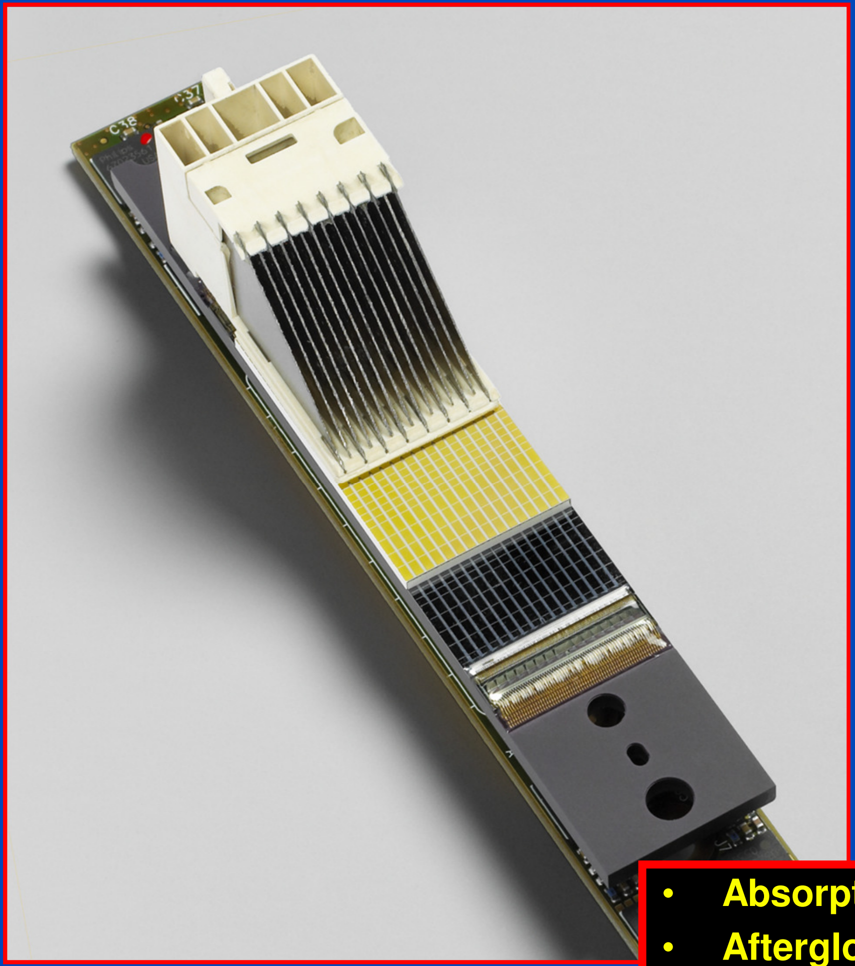
Detector Technology

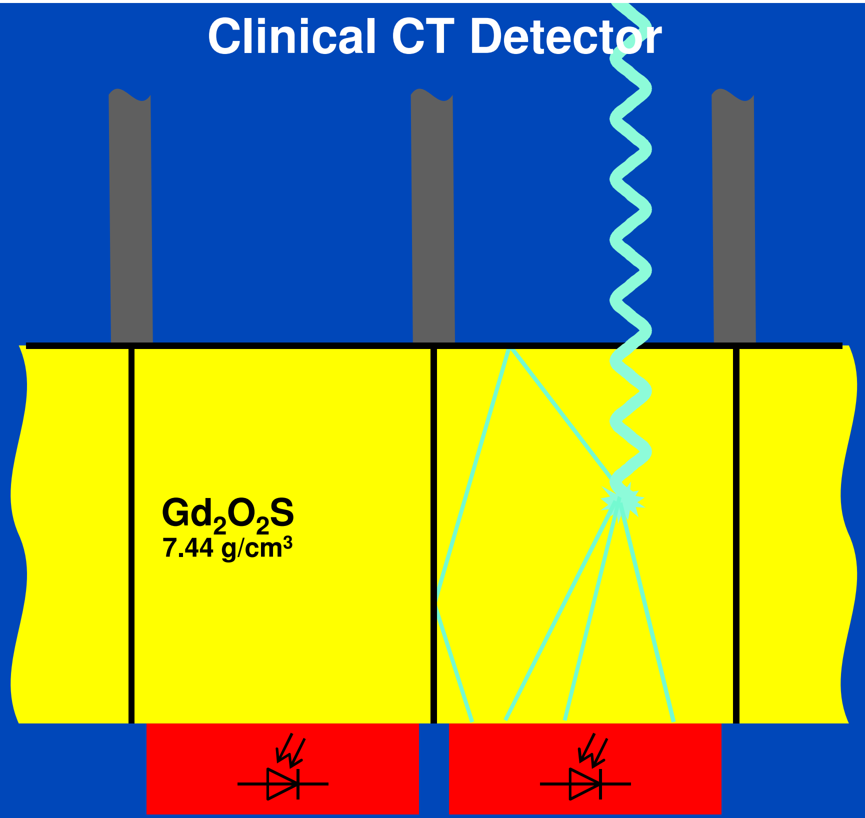
Clinical CT Detector



Flat Detector

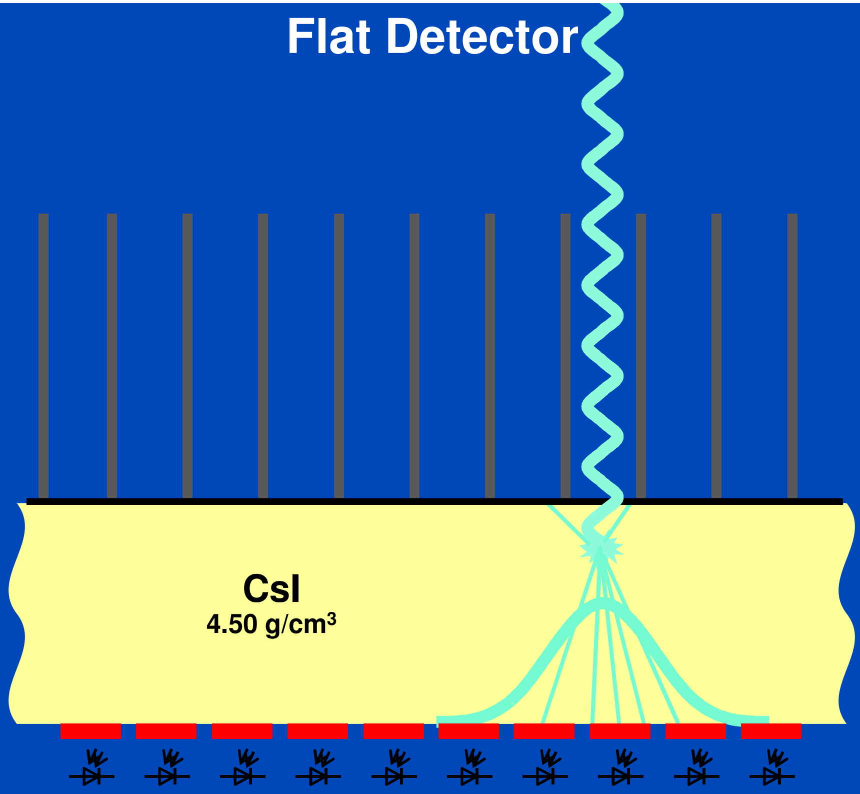
- Absorption efficiency
- Afterglow
- Dynamic range
- Cross-talk
- Framerate
- Scatter grid

Clinical CT Detector



- Anti-scatter grids are aligned to the detector pixels
- Anti-scatter grids reject scattered radiation
- Detector pixels are of about 1 mm size
- Detector pixels are structured, reflective coating maximizes light usage and minimizes cross-talk
- Thick scintillators improve dose usage
- Gd_2O_2S is a high density scintillator with favourable decay times
- Individual electronics, fast read-out (5 kHz)
- Very high dynamic range (10^7) can be realized

Flat Detector



- Anti-scatter grids are not aligned to the detector pixels
- The benefit of anti-scatter grids is unclear
- Detector pixels are of about 0.2 mm size
- Detector pixels are unstructured, light scatters to neighboring pixels, significant cross-talk
- Thick scintillators decrease spatial resolution
- CsI grows columnar and suppresses light scatter to some extent
- Row-wise readout is rather slow (25 Hz)
- Low dynamic range ($<10^3$), long read-out paths

Dose Efficiency of Flat Detectors

	Clinical CT (120 kV)			Flat Detector CT (120 kV)			Micro CT (60 kV)		
Material	Gd_2O_2S			CsI			CsI		
Density	7.44 g/cm^3			4.5 g/cm^3			4.5 g/cm^3		
Thickness	1.4 mm			0.6 mm			0.3 mm		
Manufacturer	Siemens			Varian			Hamamatsu		
Water Layer	0 cm	20 cm	40 cm	0 cm	20 cm	40 cm	0 cm	4 cm	8 cm
Photons absorbed	98.6%	97.7%	96.7%	80.0%	69.8%	62.2%	85.3%	85.6%	85.8%
Energy absorbed	94.5%	91.4%	88.7%	66.6%	55.4%	48.3%	67.1%	65.2%	64.2%

Absorption values are relative to a detector of infinite thickness.

X-Ray Exposure Dynamic Range *D*

- *D* = saturation exposure / quantum-limited exposure
  - Saturation exposure N_{max}: Exposure where the detector runs into saturation
  - Quantum-limited exposure N_{min}: Exposure where the x-ray quantum noise equals the detector‘s electronic noise.
- Measurements[1]
  - Saturation signal: Increase exposure until you obtain $\mathrm{E}(S_{\mathrm{max}})$ in the offset-corrected reading.
  - Relation $S = k \cdot N$: Evaluate an offset-corrected medium level exposure to obtain a pair of values $\mathrm{Var}(S_{\mathrm{med}})$ and $\mathrm{E}(S_{\mathrm{med}})$. Now, use the relation $\mathrm{Var}(N_{\mathrm{med}}) = \mathrm{E}(N_{\mathrm{med}})$ with $\mathrm{Var}(S_{\mathrm{med}}) = k^2 \cdot \mathrm{Var}(N_{\mathrm{med}})$ and $\mathrm{E}(S_{\mathrm{med}}) = k \cdot \mathrm{E}(N_{\mathrm{med}})$ to find $k = \mathrm{Var}(S_{\mathrm{med}}) / \mathrm{E}(S_{\mathrm{med}})$.
  - Electronic noise: Determine $\mathrm{Var}(S_{\mathrm{min}})$ from the subtraction of two dark images.
- X-ray exposure dynamic range

$$D = \frac{\mathrm{E}(N_{\mathrm{max}})}{\mathrm{E}(N_{\mathrm{min}})} = \frac{\mathrm{E}(S_{\mathrm{max}})/k}{\mathrm{Var}(S_{\mathrm{min}})/k^2} = \frac{\mathrm{E}(S_{\mathrm{max}})\mathrm{Var}(S_{\mathrm{med}})}{\mathrm{E}(S_{\mathrm{med}})\mathrm{Var}(S_{\mathrm{min}})}$$

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[1] Instead of doing this very simple procedure one may want to use statistically optimal estimates. One may use many readings, and many exposure levels. One may further determine *D* on a pixel-by-pixel basis.

Dynamic Range Required for Diagnostic Image Quality

- Soft tissue μ = 0.0192/mm object of diameter *D* between D_{min} = 200 mm and D_{max} = 500 mm with a lesion of diameter *d* = 5 mm and contrast δ = 5 HU = 0.005.
- Number of photons to be registered at the detector:

$$I(D,\delta d) = I_0 e^{-\mu D - \mu\delta d}$$

- Minimal signal difference to be detected:

$$I(D_{\text{max}},\delta d) - I(D_{\text{max}},0) \approx \mu\delta d I(D_{\text{max}},0)$$

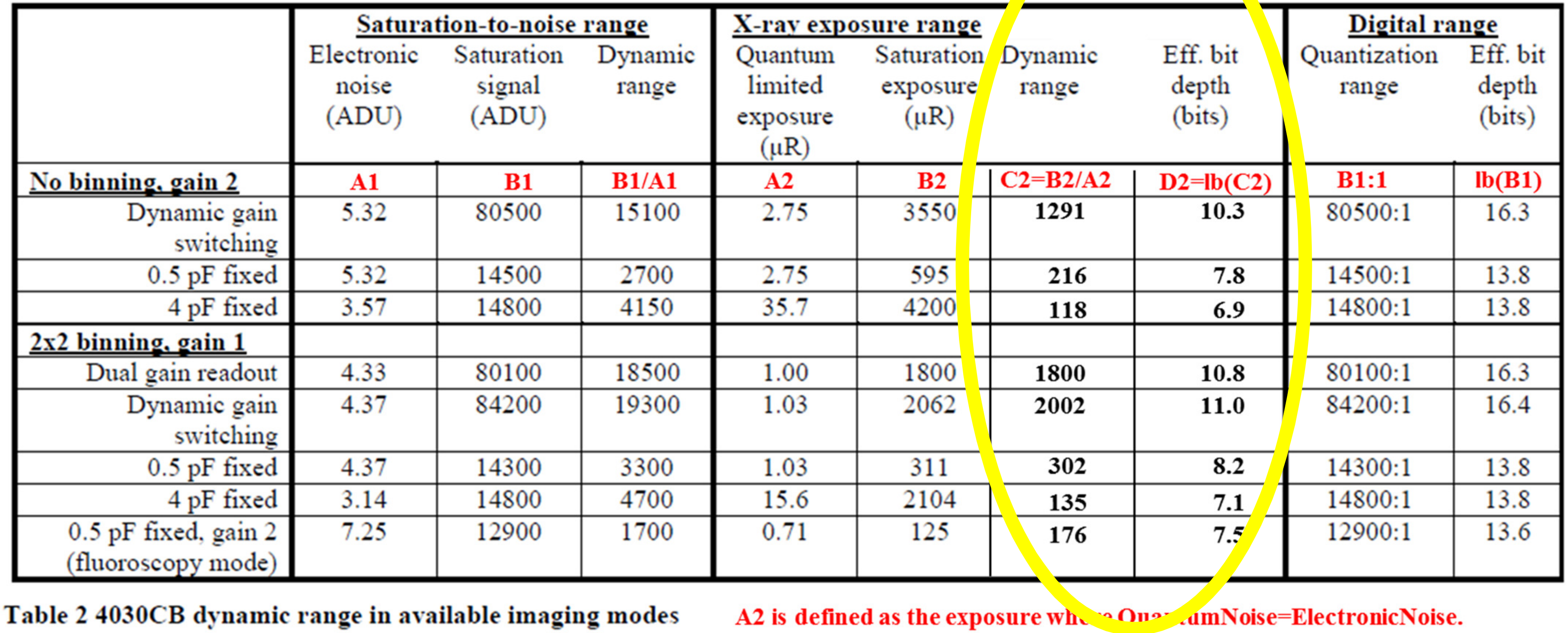
- Maximum signal to be detected:

$$I(D_{\text{min}},0)$$

- Thus, the dynamic range required in diagnostic CT is in the order of

$$\frac{I(D_{\text{min}},0)}{\mu\delta d\, I(D_{\text{max}},0)} \approx 10^6 \approx 2^{20}$$

Dynamic Range in Flat Detectors



	Saturation-to-noise range			X-ray exposure range				Digital range	
	Electronic noise (ADU)	Saturation signal (ADU)	Dynamic range	Quantum limited exposure (µR)	Saturation exposure (µR)	Dynamic range	Eff. bit depth (bits)	Quantization range	Eff. bit depth (bits)
No binning, gain 2	A1	B1	B1/A1	A2	B2	C2=B2/A2	D2=lb(C2)	B1:1	lb(B1)
Dynamic gain switching	5.32	80500	15100	2.75	3550	**1291**	**10.3**	80500:1	16.3
0.5 pF fixed	5.32	14500	2700	2.75	595	**216**	**7.8**	14500:1	13.8
4 pF fixed	3.57	14800	4150	35.7	4200	**118**	**6.9**	14800:1	13.8
2x2 binning, gain 1									
Dual gain readout	4.33	80100	18500	1.00	1800	**1800**	**10.8**	80100:1	16.3
Dynamic gain switching	4.37	84200	19300	1.03	2062	**2002**	**11.0**	84200:1	16.4
0.5 pF fixed	4.37	14300	3300	1.03	311	**302**	**8.2**	14300:1	13.8
4 pF fixed	3.14	14800	4700	15.6	2104	**135**	**7.1**	14800:1	13.8
0.5 pF fixed, gain 2 (fluoroscopy mode)	7.25	12900	1700	0.71	125	**176**	**7.5**	12900:1	13.6

Table 2 4030CB dynamic range in available imaging modes

A2 is defined as the exposure where QuantumNoise=ElectronicNoise.

Table taken from [Roos et al. "Multiple gain ranging readout method to extend the dynamic range of amorphous silicon flat panel imagers," *SPIE Medical Imaging Proc.*, vol. 5368, pp. 139-149, 2004]. Additional values were added, for convenience.

Dynamic Range in Flat Detectors

Detector		Saturation-to-noise range				X-ray exposure range					
Type	**Mode**	**Electronic noise**	**Quantum limited signal**	**Saturation signal**	**Dynamic range**	**Quantum limited exposure (µR)**	**Saturation exposure (µR)**	**Dynamic range**	**Eff. Bit depth**	**k**	**Remark**
Varian 4030CB	Dynamic gain switching	5.32	**62.89**	80500	15100	2.75	3550	1291	10.3	**0.45**	
	0.5 pF fixed	5.32	**67.39**	14500	2700	2.75	595	216	7.8	**0.42**	
	4.0 pF fixed	3.57	**283.02**	14800	4150	35.7	4200	118	6.9	**0.10**	
Perkin Elmer Dexela 2923	**?**	**4.91**	**33.01**	**13600**	**2770.6**	**?**	**?**	**412.02**	**8.69**	**0.73**	**Integration time 100 ms**
	?	**6.36**	**55.37**	**13600**	**2139.6**	**?**	**?**	**245.7**	**7.94**	**0.73**	**Integration time 1000 ms**

Table 1: Dynamic range for different detectors and imaging modes. Data for Varian 4030CB taken from [1].

- Electronic noise is measured as the pixel standard deviation of the subtraction of two dark images[1]. For the Dexela 2923 detector it was observed, that the electronic noise increases for higher integration times and thus the dynamic range decreases.
- Saturation signal is defined as the maximum signal in the linear signal range.

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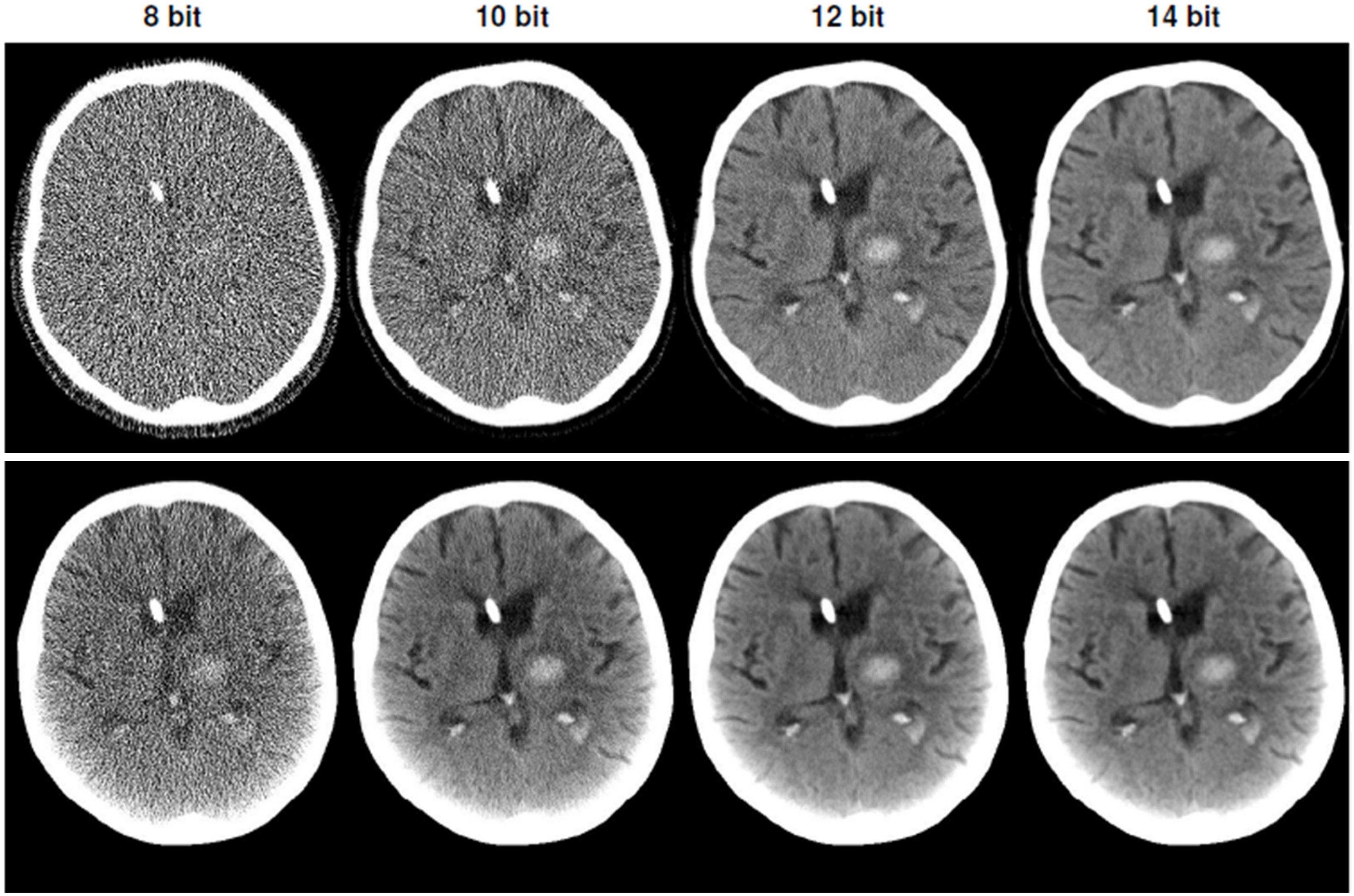
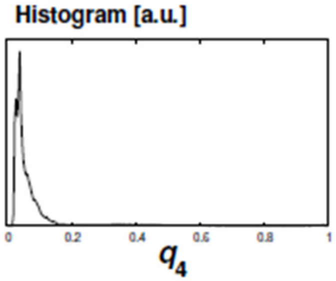
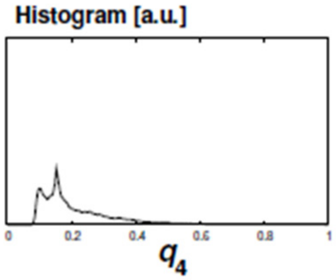
[1] Roos et al. – Multiple gain ranging readout method to extend the dynamic range of amorphous silicon flat panel imagers



No overexposure



Intended overexposure (factor 4)



The Dynamic Range of X-Ray Detectors

a) is defined as the ratio between the maximum and the minimum measureable signal or signal difference.

b) determines how often the detector can be read out per second.

c) must be very high to resolve low contrasts in small and in large cross-sections.

d) corresponds to the number of bits in the digitized detector signal.

e) is a measure of what centrifugal forces the detector can withstand.

E-Vote

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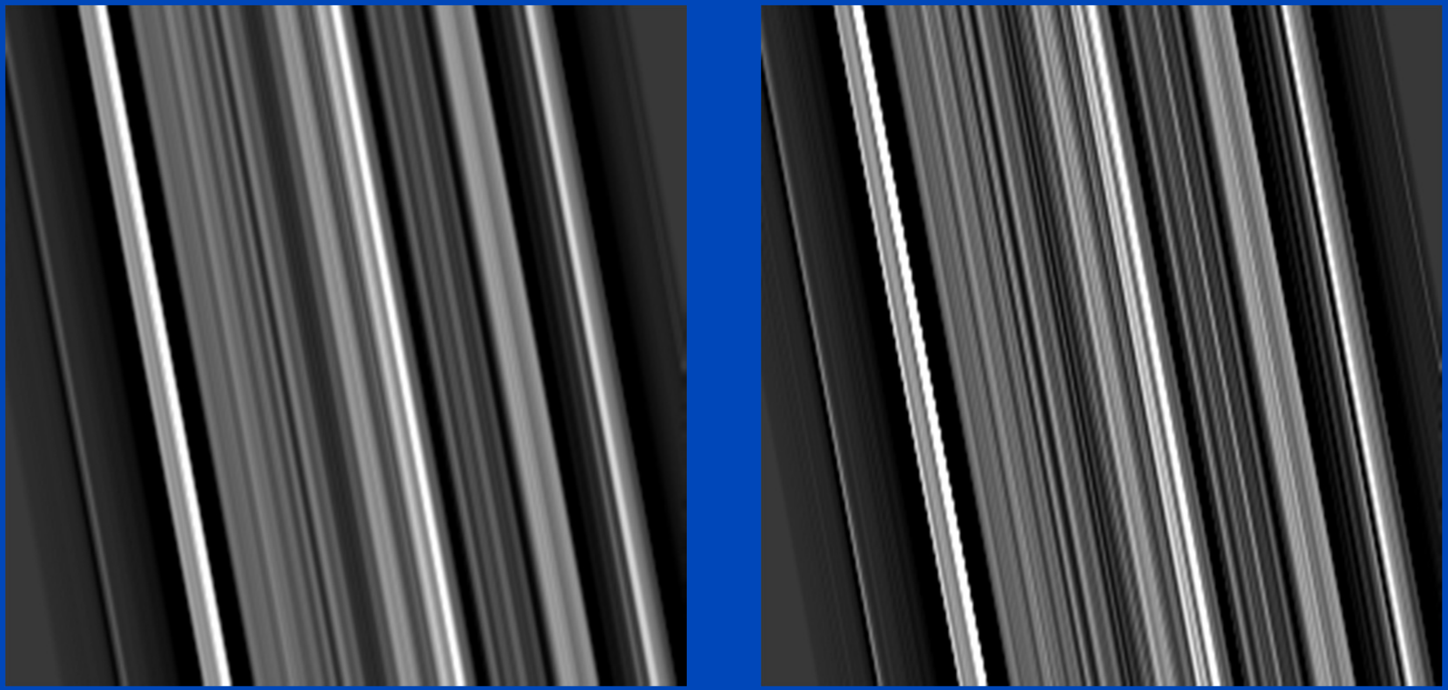
Filtered Backprojection (FBP)

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

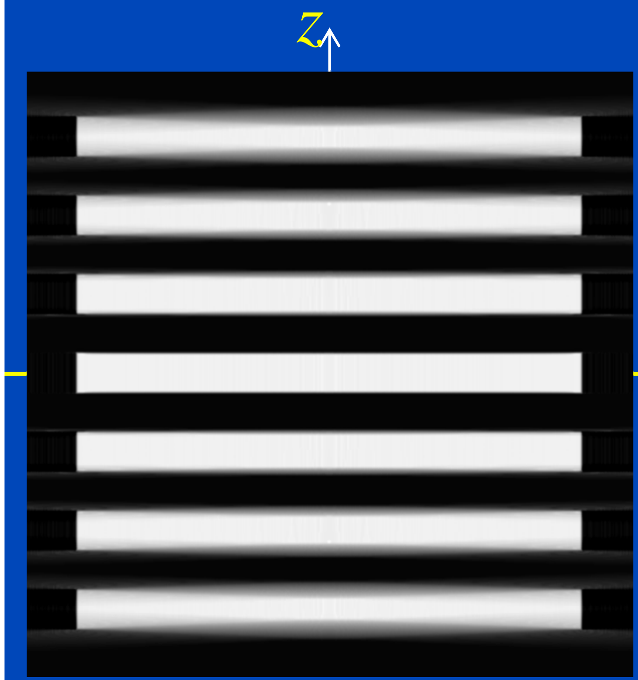


Smooth

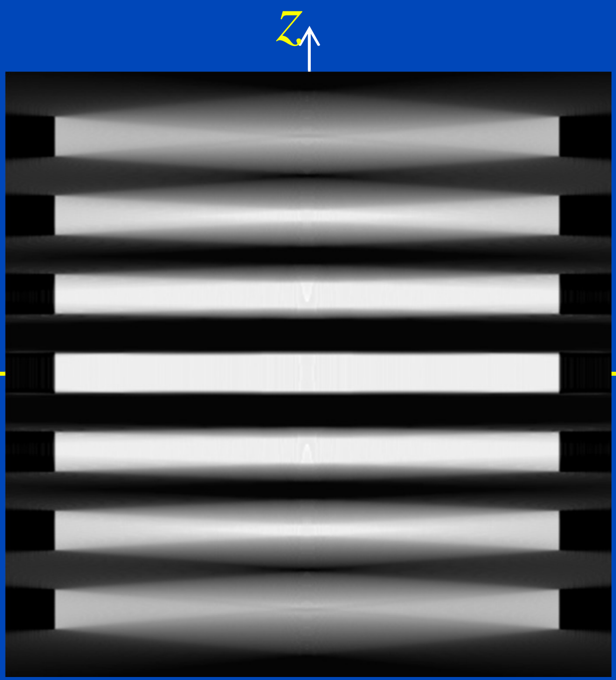
Standard

Reconstruction kernels balance between spatial resolution and image noise.

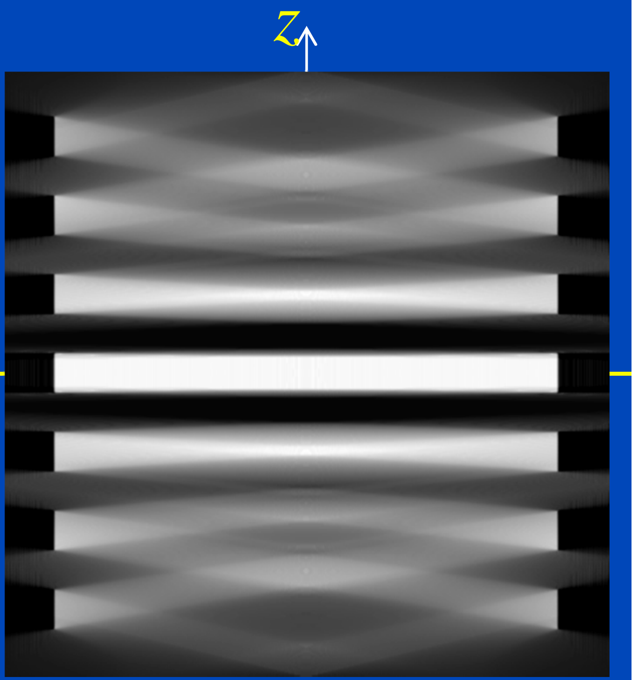
Cone-Beam Artifacts



Cone-angle Γ = 6°



Cone-angle Γ = 14°



Cone-angle Γ = 28°

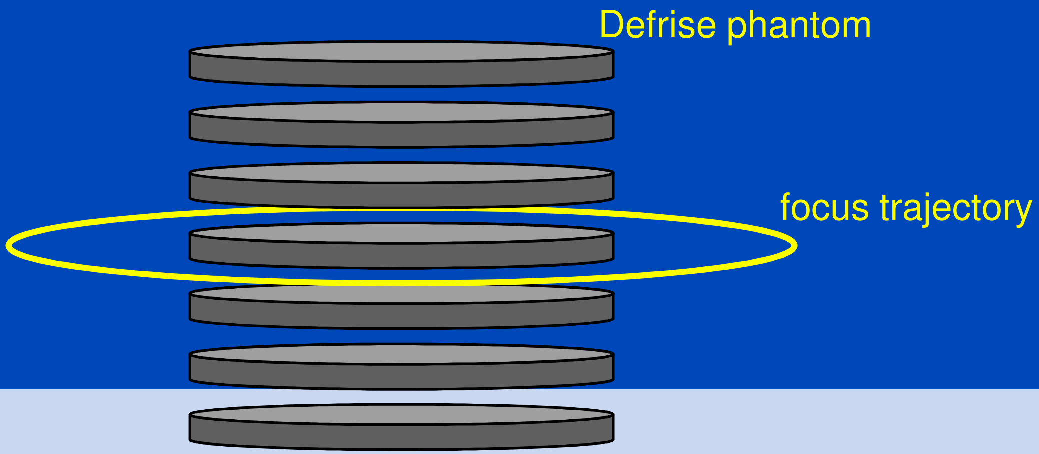


Image Quality

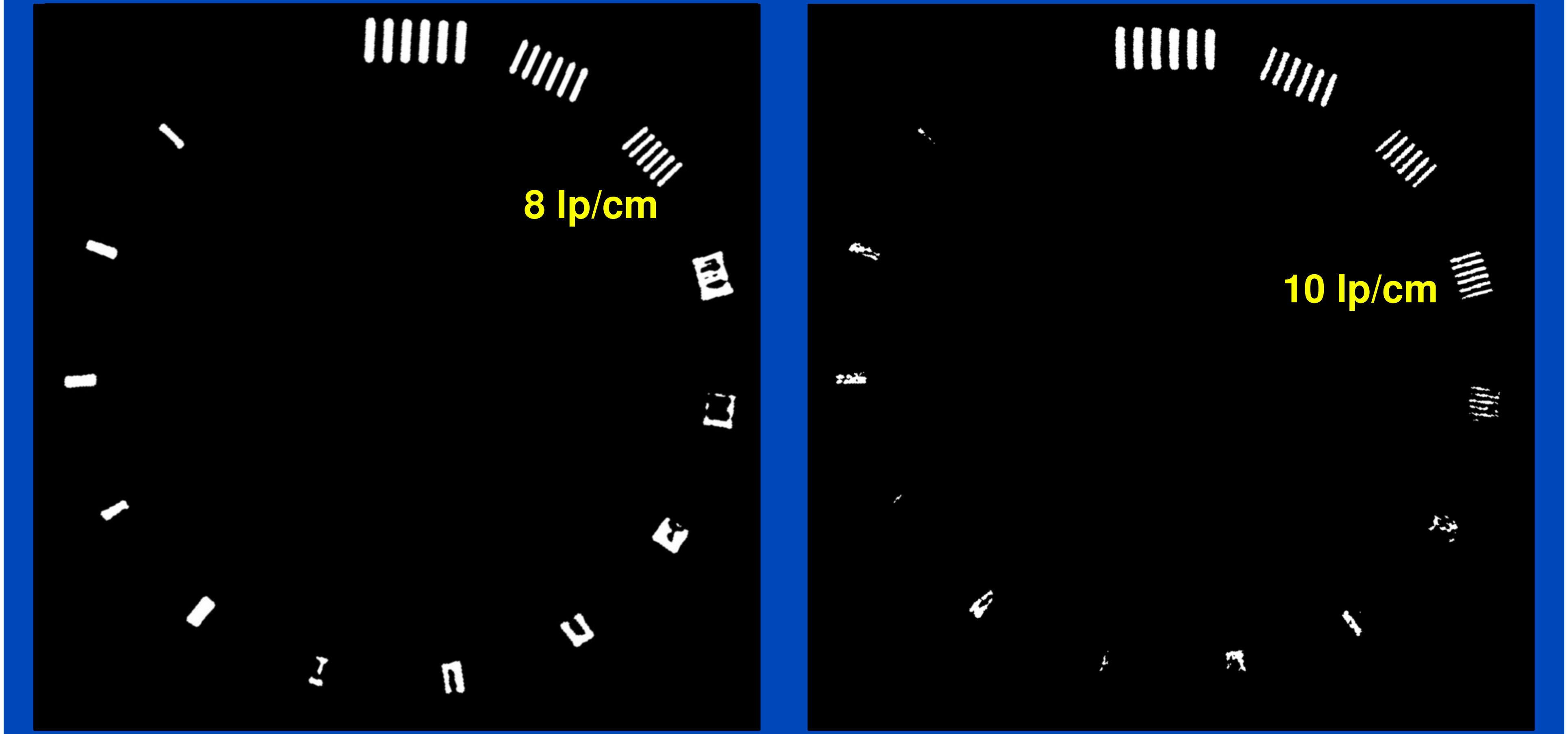
Always Relate
SNR and CNR to Unit Dose!

- SNR and CNR are useless for comparisons if these are not taken at the same dose or if SNR and CNR are not normalized to unit dose.
- The terms SNRD and CNRD are used for SNR normalized to unit dose and CNR normalized to unit dose, respectively.
- Compare only at matched spatial resolution!

$$SNRD = \frac{\text{Signal}}{\text{Noise}\sqrt{\text{Dose}}} = \frac{SNR}{\sqrt{\text{Dose}}}$$

$$CNRD = \frac{\text{Contrast}}{\text{Noise}\sqrt{\text{Dose}}} = \frac{CNR}{\sqrt{\text{Dose}}}$$

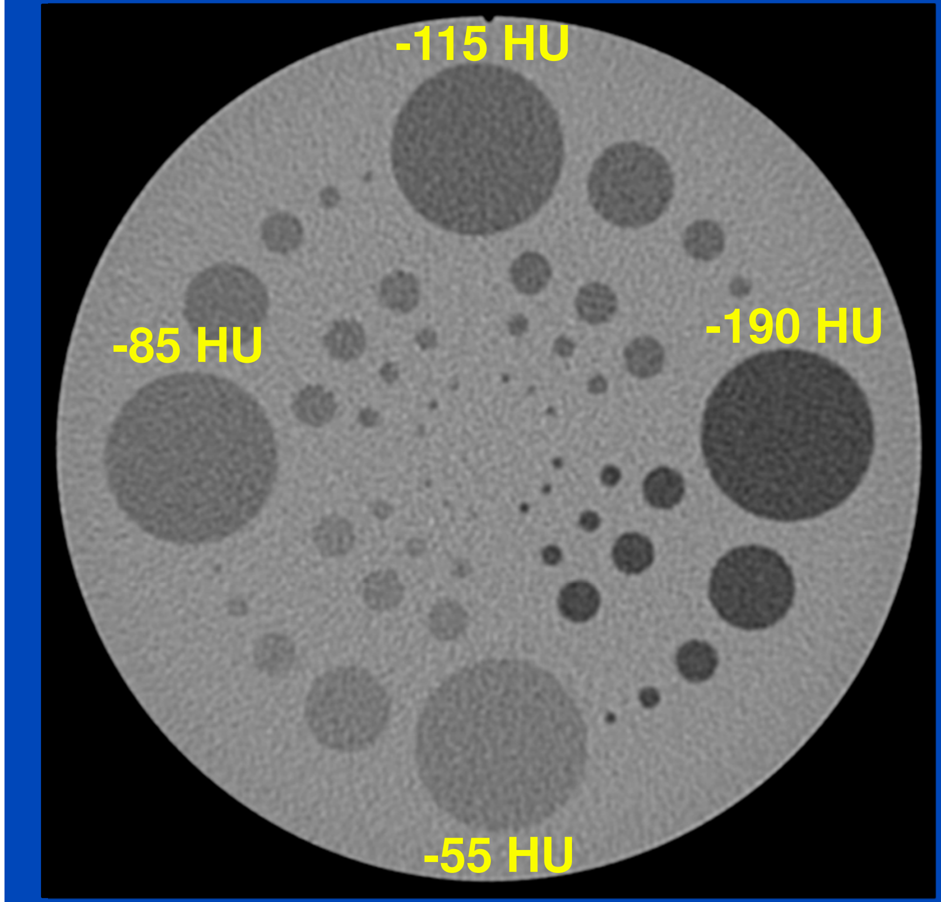
Clinical CT vs. Flat Detector CT



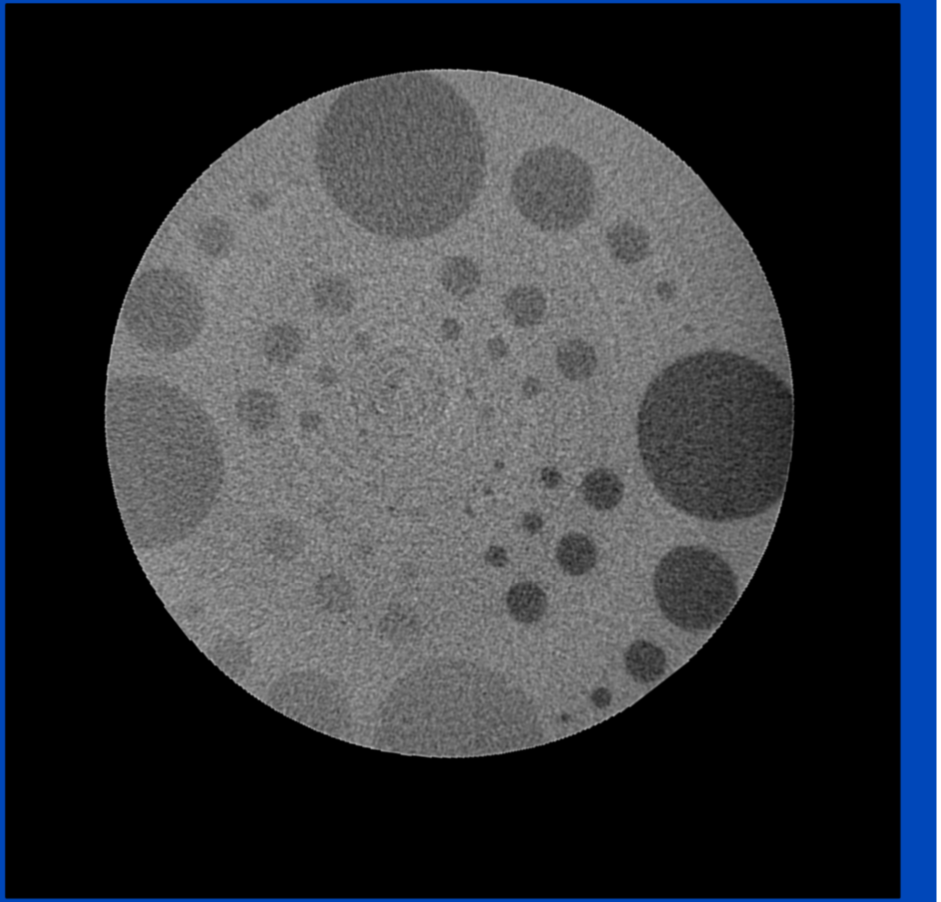
Clinical CT, Standard Kernel

Flat Detector CT, 2×2 Binning

Clinical CT vs. Flat Detector CT



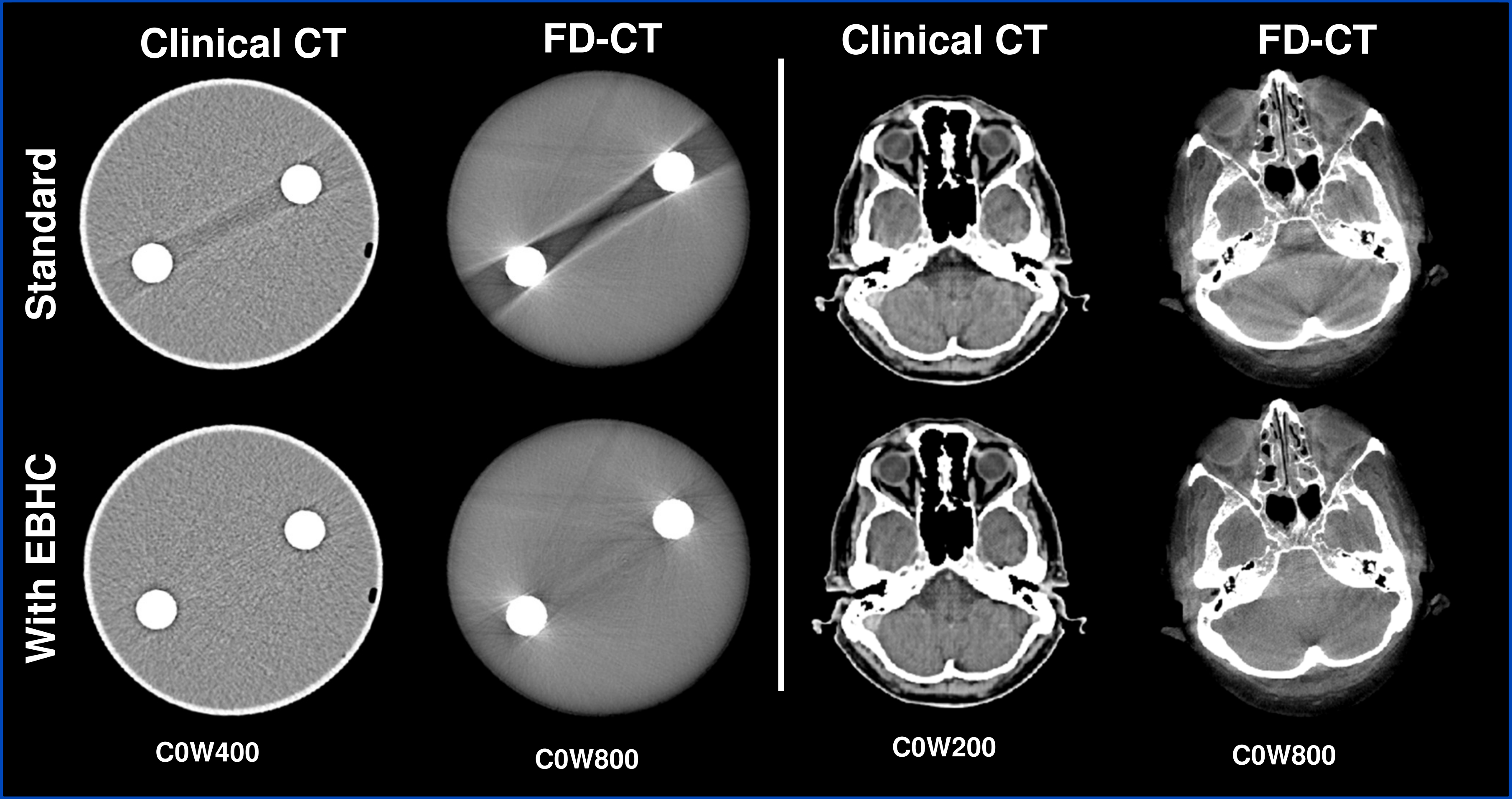
Clinical CT, Standard Kernel
C = 0 HU, *W* = 700 HU



Flat Detector CT, 2×2 Binning

Medium contrast phantom

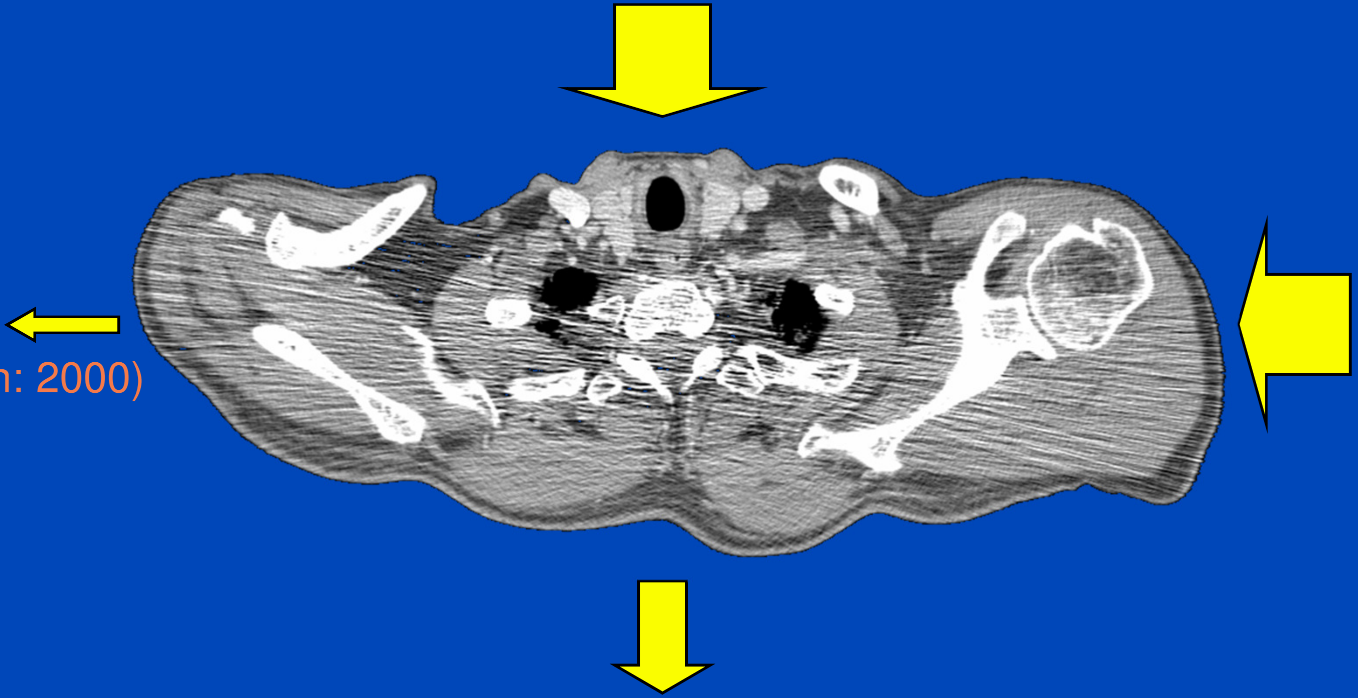
Clinical CT vs. FD-CT



Siemens Somatom Definition vs. Siemens Axiom Artis

Dose Reduction

Tube Current Modulation (TCM)



1 000 000

500
(attenuation: 2000)

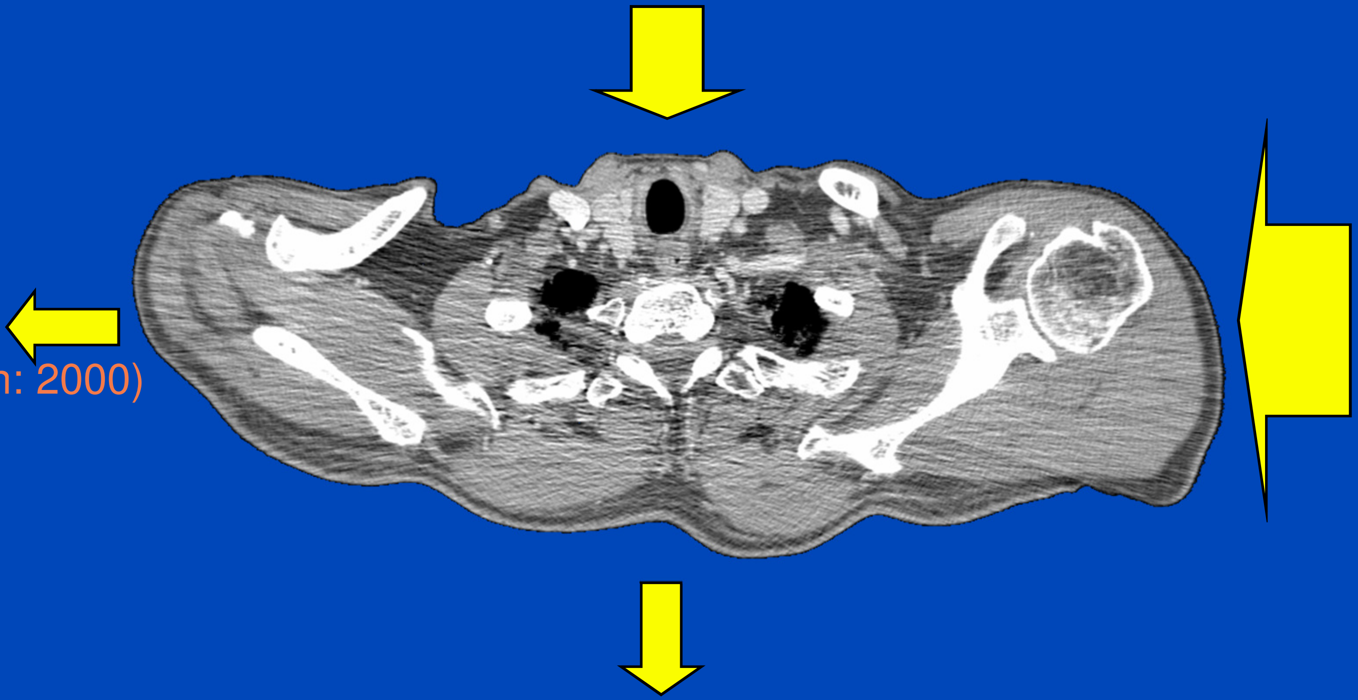
1 000 000

20 000
(attenuation: 50)

Constant tube current: High, inhomogeneous noise.

$$\sigma^2_{\text{pixel}} \propto \sum_n \sigma^2_{\text{projection } n}$$

“TCM” in Flat Detector CT



50 000

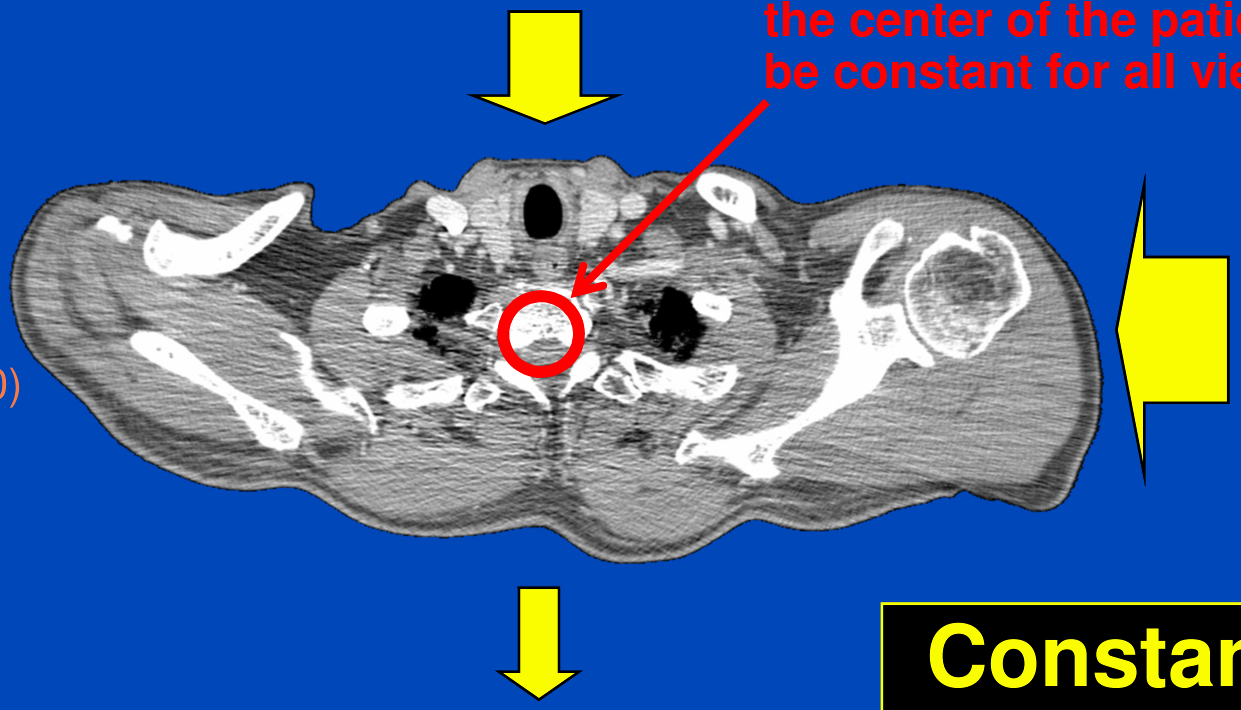
1 000
(attenuation: 2000)

2 000 000

1 000
(attenuation: 50)

Good for the detector!
Almost constant detector exposure in regions behind the patient, overexposure in peripheral regions and in non-attenuated regions.

TCM in Diagnostic CT



250 000

875
(attenuation: 2000)

1 750 000

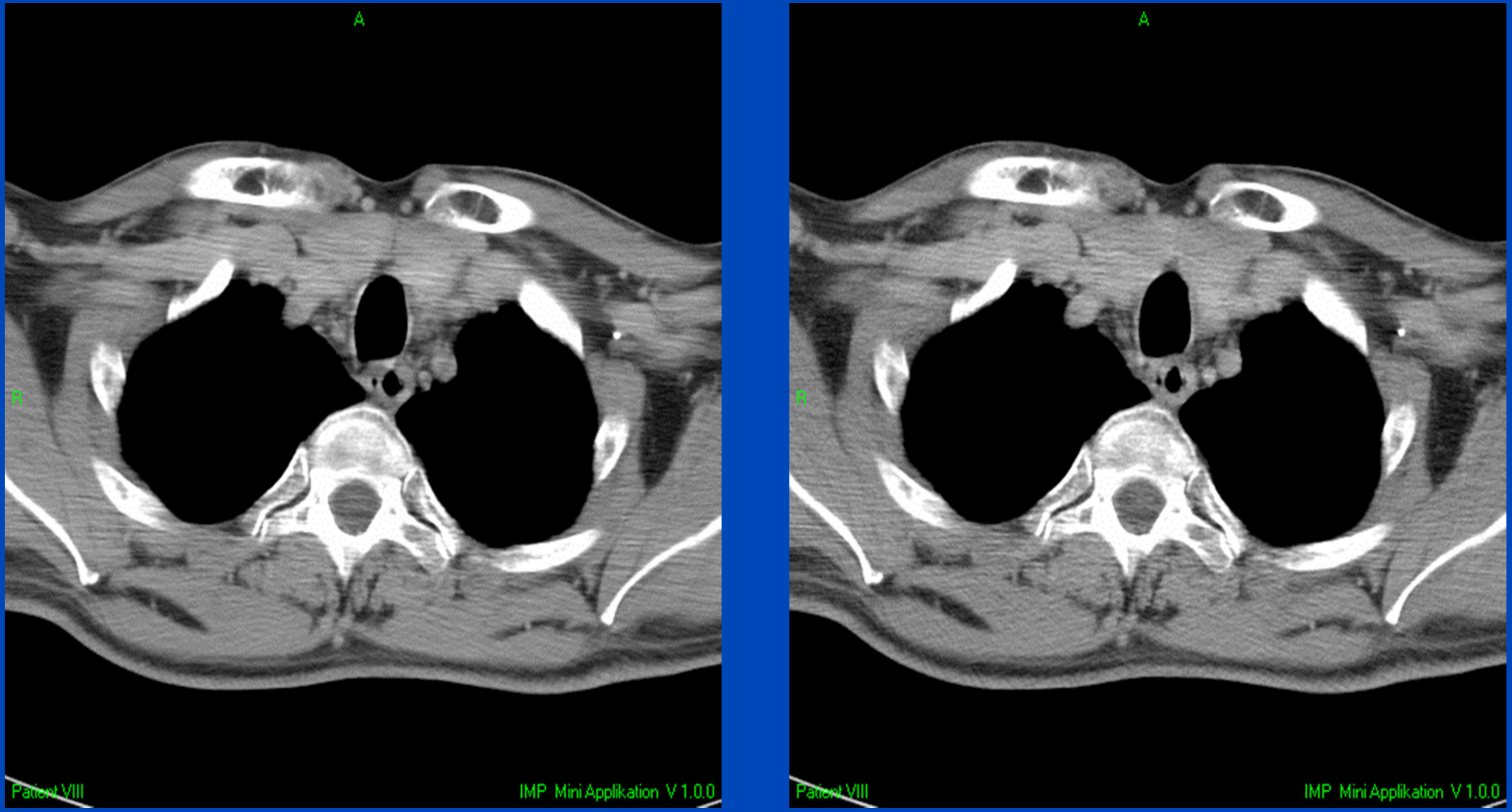
5 000
(attenuation: 50)

Rule of thumb:
The number of quanta reaching the center of the patient should be constant for all view angles.

Constant total mAs!

Good for the patient!
Almost constant exposure in the patient center means minimal patient dose at given image noise or vice versa. Low noise, homgeneous noise.

Dose Reduction by Tube Current Modulation



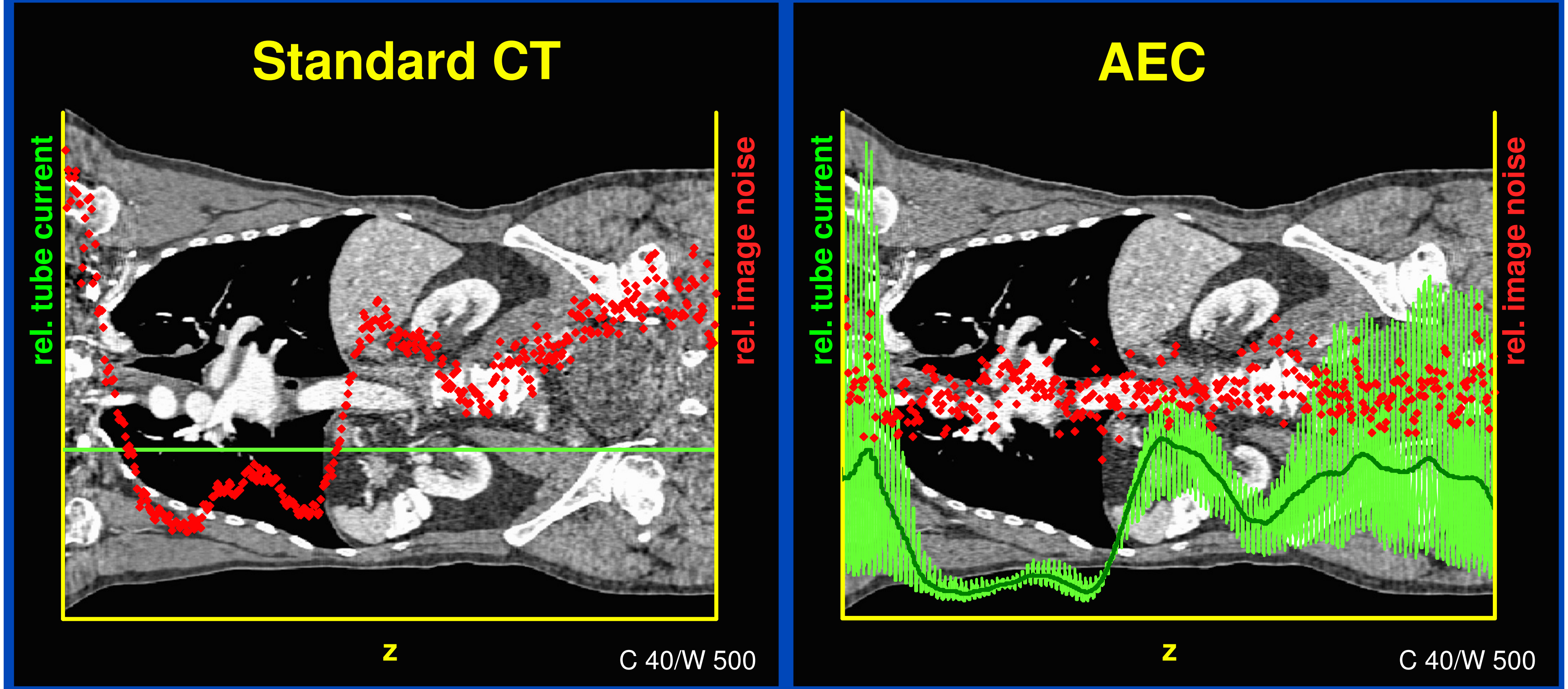
Conventional scan: 327 mAs **Online current modulation: 166 mAs**

53% dose reduction on average for the shoulder region
49% dose reduction in this case

Kalender WA et al. Med Phys 1999; 26(11):2248-2253

Automatic Exposure Control (AEC)

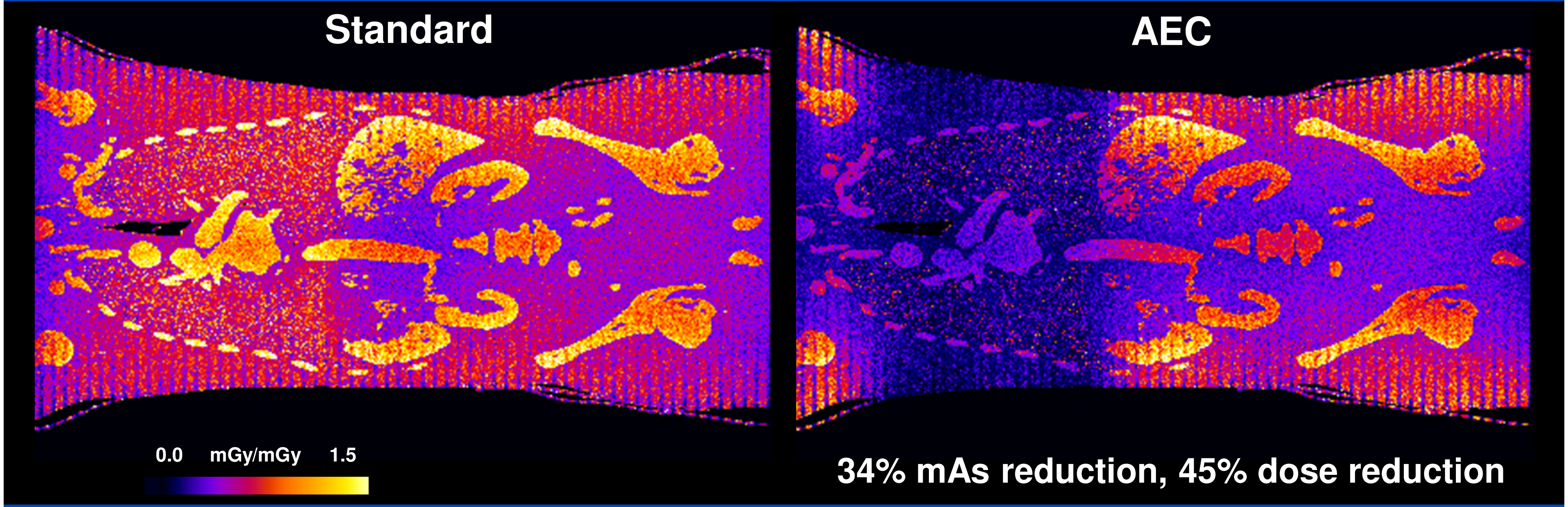
(z-dependent + angular dependent tube current modulation)



34% mAs reduction with AEC at constant image quality for that specific case

Dose Modulation: DOM, TCM, AEC, …

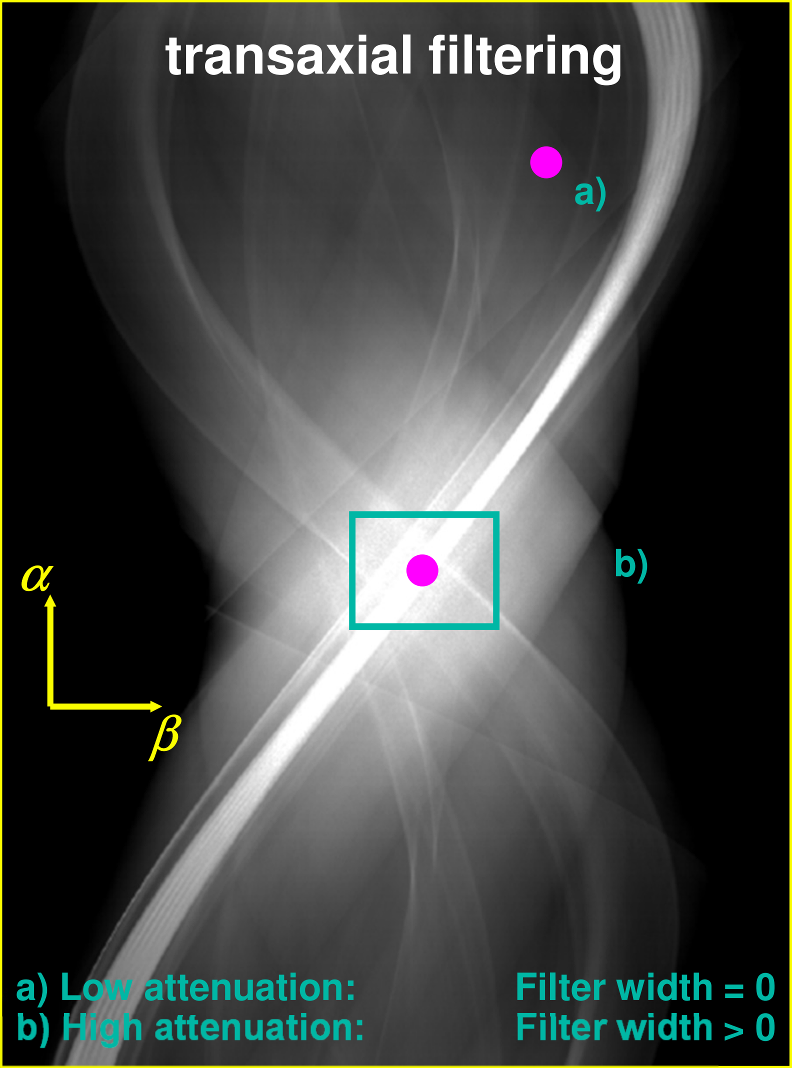
- Better dose usage
- ECG pulsing
- Avoiding organs of risc
- Specification of image quality $\sigma(z)$



34% mAs reduction, 45% dose reduction

Multidimensional Adaptive Filtering (MAF)

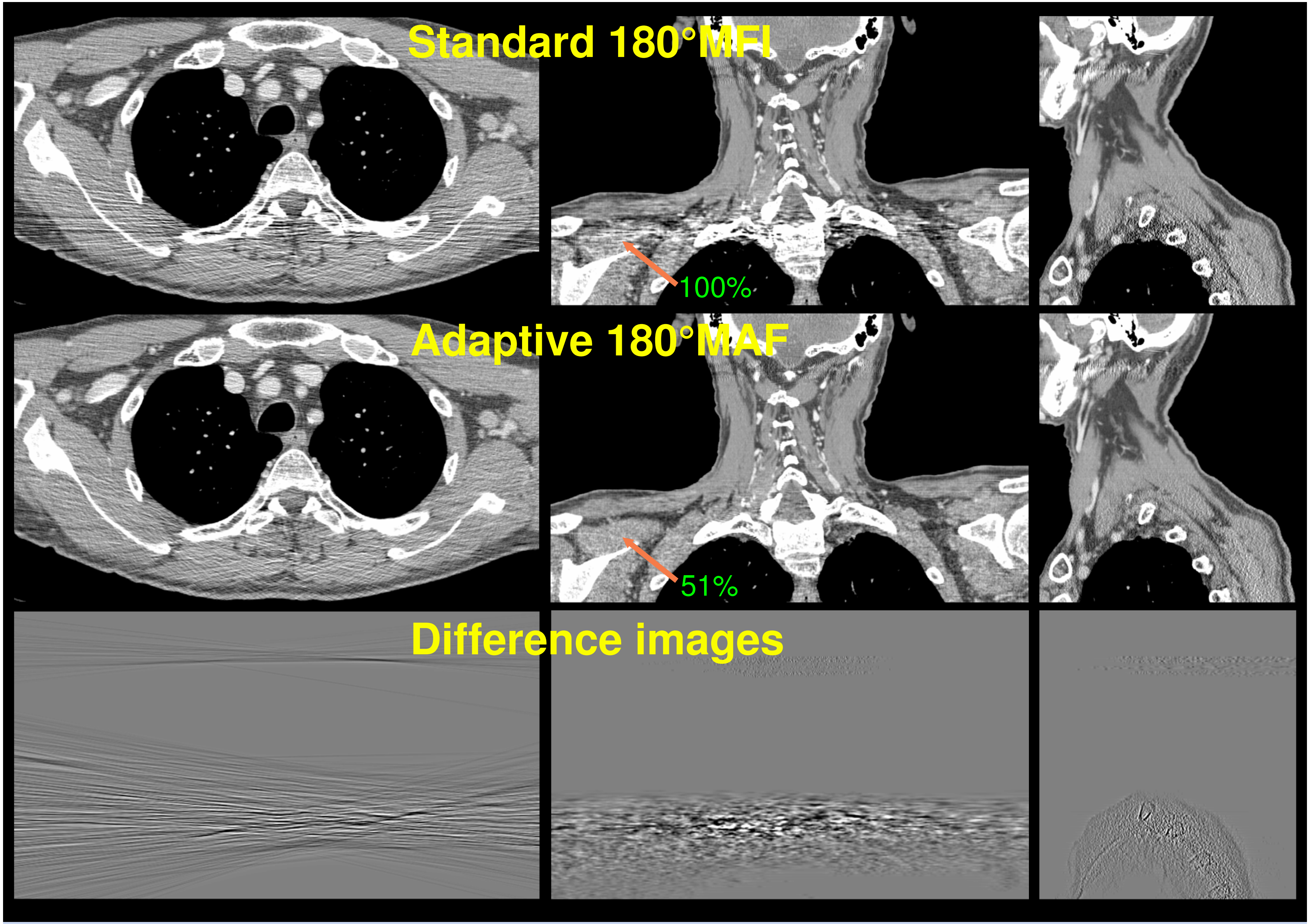
- Rawdata-based
- Local smoothing of noisy data (less than 5% modification)
- No loss of spatial resolution
- Efficient
- Noise reduction can be equivalently converted to dose reduction



a) Low attenuation: Filter width = 0
b) High attenuation: Filter width > 0

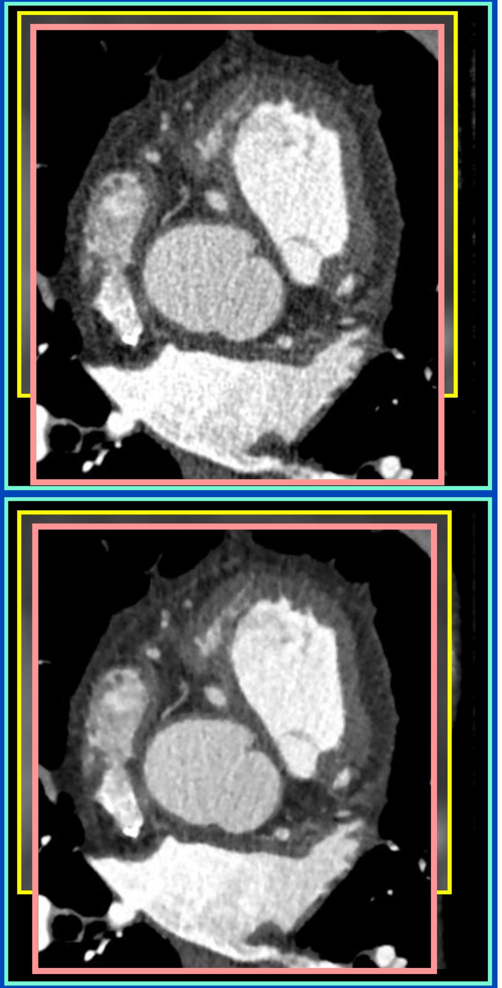
$$p_{\mathrm{MAF}}(\alpha,\beta,b) = \int d\alpha'\,d\beta'\,db'\; f_{\Delta\alpha}(\alpha-\alpha')\,f_{\Delta\beta}(\beta-\beta')\,f_{\Delta b}(b-b')\,p(\alpha',\beta',b')$$

M. Kachelrieß et al. Med Phys 2001; 28:475-490

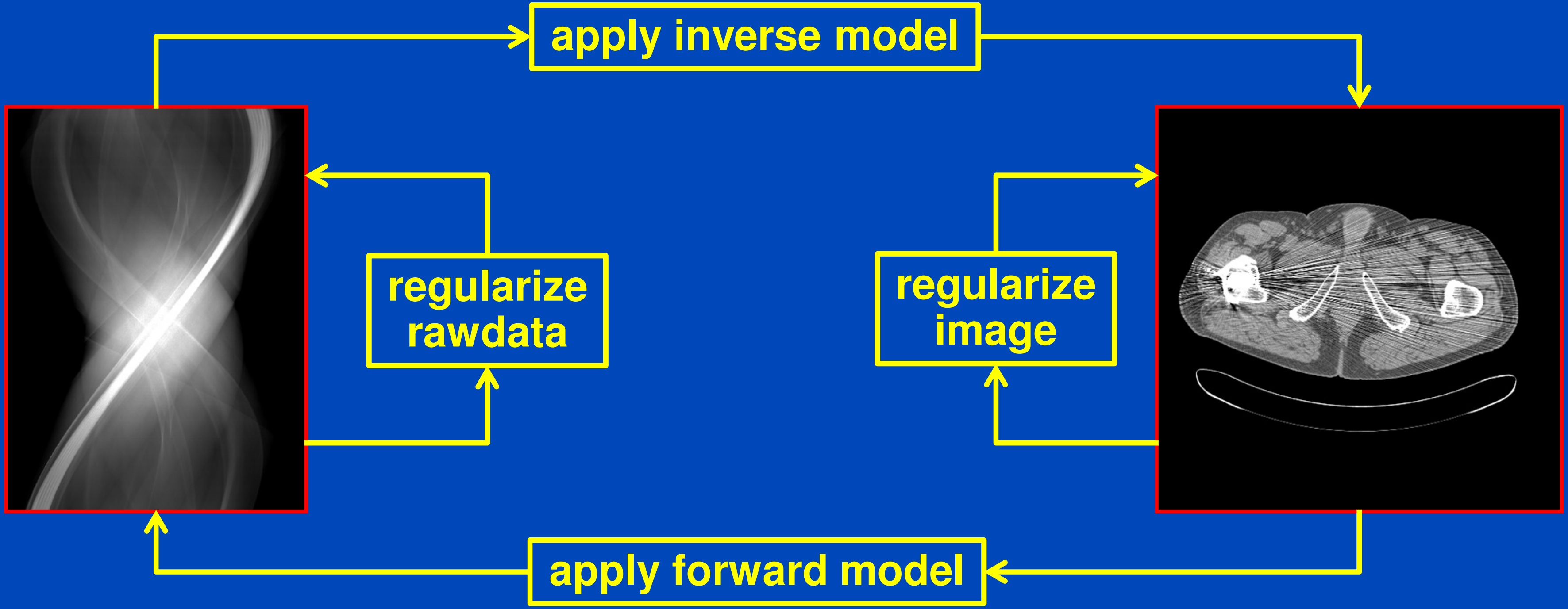


collimation 4×1 mm, d = 5 mm, (C=0 / W=500)

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!
- CT product implementations
  - AIDR 3D (adaptive iterative dose reduction, Toshiba)
  - ASIR (adaptive statistical iterative reconstruction, GE)
  - iDose (Philips)
  - IMR (iterative model reconstruction, Philips)
  - IRIS (image reconstruction in image space, Siemens)
  - VEO, MBIR (model-based iterative reconstruction, GE)
  - SAFIRE, ADMIRE (advanced model-based iterative reconstruction, Siemens)


apply inverse model

regularize rawdata

regularize image

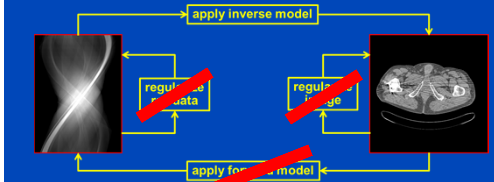
apply forward model

- Rawdata regularization: adaptive filtering[1], precorrections, filtering of update sinograms...
- Inverse model: backprojection (R^T) or filtered backprojection (R^{-1}). In clinical CT, where the data are of high fidelity and nearly complete, one would prefer filtered backprojection to increase convergence speed.
- Image regularization: edge-preserving filtering. It may model physical noise effects (amplitude, direction, correlations, …). It may reduce noise while preserving edges. It may include empirical corrections.
- Forward model (R_{phys}): Models physical effects. It can reduce beam hardening artifacts, scatter artifacts, cone-beam artifacts, noise, …

[1] M. Kachelrieß et al., Generalized Multi-Dimensional Adaptive Filtering, MedPhys 28(4), 2001

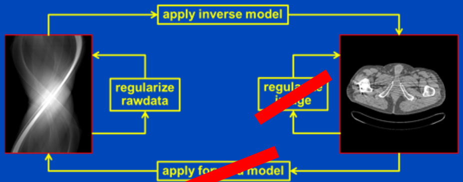
Plain FBP

σ = 26.8 HU



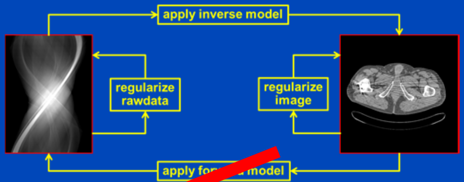
Siemens Standard

σ = 17.6 HU

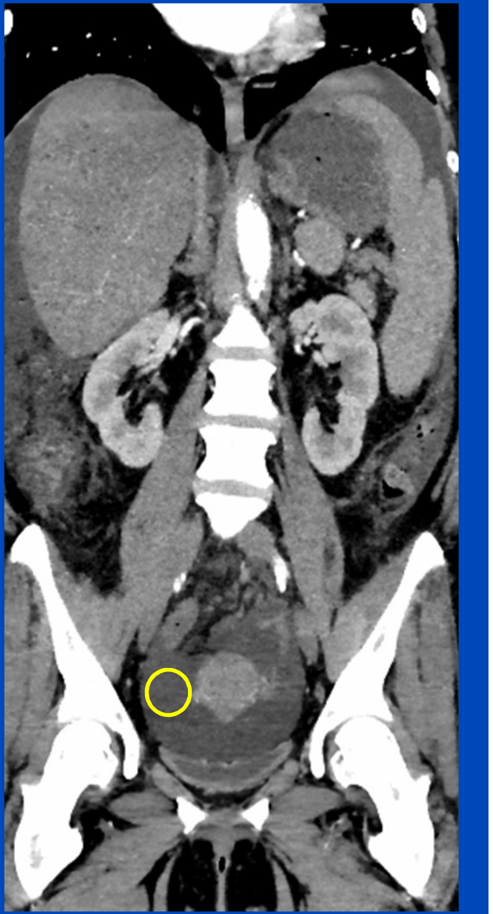


IRIS VA34

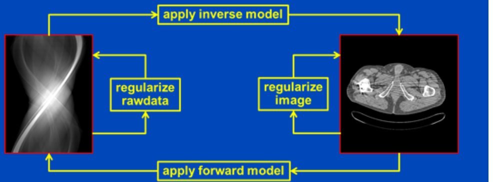
σ = 12.3 HU



SAFIRE VA40



σ = 7.8 HU



CT images provided by Siemens Healthcare, Forchheim, Germany

What makes the Differences in Image Quality and Dose between Diagnostic CT and Flat Detector CT?

a) The detector technology differs significantly in spatial resolution, absorption efficiency and dynamic range.

b) The detection principle is different. Diagnostic CT uses energy integrating systems while flat detector CT uses photon counters.

c) Dose reduction measures in clinical CT are optimized for the patient, while in flat detector CT the inferior detector properties force to optimize for the detector.

d) Due to their focussed geometry curved detectors are better suited to capture x-rays than flat detectors.

e) The diagnostic CT technology has a higher level of maturity and the soft- and hardware is more sophisticated than in flat detector CT.

E-Vote

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Clinical vs. Flat Detector CT

	Clinical CT (MDCT)	Flat Detector CT (CBCT)
Spatial resolution	0.5 mm	0.2 mm
Contrast	3 HU	30 HU
Dynamic range	≈ 20 bit	≈ 10 bit
Dose efficiency	≈ 90%	≈ 50%
Lowest rotation time	0.28 s	3 s
Temporal resolution	0.07 s	3 s
Frame rate	≈ 5000 fps	≈ 25 fps
X-ray power	100 – 120 kW	5 – 25 kW
Bow tie	optimized for patient	optimized for detector
AEC (for mAs and kV)	optimized for patient	optimized for detector
Advanced dose reduction techniques	IR, DOM, task-specific kV, organ and ECG specific AEC, dynamic collimation, …	few, if any

Thank You!

The 4th International Conference on

Image Formation in X-Ray Computed Tomography

July 18 – July 22, 2016, Bamberg, Germany

www.ct-meeting.org

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Conference Chair

Marc Kachelrieß, German Cancer Research Center (DKFZ), Heidelberg, Germany

Parts of the reconstruction software were provided by RayConStruct® GmbH, Nürnberg, Germany. This presentation will soon be available at www.dkfz.de/ct.