Optimization of Image Acquisition and Reconstruction in Multi-slice CT

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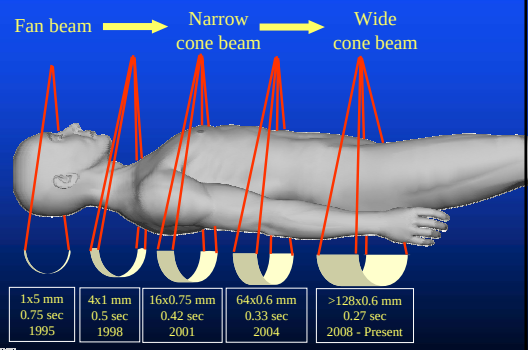
Learning Objectives

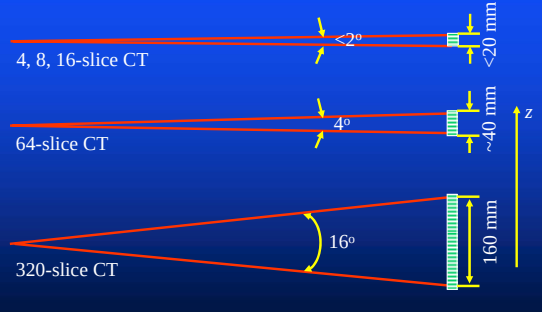
- Clinical implementations of reconstruction algorithms on multi-slice CT scanners.
- How CT acquisition parameters affect image reconstruction and image quality.
- How reconstruction parameters affect image quality.

I. Clinical Implementations of Image Reconstruction

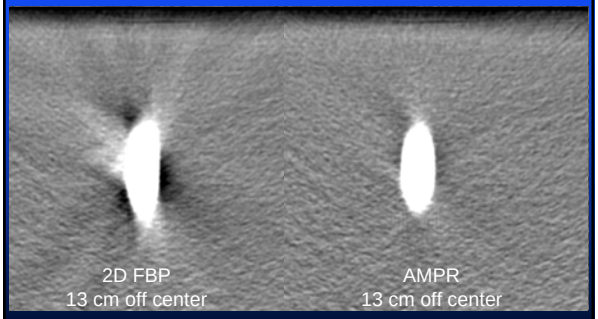
From Single-slice to Multi-slice CT



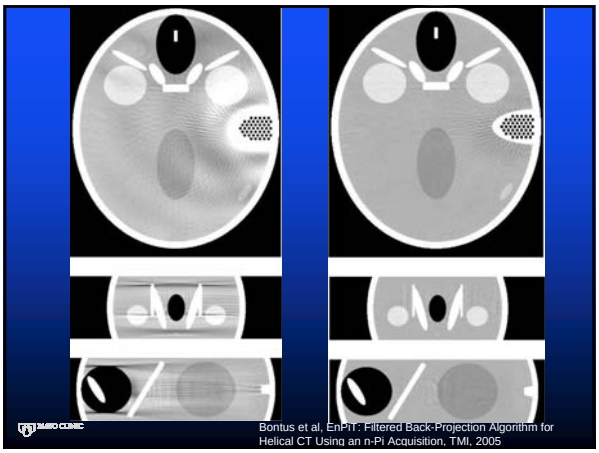
Cone-beam Effect



Cone-beam Effect

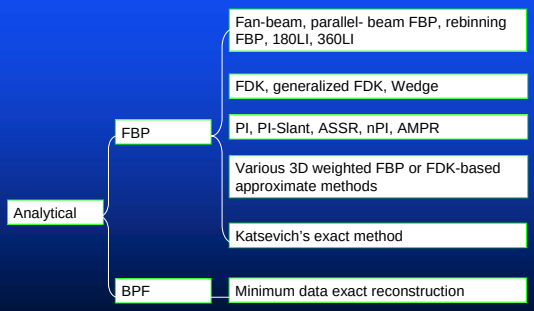


Courtesy D. Platten et al. ImPACT (RSNA 2003)



Bontus et al, EnPiT: Filtered Back-Projection Algorithm for Helical CT Using an n-Pi Acquisition, TMI, 2005

Analytical Reconstruction Methods



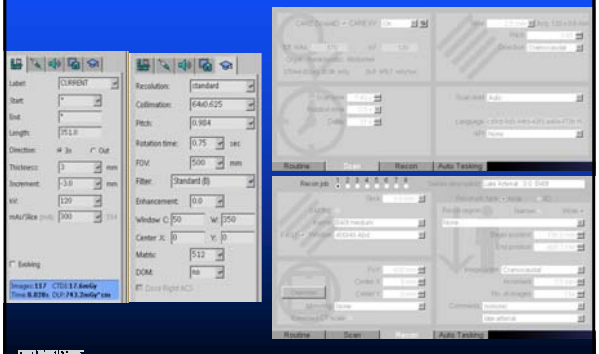
Clinical Implementations of Image Reconstruction

- All current analytical reconstructions on clinical scanners are approximate methods
- Goal is to maintain a good balance between
  - computationally efficiency
  - accuracy
  - dose efficiency – fully utilize redundant data

Vendor Variations on Analytical Reconstructions

- GE – 3D Weighted FBP (Tang et al, PMB, 2006)
- Philips – COBRA, nPI (Kohler et al, Med Phys, 2002; Bontus et al, TMI, 2005)
- Siemens – AMPR (Flohr et al, Med Phys 2003) and 3D weighted FBP (Stierstorfer et al, PMB, 2004)
- Toshiba – TCOT, modified FDK (Taguchi et al, Med Phys, 2003)

What can users control on image reconstruction?

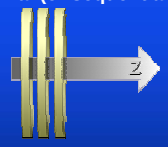


II. Acquisition parameters

- Scan mode: Axial, helical, cine, shuttle, ECG-gated
- Rotation time (e.g., 0.28 sec, 0.35 sec, 0.5 sec)
- X-ray beam filter (e.g., flat filter, bowtie filter)
- X-ray beam collimation
- Detector configuration (e.g., 64x0.625 mm, 24x1.2 mm, 128x0.6 mm)
- Helical pitch (in helical mode); Table feed (in axial mode)
- Tube potential: kV
- Tube current: mA
- mAs: Tube current (mA) x Rotation time (s)
- Effective mAs: mAs/pitch
- Automatic exposure control (AEC)

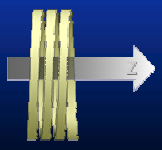
Scan Mode

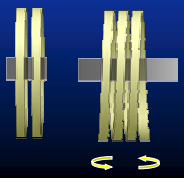
- Axial (or sequential)
- Cine



Continuous axial scan, no table move

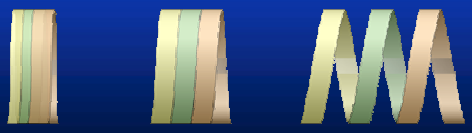
- Helical (or spiral)
- Shuttle or 4D spiral



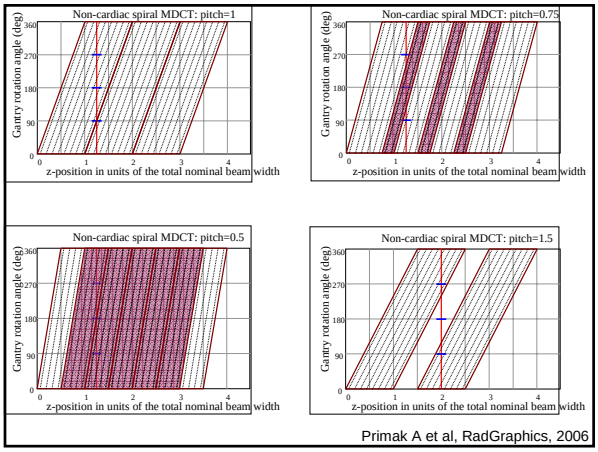


Helical Pitch

$$\text{Pitch} = \frac{\text{Table translation per rotation}}{\text{Detector collimation}}$$



Pitch < 1 | Pitch = 1 | Pitch > 1

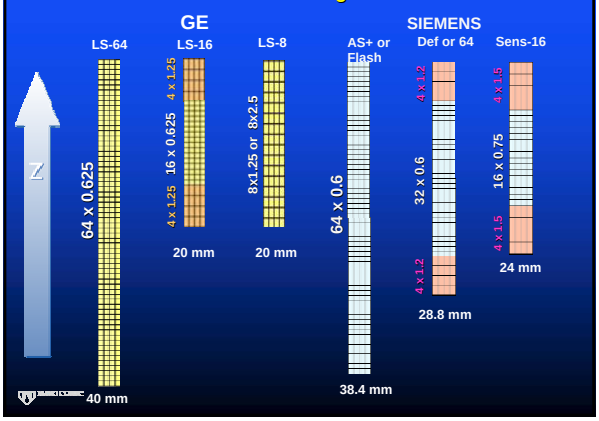


Primak A et al, RadGraphics, 2006

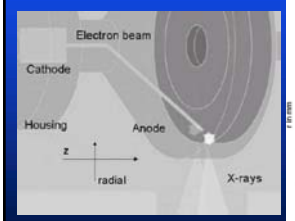
Pitch and Image Quality

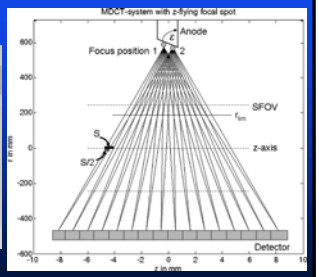
- Higher pitch – higher scan speed
- For non-cardiac exams, a higher pitch scan has a higher temporal resolution.
- Downside of higher pitch
  - Lower dose capacity
  - More severe helical artifacts (windmill)
  - More severe cone-beam artifacts If combined with wide collimation

Detector Configuration



Z-flying focal spot

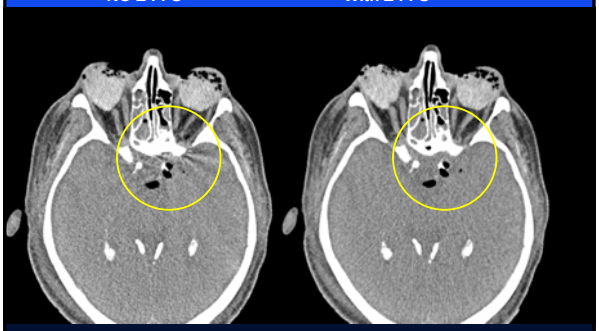




Flohr et al, Image reconstruction and image quality evaluation for a 64-slice CT scanner with z-flying focal spot. Med Phys 32, 2536, 2005.

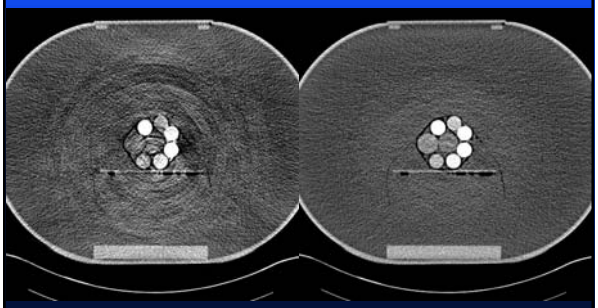
Effect of Z FFS

NO Z FFS | With Z FFS



Flohr et al, Image reconstruction and image quality evaluation for a 64-slice CT scanner with z-flying focal spot. Med Phys 32, 2536, 2005.

Detector configuration and electronic noise

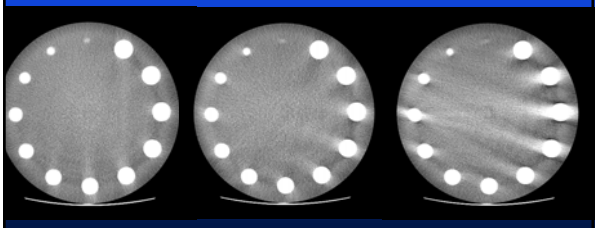


128 x 0.6* mm Equal $CTDI_{vol}$ 32 x 1.2 mm

Wider detector collimation

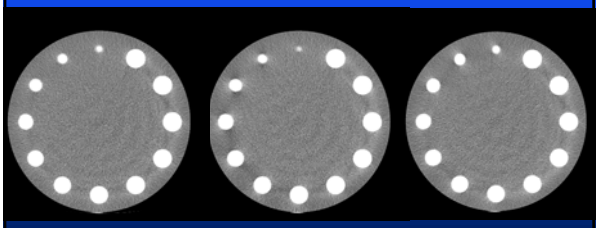
- Routinely used for faster speed and better use of tube power (64x0.625 mm, 128x0.6* mm, …)
- Tends to generate more severe artifacts if not well-corrected
  - Helical and cone-beam artifacts, particularly for high-contrast object at high pitch
- Some applications, narrower collimation is better
  - e.g., 32x0.625 mm is often a better choice for head CT than 64x0.625 mm
- Use slower pitch for wide collimation if there is no scan time concern.

Cone-beam Artifacts in Wider Collimation (4 cm)



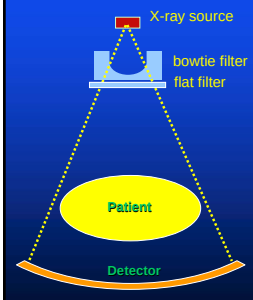
Pitch=0.516 Pitch=0.983 Pitch=1.375

Narrower Collimation (2 cm)



Pitch=0.562 Pitch=0.969 Pitch=1.375

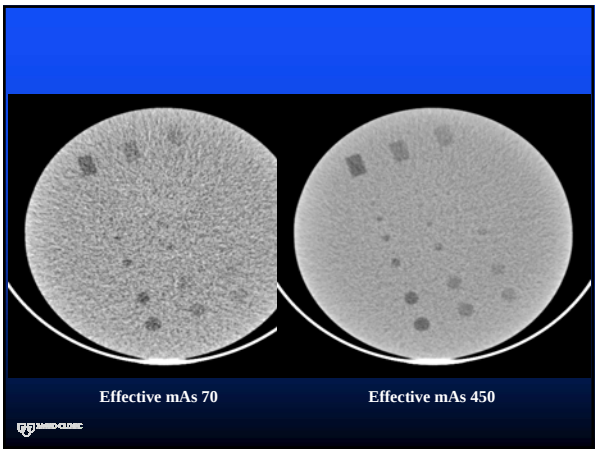
X-ray Beam Bowtie Filter

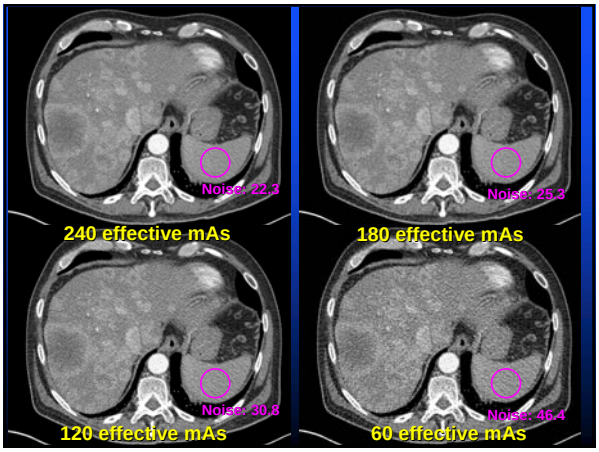


- Small bowtie
  - Ped Head, Cardiac Small, Small Head, Ped Body, Small Body
- Medium bowtie
  - Head, Cardiac Medium, Medium Body
- Large bowtie
  - Large body, Cardiac Large

mA, mAs, effective mAs, mAs/slice

- mAs: Tube current (mA) x Rotation time (s)
- Effective mAs = mAs/pitch
- Effective mAs is proportional to the radiation output (usually expressed as $CTDI_{vol}$)
- mA or mAs cannot be related to $CTDI_{vol}$ without knowing the pitch

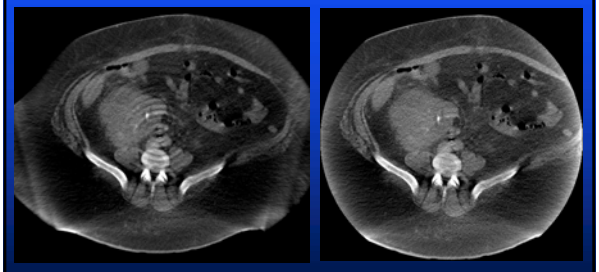




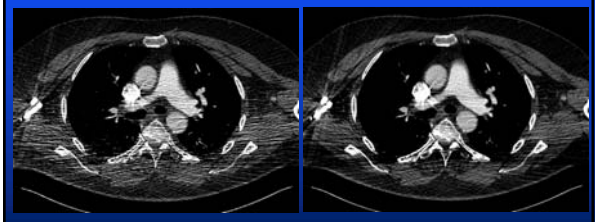
mAs vs. Noise in Cardiac and Non-cardiac Exams

- In non-cardiac exams, image noise is directly related to effective mAs (mAs/pitch).
  - Redundant data is used when pitch is less than 1
- In cardiac exams, image noise is directly related to mAs.
  - Independent of Pitch
  - Each reconstruction uses data from a half-scan range.
  - No redundant data is used even when pitch is much less than 1.

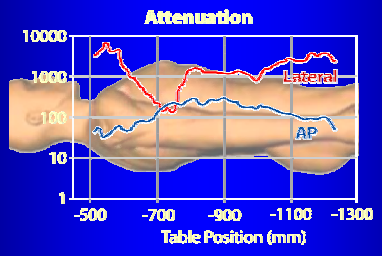
Photon Starvation



Adaptive Filter in Reconstruction



Automatic Exposure Control (AEC)



X-ray attenuation

- Varies over body region and with projection angle
- Image noise is primarily determined by noisiest projections (thick body parts)
- More photons (dose) to thinner body parts is unnecessary radiation dose

Automatic tube current modulation

- **Goal of tube current modulation is to dynamically adjust tube current according to the attenuation level.**
  - **Higher attenuation, higher tube current**
  - **Lower attenuation, lower tube current**

General strategy $I(d) = I(d_0) \cdot \left(\frac{A(d)}{A(d_0)} \right)^x$

where $A(d) = \exp(\mu d)$

d_0 is the reference attenuation path length. x is the modulation strength

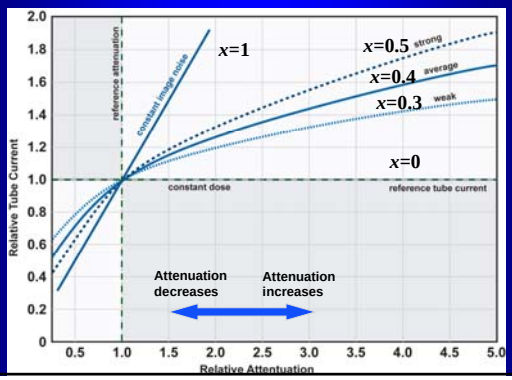
If $x = 0$ then $I(d) = I(d_0)$ Constant tube current

If $x > 0$ then tube current is modulated according to attenuation

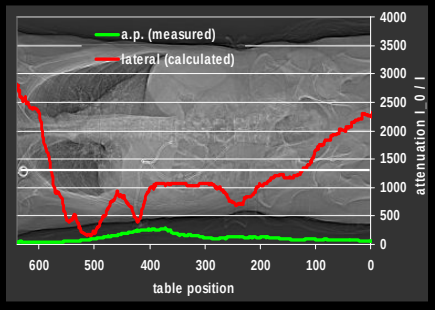
$x = 1$ equal noise for each view independent of attenuation

M. Gies, W. A. Kalender, H. Wolf and C. Suess, "Dose reduction in CT by anatomically adapted tube current modulation. I. Simulation studies," Med Phys. 26, 2235-47 (1999)

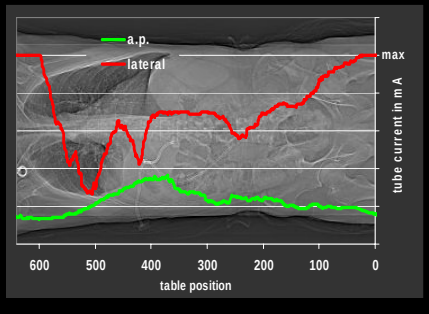
Automatic Tube Current Modulation



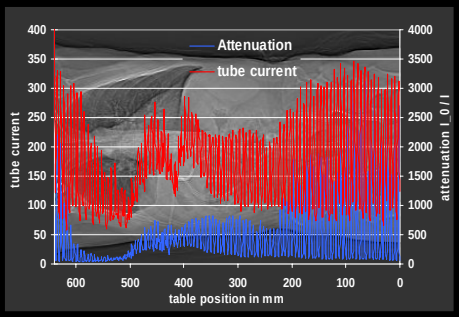
Topogram Evaluation: a.p. and lateral

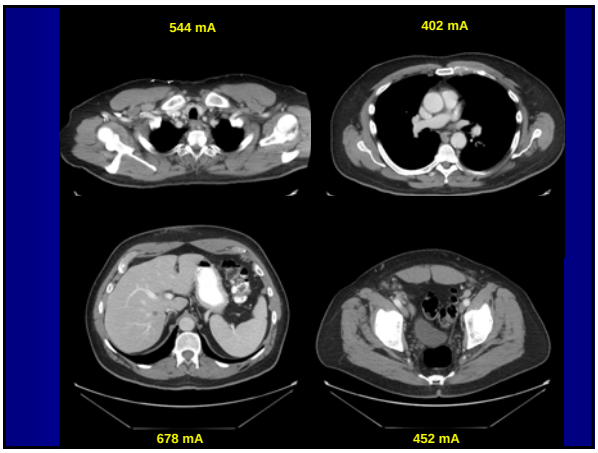


Optimal mA: a.p. and lateral

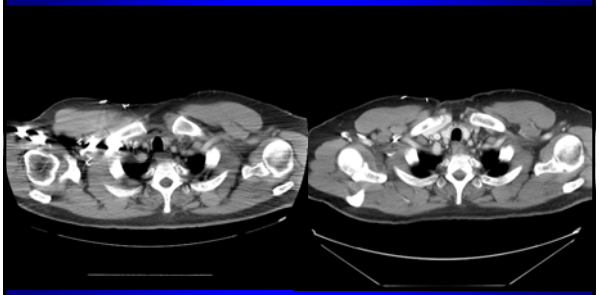


On-line tube current modulation





Streaking Artifact Reduction

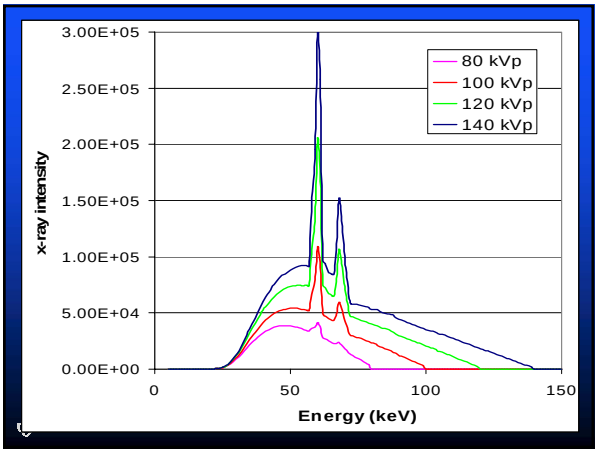


Constant Tube Current AEC

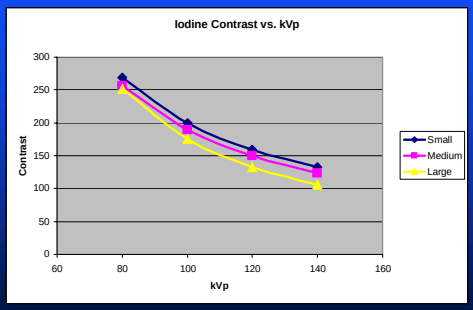
Different Versions of AEC

Manufacturer	AEC Trade name	Image Quality reference	Goal
GE	Auto mA, Smart mA	Noise Index	Maintain a constant noise level (defined in noise index), using tube currents within prescribed minimum and maximum values.
Toshiba	SureExposure	Standard deviation (high quality, standard, low dose)	Maintain a constant noise level (defined in standard deviation values for each protocol), using tube currents within preset minimum and maximum values.
Siemens	CARE Dose4D	Quality Reference mAs	Maintain the same image quality (varying noise target for different attenuation level) with reference to a target effective mAs level for a standard-sized patient.
Philips	DoseRight	Reference Image	Keep the same image quality as in the saved reference image.

Tube Potential (kV)



Iodine Contrast

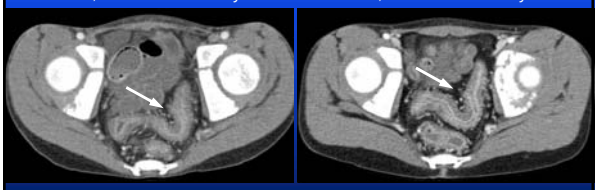


Phantom size: small 30x13 cm, medium 30x21 cm, large, 36x27 cm

Lower-kV Benefit – Increased Iodine Contrast

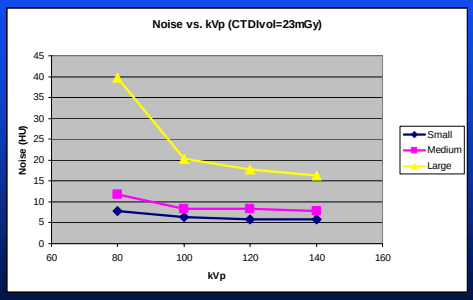
120 kV, CTDIvol=5.18 mGy

100 kV, CTDIvol=3.98 mGy



The same patient scanned with a protocol at 120 kV and a protocol at 100 kV. Note that improved contrast and visualization of mural stratification of the 100 kV image despite a 25% radiation dose reduction.

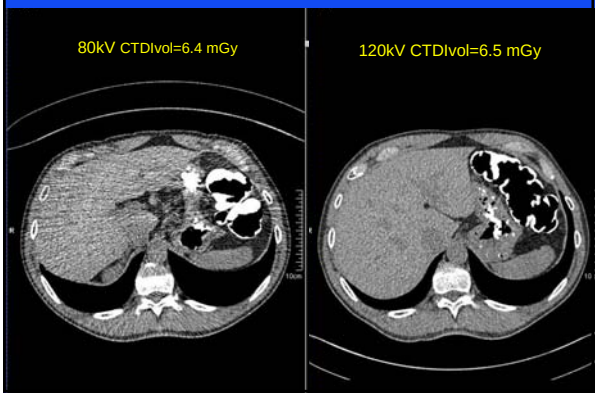
Noise



Limitation of Lower kV – increased noise

80kV CTDIvol=6.4 mGy

120kV CTDIvol=6.5 mGy



Optimal kV

- **Optimal kV is the kV that uses the minimum radiation dose to achieve the desired image quality**
  - **Dependent on patient size**
  - **Dependent on diagnostic task**

A General Strategy for Optimal kV Selection

- *Iodine CNR with a noise constraint α:*

$$CNR(\Omega_d, kV, D) \geq CNR(\Omega_d, kV_{ref}, D_{ref})$$

and

$$\sigma(\Omega_d, kV, D) \leq \alpha \cdot \sigma(\Omega_d, kV_{ref}, D_{ref})$$

- The corresponding relative dose factor (RDF) at each kV is given by

$$RDF(\Omega_d, kV) = \max\left\{\frac{C(\Omega_d, kV_{ref})}{C(\Omega_d, kV)}, \frac{1}{\alpha}\right\} \cdot \frac{k(\Omega_d, kV)}{k(\Omega_d, kV_{ref})}$$

L. Yu, H. Li, J. Fletcher, C. McCollough, Automatic selection of tube potential for radiation dose reduction in CT: A general strategy. Med Phys, 37:234-43, 2010.

Optimal kV for Abdominal CT

Modulation strength	25 cm	30 cm	35 cm	40 cm	45 cm	50 cm	55 cm
Very weak	120	120	120	120	120	120	140
Weak	100	100	100	120	120	120	140
average	100	100	100	100	120	120	140
Strong	80	80	100	100	100	120	140
Very strong	80	80	80	100	100	120	140

- Very weak: Routine non-contrast exams
- weak: Liver/pancreas
- average: Routine contrast-enhanced exams
- strong: CT enterography, CTU, stone
- Very strong: CT angiography

III. Reconstruction parameters

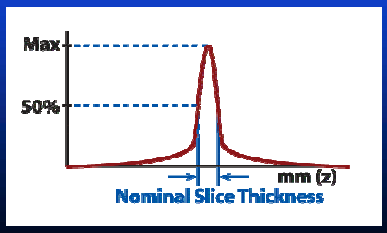
- Slice thickness/Increment
- Reconstruction kernel
- Reconstruction FOV
- 3D reformat
- Iterative reconstruction or noise reduction options

Slice Thickness

- Nominal width of image (z-direction)

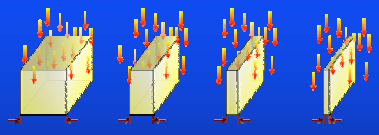
Slice sensitivity profile

(line spread function along Z)



Slice Thickness, Noise, and Dose

$$\text{Noise} \propto \frac{1}{\sqrt{\#\,\text{Photons}}}$$

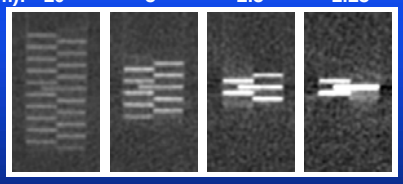


Slice thickness (mm):	5	2.5	1.25	0.625
Relative Noise:	100%	141%	200%	283%
Required mAs: (for = noise)	100%	200%	400%	800%

- Better z-resolution (less partial vol. averaging)
- Increased image noise
- *Requires* more radiation dose to get the same noise level

Slice Thickness – Partial Volume

Image (mm): 10 5 2.5 1.25

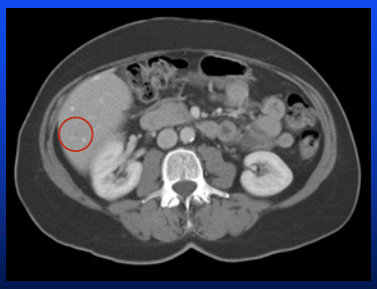


Noise (HU): 2.93 3.84 5.89 7.82

All other parameters are identical

- Thinner slices => less partial volume effect

Slice Thickness – Partial Volume



10mm image thickness

All other parameters are identical

Slice Thickness – Partial Volume

5mm image thickness

All other parameters are identical

Slice Thickness – Partial Volume

2mm image thickness

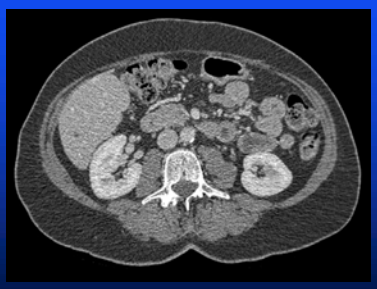
All other parameters are identical

Slice Thickness – Partial Volume

1mm image thickness

All other parameters are identical

Slice Thickness – Partial Volume



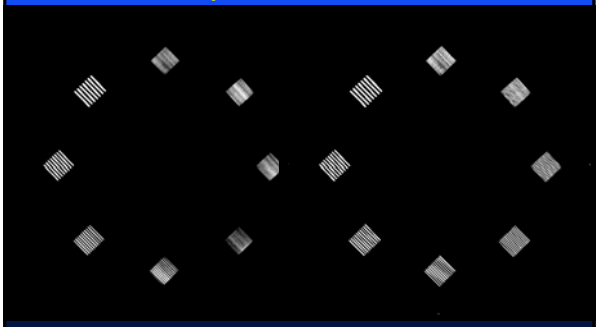
0.6mm image thickness

All other parameters are identical

Reconstruction kernel

- Reconstruction kernel (filter or algorithm) has a significant impact on spatial frequency and noise characteristics of an image
- Smooth kernels reduce high spatial-frequency information and image noise
- Sharp kernels increase high spatial-frequency information and image noise

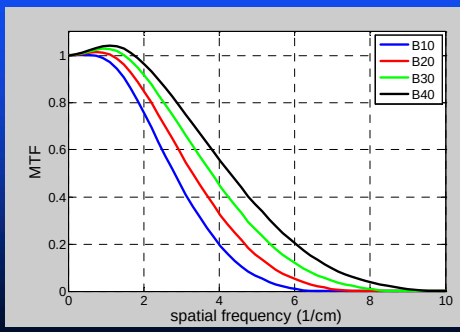
Impact of Reconstruction Kernel on In-plane spatial resolution

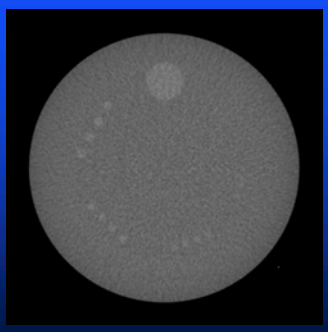


A smoother kernel

A sharper kernel

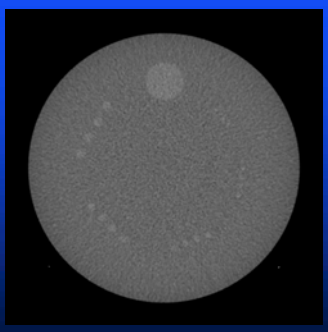
Modulation Transfer Function (MTF)





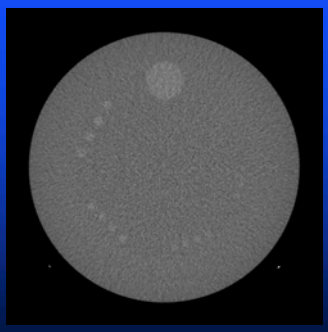
Soft

CT number: 89.7±2.2



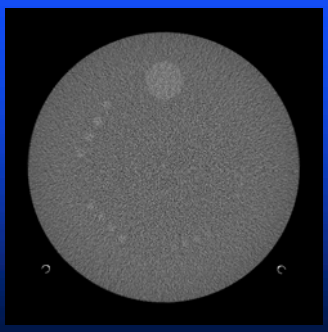
Standard

CT number: 89.7±2.6



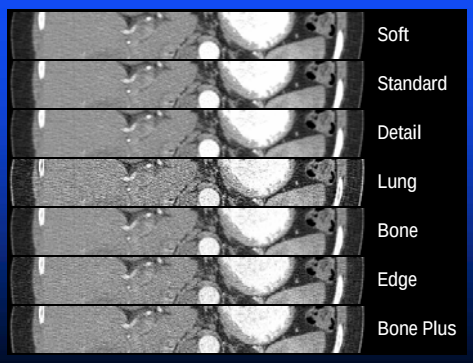
Detail

CT number: 90.7±3.5



Bone

CT number: 90.6±8.9



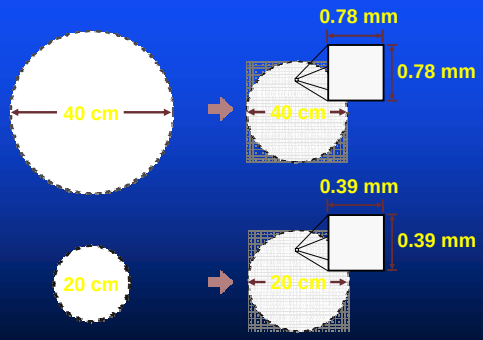
Some Considerations on GE Reconstruction algorithms

- Lung and bone plus alter CT # values
- For high res lung use bone
- Standard is most frequently used
- Soft is to recover a really grainy exam
- Use Full recon for getting the slice width you ask for
- Plus mode can help smooth noisy images or reduce some artifacts, but is up to 20% wider slice thickness

Siemens Reconstruction Kernels

- B10 → B90 Body
- H10 → H90 Head
- U30 → U90 Ultra High Resolution
- T20 → T81 Topogram
- Lower number smoother
- Higher number sharper
- Multiples of 10 are the "basic" kernels
- In between values are "special" kernels

Reconstruction FOV & Pixel Size



Pixel size & System Spatial Resolution

- Pixel size should not be confused with system spatial resolution
- Pixel size should be small enough in order not to limit the display of system resolution.

- 512 x 512 matrix

rFOV (cm)	Pixel Size (mm)	Pixel size Resolution (lp/cm)	System spatial Resolution at 10% MTF (lp/cm) B20	B30	B40
50	1.0	5.1	5.5	6.2	7.0
40	0.8	6.4			
30	0.6	8.5			
20	0.4	12.8			

3D Reformat

Reformat from 1 mm axial reconstruction

Reformat from 2 mm axial reconstruction

Iterative Reconstruction

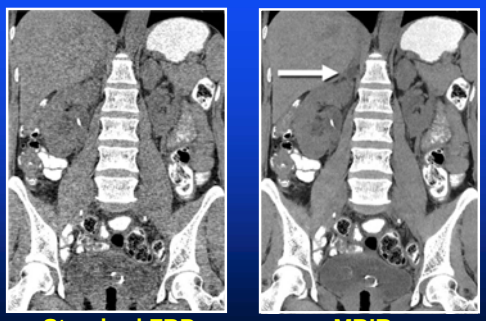
- Benefits over analytical reconstruction
  - Accurate physical models (imaging system geometry, scattering, spectrum, etc.)
    - less bias, improved quantitative accuracy, improved spatial resolution
  - Appropriate statistical models (FBP treats all data equally)
    - Lower image noise, lower radiation dose
  - Object constraints (e.g., non-negativity)
  - Flexible to handle non-standard scanning configurations
  - Potential to reconstruct images from sparse or incomplete data

Vendor implementations of Iterative Reconstruction or noise reduction methods

- GE: ASIR (adaptive statistical iterative reconstruction), MBIR (model based iterative reconstruction)
- Phillips: iDose
- Siemens: IRIS (iterative reconstruction in image space), SAFIRE (sinogram affirmed iterative reconstruction)
- Toshiba: AIDR3 (adaptive iterative dose reduction)

Source - SPIE Medical Imaging Symposium, 2011

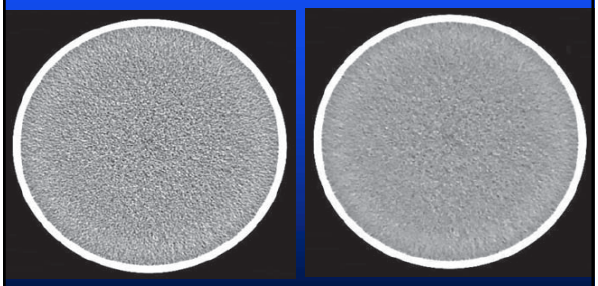
Example – Image quality improvement



Standard FBP MBIR

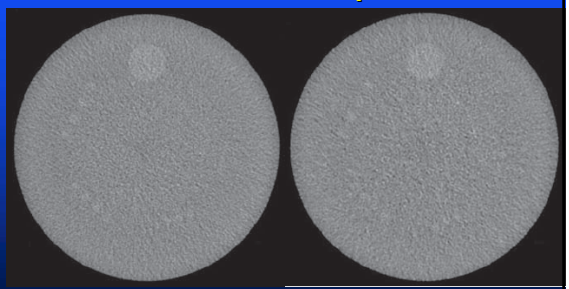
Images provided by Rendon Nelson, MD, Duke University, to AuntMinnie.com ("MBIR aims to outshine ASIR for sharpness, CT dose reduction," May 18, 2010)

Noise Texture Change



Hara, AK et al (2009). "Iterative reconstruction technique for reducing body radiation dose at CT: feasibility study." AJR Am J Roentgenol 193(3): 764-71.

Low-contrast Detectability 50% dose reduction possible?



FBP, 25 mGy ASIR, 12.5 mGy

Hara, AK et al (2009). "Iterative reconstruction technique for reducing body radiation dose at CT: feasibility study." AJR Am J Roentgenol 193(3): 764-71.

Special Considerations in Iterative Reconstruction

- Potential for dose reduction ~ 20% - 50% - task dependent
  - High-contrast allows more
  - Low-contrast allows less or none
- Noise texture could change dramatically
- Non-linear regularization term leads to non-linear noise-resolution tradeoff
  - Spatial resolution is highly dependent on contrast level
- Could reduce noise substantially without sacrificing high-contrast spatial resolution (measured in MTF)
- Low-contrast detectability needs to be evaluated more carefully.
- Quantitative evaluation and optimization of iterative reconstruction is still an active research field

Summary

- Implementations of reconstruction algorithms on clinical scanners
- How data acquisition parameters affect image quality
- How reconstruction parameters affect image quality
- Understanding of these is essential for designing dose-efficient and quality scanning protocols for various diagnostic tasks.

Thank you!

http://mayoresearch.mayo.edu/ctcic/