Basics of Clinical X-Ray Computed Tomography

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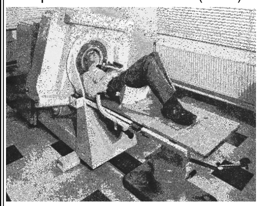
Siemens 2·2·32=128-slice
dual source cone-beam spiral CT (2005)

1536 views per rotation in 0.33 s
2·32×(672+352) 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

GE LightSpeed

Toshiba Aquilion

Philips Brilliance

Siemens Somatom Definition

Dual Source

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:**

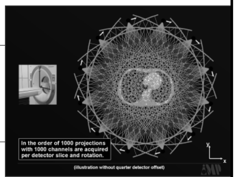
$$p(L) \approx \int dL\, \mu(\boldsymbol{r}, E_{\text{eff}})$$

with the effective energy being around 70 keV

CT Basics

From Single-Slice to Cone-Beam Spiral CT

- **Technology**
  - Basic parameters
  - Detector concepts, tube technology
  - Scan trajectories, scan modes
- **Algorithms**
  - 2D filtered backprojection
  - Spiral z-interpolation
  - ASSR and EPBP (cone-beam recon.)
  - Phase-correlated CT (e.g. cardiac CT)
- **Image quality and dose**
  - Spatial resolution (PSF, SSP, MTF)
  - Relation of noise, dose and resolution
  - Dose values (CTDI, patient dose)
  - Dose reduction techniques

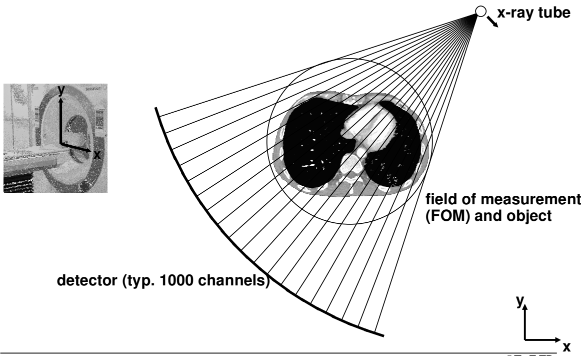


CT-Performance (Best-of Values)

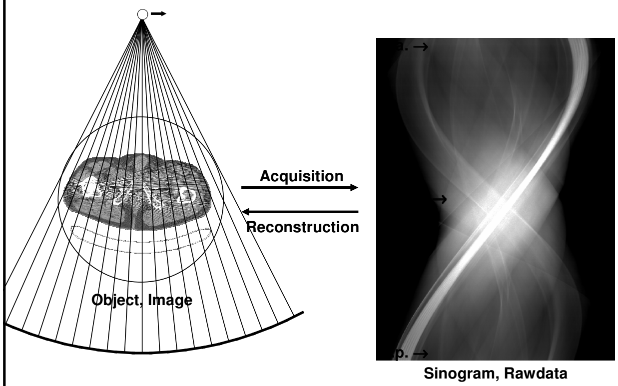
		T_{rot}	collimation	typ. 30 cm scan[1]	slices/s
	1972	300 s×4[2]	2×13 mm	---	0.007/4[2]
	1980	2 s	2 mm	20 mm, 30 s	0.5
Spiral CT	**1990**	1 s	1 mm	10 mm, 30 s	1[3]
	1995	0.75 s	1 mm	8 mm, 30 s	1.3[3]
	1998	0.5 s	4×1 mm	4×1 mm, 30 s	12[4]
	2002	0.4 s	16×0.75 mm	16×0.75 mm, 12 s	60[4]
	2004	0.33 s	64×0.5 mm	64×0.5 mm, 3 s	240[4]
	2010	0.2 s	512×0.5 mm	512×0.5 mm, 0.2 s	2500

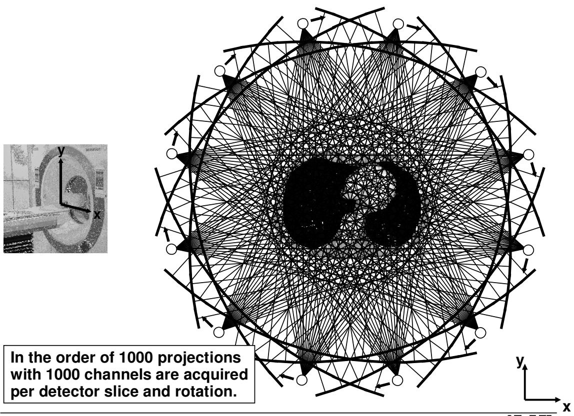
[1] assuming a breath-hold limit of 30 s
[2] factor 4 converts from head FOM to full body FOM
[3] assuming p = 1, otherwise S_{eff} is increased
[4] assuming p = 1.5 since IQ is independent of pitch for MSCT

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



(illustration without quarter detector offset)

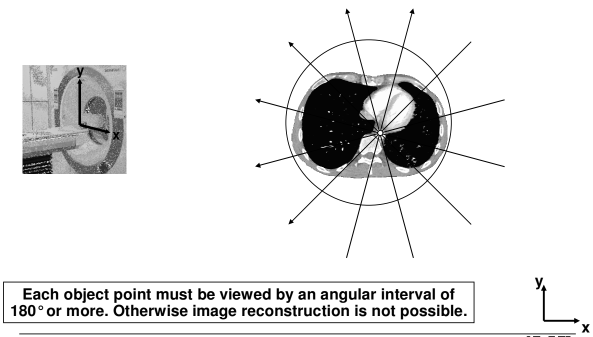




In the order of 1000 projections with 1000 channels are acquired per detector slice and rotation.

(illustration without quarter detector offset)

Data Completeness



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

Basic Parameters

(best-of values typical for modern scanners)

- In-plane resolution: 0.4 … 0.7 mm
- Nominal slice thickness: S = 0.5 … 1.5 mm
- Effective slice thickness: S_{eff} = 0.5 … 10 mm
- Tube (max. values): 100 kW, 140 kV, 800 mA
- Effective tube current: mAs_{eff} = 10 mAs … 1000 mAs
- Rotation time: T_{rot} = 0.33 … 0.5 s
- Simultaneously acquired slices: M = 4 … 64
- Table increment per rotation: d = 2 … 50 mm
- Pitch value: p = 0.3 … 1.5
- Scan speed: up to 16 cm/s
- Temporal resolution: 50 … 250 ms

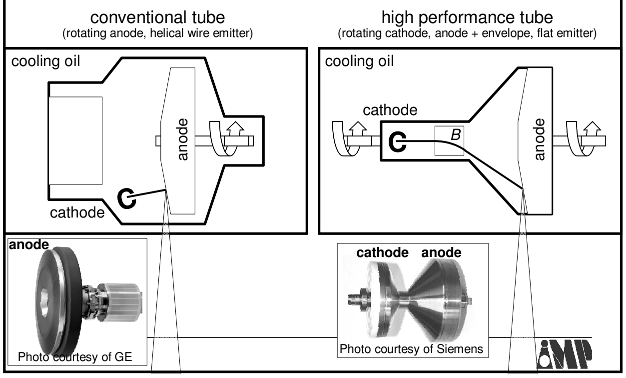
Demands on the Mechanical Design

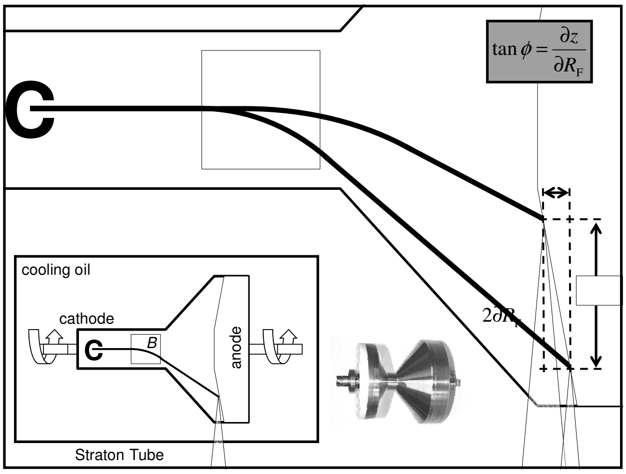
- Continuous data acquisition in spiral scanning mode
- Able to withstand very fast rotation
  - Centrifugal force at 550 mm with 0.5 s: F = 9 g
  - with 0.4 s: F = 14 g
  - with 0.3 s: F = 25 g
- Mechanical accuracy better than 0.1 mm
- Compact and robust design
- Short installation times
- Long service intervals
- Low cost

Demands on X-Ray Sources

- High instantaneous power levels (typ. 50-100 kW)
- Increasing with rotation speed
- High continuous power levels (typ. >5 kW)
- High cooling rates (typ. >1 MHU/minute)
- High tube current variation (low inertia)
- Compact and robust design

Tube Technology



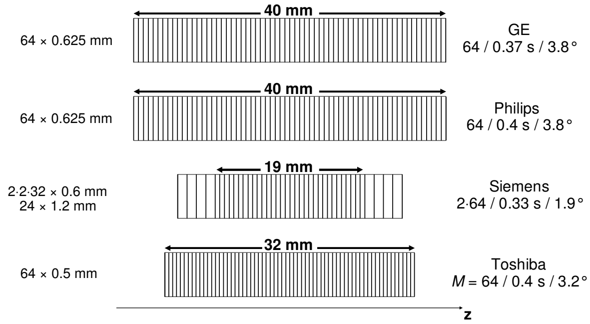


Demands on CT Detector Technology

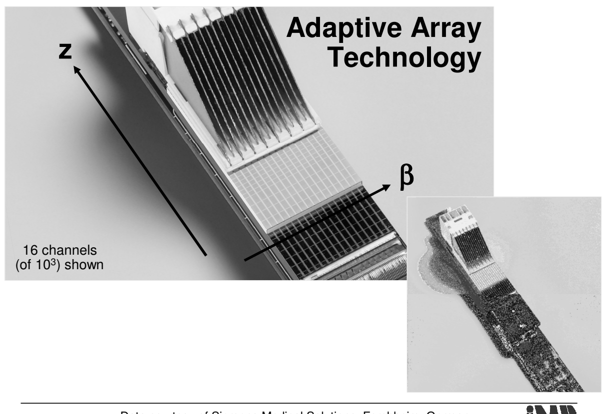
- Available as multi-row arrays
- Very fast sampling (typ. 300 µs)
- Favourable temporal characteristics (decay time < 10 µs)
- High absorption efficiency
- High geometrical efficiency
- High count rate (up to 10^8 cps*)
- Adequate dynamic range (at least 20 bit)

* in the order of 10^5 counts per reading and 10^3 readings per second

Multirow Detectors for Multi-Slice CT 2006

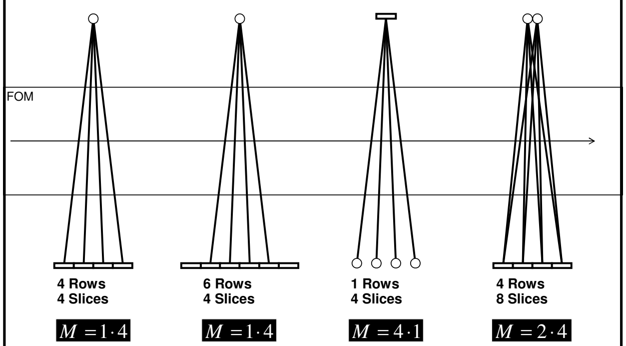


Number of simultaneously acquired slices M / Rotation time t_{rot} / Cone-angle Γ

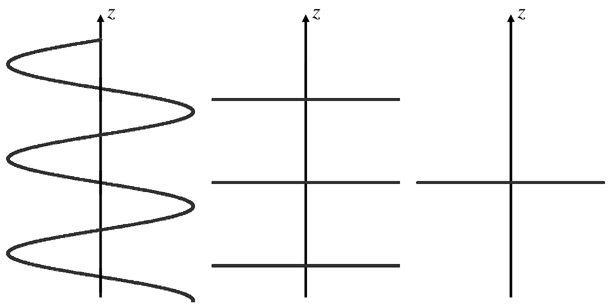


Data courtesy of Siemens Medical Solutions, Forchheim, Germany

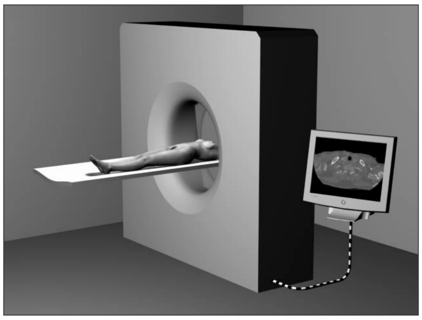
Rows vs. Slices



Scan Trajectories



Spiral **Sequence** **Circle**



Animation by Udo Buhl, Aachen

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Emission vs. Transmission

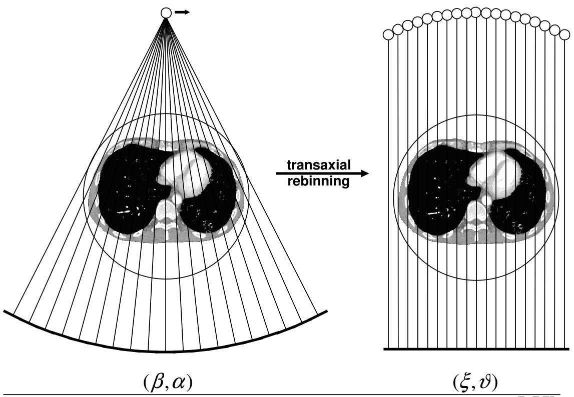
Emission tomography
- Infinitely many sources
- No source trajectory
- Detector trajectory may be an issue
- 3D reconstruction relatively simple

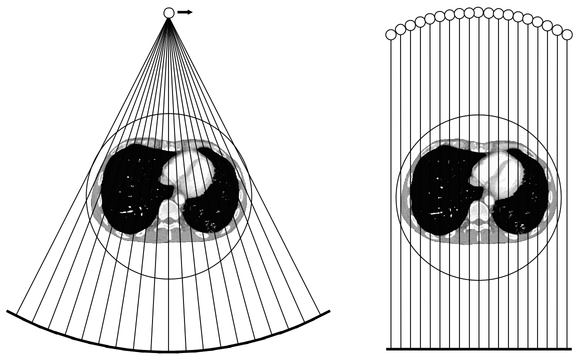
Transmission tomography
- A single source
- Source trajectory is the major issue
- Detector trajectory is an important issue
- 3D reconstruction extremely difficult

2D: In-Plane Geometry

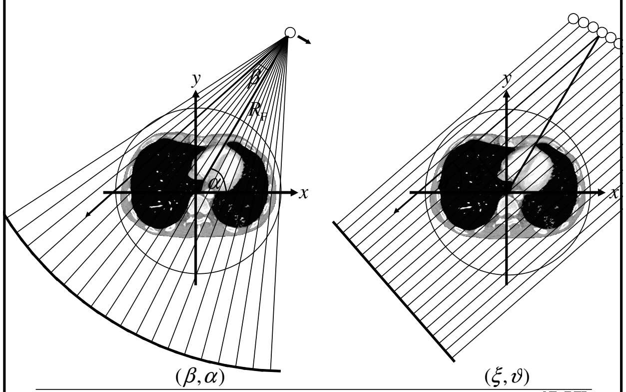
- Decouples from longitudinal geometry
- Useful for many imaging tasks
- Easy to understand
- 2D reconstruction
  - Rebinning = resampling, resorting
  - Filtered backprojection

Fan-beam geometry **Parallel-beam geometry**

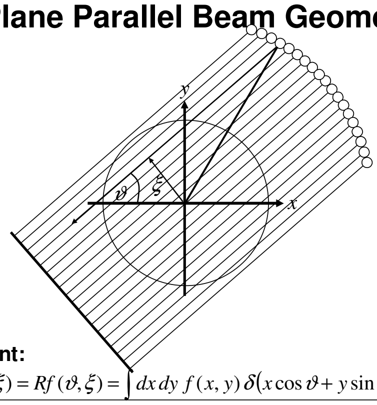




Fan-beam geometry **Parallel-beam geometry**



In-Plane Parallel Beam Geometry



Measurement:

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

FBP (Filtered Backprojection)

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

1D FT: $\int d\xi\, p(\vartheta, \xi)\, e^{-2\pi i \xi u} = \int dx\, dy\, f(x, y)\, e^{-2\pi i u (x \cos\vartheta + y \sin\vartheta)}$

Central slice theorem: $P_2(\vartheta, u) = F(u \cos\vartheta, u \sin\vartheta)$

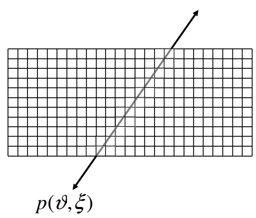
Inversion:

$$f(x, y) = \int_0^\pi d\vartheta \int_{-\infty}^{\infty} du\, |u|\, P_2(\vartheta, u)\, e^{2\pi i u (x \cos\vartheta + y \sin\vartheta)}$$

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

2D Backprojection

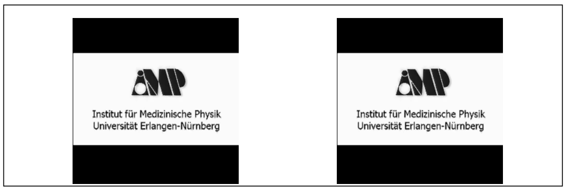
(Discrete Version of the Transpose Radon Transform)



Add ray value to each pixel in the "vicinity" of the ray.

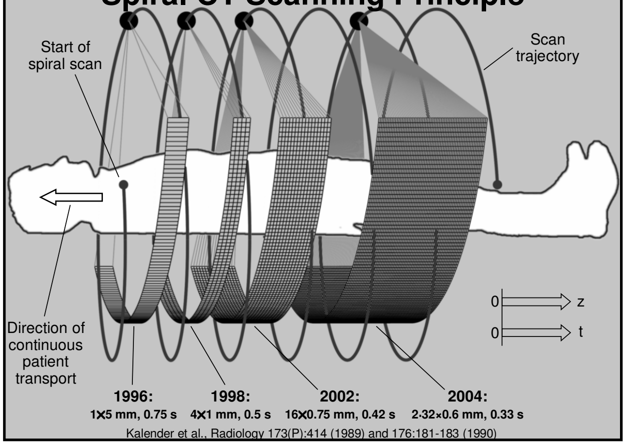
Filtered Backprojection (FBP)

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



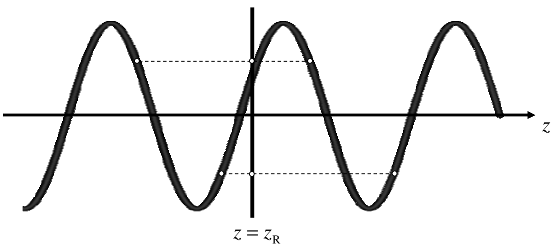
Reconstruction kernels balance between spatial resolution and image noise.

Spiral CT Scanning Principle

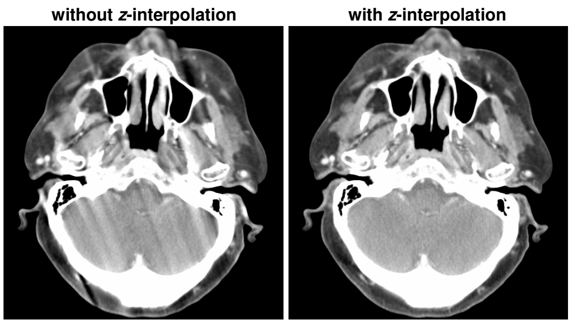


Kalender et al., Radiology 173(P):414 (1989) and 176:181-183 (1990)

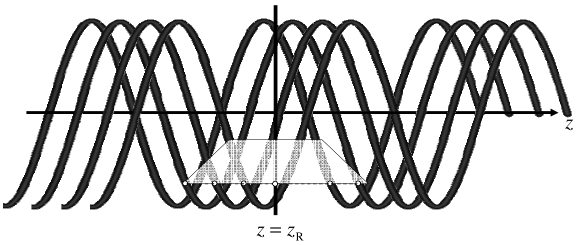
Spiral z-Interpolation for Single-Slice CT M=1



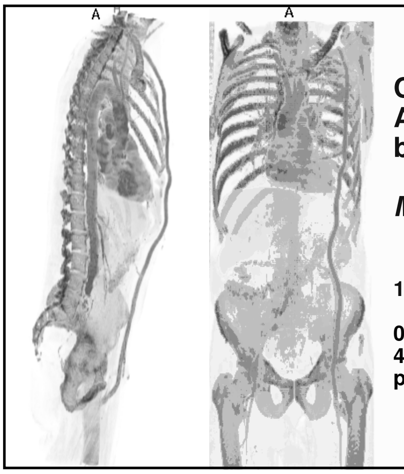
Spiral z-interpolation is typically a linear interpolation between points adjacent to the reconstruction position to obtain circular scan data.



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



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

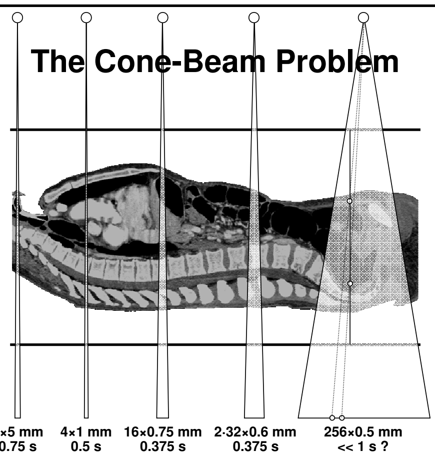


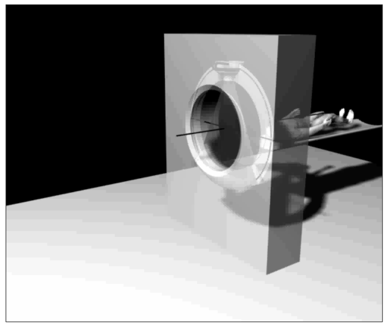
CT Angiography: Axillo-femoral bypass

M = 4

120 cm in 40 s

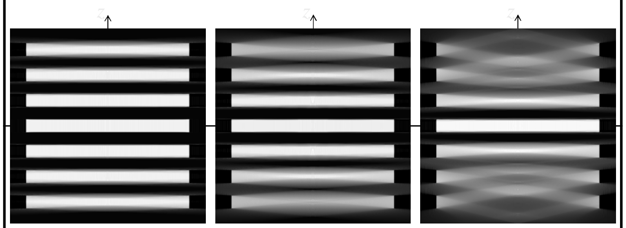
0.5 s per rotation
4×2.5 mm collimation
pitch 1.5



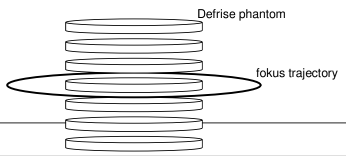


Animation by Siemens

Cone-Beam Artifacts



Cone-angle Γ = 6° | Cone-angle Γ = 14° | Cone-angle Γ = 28°



Advanced Single-Slice Rebinning (ASSR)

3D and 4D Image Reconstruction for Small Cone Angles

- **First practical solution to the cone-beam problem in medical CT**
- **Reduction of 3D data to 2D slices**
- **Commercially implemented as AMPR**
- **ASSR is recommended for up to 64 slices**

Do not confuse the transmission algorithm ASSR with the emission algorithm SSRB!

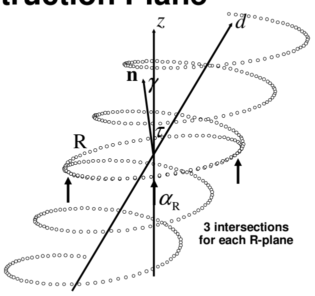
Kachelrieß M, Schaller S, Kalender WA. Med Phys 2000; 27(4):754-772

The Reconstruction Plane

For each reconstruction position α_R minimize the mean deviation of the R–plane and the spiral segment around α_R.

$$\mathrm{R}: \mathbf{n}\cdot\mathbf{r} - c = 0$$

$$\mathbf{n} = \begin{pmatrix} \sin\gamma\cos\varphi \\ \sin\gamma\sin\varphi \\ \cos\gamma \end{pmatrix}$$

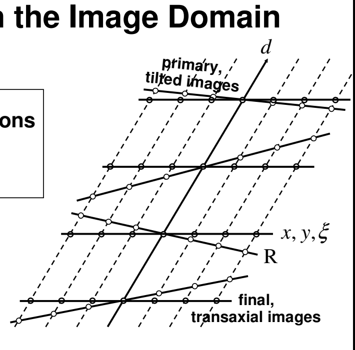


Resulting mean deviation at R_F:	$\Delta_{mean} \approx 0.014\,d$
at R_M:	$\Delta'_{mean} \approx 0.007\,d$

Kachelrieß M, Schaller S, Kalender WA. Med Phys 2000; 27(4):754-772

d–Filtering in the Image Domain

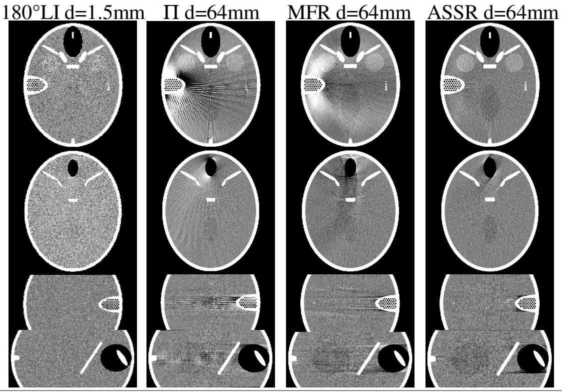
- **No in-plane interpolations**
- **Interpolation along *d***
- **Arbitrary *d*-filter width**



Kachelrieß M, Schaller S, Kalender WA. Med Phys 2000; 27(4):754-772

Comparison to Other Approximate Algorithms

180°LI d=1.5mm Π d=64mm MFR d=64mm ASSR d=64mm

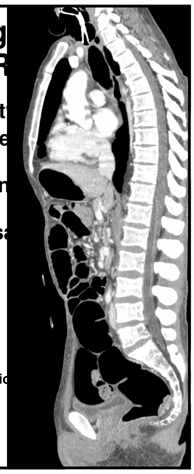


Bruder H, Kachelrieß M, Schaller S. SPIE Med. Imag. Conf. Proc., 3979, 2000

atient Imag with ASSR

- gh image qualit
- gh performance
- se of available) reconstruction rdware
-)0% detector us
- rbitrary pitch

- Sensation 16 at IMP
- 0.5 s rotation
- 16×0.75 mm collimatio
- pitch 1.0
- 70 cm in 29 s
- 1.4 GB rawdata
- 1400 images

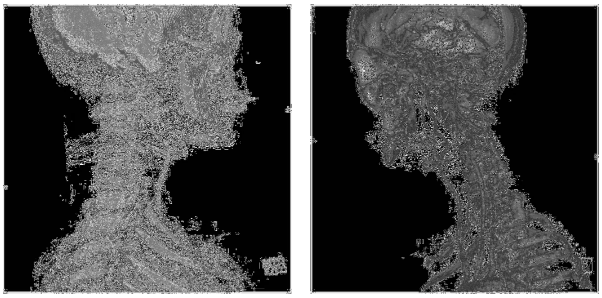


CTA, Sensation 16 at IMP

Data courtesy of Dr. Michael Lell, Erlangen, Germany

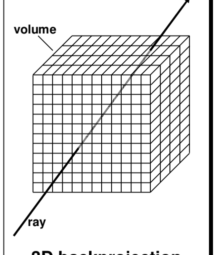
CT-Angiography

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



Feldkamp-Type Reconstruction

- Approximate
- Similar to 2D reconstruction:
  - row-wise filtering of the rawdata
  - followed by backprojection
- True 3D volumetric backprojection along the original ray direction
- Compared to ASSR:
  - larger cone-angles possible
  - lower reconstruction speed
  - requires 3D backprojection hardware



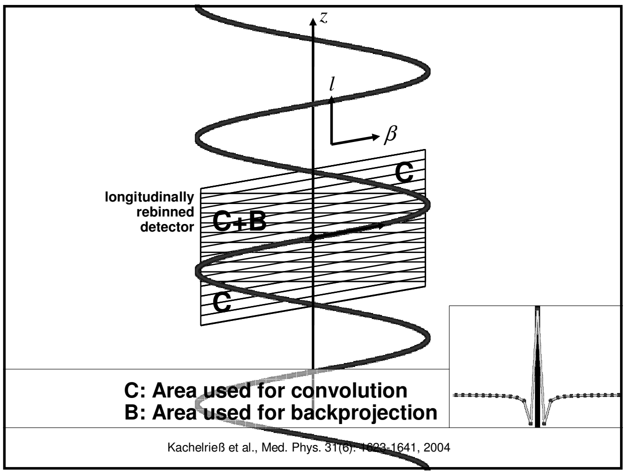
3D backprojection

Extended Parallel Backprojection (EPBP)

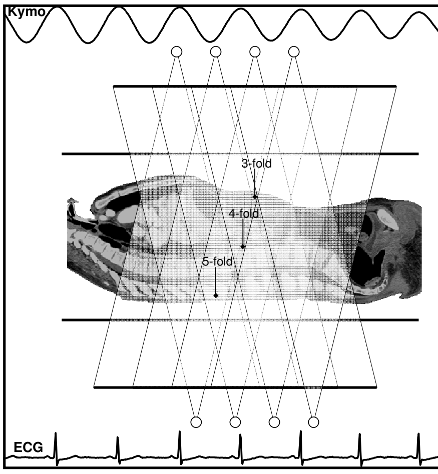
3D and 4D Feldkamp-Type Image Reconstruction for Large Cone Angles

- Trajectories: circle, sequence, spiral
- Scan modes: standard, phase-correlated
- Rebinning: azimuthal + longitudinal + radial
- Feldkamp-type: convolution + true 3D backprojection
- 100% detector usage
- Fast and efficient

Kachelrieß et al., Med. Phys. 31(6): 1623-1641, 2004



Kachelrieß et al., Med. Phys. 31(6): 1623-1641, 2004



The complicated pattern of overlapping data …

… will become even more complicated with phase-correlation.

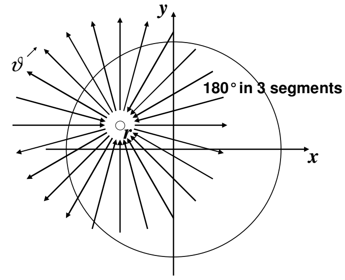
⇒ Individual voxel-by-voxel weighting and normalization.

The 180° Condition

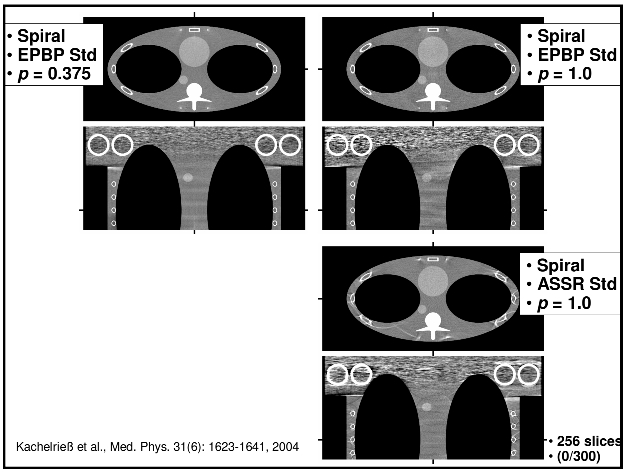
$$\int d\vartheta\, w(\vartheta) = \pi$$

and

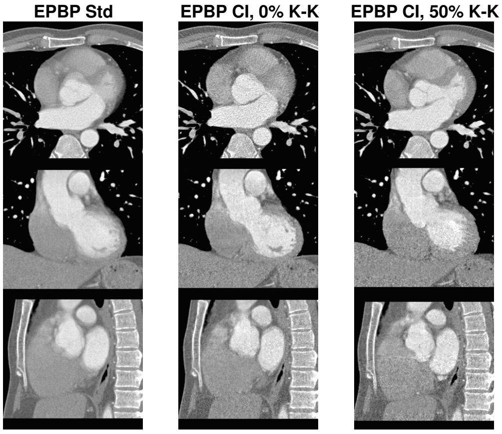
$$\sum_k w(\vartheta + k\pi) = 1$$



The (weighted) contributions to each object point must make up an interval of 180° and weight 1.



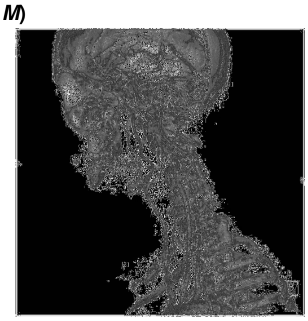
Kachelrieß et al., Med. Phys. 31(6): 1623-1641, 2004



Patient example, 32x0.6 mm, z-FFS, p=0.23, t_{rot}=0.375 s.

Advantages of Multi-Slice Spiral CT

- Image quality independent of scan parameters
- Increase (up to a factor of *M*)
  - of scan speed
  - of z-resolution
- New applications
  - CT angiography
  - dynamic studies
  - virtual endoscopy
  - cardiac CT
  - …



Today, complete anatomical regions are routinely scanned with MSCT within a few seconds with isotropic sub-millimeter spatial resolution.

Motion Artifacts of the Heart

Cardiac CT

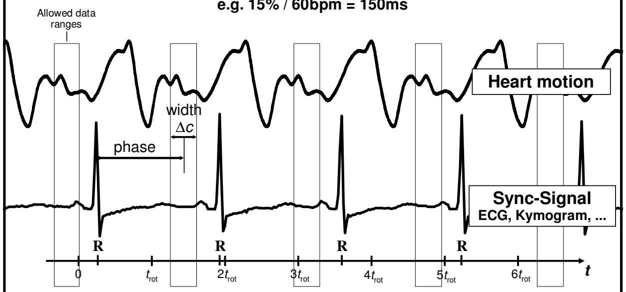
- Periodic motion
- Synchronisation needed (ECG, Kymogram, others)
- Prospective Gating
- Phase-correlated reconstruction = Retrospective Gating
  - Single-phase (partial scan) approaches, e.g. 180°MCD
  - Bi-phase approaches, e.g. ACV (Flohr et al.)
  - Multi-phase Cardio Interpolation methods, e.g. 180°MCI (gold-standard)
- Generations
  - Single-slice spiral CT: 180°CD, 180°CI (introduced 1996*)
  - Multi-slice spiral CT: 180°MCD, 180°MCI (introduced 1998*)
  - Cone-beam spiral CT: ASSR CD, ASSR CI (introduced 2000*)
  - Wide cone-beam CT: EPBP (introduced 2002*)

*Med. Phys. 25(12) 1998, Med. Phys. 27(8) 2000, Proc. Fully 3D 2001, Med. Phys. 31(6) 2004

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.

Kachelrieß et al., Radiology 205(P):215, (1997)

Maximum Pitch for Full Phase Selectivity

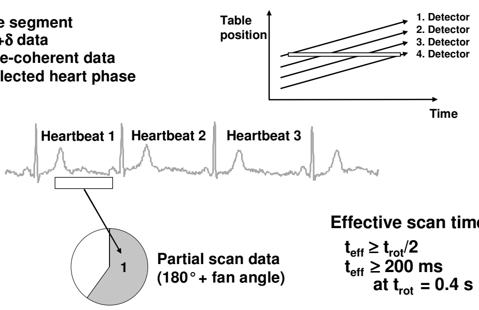
- Voxel illumination must exceed one motion cycle
- Table increment per motion cycle must not exceed collimation

$$p \le f_{H}\, t_{rot}$$

- E.g. t_{rot} = 0.5 s and f_H = 60 bpm implies $p < 0.5$
- The smaller the pitch value the more segments can be combined

Partial Scan Reconstruction

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



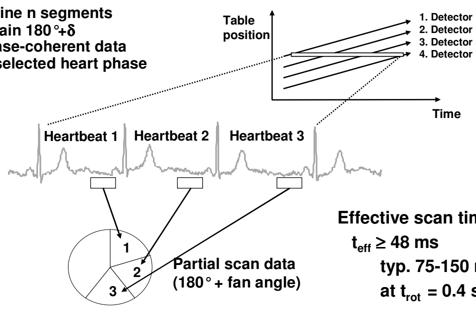
Effective scan time

$t_{eff} \ge t_{rot}/2$
$t_{eff} \ge 200$ ms
at $t_{rot} = 0.4$ s

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

Multi-Segment Reconstruction

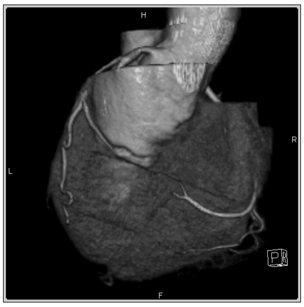
Combine n segments to obtain 180°+δ of phase-coherent data for a selected heart phase



Effective scan time

$t_{eff} \ge 48$ ms
typ. 75-150 ms
at $t_{rot} = 0.4$ s

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



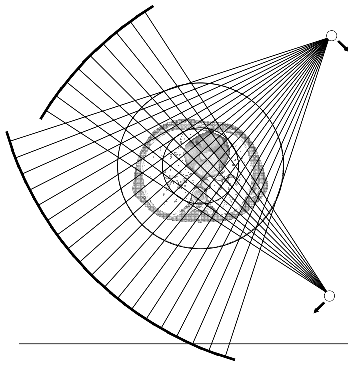
Volume Zoom, 4 × 2.5 mm, 0.5 s, 1998

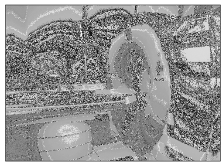
Multi-segment 180°MCI reconstruction, 90 bpm

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

Data courtesy of Stephan Achenbach

Multi-Threaded CT, Dual Source CT



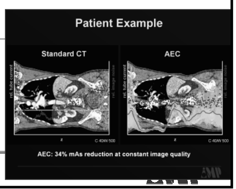


Siemens SOMATOM Definition dual source cone-beam spiral CT at the IMP

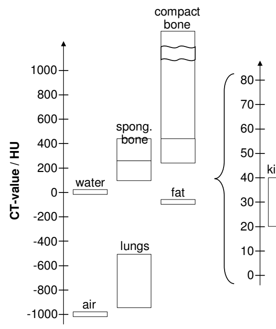
CT Basics

From Single-Slice to Cone-Beam Spiral CT

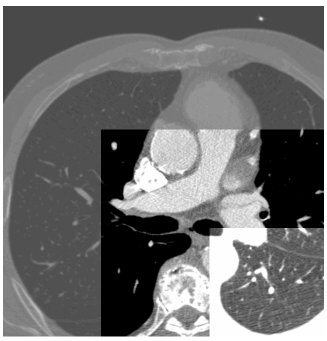
- **Technology**
  - **Basic parameters**
  - **Detector concepts, tube technology**
  - **Scan trajectories, scan modes**
- **Algorithms**
  - **2D filtered backprojection**
  - **Spiral z-interpolation**
  - **ASSR and EPBP (cone-beam recon.)**
  - **Phase-correlated CT (e.g. cardiac CT)**
- **Image quality and dose**
  - **Spatial resolution (PSF, SSP, MTF)**
  - **Relation of noise, dose and resolution**
  - **Dose values (CTDI, patient dose)**
  - **Dose reduction techniques**

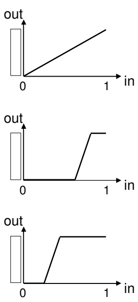


What is Displayed?



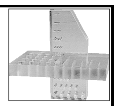
$$CT(\boldsymbol{r}) = \frac{\mu(\boldsymbol{r}) - \mu_{\mathrm{Water}}}{\mu_{\mathrm{Water}}} \cdot 1000\,\mathrm{HU}$$



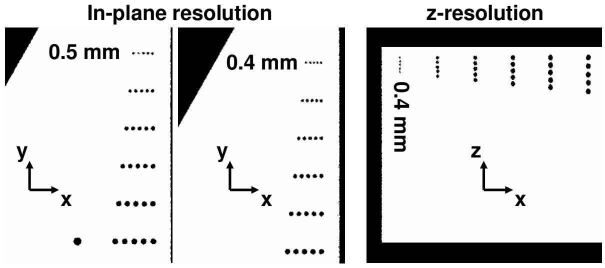


(0, 5000) (0, 1000) (-750, 1000)

Spatial Resolution 1



In-plane resolution **z-resolution**

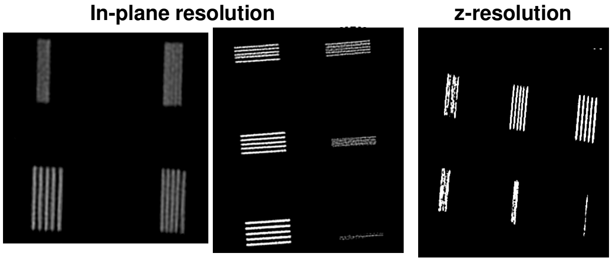


Std. scan, x/y **UHR scan, x/y** **Std. or UHR scan, x/z**

Sensation 64, collimation: 2·32×0.6 mm

Spatial Resolution 2

In-plane resolution | **z-resolution**

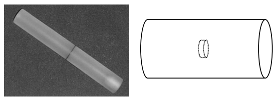


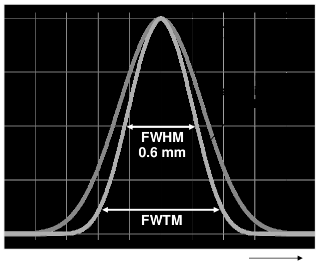
Std. scan, x/y | UHR scan, x/y | Std. or UHR scan, x/z

Sensation 64, collimation: 2·32×0.6 mm

Spatial Resolution 3

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





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

Sensation 64, collimation: 2·32×0.6 mm

Tricks to Improve Resolution

- Sharp reconstruction kernels
- Lowest possible S_{eff}
- Decrease the size of the detector pixels
- Oversampling
  - zFFS
  - αFFS
  - Detector quarter offset
- Use of detector combs

However, image noise becomes crucial!

Dependencies of IQ and Dose

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

$$\sigma^2 \propto \frac{1}{D} \propto \frac{1}{\mathrm{mAs_{eff}}}$$

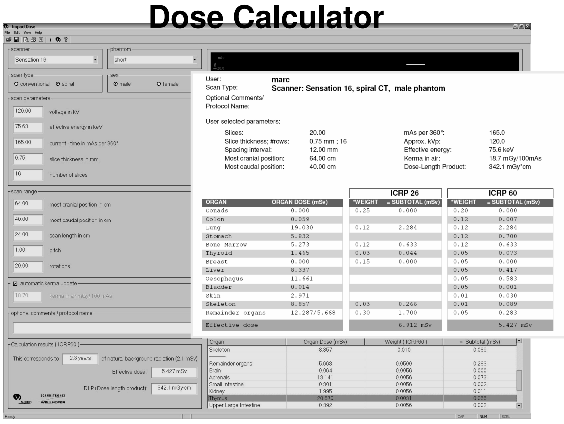
- Dose increases with the fourth power of the spatial resolution for a given object and image noise

$$(\sigma/\mu)^2 \propto \frac{e^{\mu 2R}+1}{\mu^2 dx^4}$$

Noise relative to the background (=1/SNR)

Fourth power of the resolution element size

Dose Calculator



Demo version of ImpactDose available at www.vamp-gmbh.de

Patient Dose in CT

Typical Values for 16-Slice Scanners

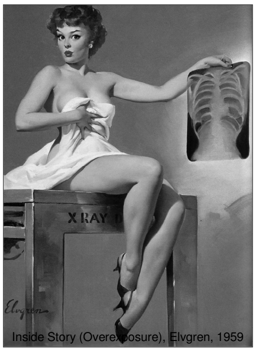
	Head	Thorax	Abdomen	Pelvis
Scan range / mm	120	300	400	200
Scan time / s	0.75	0.5	0.5	0.5
Collimation / mm	16×0.75	16×0.75	16×0.75	16×0.75
Eff. mAs / mAs	320	100	160	160
Critical organ	Brain	Lung	Stomach	Colon
Organ dose / mSv	54.6	11.2	15.7	11.7
Eff. dose / mSv	2.9	3.8	8.2	3.9
Eff. dose / 2.1 mSv	1.4	1.8	3.9	1.9

Routine protocols, 120 kV, male phantom.

Strategies for Dose Reduction

Potential reasons for an increase:
- **Higher volume coverage**
- **Multiphasic examinations**
- **More examinations**
- **Higher spatial resolution**
- **New special applications**

Potential ways to decrease dose:
- **New display techniques**
- **Advanced reconstruction (MAF)**
- **Automatic exposure control (AEC)**
- **Optimized spectra**
- **Dose training (dose tutor)**

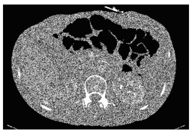


Inside Story (Overexposure), Elvgren, 1959

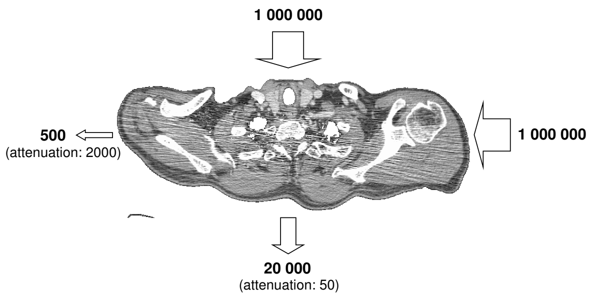
Standard Display


0,5×0,5×0,5 mm³
C = 50 HU, W = 400 HU


Sliding Thin Slab (STS) Display


0,5×0,5×10 mm³
C = 50 HU, W = 400 HU

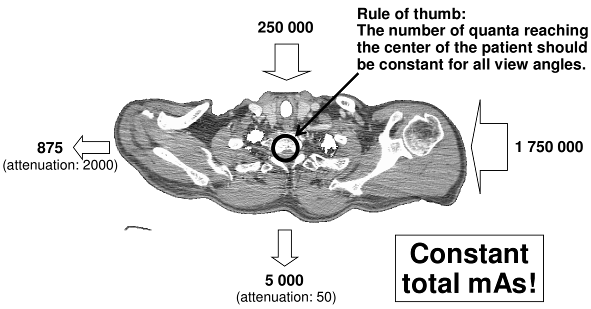

Tube Current Modulation



Constant tube current: High, inhomogeneous noise.

$$\sigma^2_{\text{pixel}} = const. \cdot \sum_n \sigma^2_{\text{projection},n}$$

Tube Current Modulation



Modulated tube current: Low, homogeneous noise.

$$\sigma^2_{\text{pixel}} = const. \cdot \sum_n \sigma^2_{\text{projection},n}$$

Dose Reduction by Tube Current Modulation


Conventional scan: 327 mAs

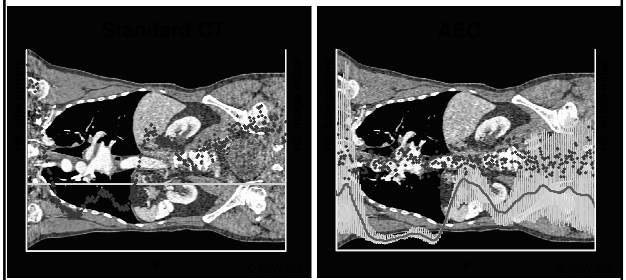

Online current modulation: 166 mAs

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

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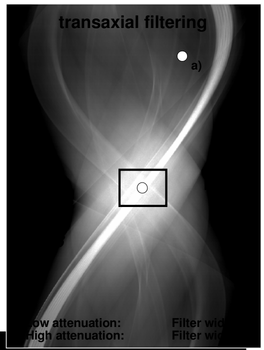
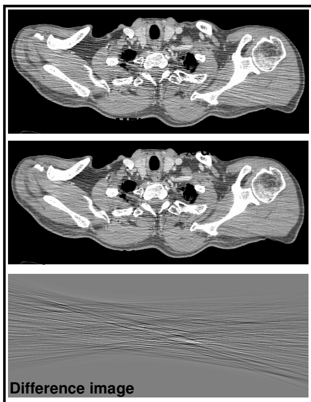
Multidimensional Adaptive Filtering (MAF)

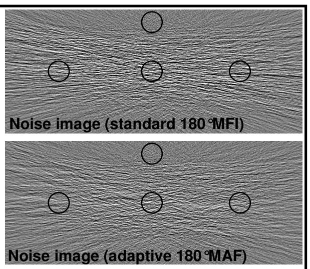
- Rawdata based
- Local smoothing of noisy data (less than 5% modification)
- No loss of spatial resolution
- Efficient
- Noise reduction can be equivalently converted to dose reduction



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

Kachelrieß M, Watzke O, Kalender WA. Med Phys 2001; 28:475-490



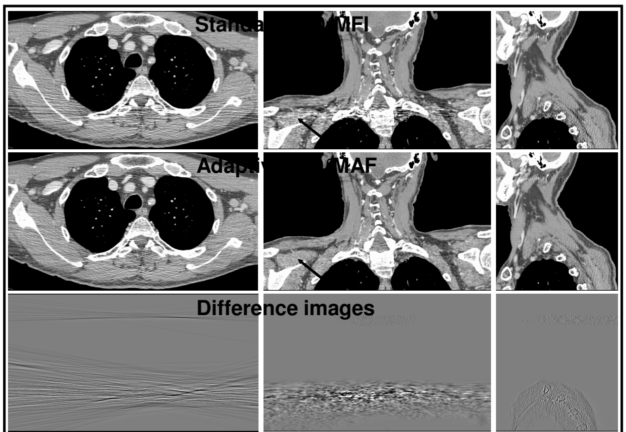


180°MAF relative to 180°MFI:

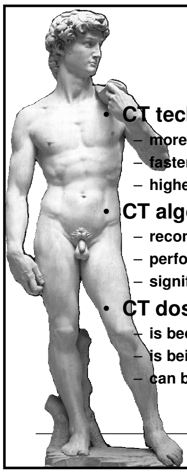
	Noise	Dose	Resolution
left:	61%	37%	97%
center:	63%	40%	97%
right:	60%	36%	97%
upper:	100%	100%	100%

Noise in the shoulder region typically reduced to 50%...70%.

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



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



Summary

- **CT technology is further evolving towards**
  - more slices
  - faster rotation times
  - higher spatial resolution
- **CT algorithms**
  - reconstruct cone-beam data for any trajectory
  - perform phase-correlated imaging (4D)
  - significantly reduce artifacts (beam hardening, truncation …)
- **CT dose**
  - is becoming more and more an important issue (also in the US)
  - is being reduced by manufacturers‘ efforts (e.g. MAF, AEC)
  - can be most significantly reduced by user training

Thank You!