Flat panel CT: A new spin on CT

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Learning objectives

1. Understand the key differences between conventional CT and FPCT.
2. Understand the use of FPCT in the clinic and the role it plays.
3. Understand the operation of FPCT systems.
4. Understand methods for assessing dose and image quality at FPCT.

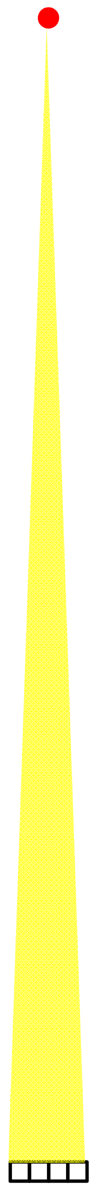
CBCT or FPCT?

- Cone beam CT (CBCT) describes *any* CT system where the cone angle is large
  - E.g., Toshiba Aquilion 1 (320 slice, 160 mm)
- Flat panel CT (FPCT) describes a CT system that uses a flat panel detector
  - E.g., DynaCT (40 x 30 cm FP)
- FPCT is a subset of CBCT

SDD = 105 cm

4 x 2.5 mm

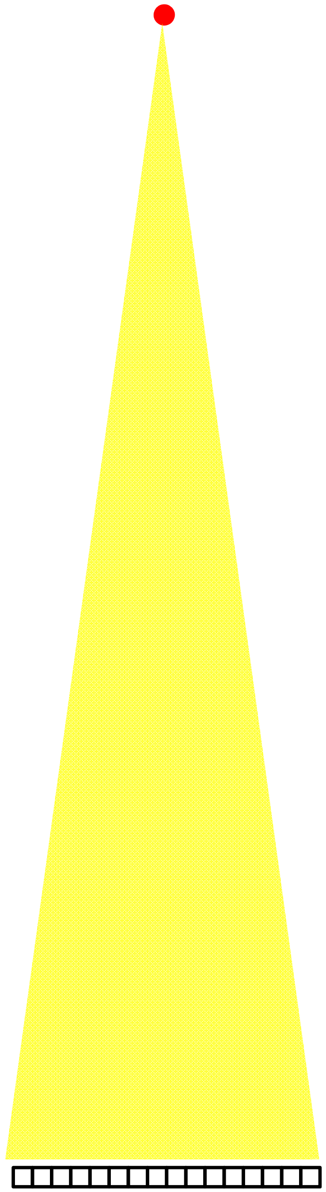
$\phi_{cone} \sim 0.6°$



SDD = 105 cm

16 x 1.25 mm

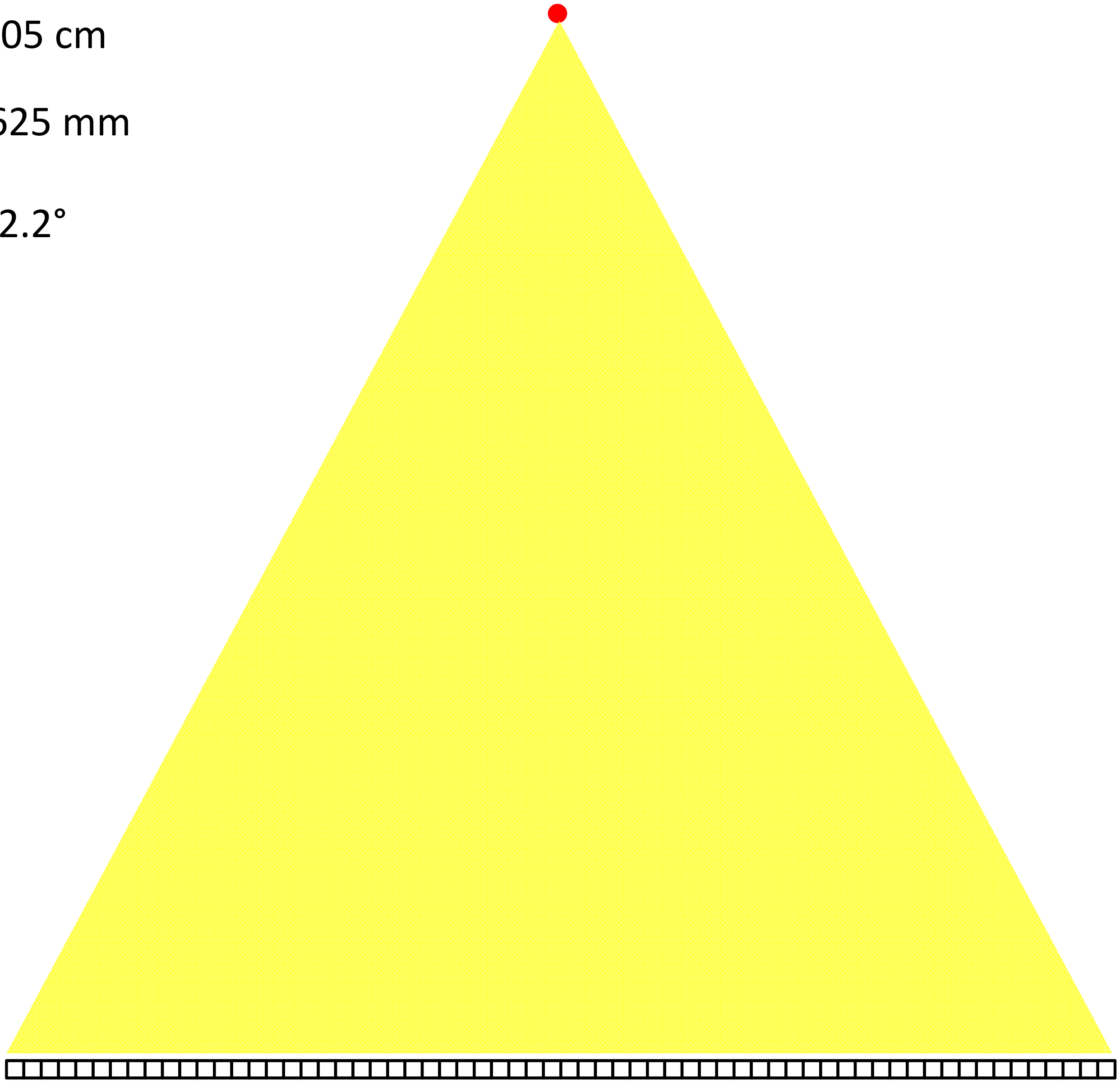
$\phi_{cone} \sim 1.1°$



SDD = 105 cm

64 x 0.625 mm

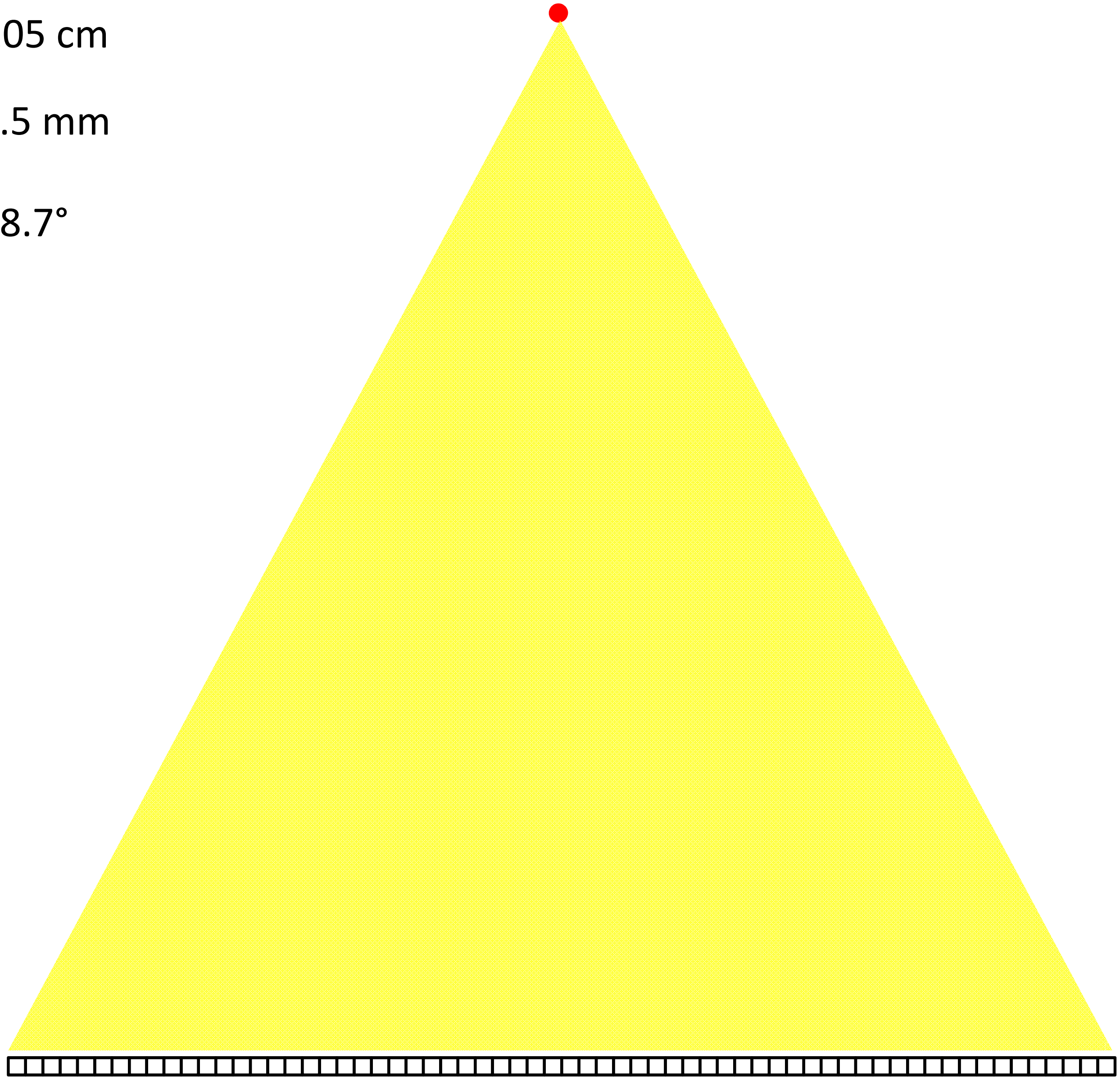
$\phi_{cone} \sim 2.2°$



SDD = 105 cm

320 x 0.5 mm

$\phi_{cone} \sim 8.7°$

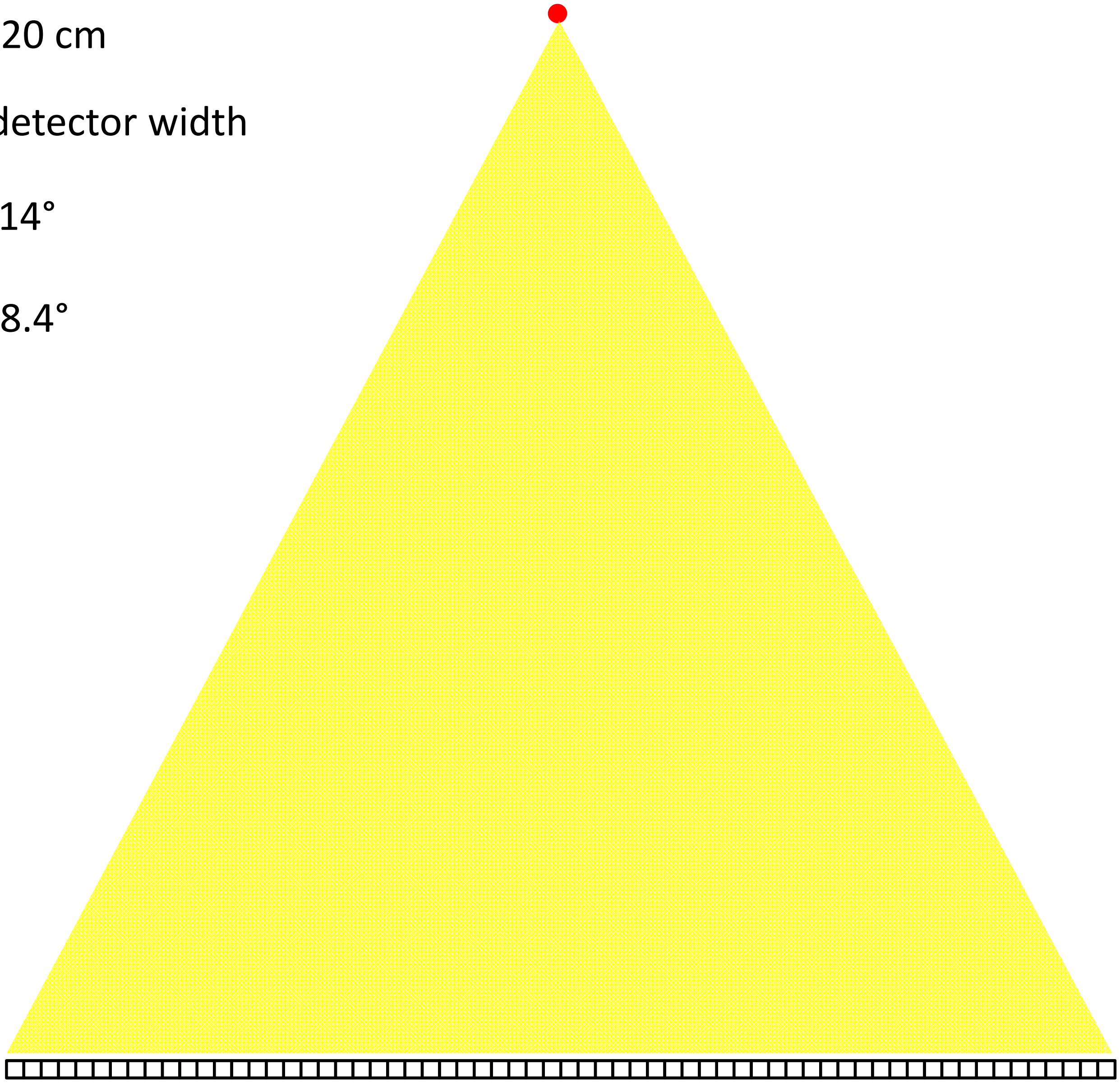


SDD = 120 cm

30 cm detector width

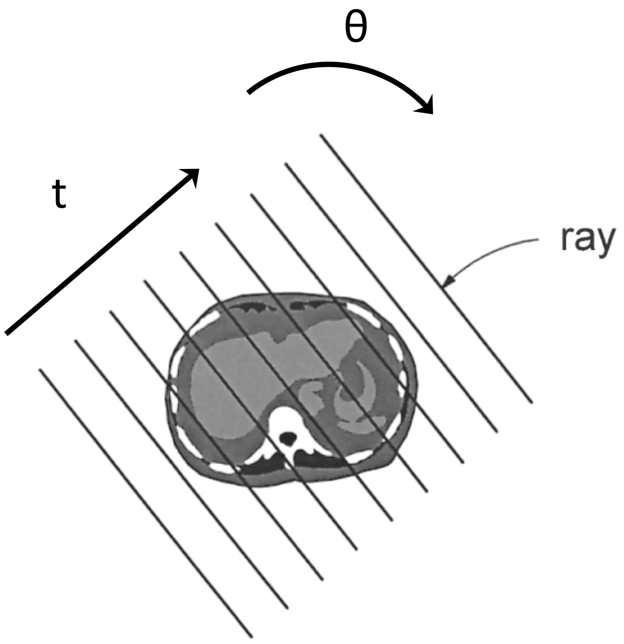
$\phi_{cone} \sim 14°$

$\phi_{fan} \sim 18.4°$



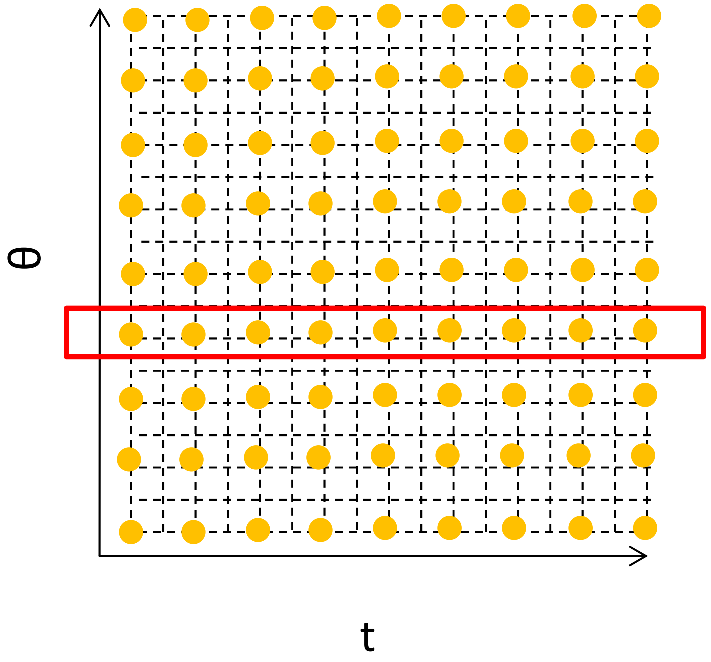
FUNDAMENTALS OF FPCT

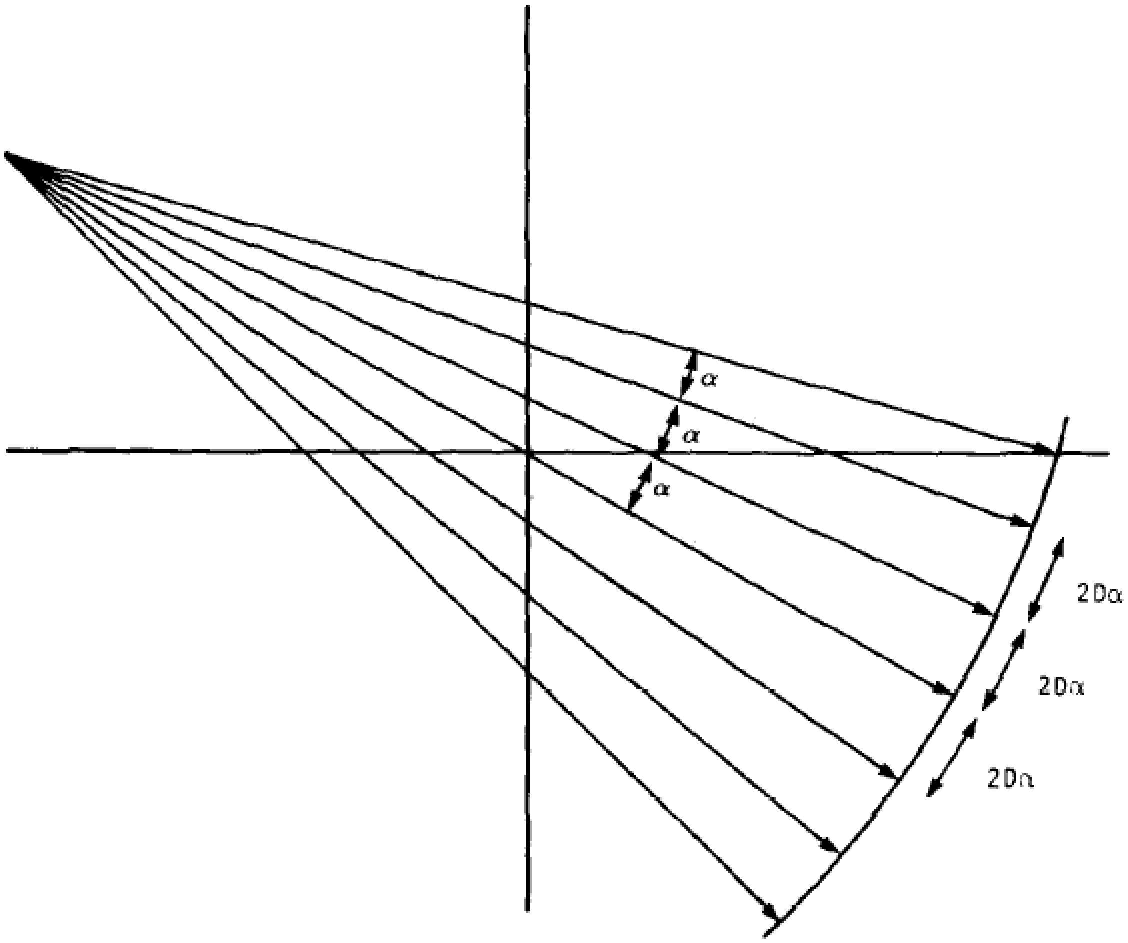
Reconstruction from projections



parallel beam projection

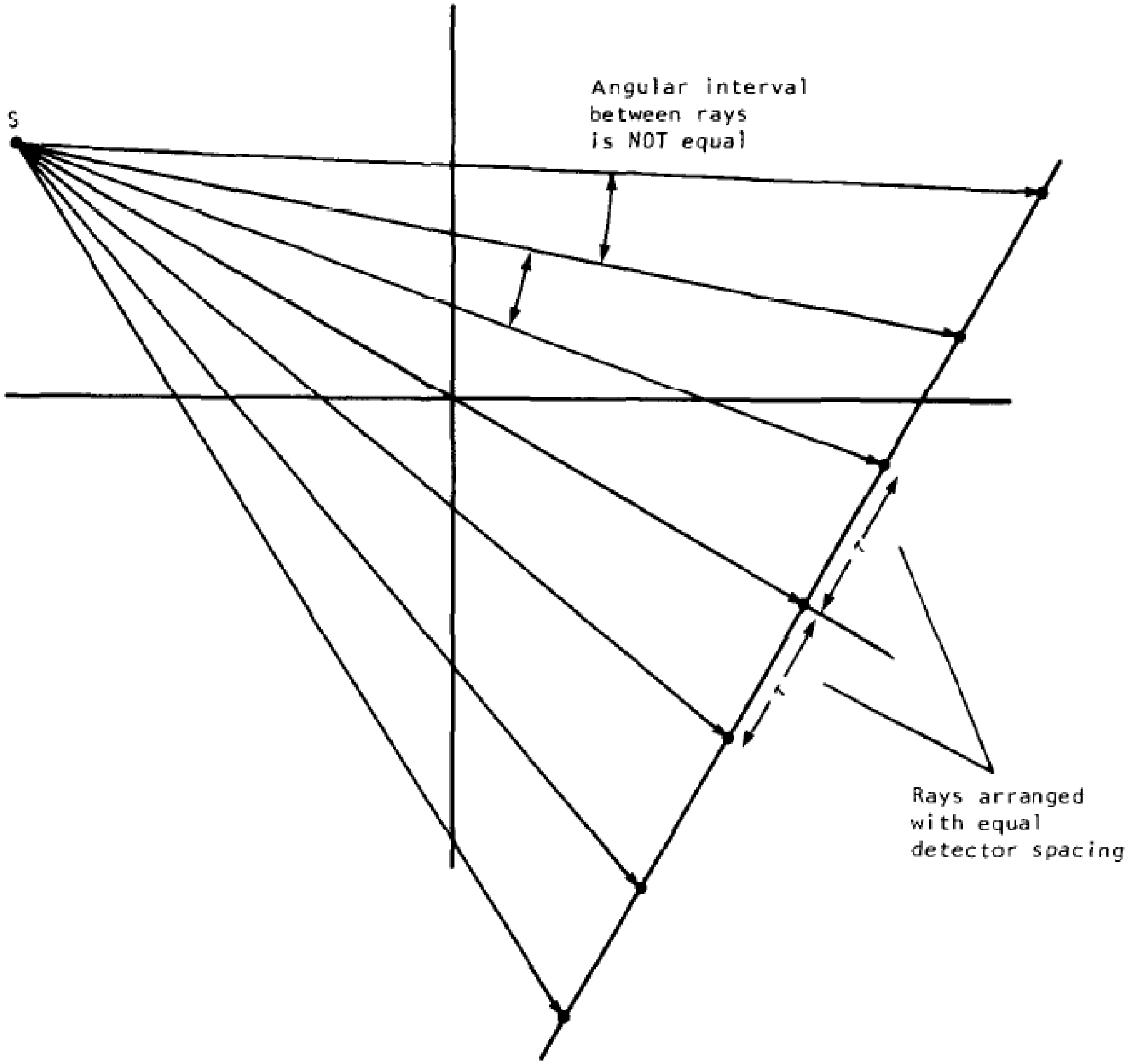
Bushberg, 2nd ed.





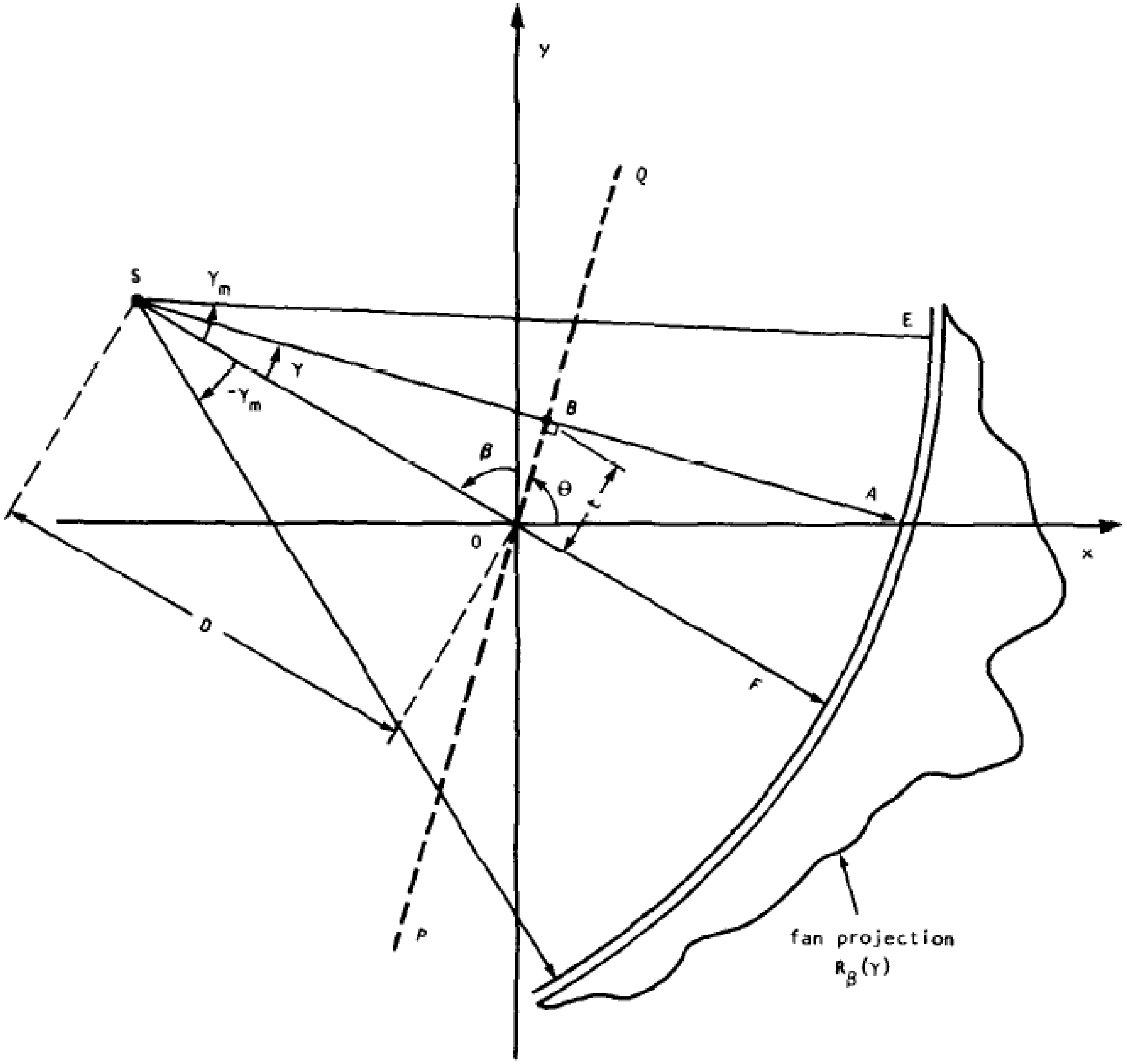
(b)

A. C. Kak and Malcolm Slaney, *Principles of Computerized Tomographic Imaging,* IEEE Press, 1988

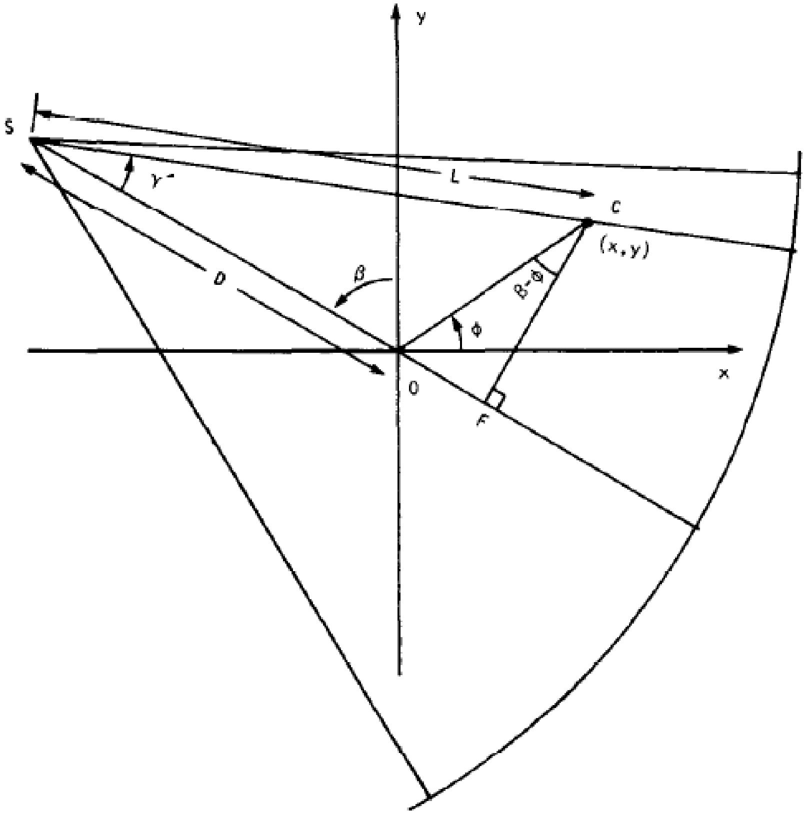


(c)

A. C. Kak and Malcolm Slaney, *Principles of Computerized Tomographic Imaging,* IEEE Press, 1988



A. C. Kak and Malcolm Slaney, *Principles of Computerized Tomographic Imaging,* IEEE Press, 1988



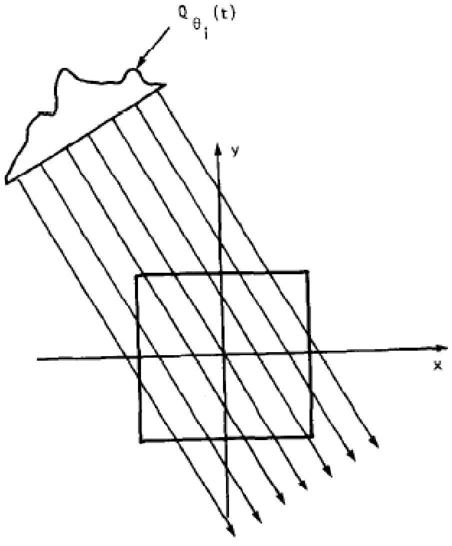
$$f(r, \phi) = \int_0^{2\pi} \frac{1}{L^2} \int_{-\gamma_m}^{\gamma_m} R_\beta(\gamma) g(\gamma' - \gamma) D \cos \gamma \, d\gamma \, d\beta$$

$$R'_\beta(\gamma) = R_\beta(\gamma) \cdot D \cdot \cos \gamma.$$

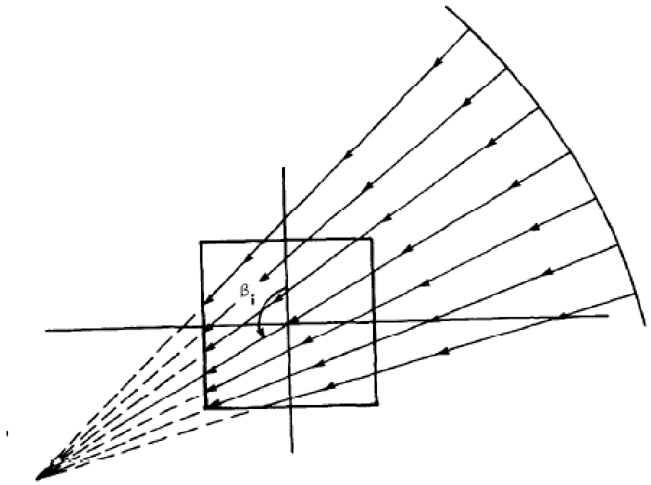
$$Q_\beta(\gamma) = R'_\beta(\gamma) * g(\gamma)$$

$$f(r, \phi) = \int_0^{2\pi} \frac{1}{L^2} Q_\beta(\gamma') \, d\beta$$

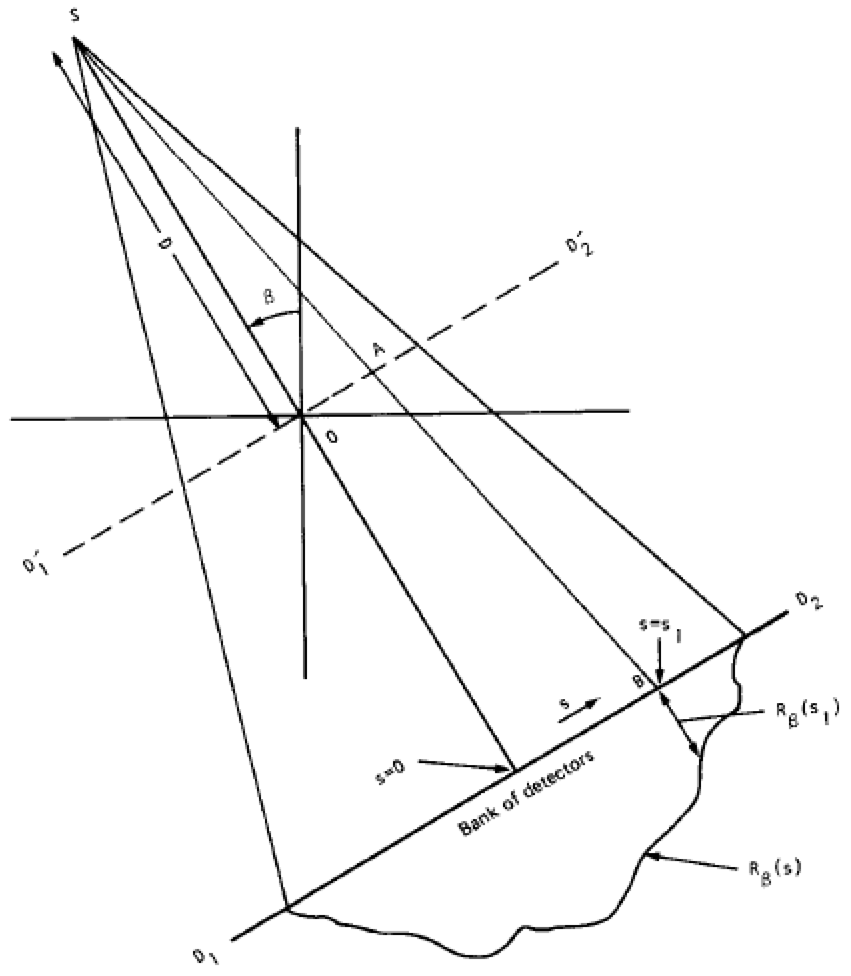
A. C. Kak and Malcolm Slaney, *Principles of Computerized Tomographic Imaging,* IEEE Press, 1988



(a)

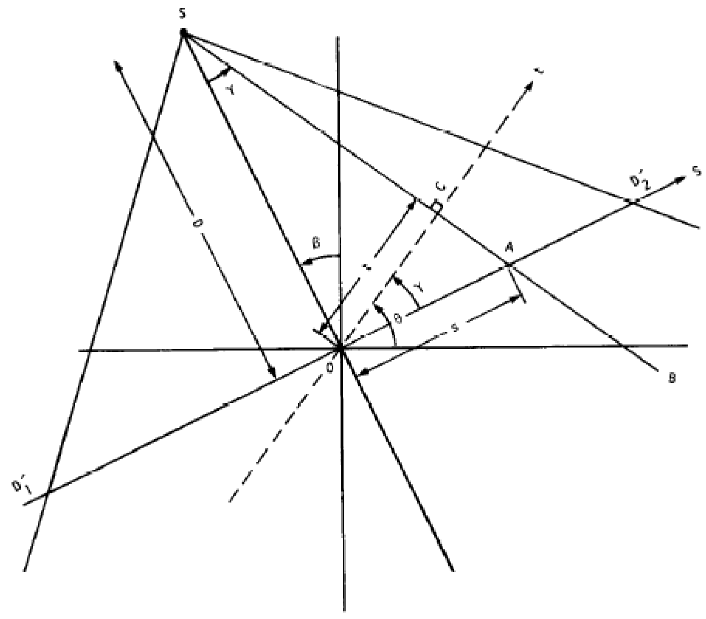


A. C. Kak and Malcolm Slaney, *Principles of Computerized Tomographic Imaging,* IEEE Press, 1988



A. C. Kak and Malcolm Slaney, *Principles of Computerized Tomographic Imaging,* IEEE Press, 1988

$$f(r,\phi)=\int_0^{2\pi}\frac{1}{U^2}\int_{-\infty}^{\infty}R_\beta(s)g(s'-s)\,\frac{D}{\sqrt{D^2+s^2}}\,ds\,d\beta$$



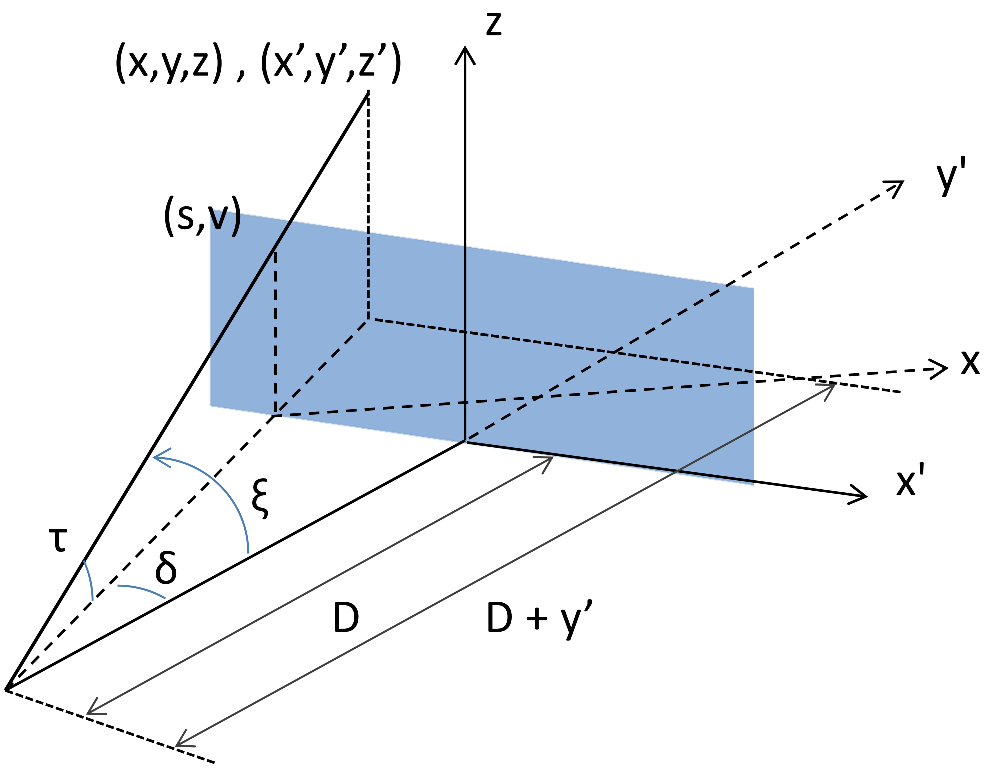
$$R'_\beta(s)=R_\beta(s)\cdot\frac{D}{\sqrt{D^2+s^2}}\,.$$

$$Q_\beta(s)=R'_\beta(s)*g(s)$$

$$f(r,\phi)=\int_0^{2\pi}\frac{1}{U^2}\,Q_\beta(s')\,d\beta$$

A. C. Kak and Malcolm Slaney, *Principles of Computerized Tomographic Imaging,* IEEE Press, 1988

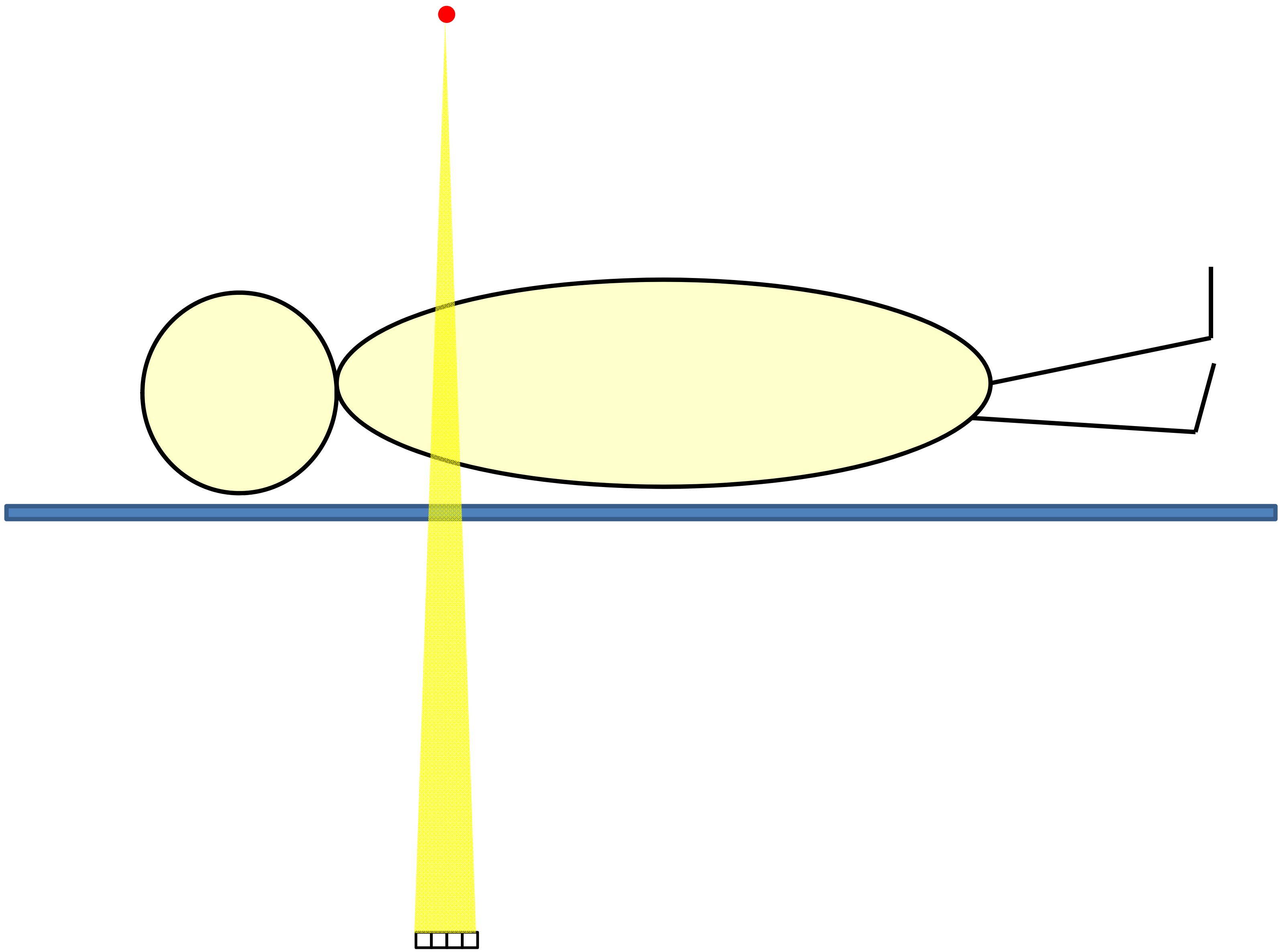
Cone beam reconstruction

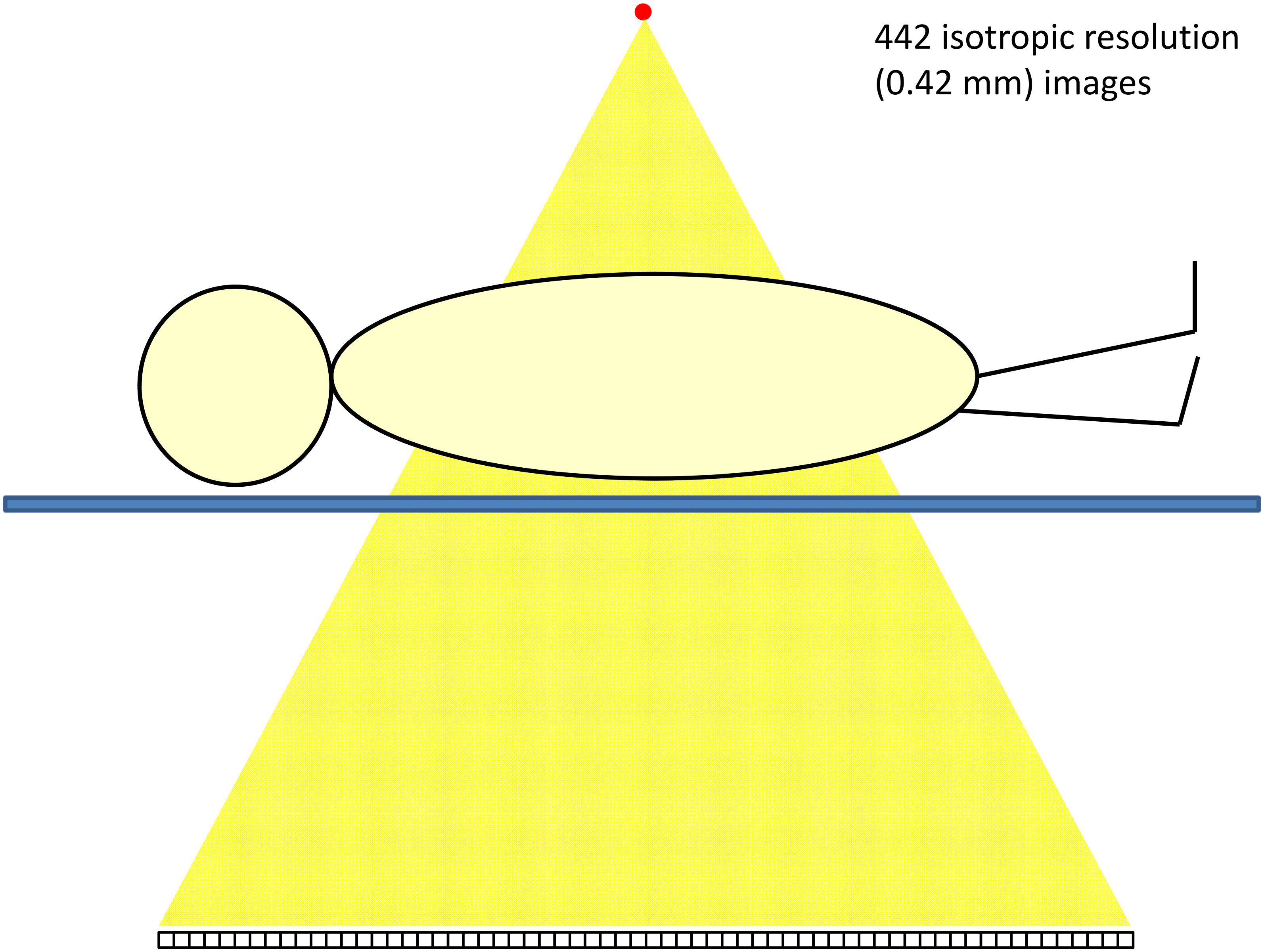


Projection samples scaled by cos(ξ)

Projection samples weighted by

$$\left(\frac{D}{D+yx'}\right)^2$$





442 isotropic resolution
(0.42 mm) images

Current implementations of FDCT

- O-arm system (Medtronic)
- iCAT dental system (ISI)
- IR lab
  - DynaCT (Siemens)
  - Allura 3D-RA (Philips)
  - Allura XperSwing (Philips)
  - Innova 3D/CT (GE)
  - 3D-Angio (Shimadzu)
- Dedicated breast CT (Koning Corporation)
- Vision FD Vario 3D (Ziehm)
- Varian OBI, etc

Key differences between FPCT and CT

- Conventional dose measurement fails
- Certain characteristics of FPCT images differ from CT
  - Differences in image reconstruction process
- Acquisition of basis (projection) images differs from CT
  - Technique factors may change during the scan
    - Old CT: constant kVp, mA, constant time
    - New CT: constant kVp (or 2), Δ mA, constant time
    - FPCT: Δ kVp, Δ mA, constant time

CT reconstruction

Projection data measured

Conditioning/ projection space corrections

-ln

Filter

Backproject

Image space corrections

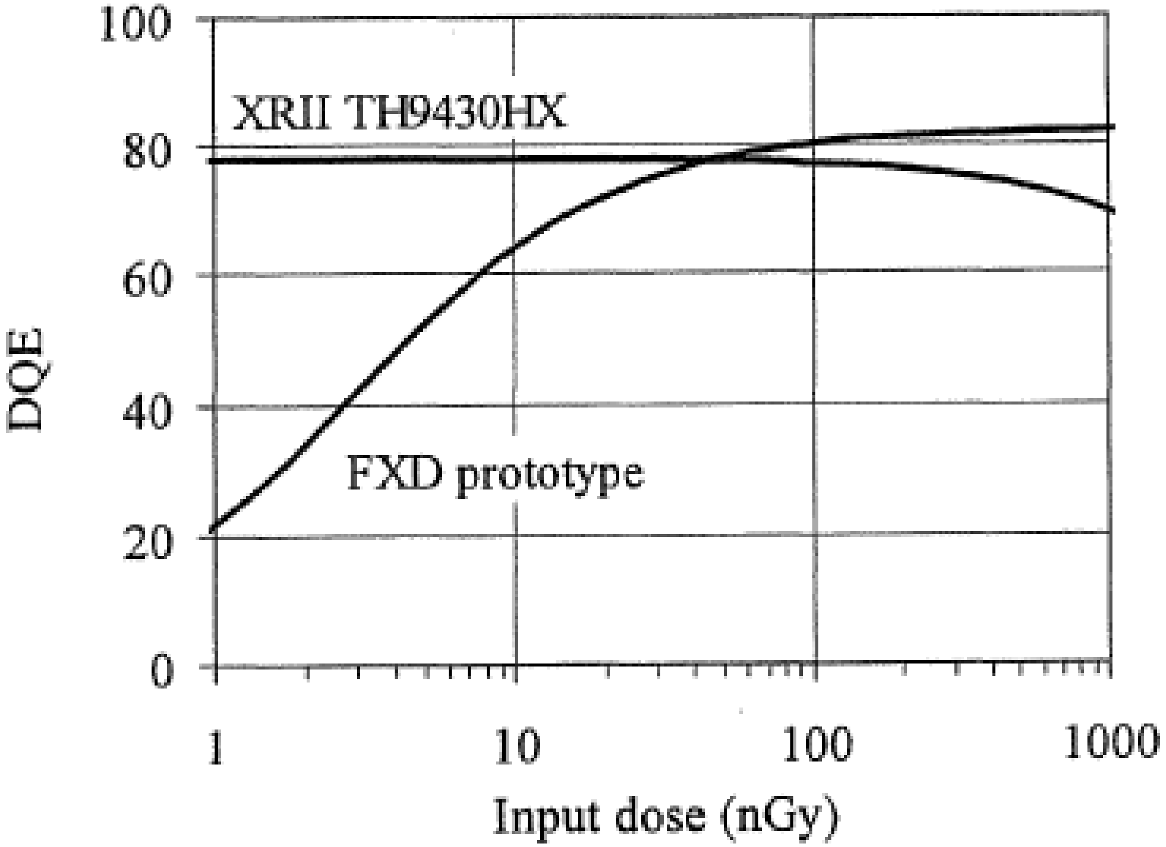
CT# cal.

Technological challenges in CBCT

- Reconstruction implementation
  - Cone beam reconstruction is exact only in plane of rotation
  - Artifacts in the axial direction (blurring)
- Data acquisition and processing
  - Many channels simultaneously
    - Increase in the number of channels in the z direction

Technological challenges in FPCT

- All of the above
- Detector readout/data acquisition and processing
  - CT scanner: 1344 x 64 channels, 1160 projections
    - 86016 samples per projection, 99.8 M samples per rotation
  - FP: 3 MP, 480 projections
    - 3 M samples per projection, 1.44 B samples per rotation
    - In 8 sec
    - Binning: 2x2, reduces data by 4x (0.75 M samples per projection, 360 M samples per rotation)
    - Binning: 4x4, reduces data by 16x (0.1875 M samples per projection, 90 M samples per projection)
  - Readout noise at low input exposures



Koch et al. Detective quantum efficiency of an x-ray image intensifier chain as a benchmark for amorphous silicon flat panel detectors, Proc SPIE Med Imag 2(25):115—120, 2001.

Technological challenges to FPCT

- Glass substrate is fragile and limits rotation speed
- C-arms flex and therefore calibration is extremely important and must be performed for many different configurations

Advantages of FPCT

- Surgical guidance
  - In conjunction with RF or other guidance systems
- CT-like images immediately available in IR lab
  - Bleeds, embolization, etc.
  - Significant impact
- Inexpensive alternative to conventional CT
  - No need for room modification
  - Small footprint
  - RO – can be installed on the front of a linac gantry
- Dedicated breast CT
  - Axial images, improved image quality?
- MicroCT/small animal imaging
- Includes radiography/fluoroscopy capability

TECHNICAL ASPECTS OF FPCT

Operating modes

- Depending on the manufacturer and the use, FPCT systems operate differently
  - iCAT: fixed kVp, fixed mA, dose level selected
  - O-arm: fixed kVp, fixed mA, dose level selected
  - DynaCT: selected kVp, variable mA, selected dose level, selected filtration
  - Varian: fixed kVp, fixed mA, fixed pulse width (based on calibration)
- These operating modes will impact the final image quality and patient dose

AEC ~ TCM

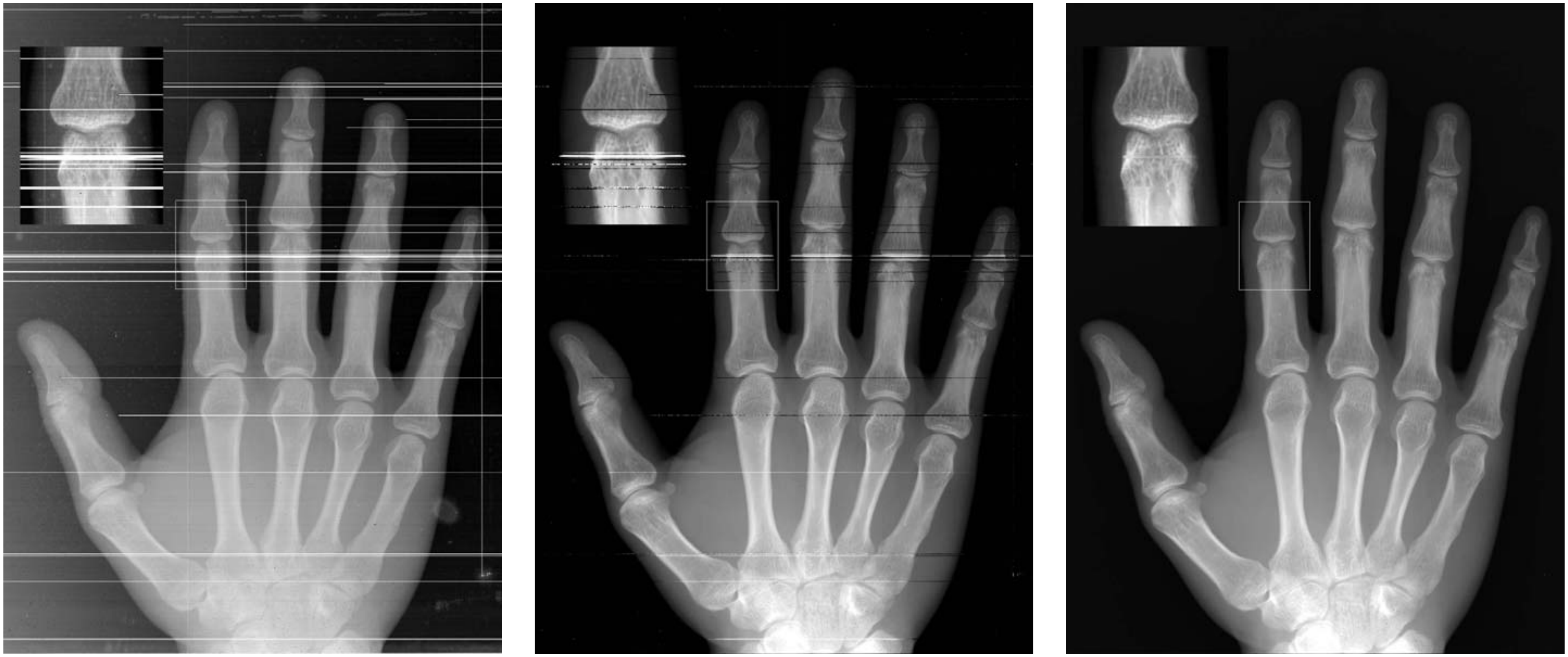
- Modern fluoroscopic systems operate under automatic exposure control
- Technical parameters are varied in a set sequence to maintain IAKRD
  - mA, pw, kVp, Cu, repeat
- Some FPCTs operate under the same principle
- This is similar to tube current modulation
  - Only mA varied in TCM
  - Similar advantage in that input dose/view should remain constant, no noisy views
  - Now conventional CT has CarekV…

Bowtie filter

- One or more bowtie filters is used in conventional CT scanners to flatten the x-ray fluence at the detector array
  - Improves image quality
    - Beam hardening reduction, scatter reduction, flatness
  - Reduces patient dose
    - Especially at periphery of patient
- Bowtie filters may not be used, or may be removable, in FPCT systems
  - Varian OBI uses removable bowtie filter
  - DynaCT uses no bowtie filter
- Potential drawbacks include
  - Need for multiple filters based on scan mode
  - Potential negative impact on dose and IQ if patient not centered
  - Artifacts from slight misalignments of bowtie filter

FPCT calibration – receptor

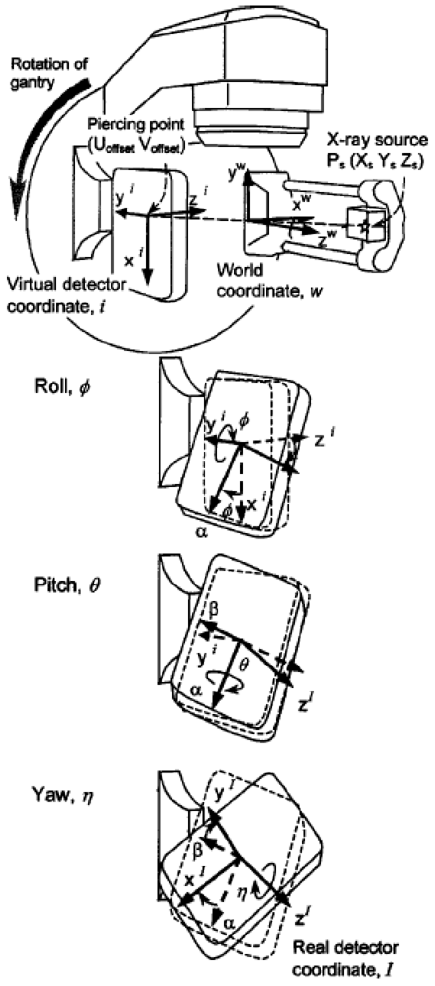
- Each pixel in a FP detector is characterized by its own sensitivity and dark current
  - Gain and offset calibration
  - Same in CT
- For FPCT applications, small gain variations in hundreds of projections will cause artifacts
  - Calibration performed at *each* available dose level
  - Ring corrections also applied
- Bad pixel correction also applied to raw readout from detector – separate issue



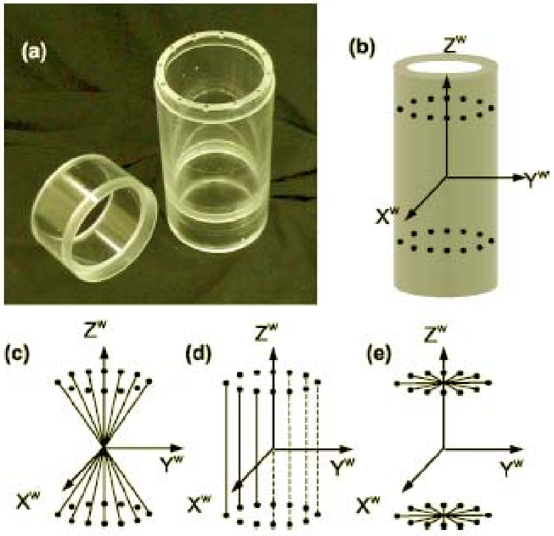
Images courtesy of Ian Yorkston, Ph.D.

FPCT calibration – geometric

- CT reconstruction algorithms need to know the scan geometry *a priori*
  - Rigidly fixed in conventional CT
  - Remember recon. algorithm
- C-arms and linacs, by necessity, have many degrees of freedom of movement
- And the C-arm can flex
- All of these movements must be accounted for and corrected for in the reconstruction



Cho Y, et al. Accurate technique for complete geometric calibration of cone-beam computed tomography systems. Med Phys 32(4): 968-983, 2005.



Cho Y, et al. Accurate technique for complete geometric calibration of cone-beam computed tomography systems. Med Phys 32(4): 968-983, 2005.

Daly MJ, et al. Geometric calibration of a mobile C-arm for intraoperative cone-beam CT . Med Phys 35(5):2124-2136, 2008.

G.H. Chen, AAPM 2009

http://www.aapm.org/meetings/amos2/pdf/42-11961-89149-125.pdf

G.H. Chen, AAPM 2009

http://www.aapm.org/meetings/amos2/pdf/42-11961-89149-125.pdf

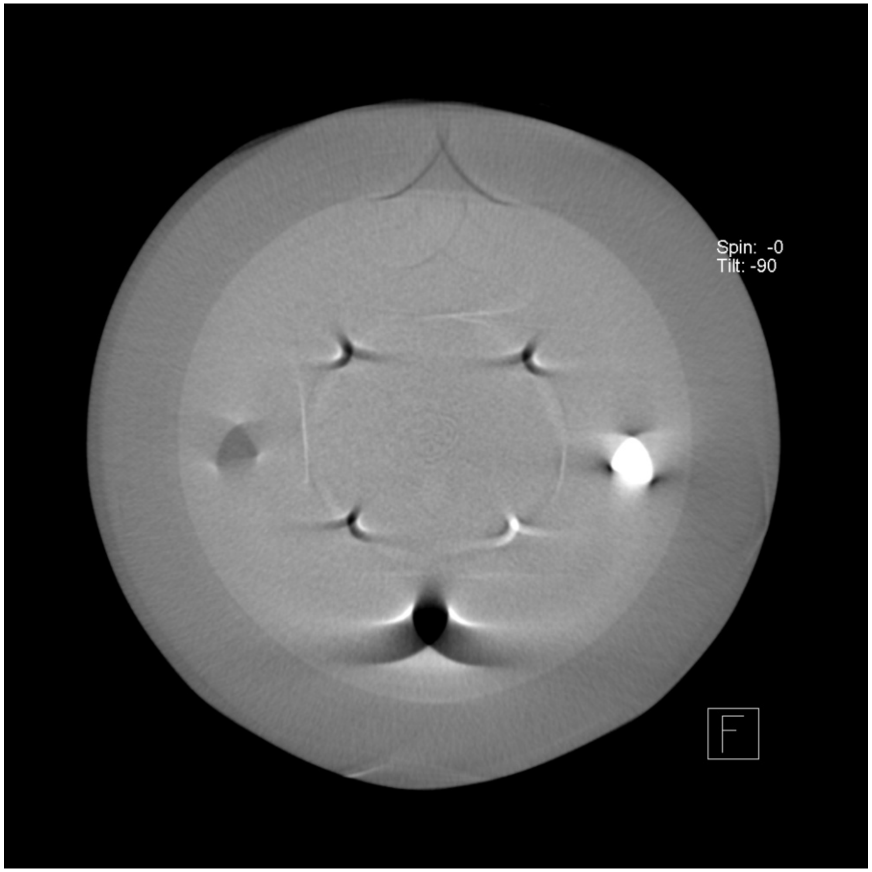
Other approaches

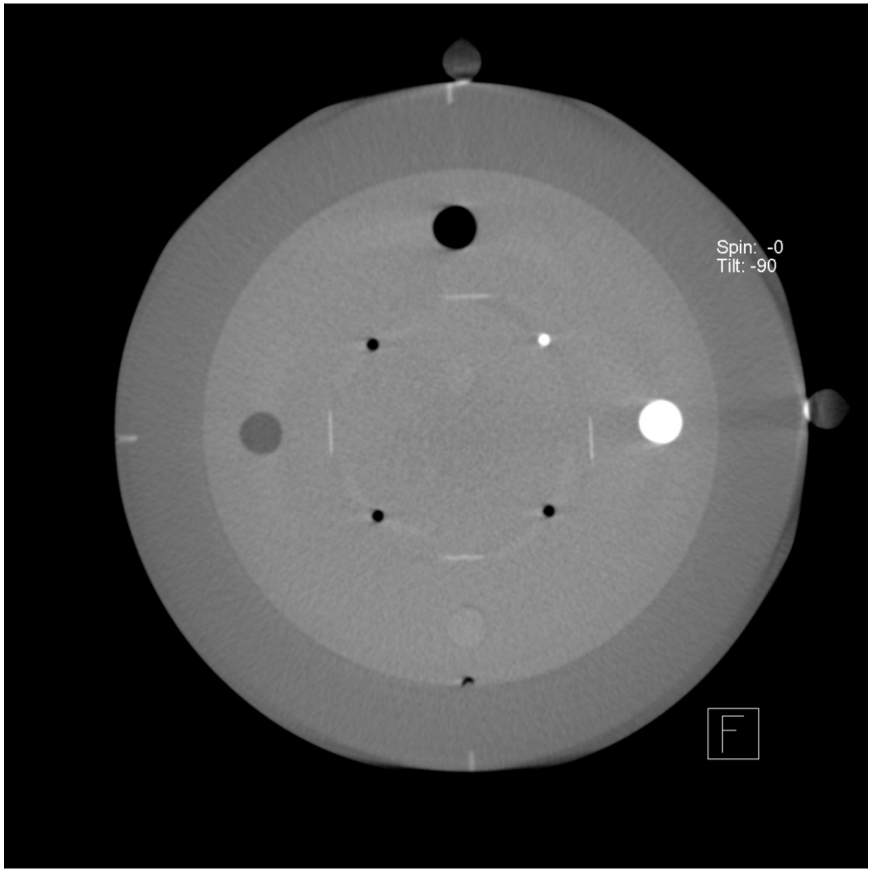
- Varian Truebeam incorporates Isocal which maps the flex of the gantry and arms as a function of rotation angle
  - MV vs kV vs treatment iso offset
  - FPCT calibration

Impact of calibration sensitivity

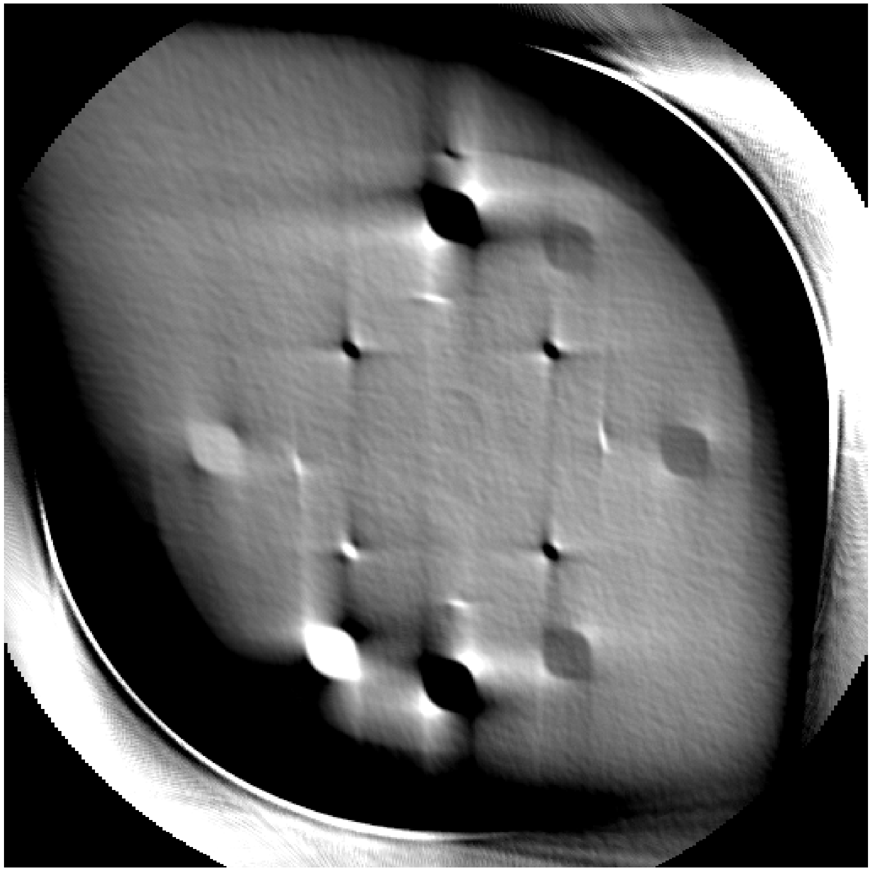
- 3D no longer supported on XRII systems
- Extensive series of calibrations may need to be performed
  - Can mean lots of downtime after component replacement
- Modes that are not calibrated will not be available for reconstruction
- Scan errors (e.g. wrong start position) will result in significant artifacts
- Each combination of factors should be evaluated for artifacts
  - Minimize number of protocols used

Wrong orientation of FP during DynaCT calibration





angles differs from calibration



For example…

- Siemens Axiom Artis DynaCT (ceiling mounted)
- Available dose settings: 0.1, 0.12, 0.14, 0.17, 0.2, 0.24, 0.36, 0.54, 0.81, 1.2, 1.82, 2.4, 3.6, 5.4 μGy/frame
- Available positions: HS, LS, RS
- Available scan times: 2 s – 25 s
- Angular sampling: 0.4, 0.5, 0.6, 0.8, 1, 1.2, 1.5, 2, 2.5, 3, 4.5, 6, 7.5, 12, 15
- Total # of calibrations for all modes: 15,120

DOSE IN FPCT

Back of the envelope

FPCT (clinical protocol)

- 18.7 cm of coverage along z-axis
- 90 kVp (target)
- 0.36 µGy/frame
- $CTDI_{vol}$ (integral) = 10.6 mGy

- One phantom: 942 mR (C)
- Two phantoms: 977.4 mR (C) (-4%)

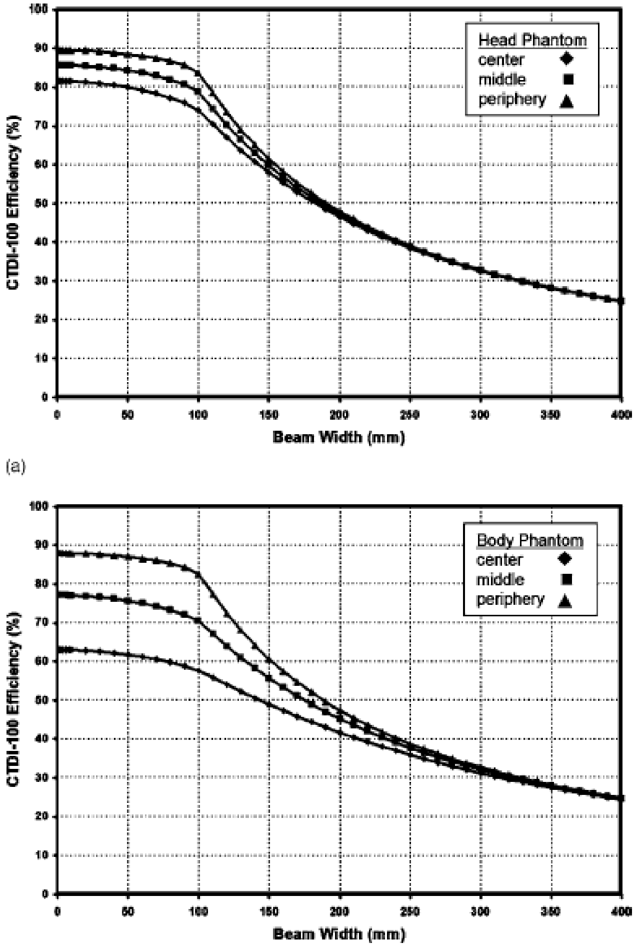
Conventional CT

- Scan length = 18.7 cm
- 120 kV, 100 mAs
- $CTDI_{vol}$ (integral) = 16.4 mGy

16 cm CTDI phantoms

Conventional dose metrics fail

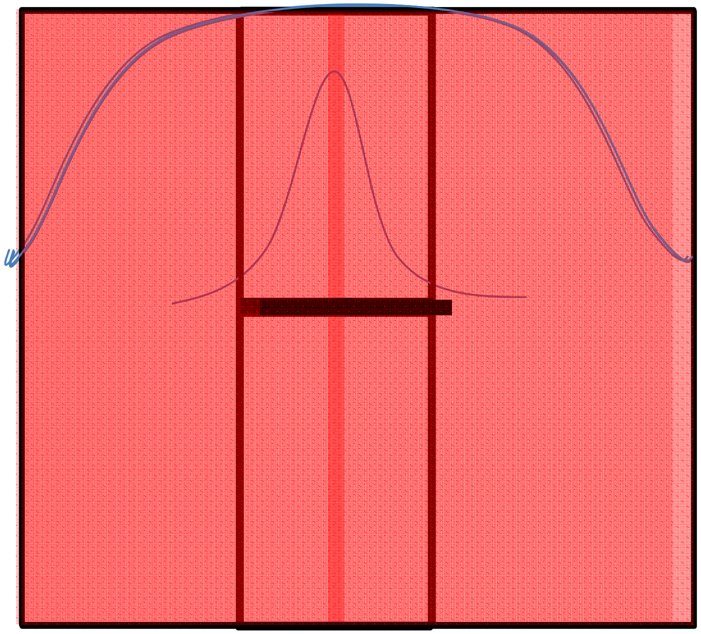
- What are we measuring with CTDI_{100} ?
  - Collection efficiency is poor for wide beams
- In FPCT, beam widths in the z-direction can be as large as 190 mm (ϕ_{cone} = 14°)
  - Can't even measure all the primary radiation
  - With pencil chamber, dose is integrated over arbitrary length of 100 mm



Boone JM. The trouble with CTDI100. Med Phys 34(4):1364-1371, 2007.

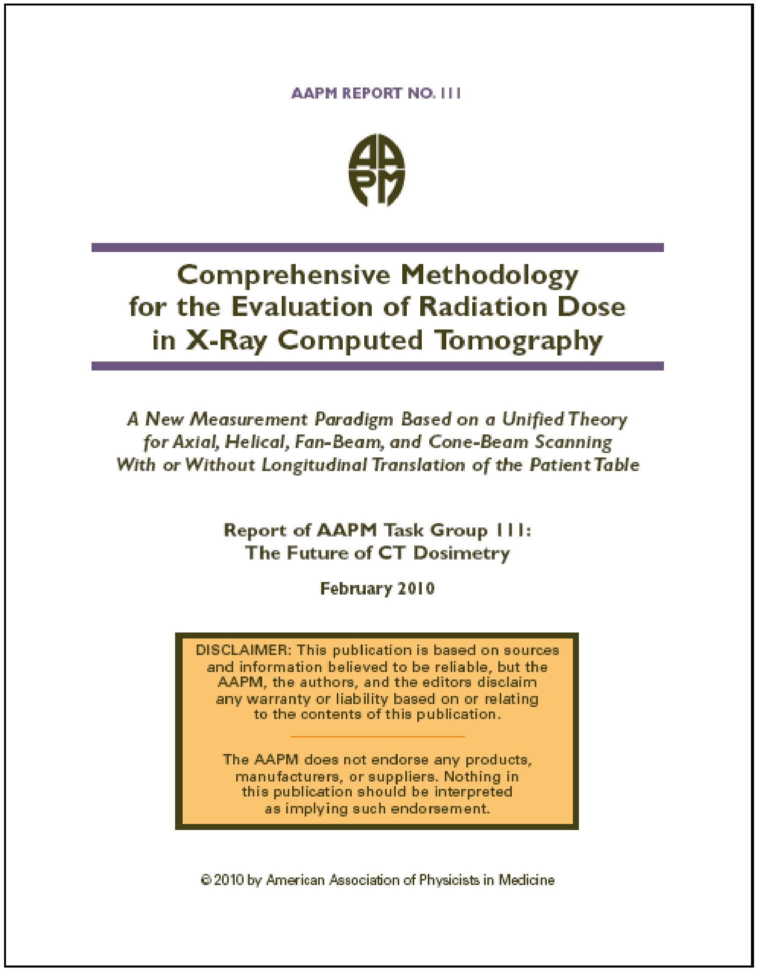
Measuring dose in FPCT

- Integrating the dose over an (arbitrary) 100 mm length is *not* an analog of CTDI
- Measuring the (point) dose at the center of the scan volume *is* an analog
  - Similar situation for helical CT scanning
  - MSAD



Changes to CT dosimetry

- Changes are happening on several fronts
  - TG 111
  - ICRU
  - Literature
- Fahrig and Dixon have proposed a formalism for dose in FPCT



AAPM REPORT NO. 111

Comprehensive Methodology for the Evaluation of Radiation Dose in X-Ray Computed Tomography

A New Measurement Paradigm Based on a Unified Theory for Axial, Helical, Fan-Beam, and Cone-Beam Scanning With or Without Longitudinal Translation of the Patient Table

Report of AAPM Task Group 111:
The Future of CT Dosimetry

February 2010

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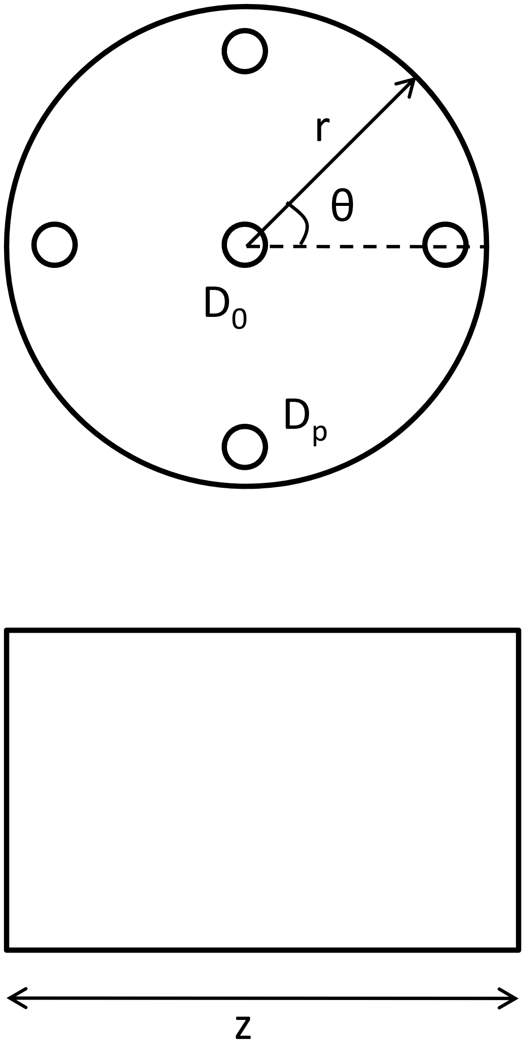
Fahrig-Dixon formalism

- We have a non-uniform, position-dependent dose distribution D(r,θ,z)
- We desire to measure the average dose at the center of the scan plane (CTDI/MSAD analog)
- Define

$$\boxed{\bar{D}(0) = \frac{1}{3} D_0 + \frac{2}{3} \bar{D}_p}$$ *



Measuring point doses

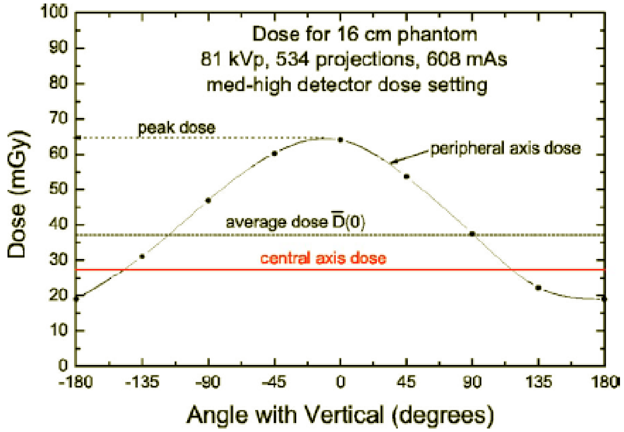
- Point doses can be measured using a variety of dosimeters
  - TLD
  - OSL
  - Ionization chamber
- The easiest solution is the use of a Farmer chamber

Fahrig-Dixon formalism

- As you can see, this is really very similar to measuring CTDI_{100}
  - Center + periphery measurements
  - Point measurement (0.6 cc not 3 cc/100 mm)
- If data are not collected over 360°, your results will vary based on the starting and ending positions of the scan

Dose distribution in C-arm FPCT

- Data acquisition range is limited by mechanical considerations
  - 180° + fan beam (~ 200° total) for LS + RS
  - 360° option for HS
- For LS + RS, peripheral dose will be highest at the midpoint of the scan
- Center dose usually unaffected by starting point

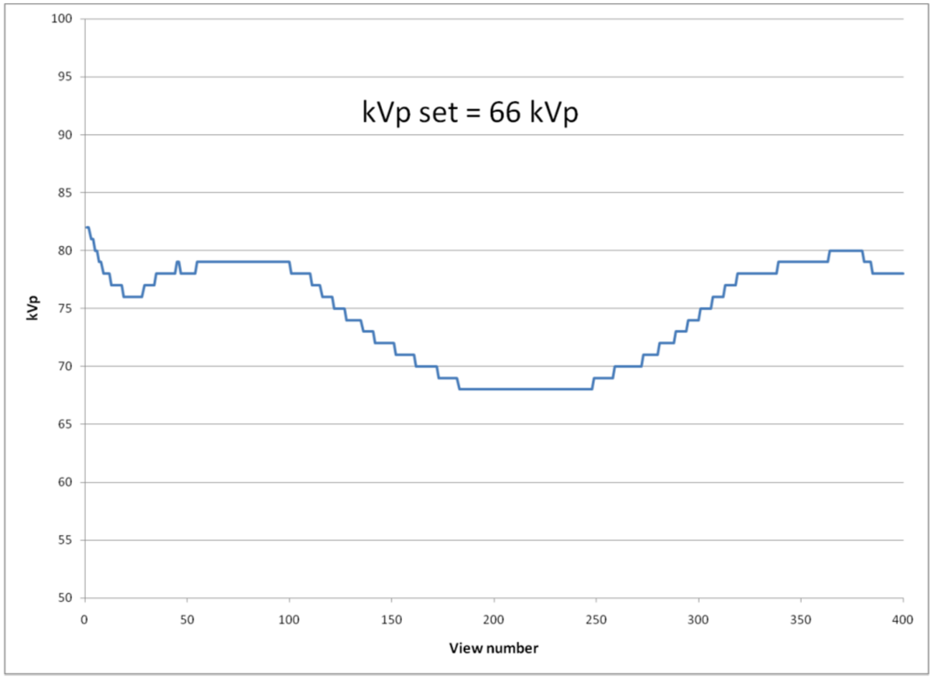


Fahrig R, et al. Dose and image quality for a cone-beam C-arm CT system. Med Phys 33(12):4541-4550, 2006.

Jeff Siewerdsen, AAPM 2009

Calculating patient or reference doses

- If your system operates under ADRC, corrections must be made to the proposed metric
  - kVp changes
- Jon Anderson et al. discussed this in their presentation at RSNA 2004



The Anderson formalism

$$\mathrm{mAs}_{\mathrm{equiv}}^{\mathrm{70\ kVp}} = \mathrm{mAs}_{\mathrm{frame}} \times \left(\frac{\mathrm{kVp}_{\mathrm{frame}}}{70} \right)^{\alpha}$$

$$\mathrm{D}_{\mathrm{scan\ X}} = \mathrm{D}_{\mathrm{phantom}}^{\mathrm{70\ kVp}} \times \frac{\sum_{\mathrm{all\ frames}} \mathrm{mAs}_{\mathrm{equiv}}^{\mathrm{70\ kVp}}}{\sum_{\mathrm{all\ frames}} \mathrm{mAs}_{\mathrm{scan\ X}}}$$

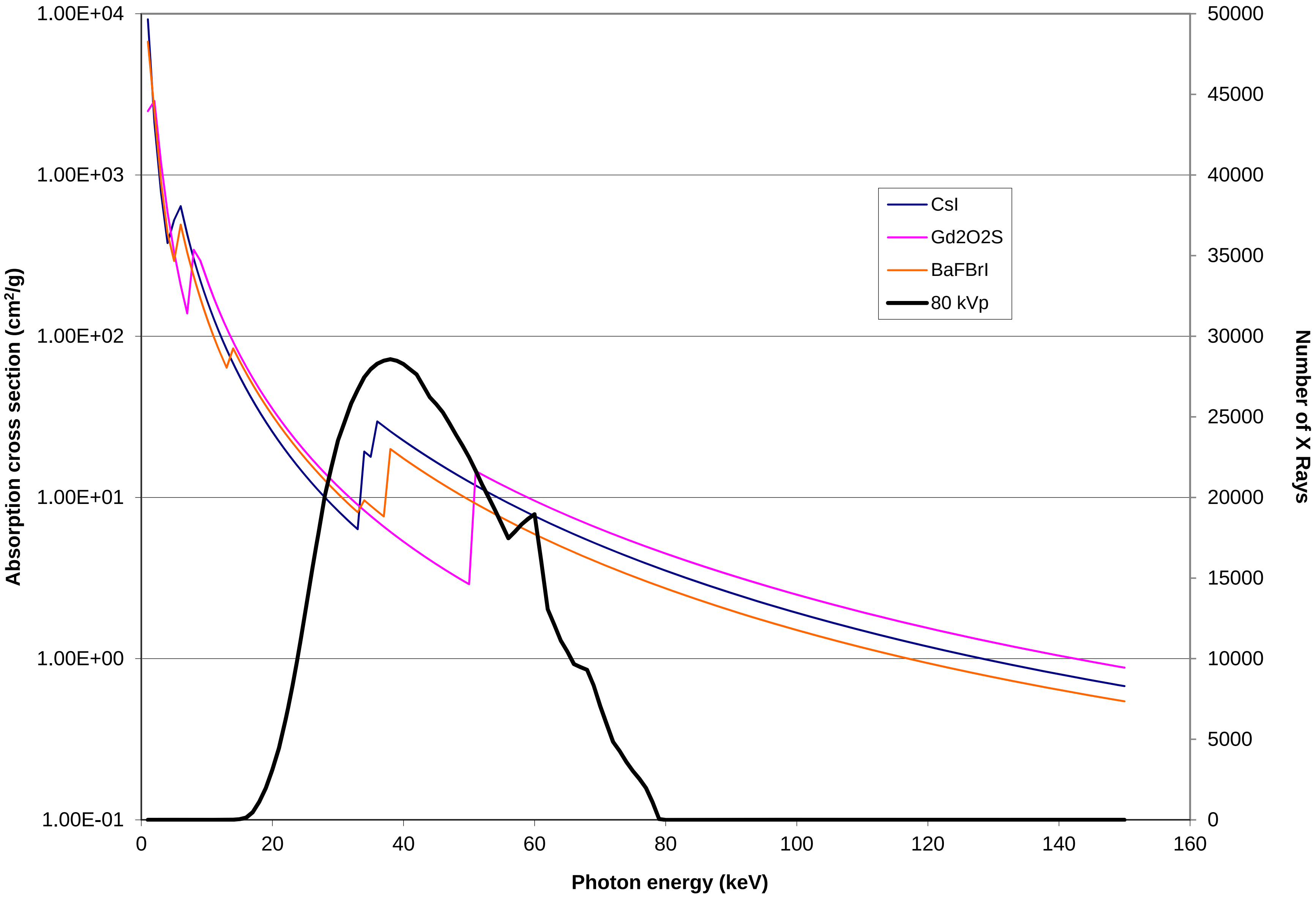
Jon Anderson et al RSNA 2004

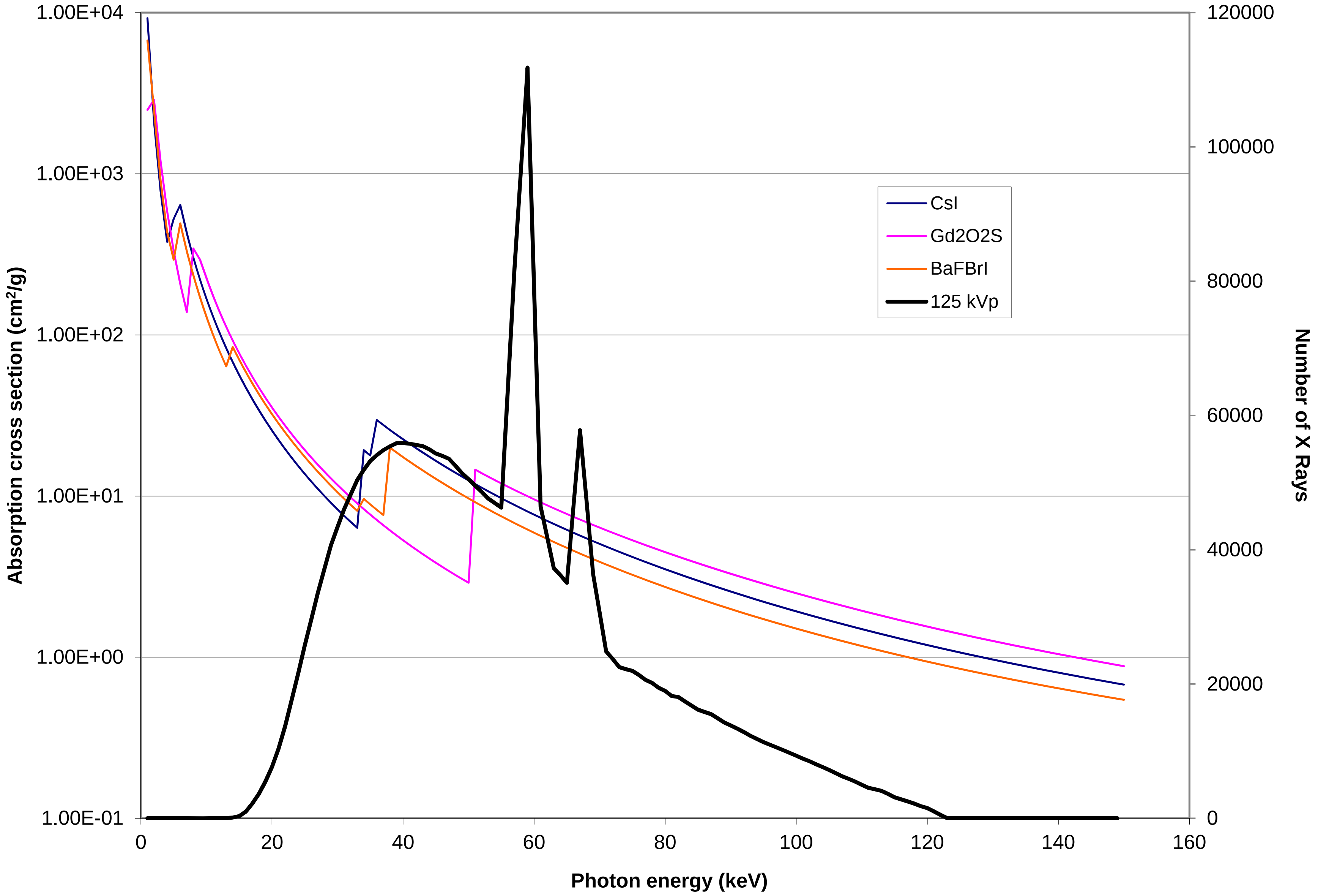
Measuring reference doses

- While dose for small body parts can be determined easily (16 cm CTDI phantom), the SFOV (fan angle) is too small to fully encompass the 32 cm CTDI phantom
  - Cutoff in peripheral measurements
  - Could use intermediate holes if available
  - Slight variations in phantom position could result in large measurement variation
  - Center measurement is reliable
  - What is the appropriate weighting function?
    - More precise = more measurements

Optimizing dose

- The optimal dose will be achieved by:
    1. Selecting the minimum dose per frame necessary for the imaging task
        - Rendering methods must be considered as well
    2. Knowing your equipment (read the manual!)
    3. Choosing the appropriate kVp setting based on the attenuation of the body part imaged*
    4. Choosing the appropriate angular sampling*
    5. Removing unnecessary body parts from the FOV

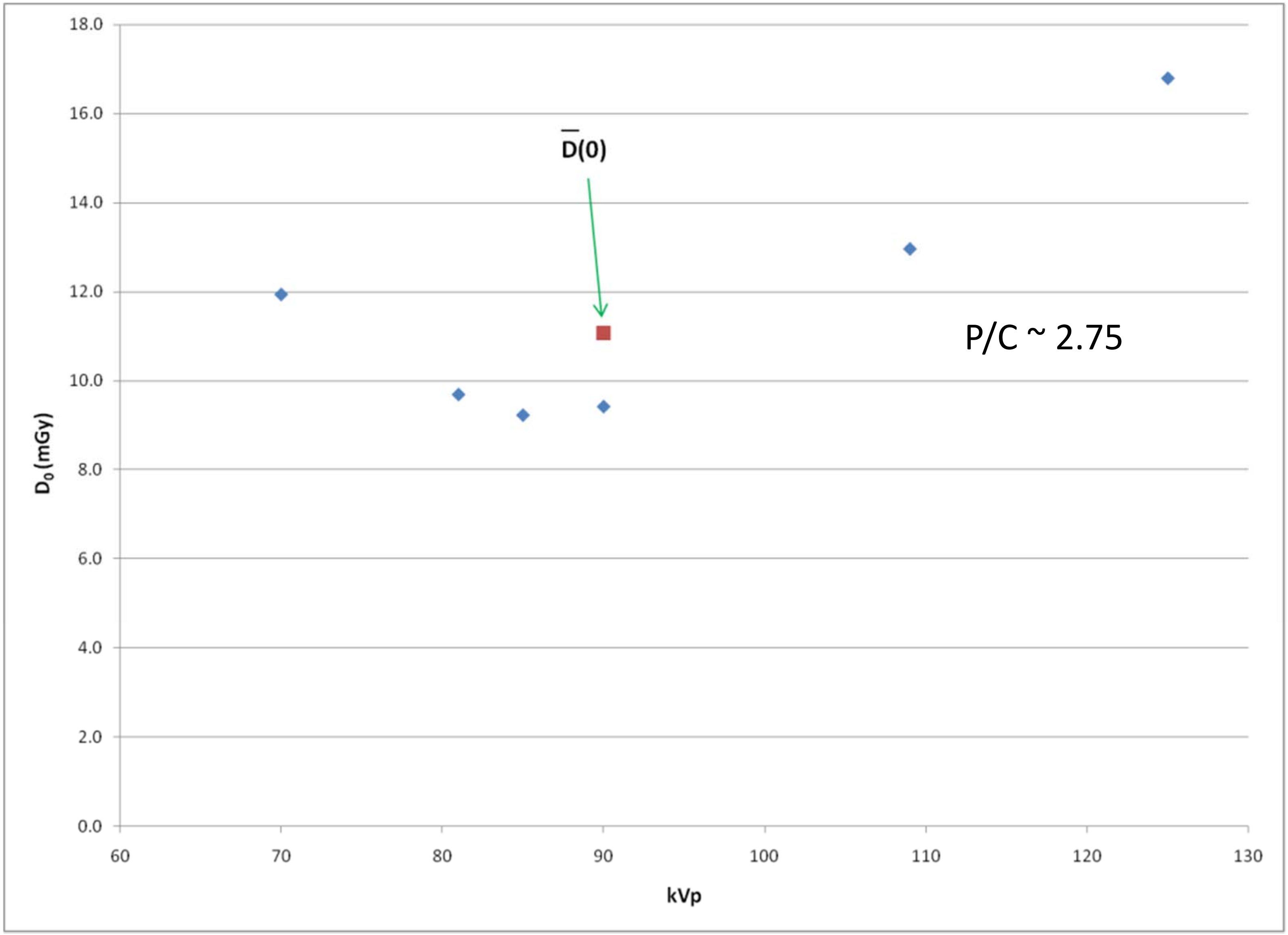

1.00E+04
1.00E+03
1.00E+02
1.00E+01
1.00E+00
1.00E-01
Absorption cross section (cm²/g)
0
20
40
60
80
100
120
140
160
Photon energy (keV)
50000
45000
40000
35000
30000
25000
20000
15000
10000
5000
0
Number of X Rays
CsI
Gd2O2S
BaFBrI
80 kVp


1.00E+04
1.00E+03
1.00E+02
1.00E+01
1.00E+00
1.00E-01
Absorption cross section (cm²/g)
0
20
40
60
80
100
120
140
160
Photon energy (keV)
120000
100000
80000
60000
40000
20000
0
Number of X Rays
CsI
Gd2O2S
BaFBrI
125 kVp

Head dose vs. kVp setting

- The operator can select the desired input dose per frame in the configuration settings
- Siemens, in the settings for DynaCT, also allows the operator to select the *target* kVp value
  - May or may not be maintained based on attenuation by the patient

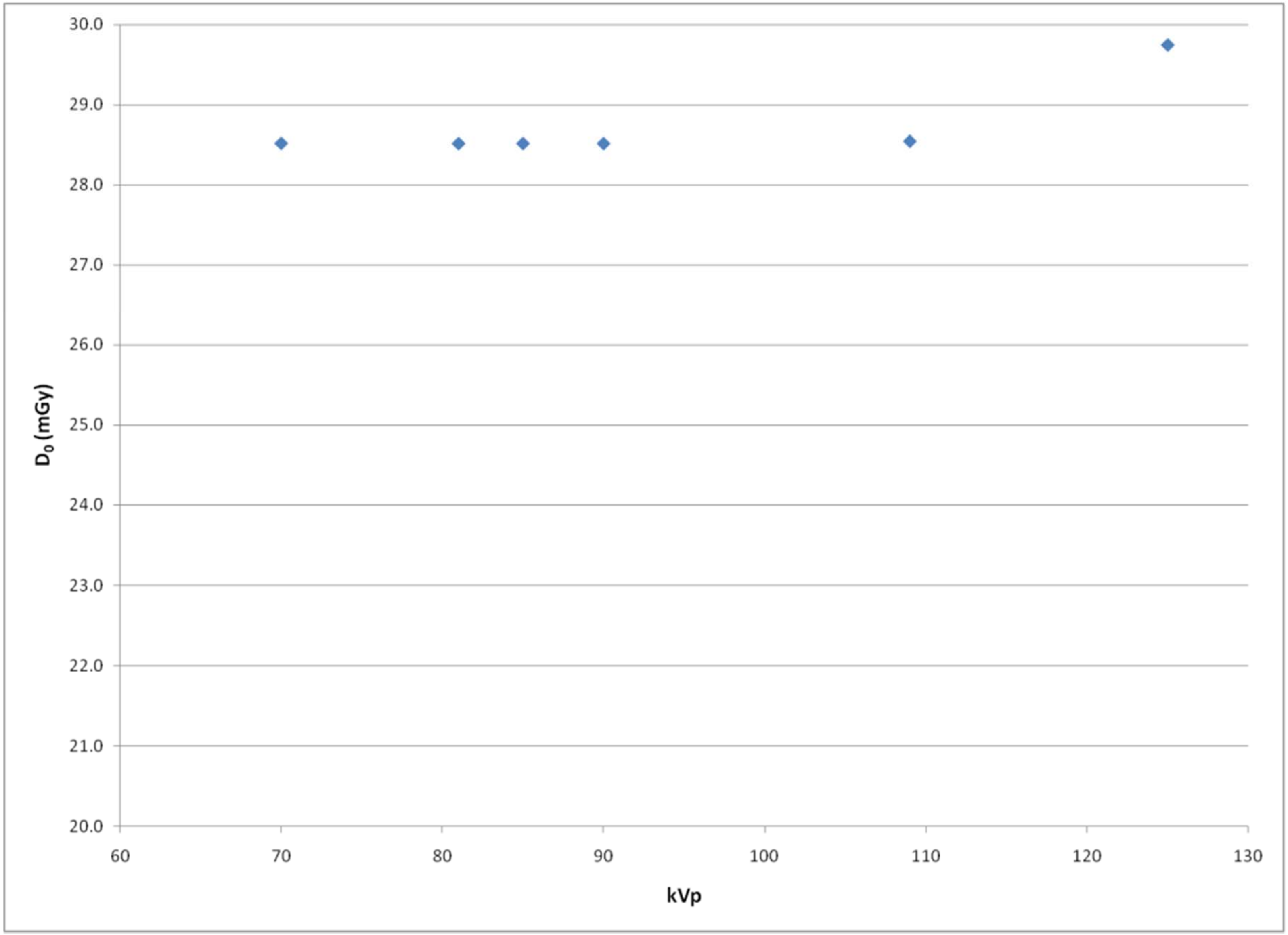
16 cm CTDI phantom – 0.36 µGy/frame



Body dose vs kVp setting

- In terms of center dose, attenuation is such that there is little dose savings by *selecting* lower kVp values

32 cm CTDI phantom – 0.36 uGy/frame



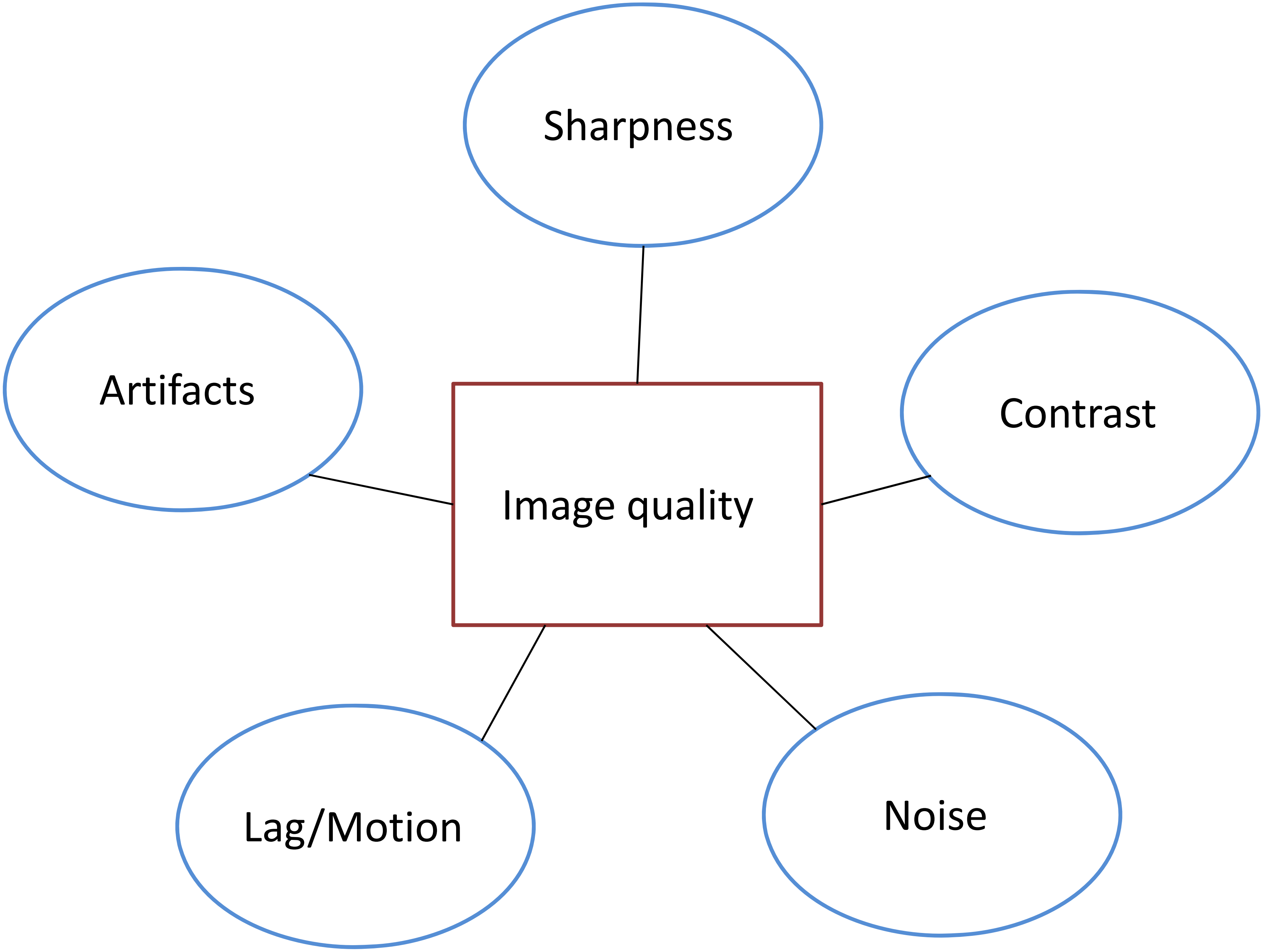
Additional resources

- Several authors have already proposed metrics and methods for assessing dose in FPCT
- Fahrig R, et al. Dose and image quality for a cone-beam C-arm CT system. Med Phys 33(12):4541-4550, 2006.
- Anderson JA, et al. CT dosimetry and the new modalities: Cone-beam and wide area CT. RSNA 2004.
- Kyriakou Y, et al. Concepts for dose determination in flat-detector CT. Phys Med Biol 53:3551-3566, 2008.

IMAGE QUALITY IN FPCT

What are your goals?

- To ensure constancy?
- To evaluate inter-system differences?
- To create/benchmark imaging protocols?
- Standardization of clinical research?
- To compare against manufacturer’s specifications (if there are any), i.e. acceptance testing?
- To evaluate adequacy for performing a specific task?
  - Replanning
  - Quantification
- To examine the influence of reconstruction and acquisition parameters on image quality?


Sharpness
Artifacts
Contrast
Image quality
Lag/Motion
Noise

Jeff Siewerdsen AAPM 2011 Image Recon. 1

http://www.aapm.org/meetings/amos2/pdf/59-17243-37526-878.pdf

Image corrections

Conventional CT

- Daily air cals/defect map
- CT# calibration
- Beam hardening corrections
- Reference channels
- To name a few

FPCT

- Yearly (maybe) I_0 cals
- Geometry cal

- No reference channels
- No daily cals
- No CT# calibration

Artifacts in FPCT

- Cone beam artifacts
  - Wide cone angle
- Streak artifacts
  - View aliasing
- Truncation
  - Limited FOV
- Scatter/cupping
  - Scatter
  - Lack of bowtie filter
  - Beam hardening
- Ring

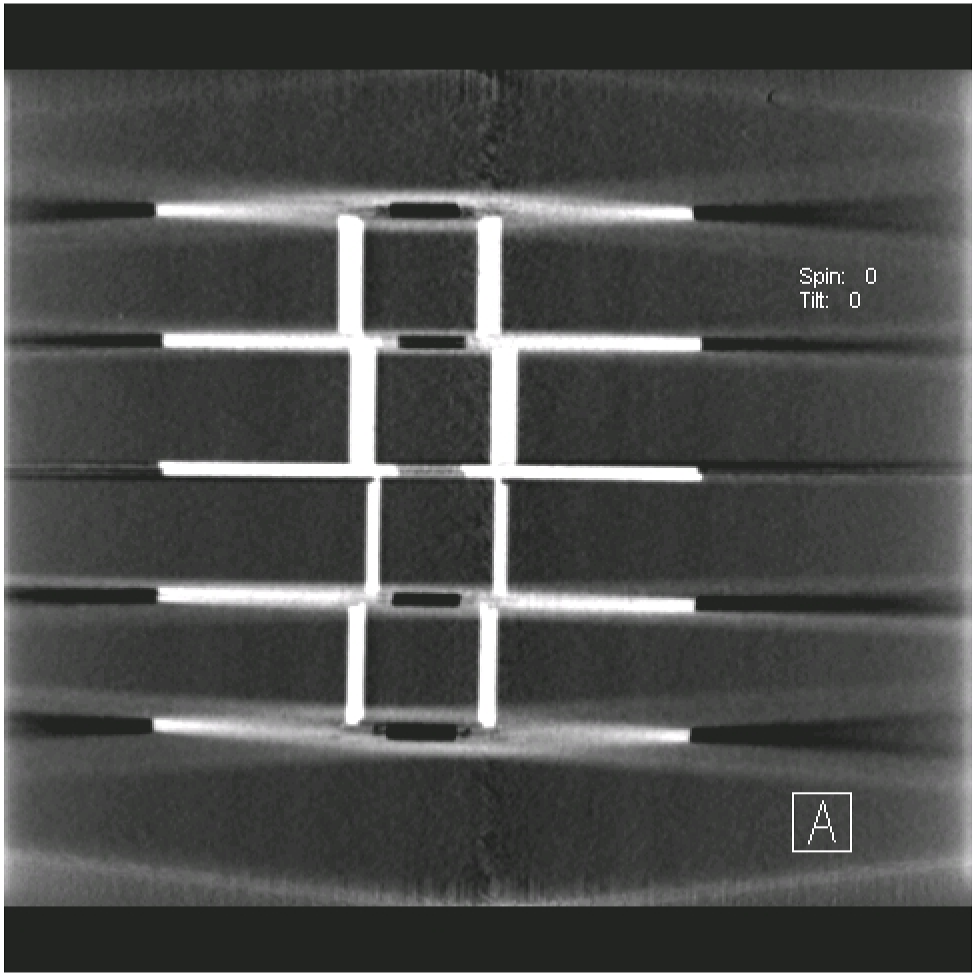
http://www.aapm.org/meetings/amos2/pdf/42-11961-35729-166.pdf

Cone beam artifact

- Image quality can vary substantially across the FOV, particularly in the cone beam direction
  - Even with corrections
- You may "unintentionally" measure this effect

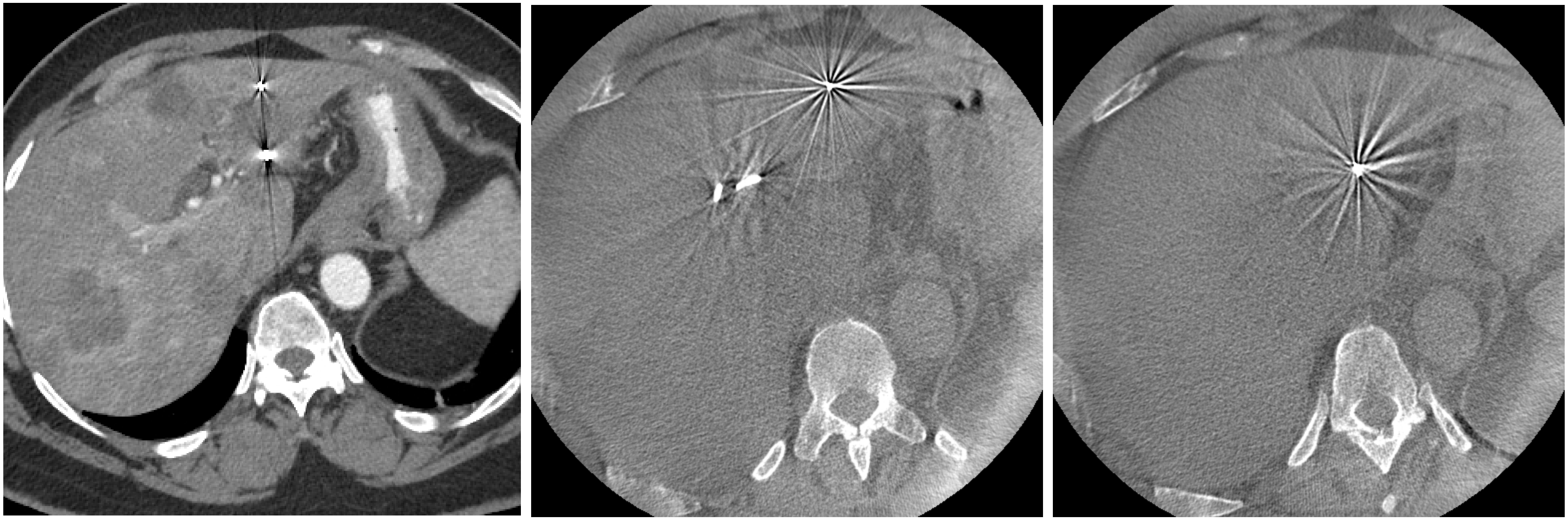
Jeff Siewerdsen AAPM 2009

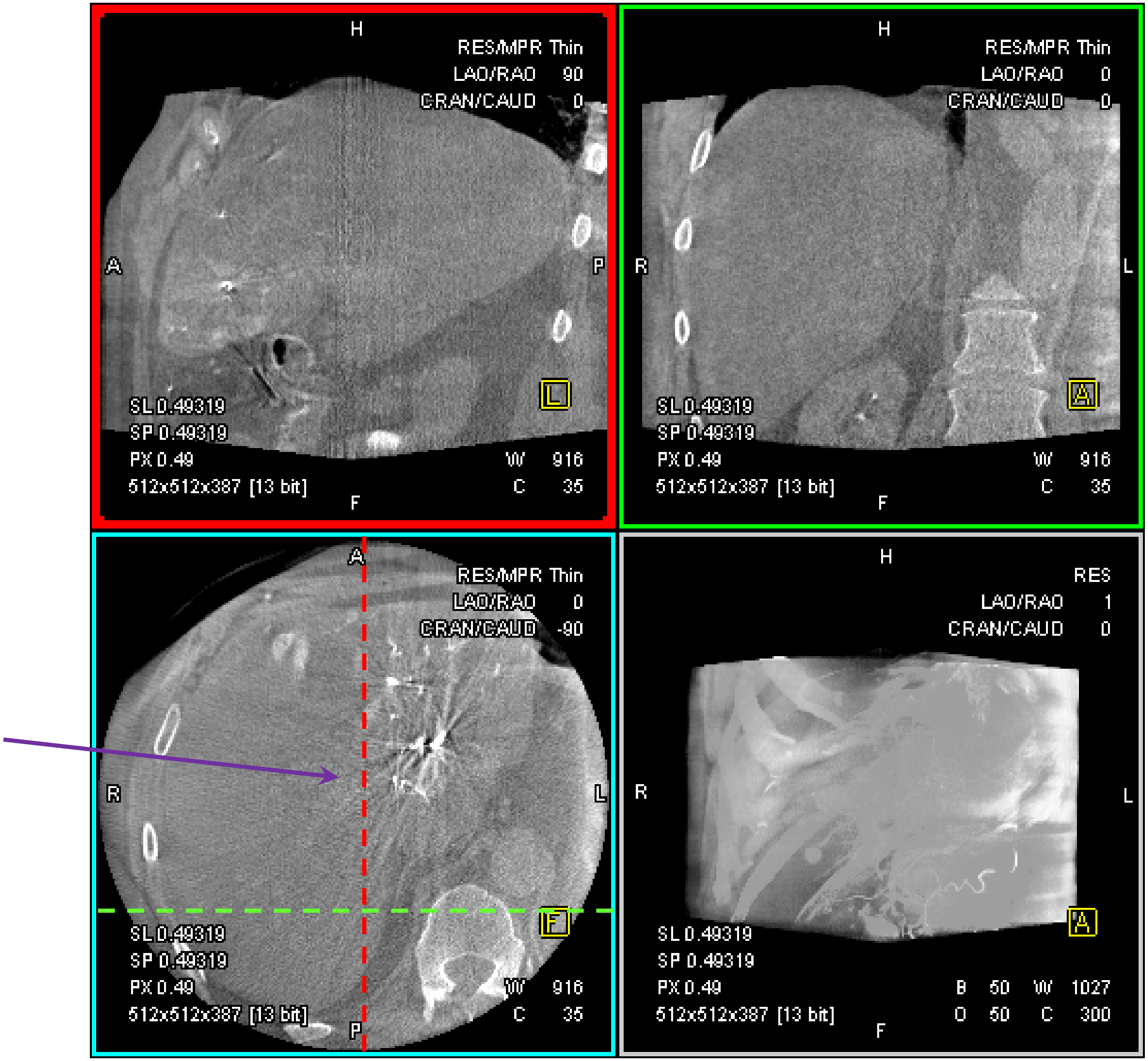
Homemade (very poorly) DeFrise



Streaks

- Undersampling causes streak artifacts
  - View aliasing
  - Half fan scanning results in loss of conjugate rays

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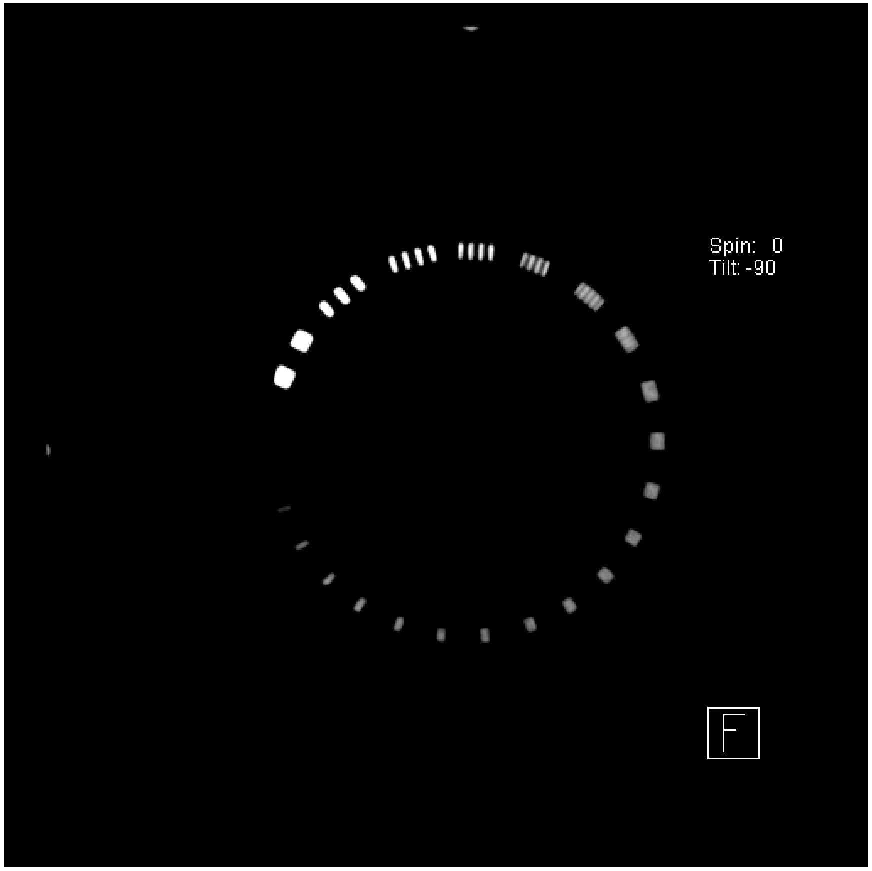
RES/MPR Thin
LAO/RAO 90
CRAN/CAUD 0
A
P
L
SL 0.49319
SP 0.49319
PX 0.49
512x512x387 [13 bit]
W 916
C 35
F
H
RES/MPR Thin
LAO/RAO 0
CRAN/CAUD 0
R
L
A
SL 0.49319
SP 0.49319
PX 0.49
512x512x387 [13 bit]
W 916
C 35
F
A
RES/MPR Thin
LAO/RAO 0
CRAN/CAUD -90
R
L
F
SL 0.49319
SP 0.49319
PX 0.49
512x512x387 [13 bit]
W 916
C 35
P
H
RES
LAO/RAO 1
CRAN/CAUD 0
R
L
A
SL 0.49319
SP 0.49319
PX 0.49
512x512x387 [13 bit]
B 50
O 50
W 1027
C 300
F

Sharpness in FPCT

- Sharpness in CT is influenced by many factors
  - Matrix size (512 vs. 1024)
  - DFOV
  - Kernel
  - Detector configuration
  - Reconstruction algorithm
  - Motion
  - Artifacts
  - Focal spot
- Binned pixel size is ~ 0.42 mm
  - Isotropic resolution for MPR/3D
  - This is equivalent to 21.5 cm FOV in CT
  - Changing DFOV depends on manufacturer



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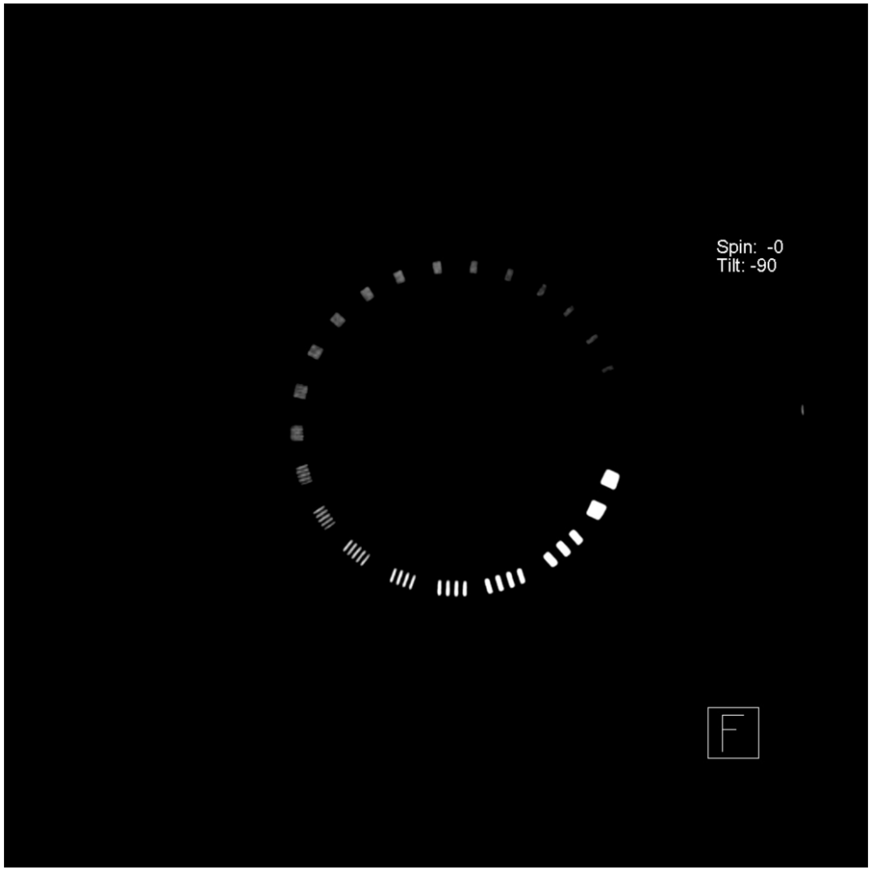

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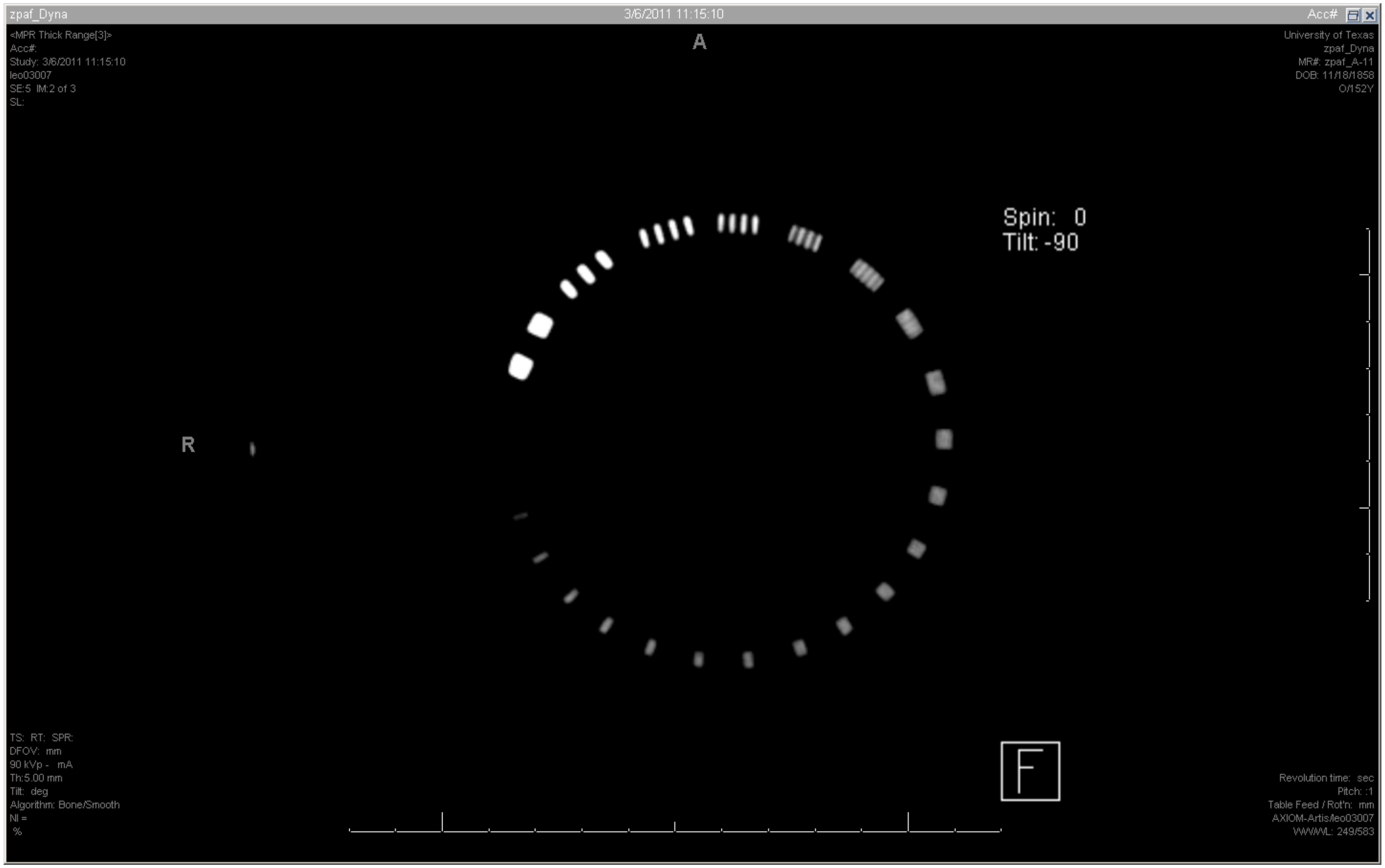


zpaf_Dyna
3/6/2011 11:15:10
Acc#

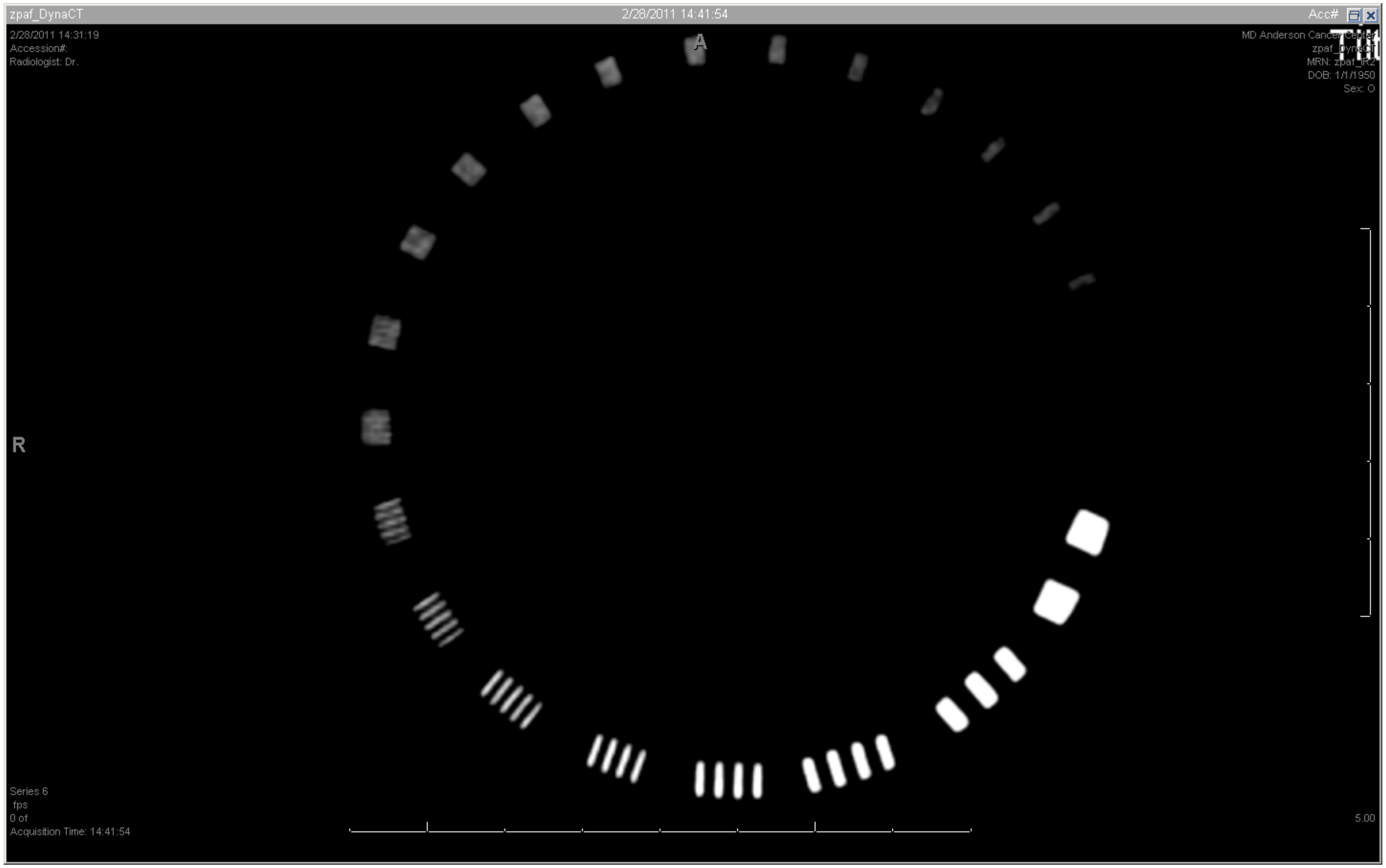
<MPR Thick Range[3]>
Acc#:
Study: 3/6/2011 11:15:10
leo03007
SE:5 IM:2 of 3
SL:

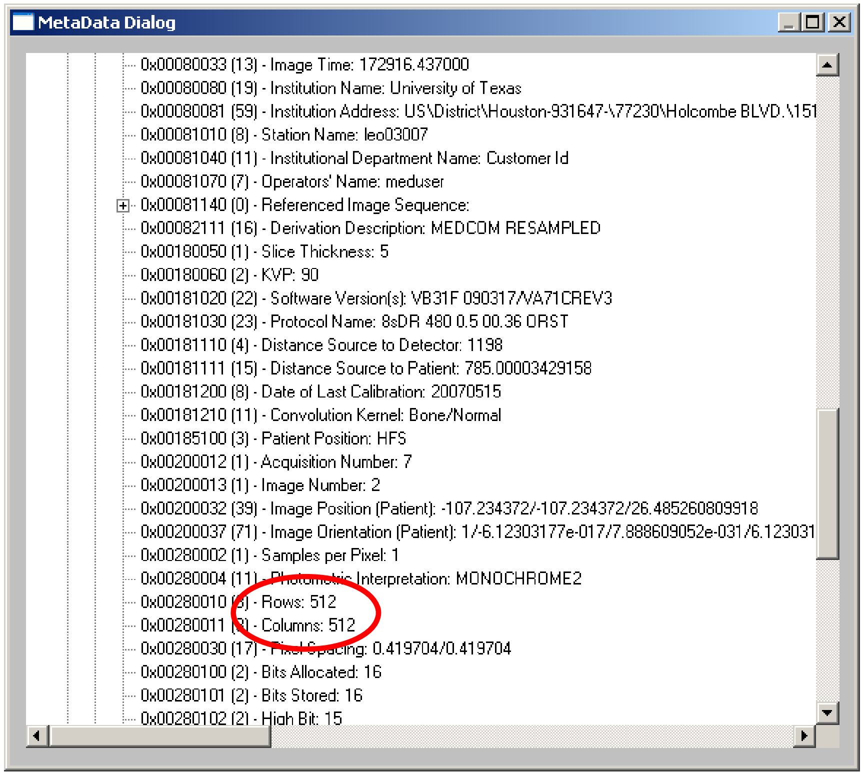
University of Texas
zpaf_Dyna
MR#: zpaf_A-11
DOB: 11/18/1858
O/152Y

Spin: 0
Tilt: -90

TS: RT: SPR:
DFOV: mm
90 kVp - mA
Th:5.00 mm
Tilt: deg
Algorithm: Bone/Smooth
NI =
%

Revolution time: sec
Pitch: :1
Table Feed / Rot'n: mm
AXIOM-Artis/leo03007
WW/WL: 249/583





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0x00080080 (19) - Institution Name: University of Texas
0x00080081 (59) - Institution Address: US\District\Houston-931647-\77230\Holcombe BLVD.\151
0x00081010 (8) - Station Name: leo03007
0x00081040 (11) - Institutional Department Name: Customer Id
0x00081070 (7) - Operators' Name: meduser
0x00081140 (0) - Referenced Image Sequence:
0x00082111 (16) - Derivation Description: MEDCOM RESAMPLED
0x00180050 (1) - Slice Thickness: 5
0x00180060 (2) - KVP: 90
0x00181020 (22) - Software Version(s): VB31F 090317/VA71CREV3
0x00181030 (23) - Protocol Name: 8sDR 480 0.5 00.36 ORST
0x00181110 (4) - Distance Source to Detector: 1198
0x00181111 (15) - Distance Source to Patient: 785.00003429158
0x00181200 (8) - Date of Last Calibration: 20070515
0x00181210 (11) - Convolution Kernel: Bone/Normal
0x00185100 (3) - Patient Position: HFS
0x00200012 (1) - Acquisition Number: 7
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0x00200037 (71) - Image Orientation (Patient): 1/-6.12303177e-017/7.888609052e-031/6.123031
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0x00280004 (11) - Photometric Interpretation: MONOCHROME2
0x00280010 (2) - Rows: 512
0x00280011 (2) - Columns: 512
0x00280030 (17) - Pixel Spacing: 0.419704/0.419704
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0x00280101 (2) - Bits Stored: 16
0x00280102 (2) - High Bit: 15
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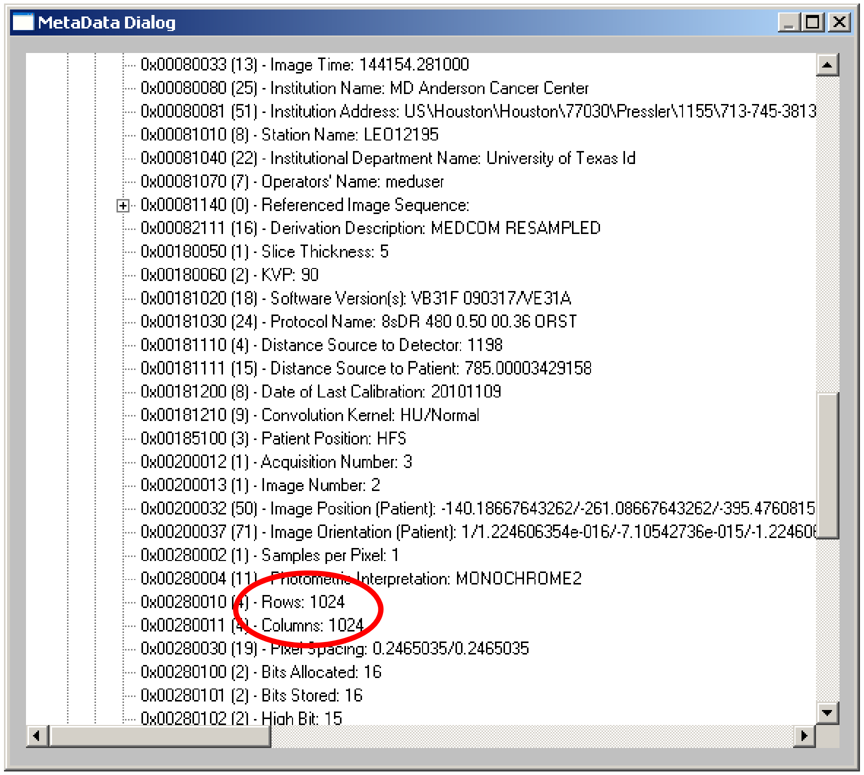

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0x00081070 (7) - Operators' Name: meduser
0x00081140 (0) - Referenced Image Sequence:
0x00082111 (16) - Derivation Description: MEDCOM RESAMPLED
0x00180050 (1) - Slice Thickness: 5
0x00180060 (2) - KVP: 90
0x00181020 (18) - Software Version(s): VB31F 090317/VE31A
0x00181030 (24) - Protocol Name: 8sDR 480 0.50 00.36 ORST
0x00181110 (4) - Distance Source to Detector: 1198
0x00181111 (15) - Distance Source to Patient: 785.00003429158
0x00181200 (8) - Date of Last Calibration: 20101109
0x00181210 (9) - Convolution Kernel: HU/Normal
0x00185100 (3) - Patient Position: HFS
0x00200012 (1) - Acquisition Number: 3
0x00200013 (1) - Image Number: 2
0x00200032 (50) - Image Position (Patient): -140.18667643262/-261.08667643262/-395.4760815
0x00200037 (71) - Image Orientation (Patient): 1/1.224606354e-016/-7.10542736e-015/-1.22460
0x00280002 (1) - Samples per Pixel: 1
0x00280004 (11) - Photometric Interpretation: MONOCHROME2
0x00280010 (4) - Rows: 1024
0x00280011 (4) - Columns: 1024
0x00280030 (19) - Pixel Spacing: 0.2465035/0.2465035
0x00280100 (2) - Bits Allocated: 16
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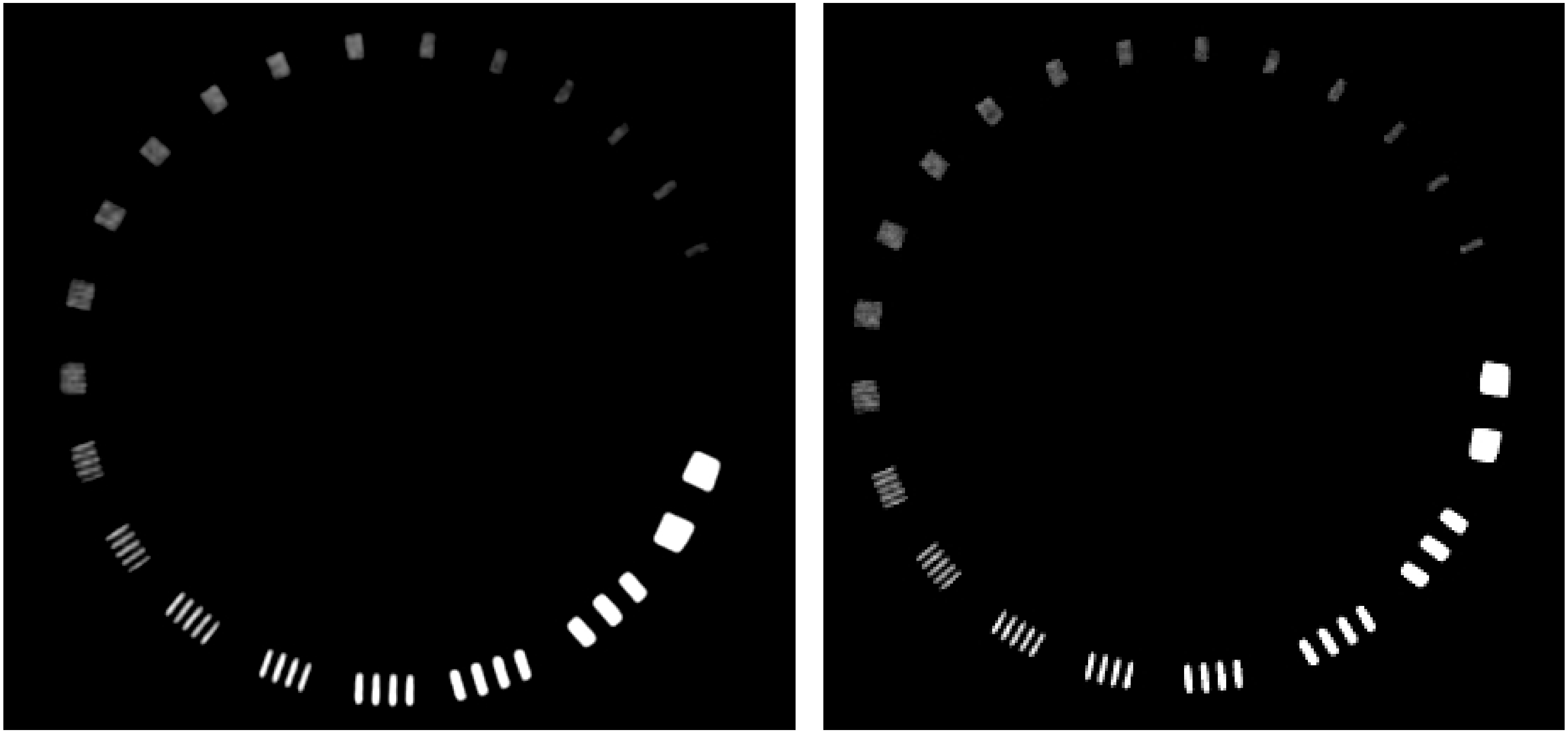
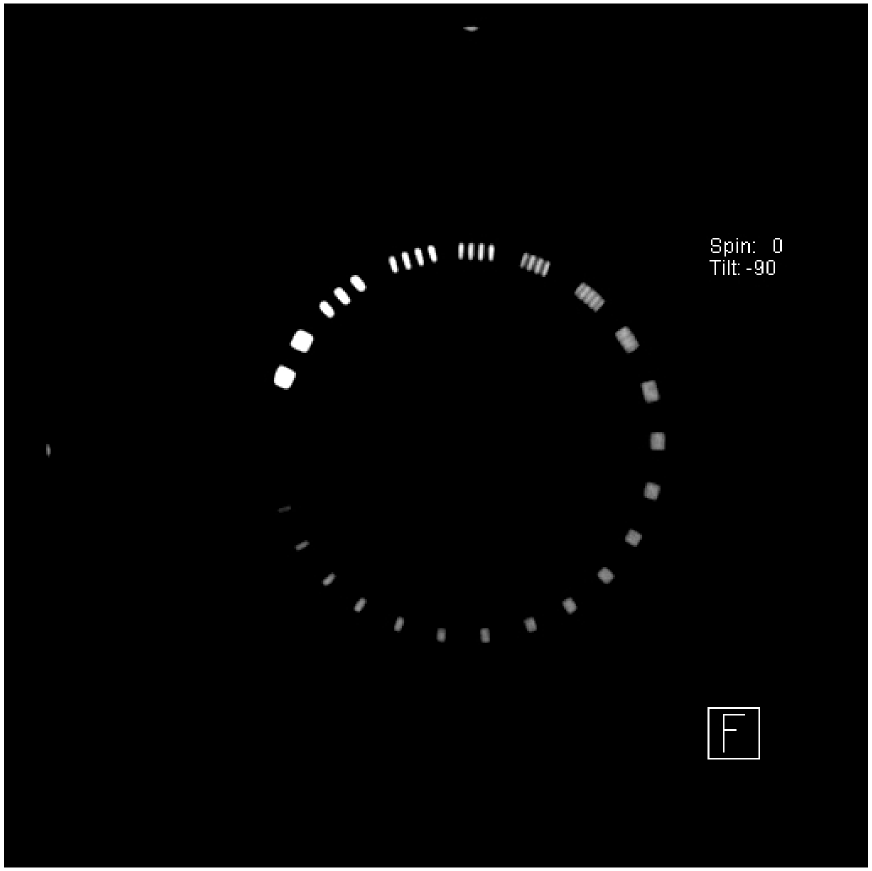
0x00280102 (2) - High Bit: 15
```

Guess the difference



Resolution limited by focal spot size?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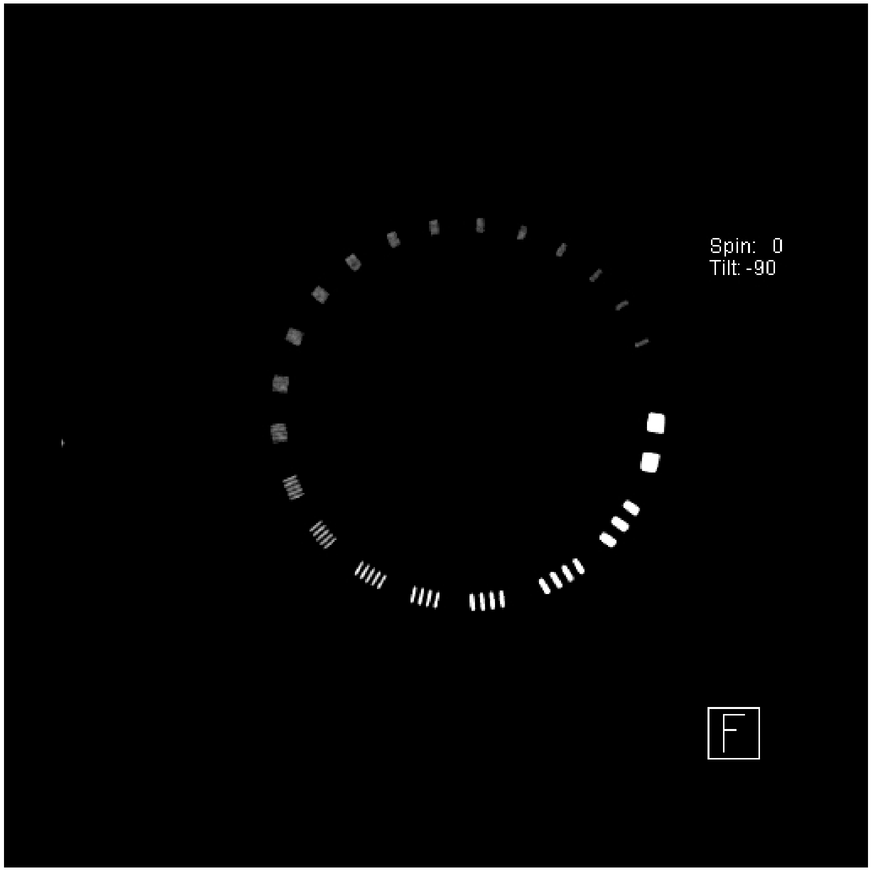
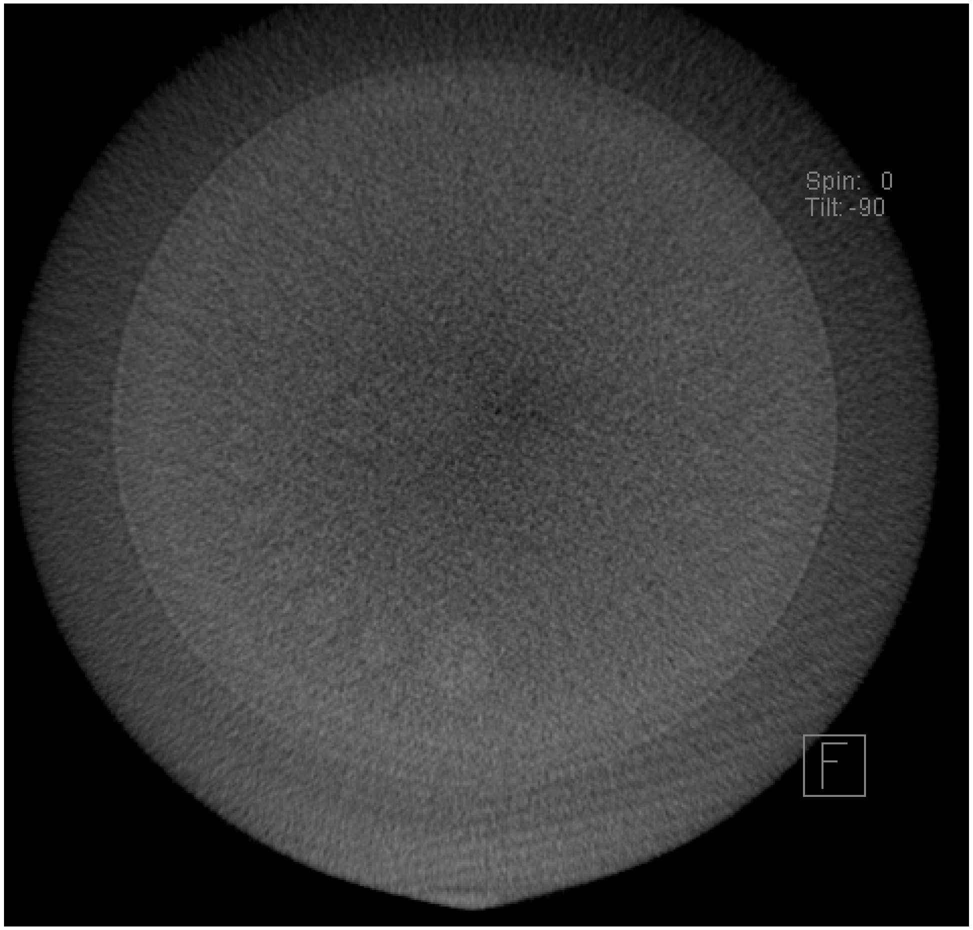

Spin: 0
Tilt: -90
F

Sharpness in FPCT

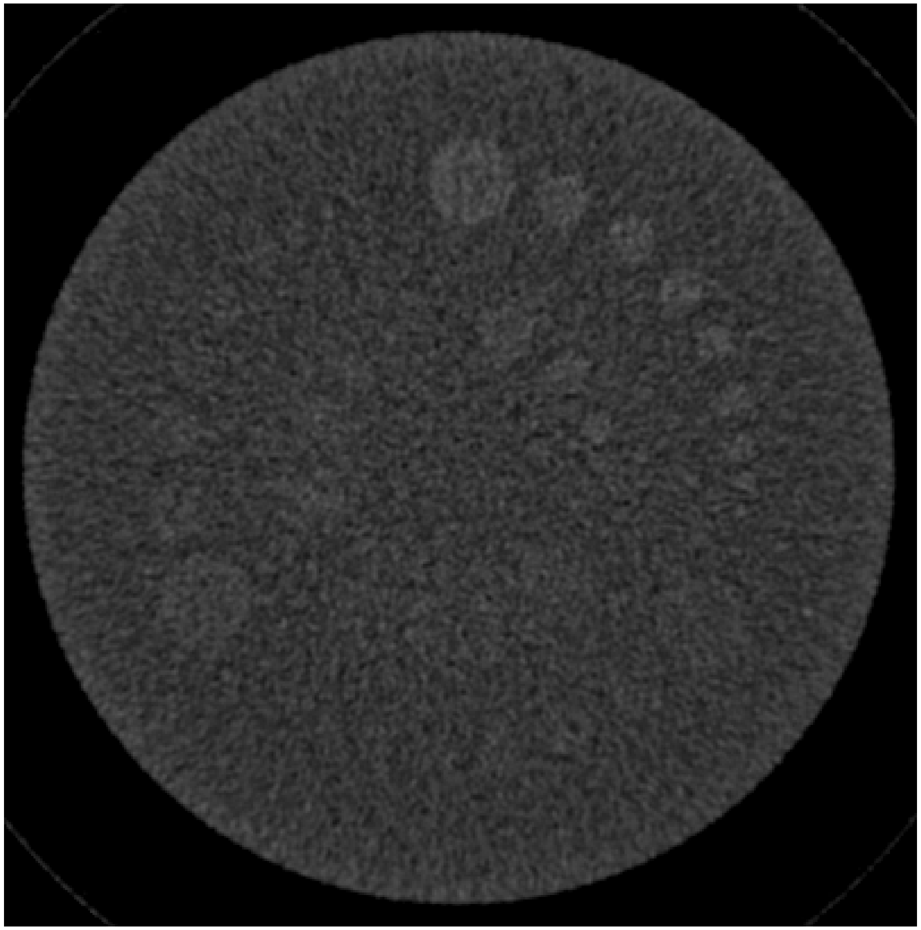
- The cone beam artifact also blurs images near the ends of the FOV
- Patient motion (e.g., breathing) can also be a problem
  - 8 sec scan time for DynaCT
  - Image-guided surgery – breathing can be controlled

Noise and low contrast resolution

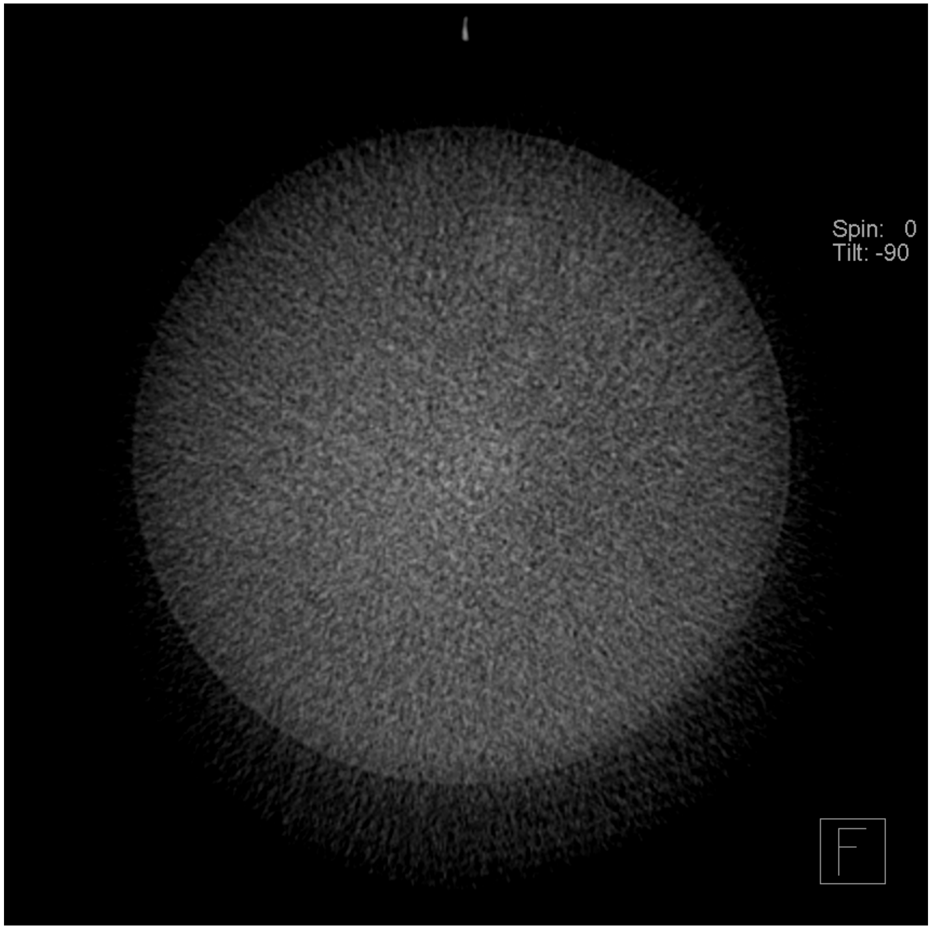
- Contrast resolution is much poorer in FPCT than in CT, and noise is less homogeneous
- Several reasons for this difference
  - FP detectors suffer from more additive noise
  - Del is smaller
  - Scattered radiation
  - Absence of bowtie filter
  - Less than 360 rotation
- We have to ask the question – For what are we using these images?



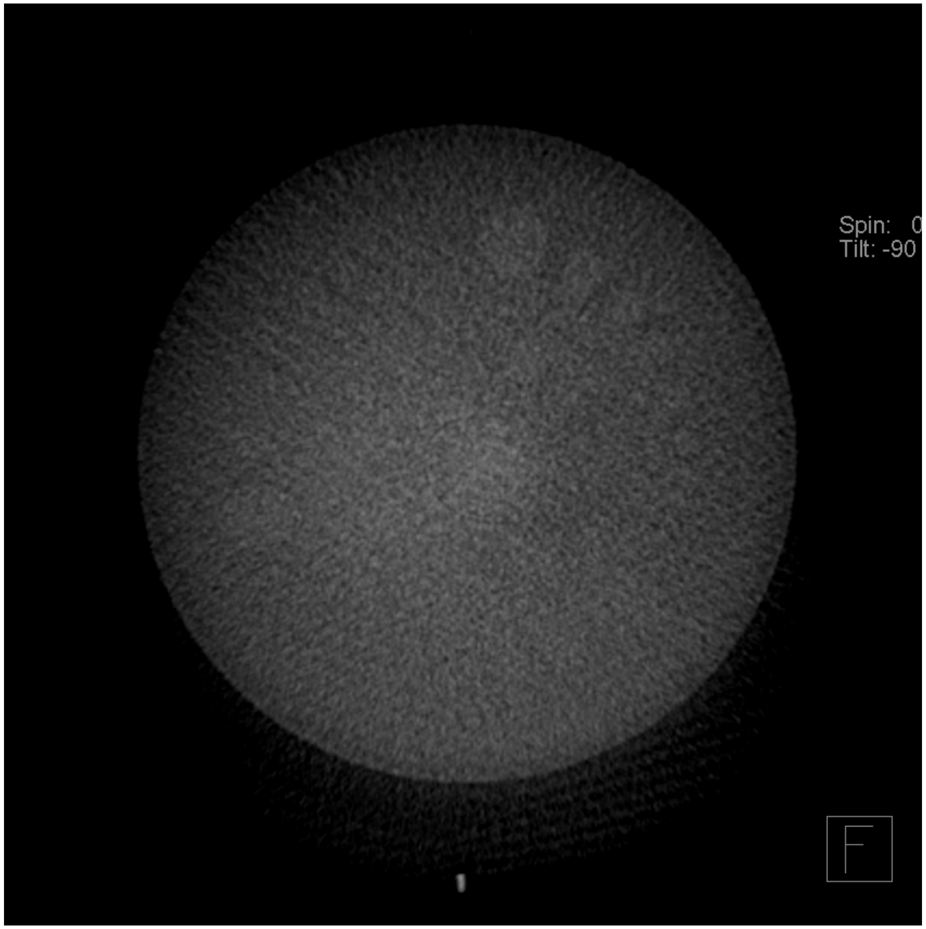
1.2 µGy/frame
(~ 75% CT dose)



CT 240 mAs



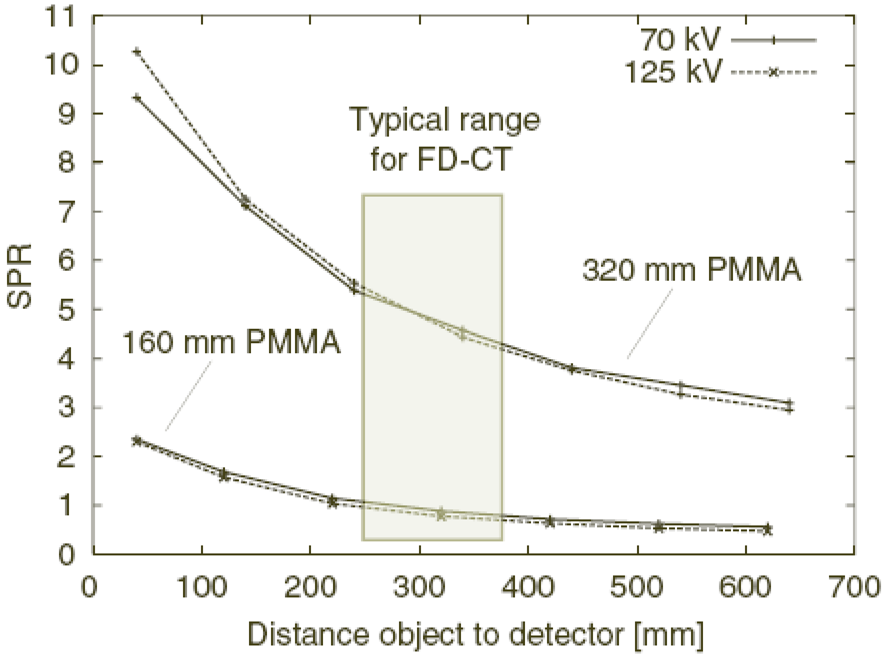
0.36 µGy/frame



1.2 µGy/frame

Scatter

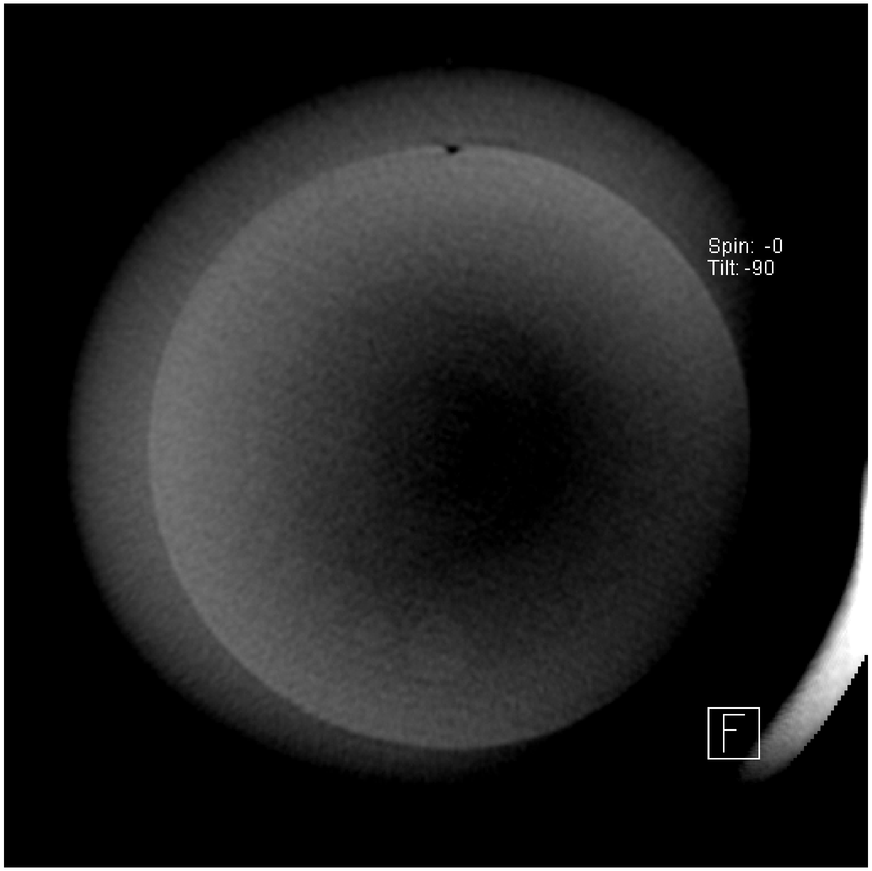
- A CT scanner is fairly efficient at rejecting scattered radiation
  - Collimated x-ray beam
  - Anti-scatter grid (~ 13:1)
  - Air gap
- A FPCT scanner is not very efficient at rejecting scattered radiation
  - CON: Very large x-ray beam (30 x 40 cm)
  - PRO: Air gap, anti-scatter grid



Kyriakou Y, Kalendar W. Efficiency of anti-scatter grids for flat-detector CT. Phys Med Biol 52:6275-6293, 2007.

Scatter

- Scatter
  - Reduces contrast
  - Introduces artifacts
- Anti-scatter grids are used to reduce the S/P ratio

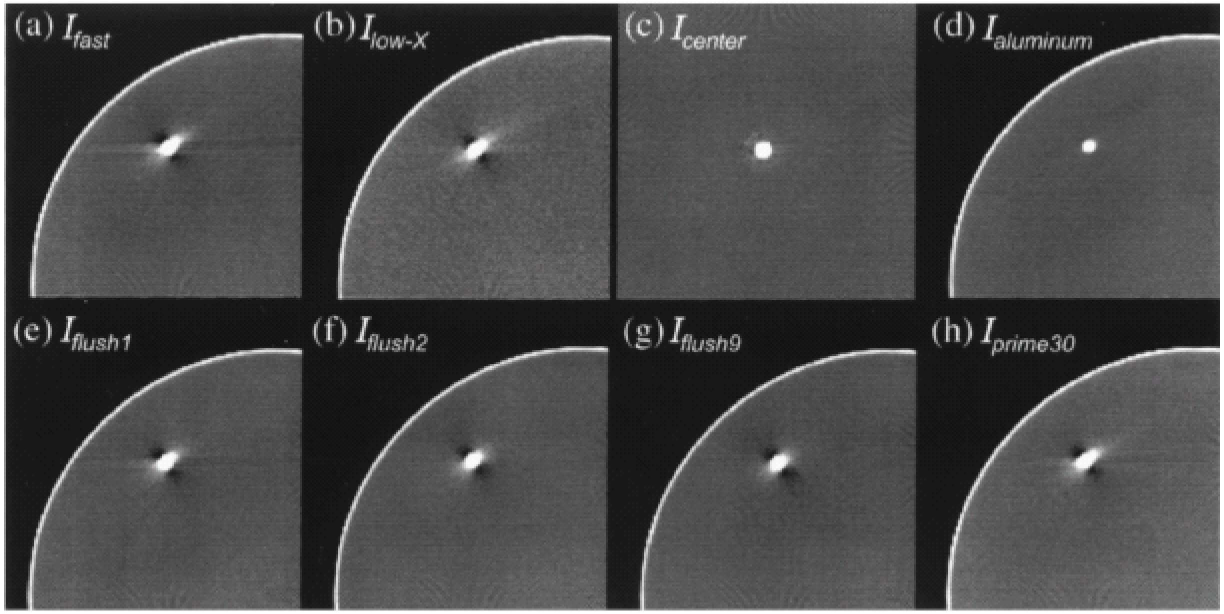


Motion

- Motion is a problem in many implementations of tomographic imaging
- The likelihood of motion increases with the total scan time
- In FPCT, scan time can be reduced by
  - Reducing angular sampling
  - Increasing rotation speed (this also requires increasing readout capability)
  - Always need 180° + fan beam
  - There are always tradeoffs

Lag

- CT detectors are ultrafast and have little to no afterglow
- FP image receptors suffer from multiple sources of lag
  - Afterglow
  - Charge trapping
- This can cause artifacts in sequences of rapidly acquired images



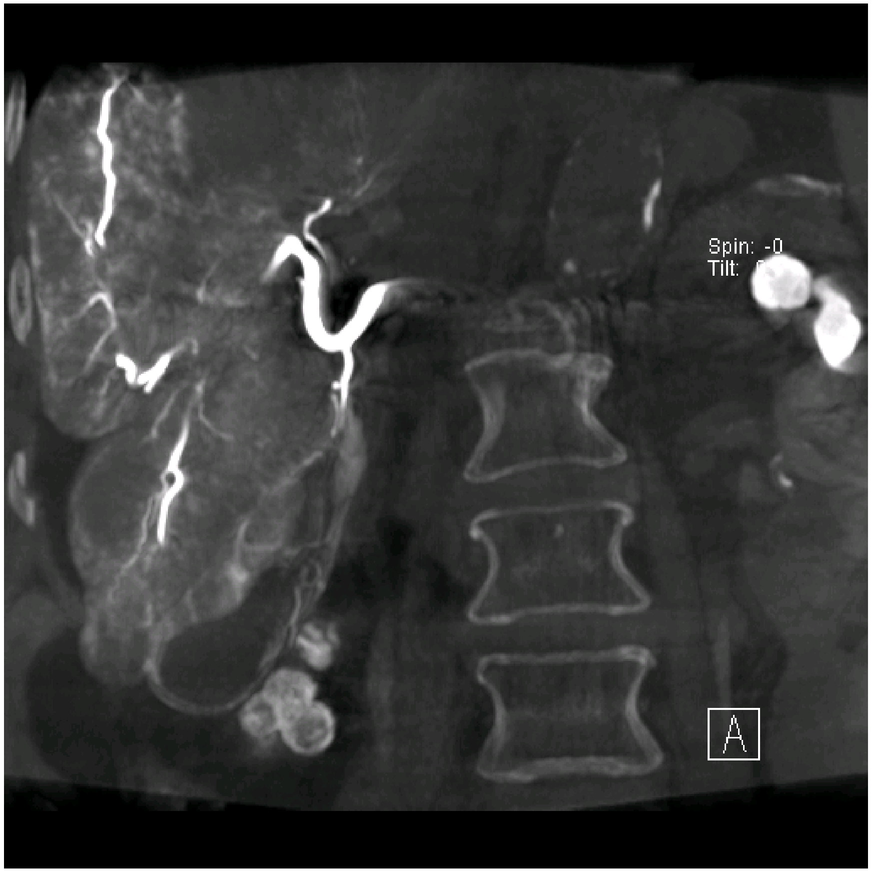
Siewerdsen JH and Jaffray DA. Cone-beam computed tomography with a flat-panel imager: Effects of image lag. Med Phys 26(12):2635-2647, 1999.

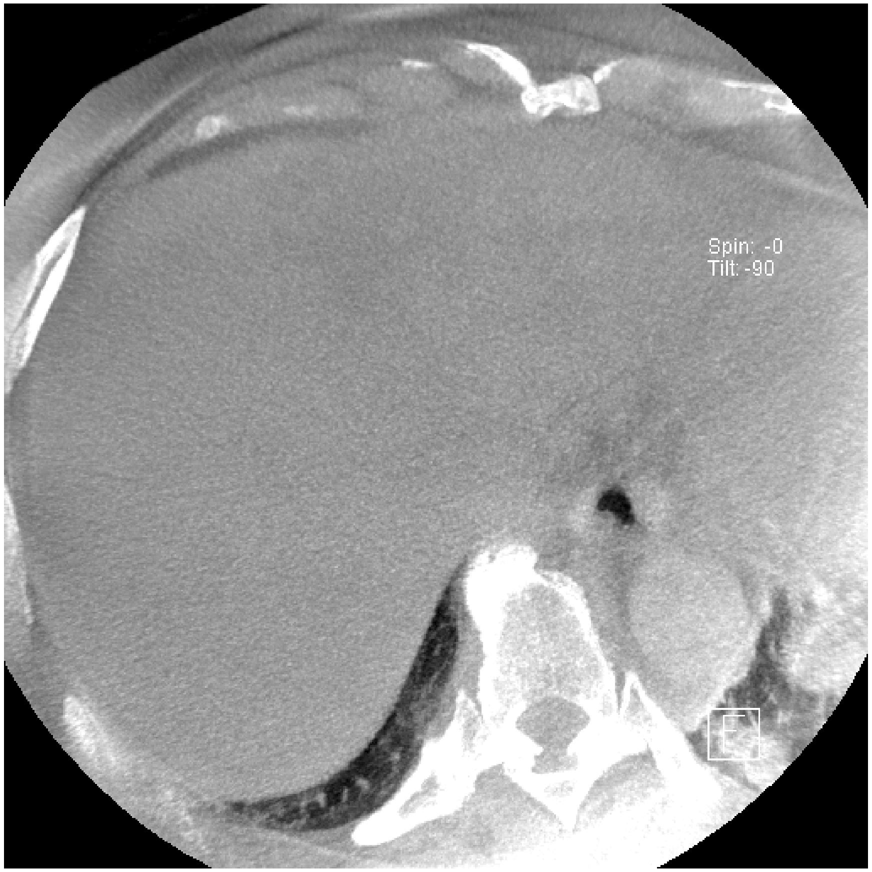
Lag

- In fact, the OBI maintenance manual mentions this in several places
  - Many irradiations of the panel for I_0 calibration
  - Recommends panel has some recent irradiation history for other calibrations

Contrast

- Contrast is influenced by a variety of factors
  - kVp/filtration
  - Scatter
  - Contrast agents
  - Injection protocols
  - Bit depth
  - Image processing
  - Image rendering

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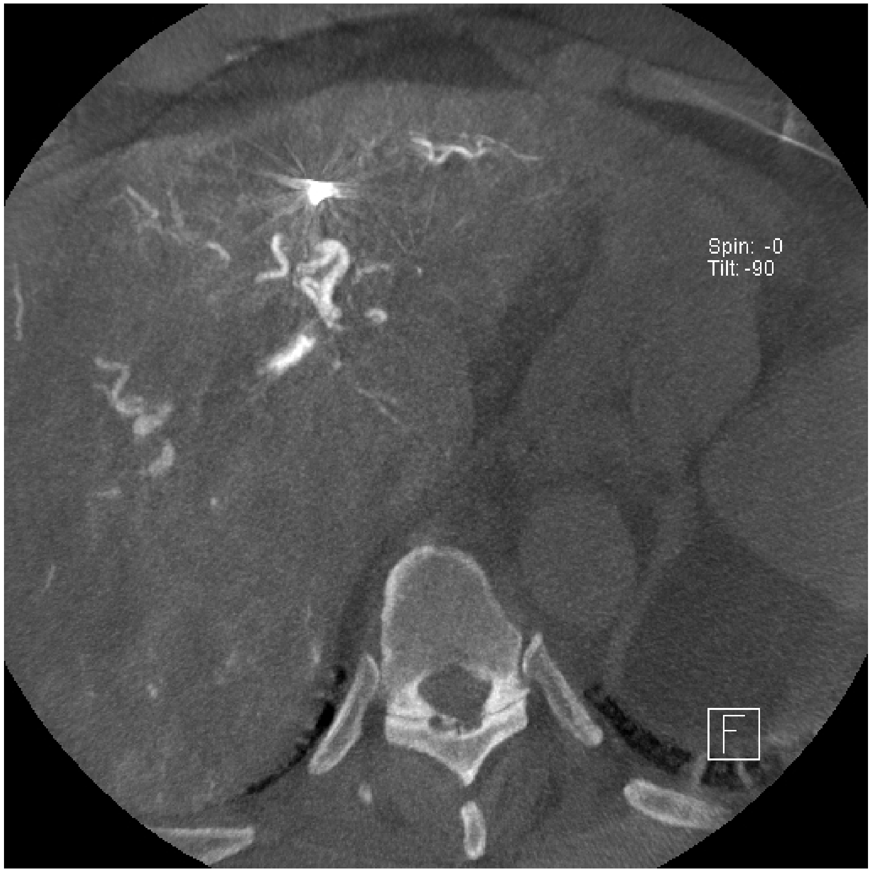
Spin: -0
Tilt: -90
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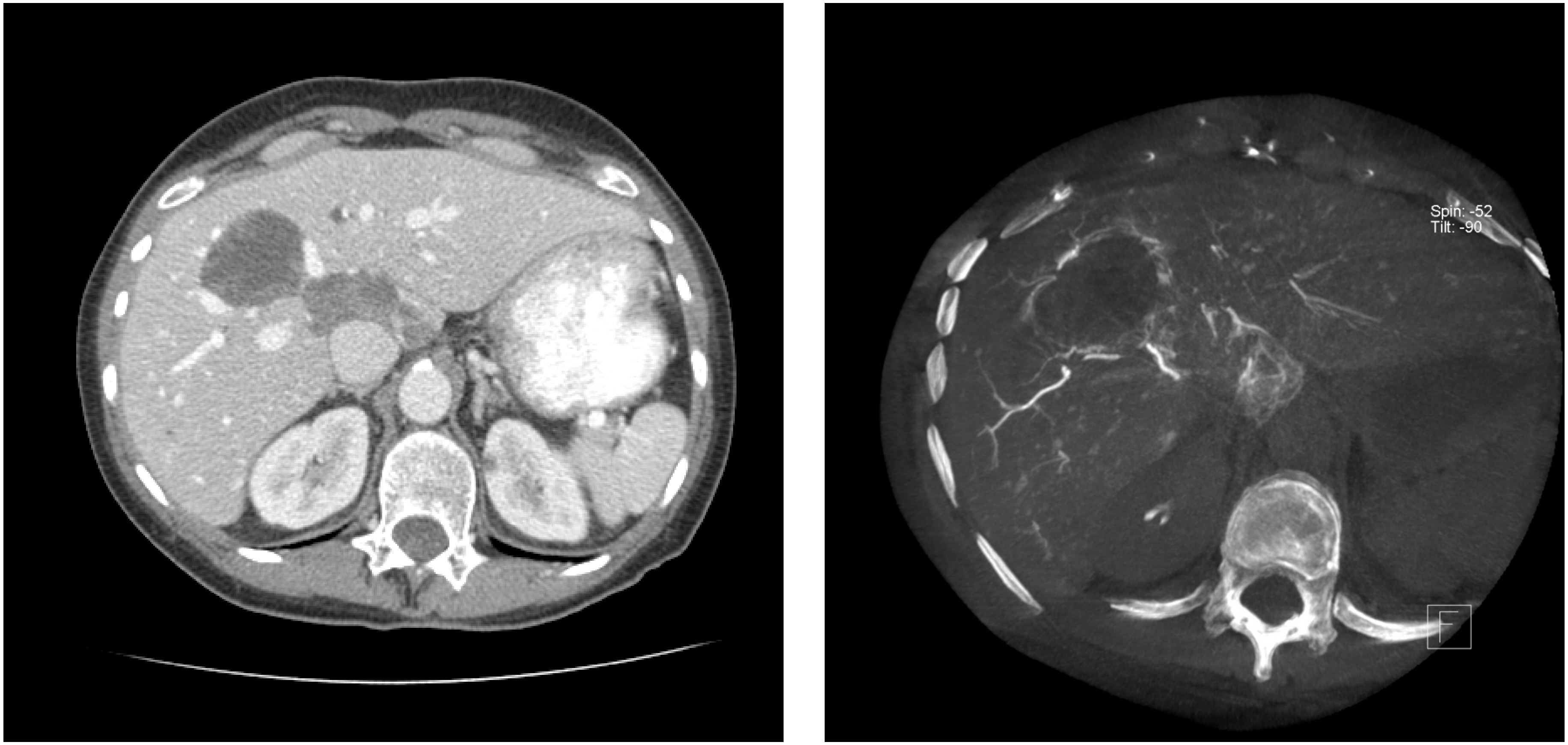
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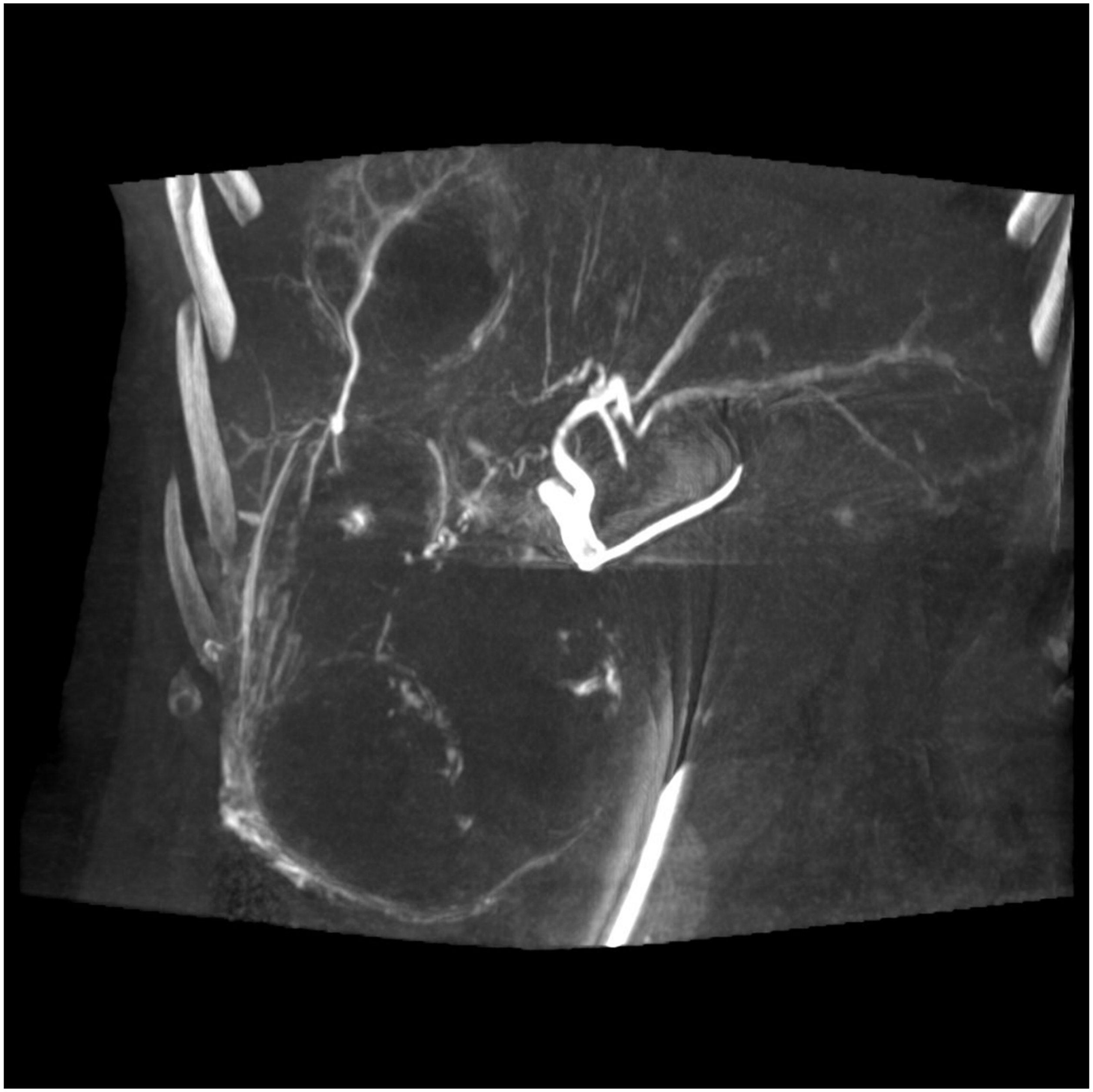
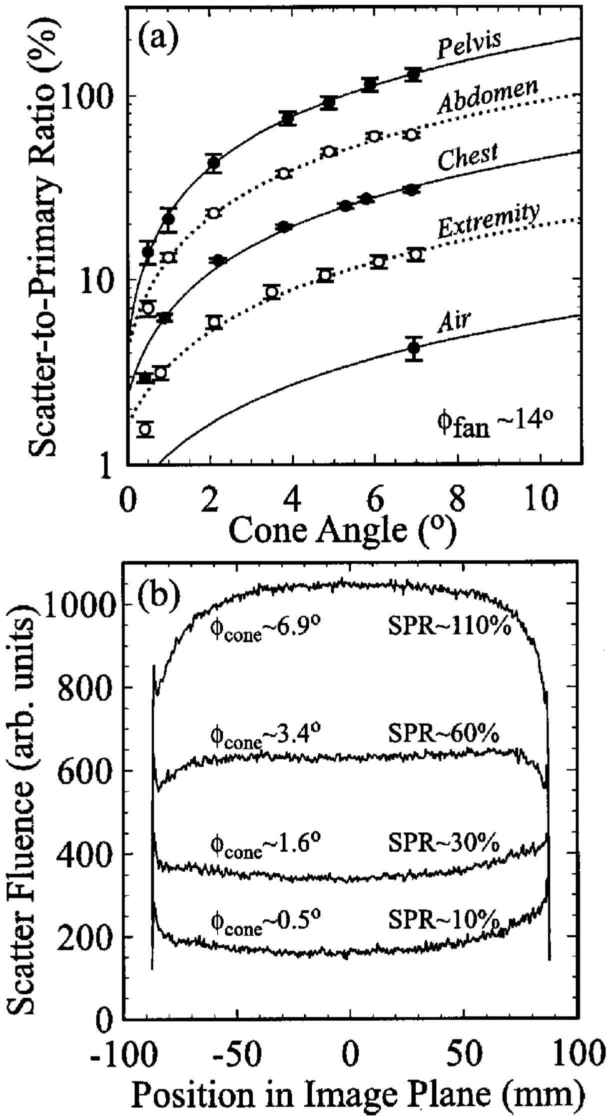
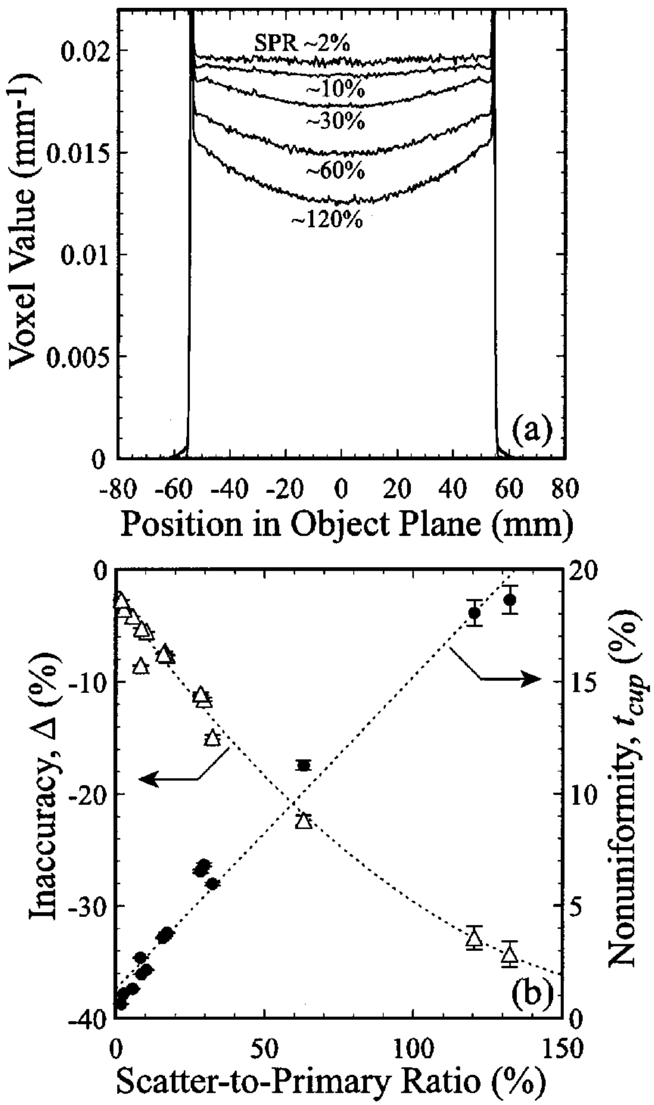


Quantitative accuracy in FPCT

- CT numbers (HU) are affected by many factors
  - Reconstruction method
  - Kernel
  - kVp
  - Artifacts
  - Calibrations
  - Position in FOV
- Many implementations of FPCT do not rely on quantification using absolute CT numbers
  - Dental
  - MIP Dyna
  - Positioning verification in RO
- Caution must be exercised if using CT numbers for quantification

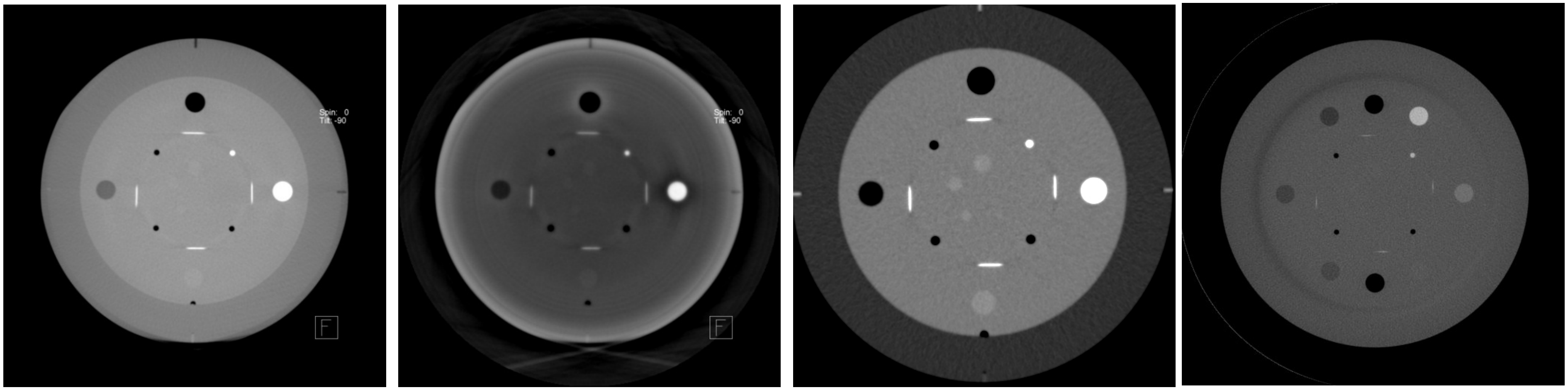
Varian OBI Maintenance Manual



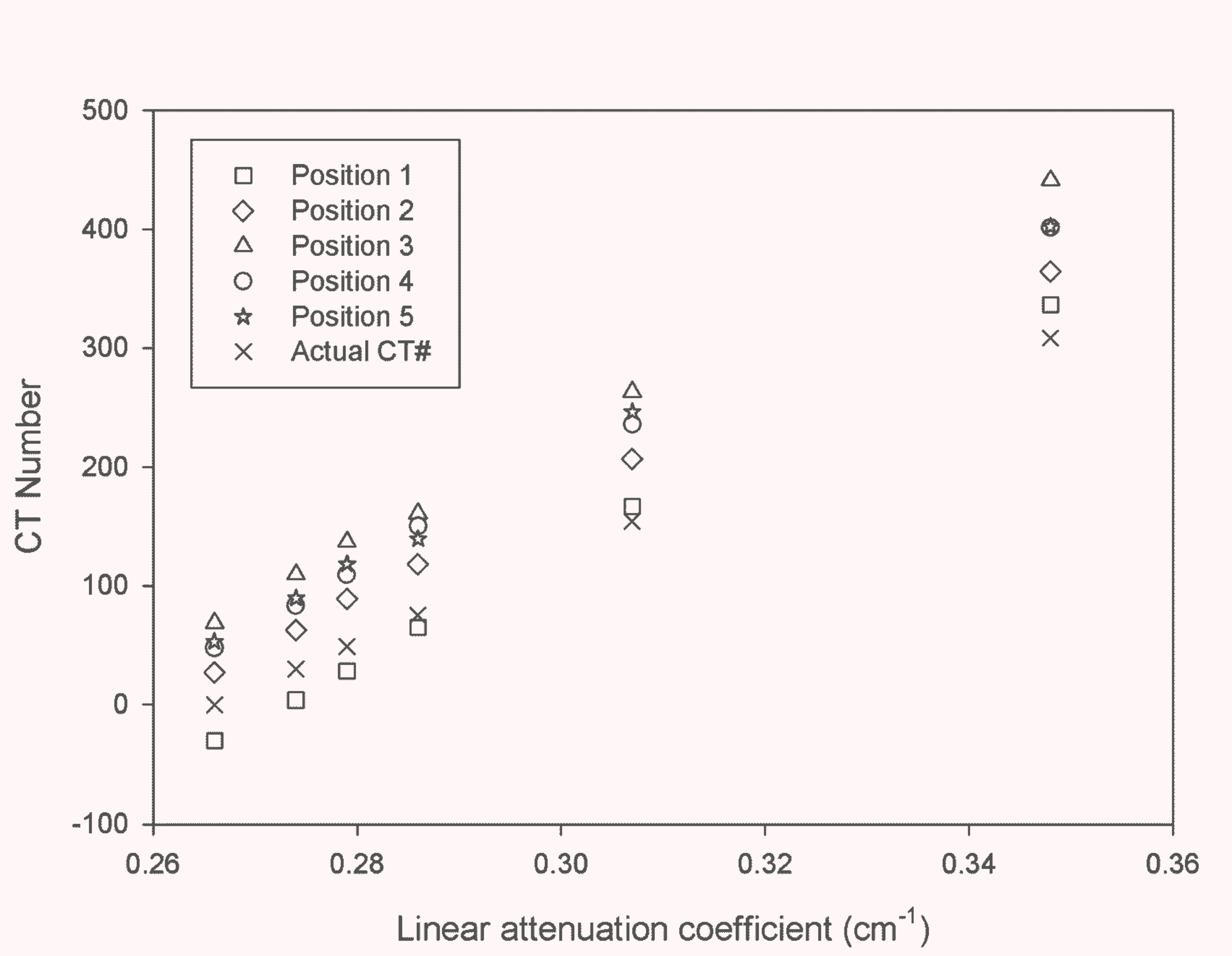


Jeffrey H. Siewerdsen and David A. Jaffray. Cone-beam computed tomography with a flat-panel imager: Magnitude and effects of x-ray scatter. Med Phys 28:220-231, 2001.

Reconstruction parameters affect HU



Material	CT number (HU/normal)	CT number (EE/normal)	CT number (Sensation 64)	Varian OBI (Full)	Actual
Air	-859	-1018	-998	-996	-1000
LDPE	-84	-561	-90	-119	-94
PMMA	117	-338	124	101	120
PTFE	865	598	931	963	891
Slope	4742	4406	5310	5384	5208


500
400
300
200
100
0
-100
CT Number
0.26
0.28
0.30
0.32
0.34
0.36
Linear attenuation coefficient (cm^{-1})
□ Position 1
◇ Position 2
△ Position 3
○ Position 4
☆ Position 5
× Actual CT#

Relative contrast enhancement

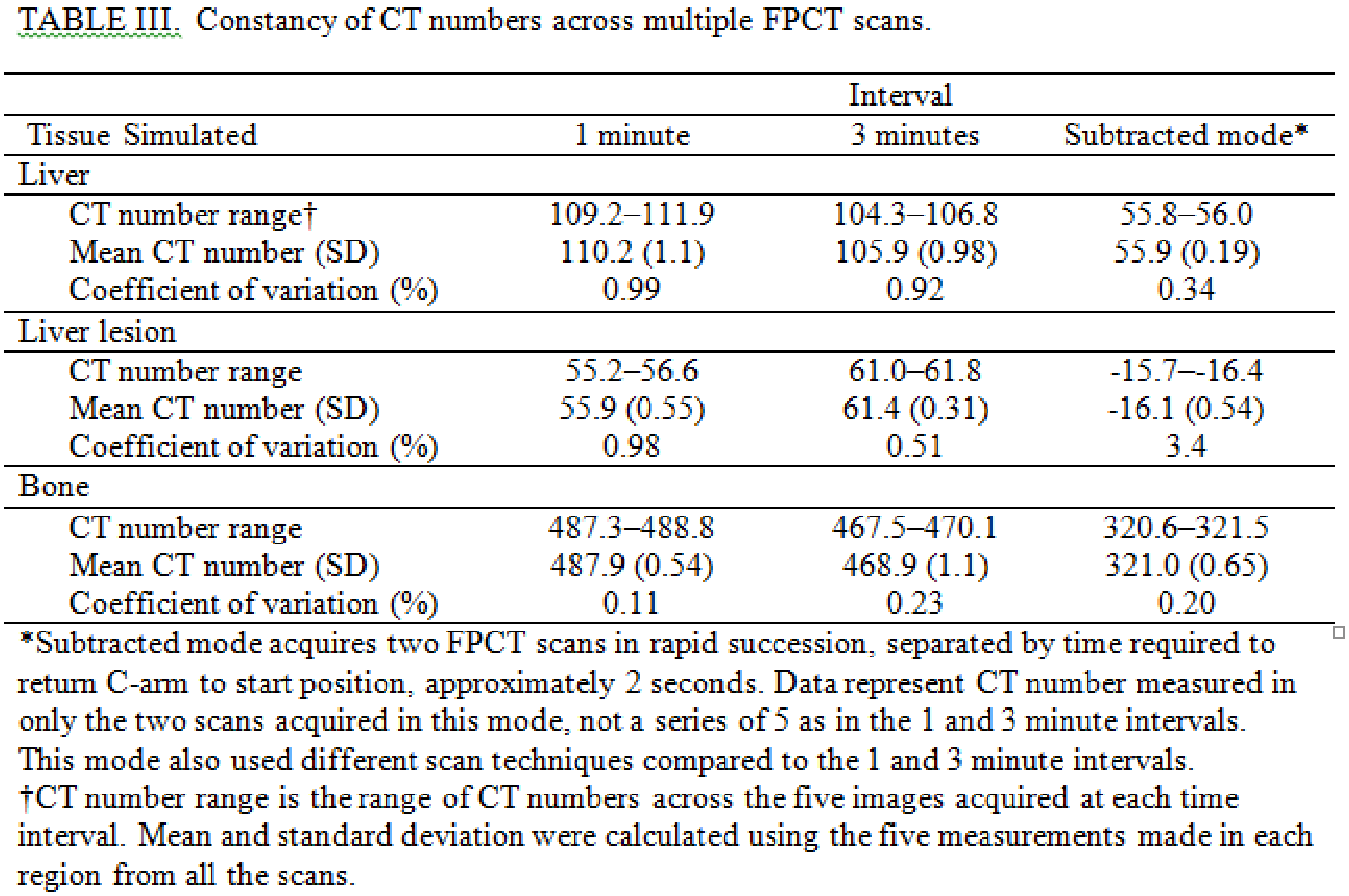


TABLE III. Constancy of CT numbers across multiple FPCT scans.

	Interval		
Tissue Simulated	1 minute	3 minutes	Subtracted mode*
Liver			
CT number range†	109.2–111.9	104.3–106.8	55.8–56.0
Mean CT number (SD)	110.2 (1.1)	105.9 (0.98)	55.9 (0.19)
Coefficient of variation (%)	0.99	0.92	0.34
Liver lesion			
CT number range	55.2–56.6	61.0–61.8	-15.7–-16.4
Mean CT number (SD)	55.9 (0.55)	61.4 (0.31)	-16.1 (0.54)
Coefficient of variation (%)	0.98	0.51	3.4
Bone			
CT number range	487.3–488.8	467.5–470.1	320.6–321.5
Mean CT number (SD)	487.9 (0.54)	468.9 (1.1)	321.0 (0.65)
Coefficient of variation (%)	0.11	0.23	0.20

*Subtracted mode acquires two FPCT scans in rapid succession, separated by time required to return C-arm to start position, approximately 2 seconds. Data represent CT number measured in only the two scans acquired in this mode, not a series of 5 as in the 1 and 3 minute intervals. This mode also used different scan techniques compared to the 1 and 3 minute intervals.

†CT number range is the range of CT numbers across the five images acquired at each time interval. Mean and standard deviation were calculated using the five measurements made in each region from all the scans.

Jones AK and Mahvash A. Evaluation of the potential utility of flat panel CT for quantifying relative contrast enhancement. Med Phys 39:4149-4154, 2012.

Relative contrast enhancement

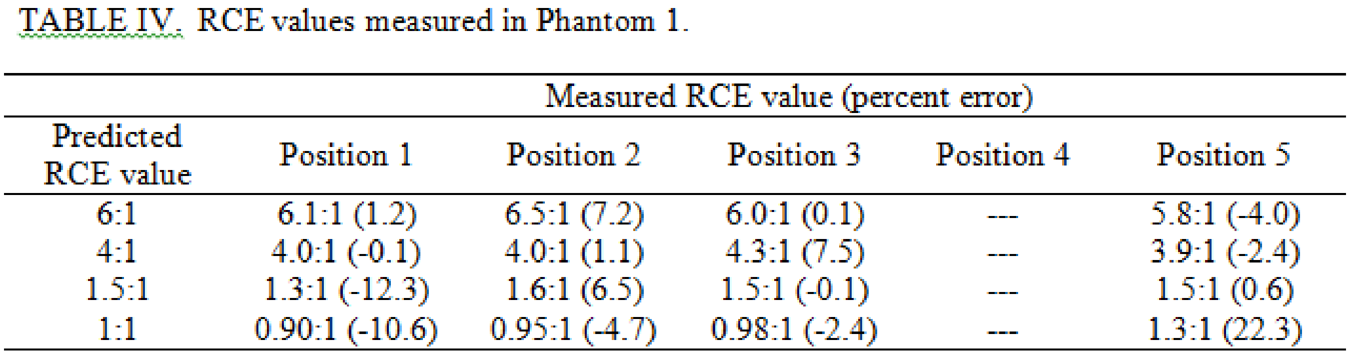


TABLE IV. RCE values measured in Phantom 1.

	Measured RCE value (percent error)				
Predicted RCE value	Position 1	Position 2	Position 3	Position 4	Position 5
6:1	6.1:1 (1.2)	6.5:1 (7.2)	6.0:1 (0.1)	---	5.8:1 (-4.0)
4:1	4.0:1 (-0.1)	4.0:1 (1.1)	4.3:1 (7.5)	---	3.9:1 (-2.4)
1.5:1	1.3:1 (-12.3)	1.6:1 (6.5)	1.5:1 (-0.1)	---	1.5:1 (0.6)
1:1	0.90:1 (-10.6)	0.95:1 (-4.7)	0.98:1 (-2.4)	---	1.3:1 (22.3)

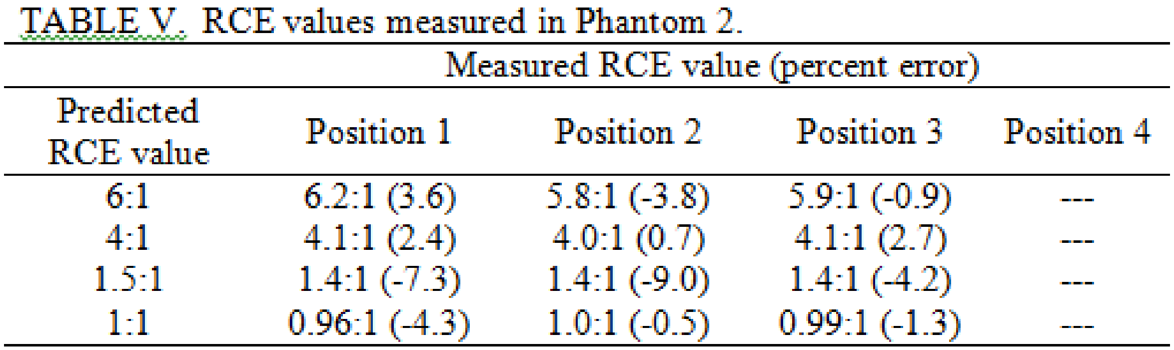


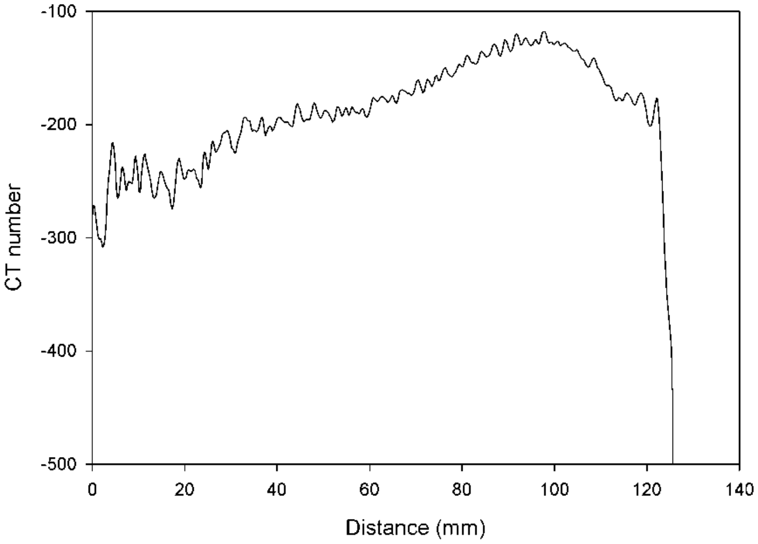
TABLE V. RCE values measured in Phantom 2.

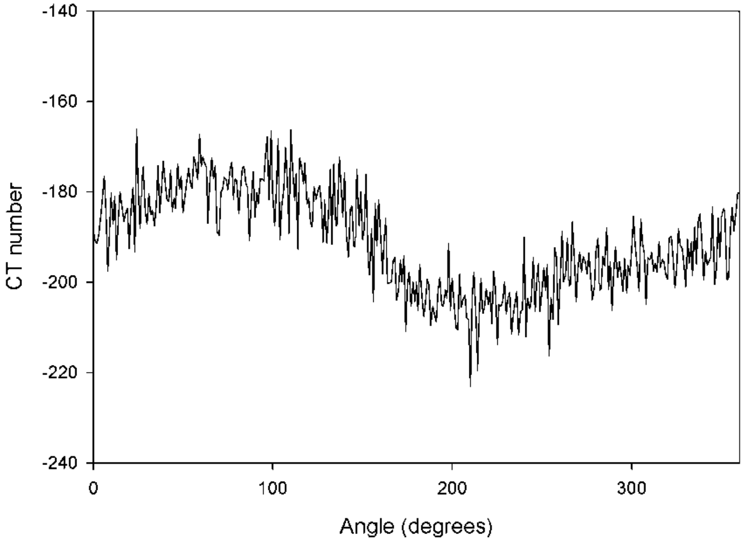
	Measured RCE value (percent error)			
Predicted RCE value	Position 1	Position 2	Position 3	Position 4
6:1	6.2:1 (3.6)	5.8:1 (-3.8)	5.9:1 (-0.9)	---
4:1	4.1:1 (2.4)	4.0:1 (0.7)	4.1:1 (2.7)	---
1.5:1	1.4:1 (-7.3)	1.4:1 (-9.0)	1.4:1 (-4.2)	---
1:1	0.96:1 (-4.3)	1.0:1 (-0.5)	0.99:1 (-1.3)	---

Jones AK and Mahvash A. Evaluation of the potential utility of flat panel CT for quantifying relative contrast enhancement. Med Phys 39:4149-4154, 2012.

Uniformity

- Beam hardening, scatter, incomplete rotation, lack of bowtie filter, etc. all mean that signal (and therefore CT number) is non-uniform across the image

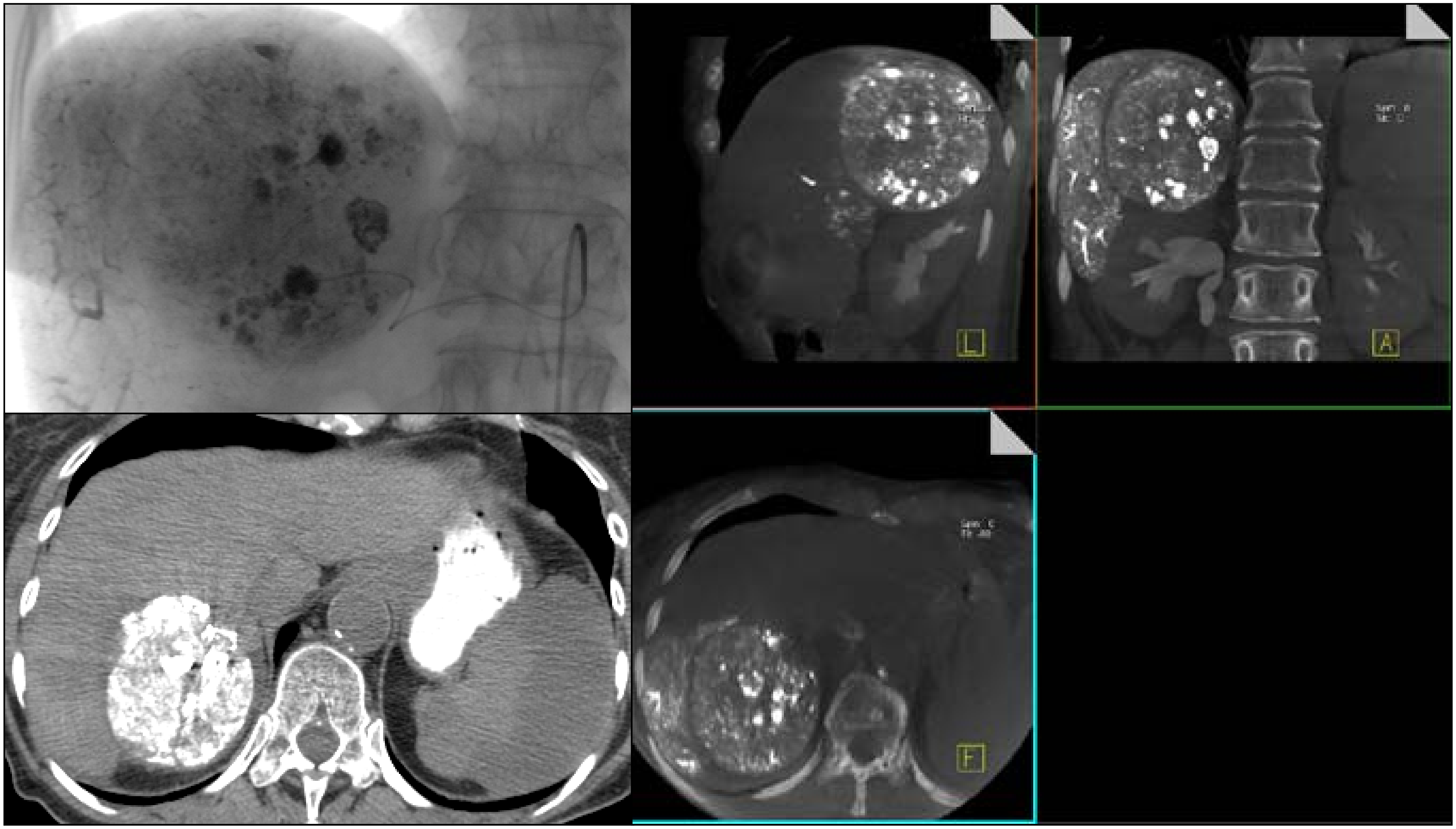




CLINICAL USE OF FPCT

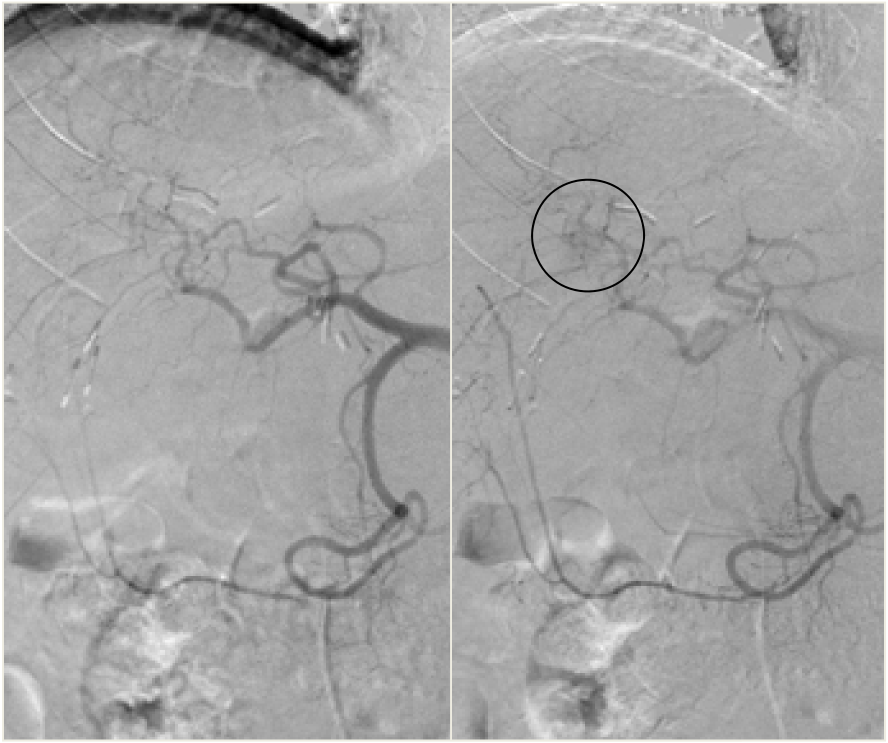
C-arm FPCT

Verification of therapeutic endpoint



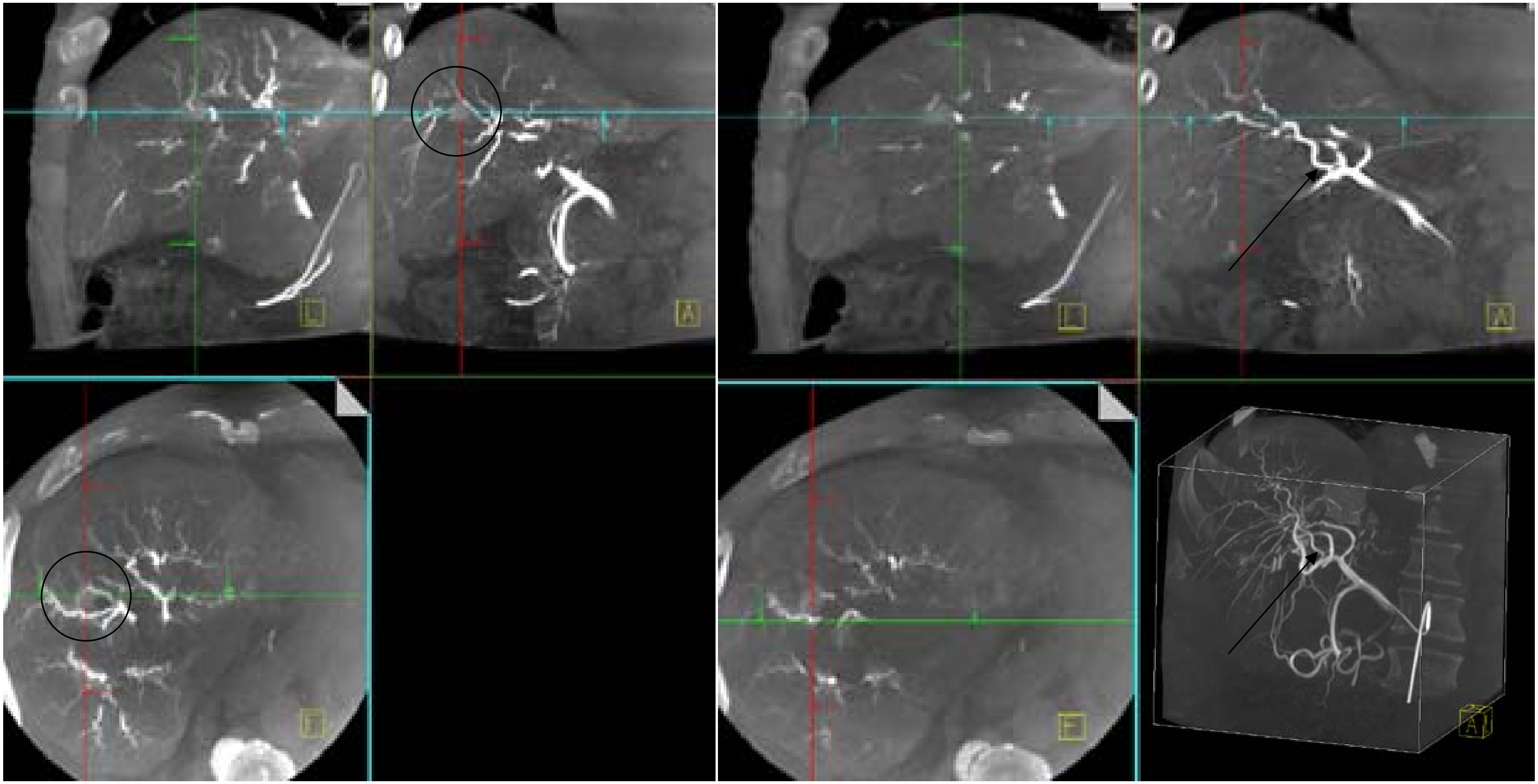
C-arm FPCT

Angiographically occult lesions



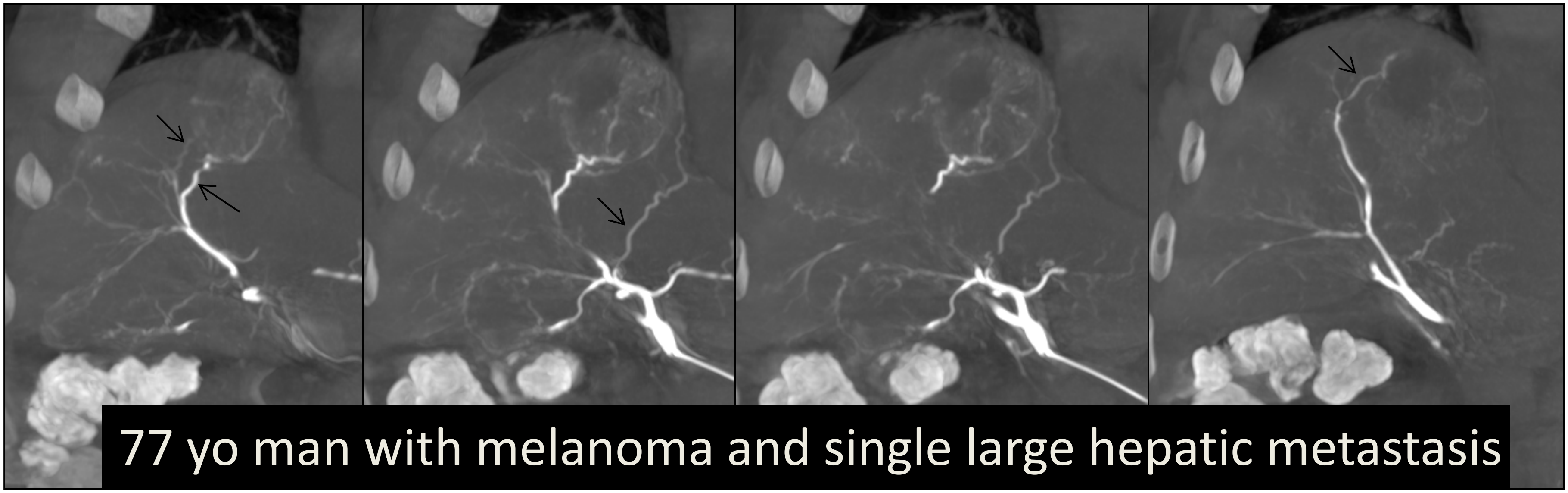
C-arm FPCT

Angiographically occult lesions



C-arm FPCT

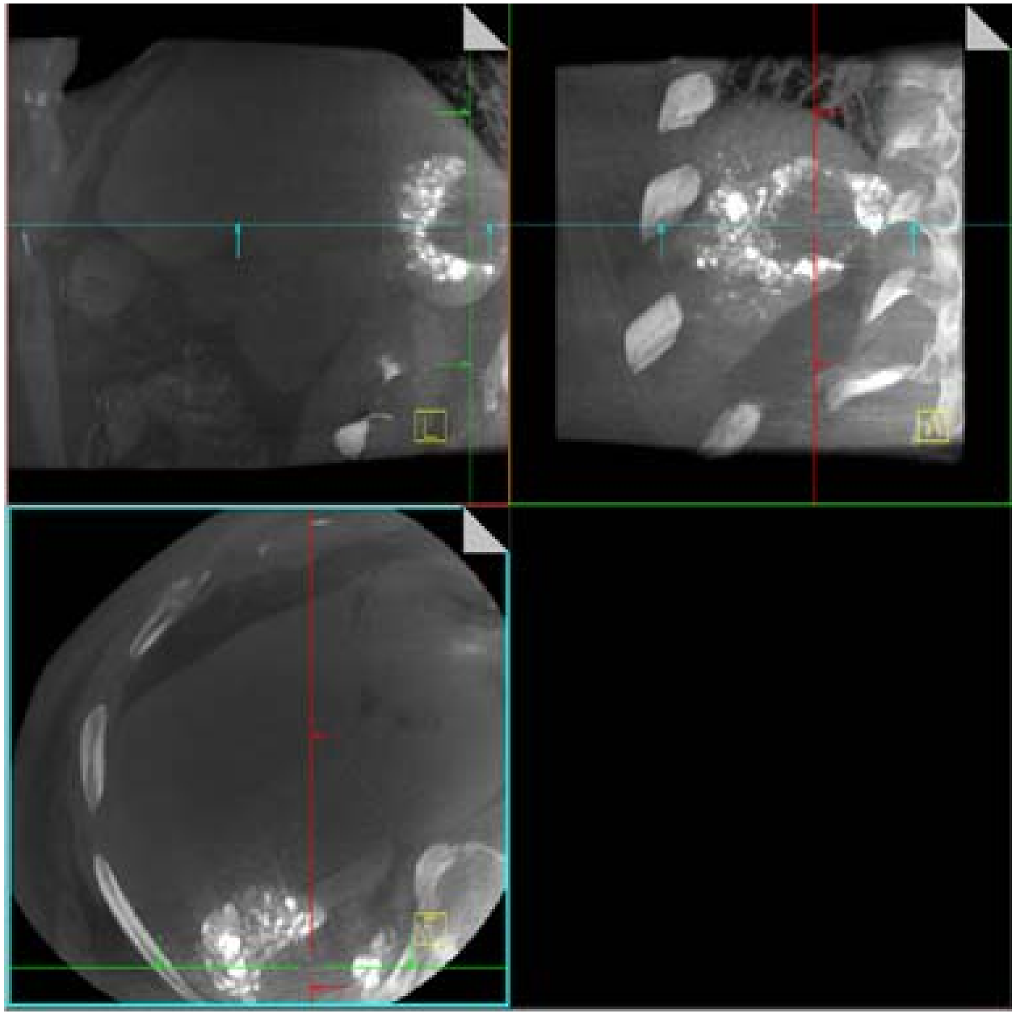
Pre-planning - Large lesions



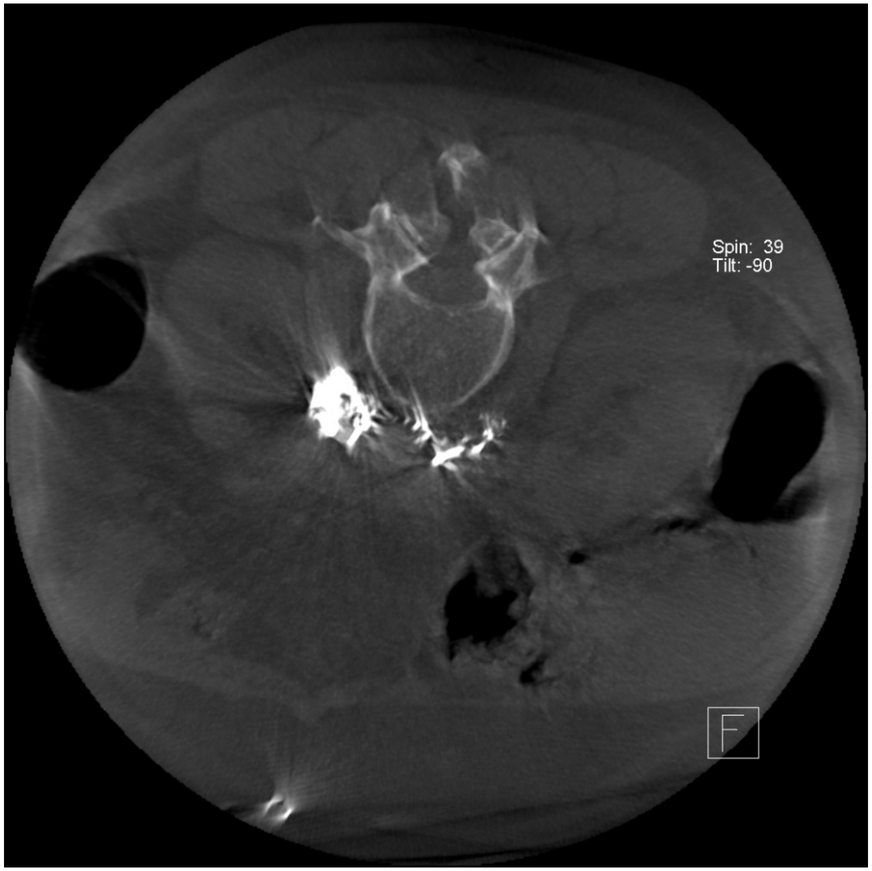
77 yo man with melanoma and single large hepatic metastasis

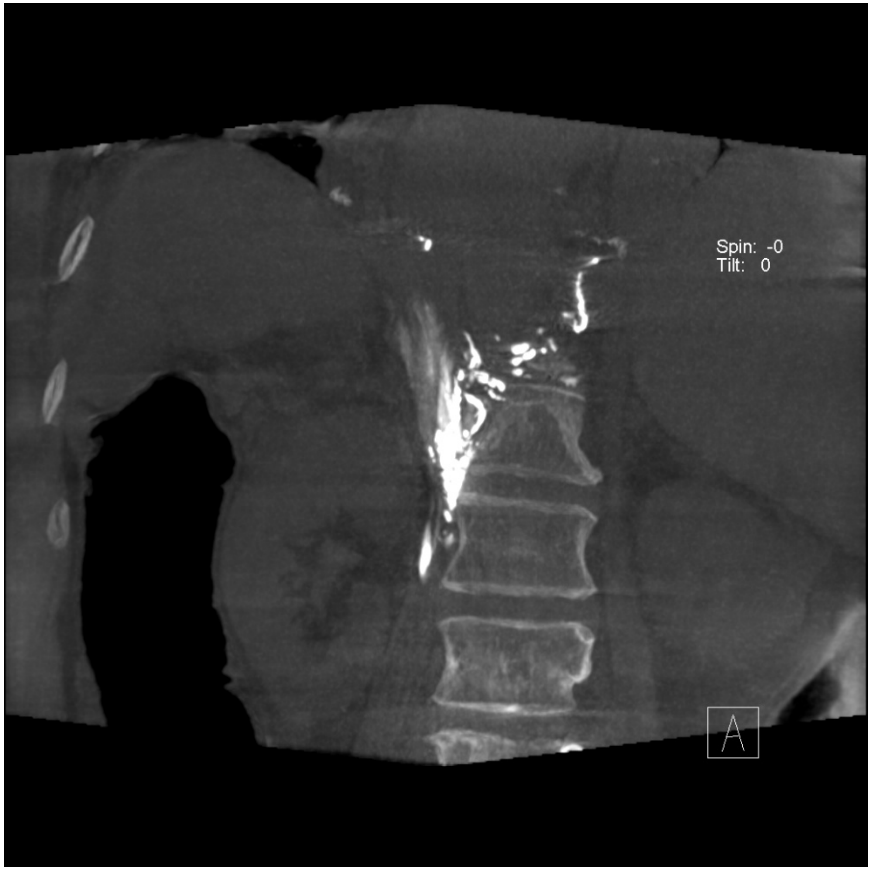
C-arm FPCT

Endpoint assessment - Incomplete therapy

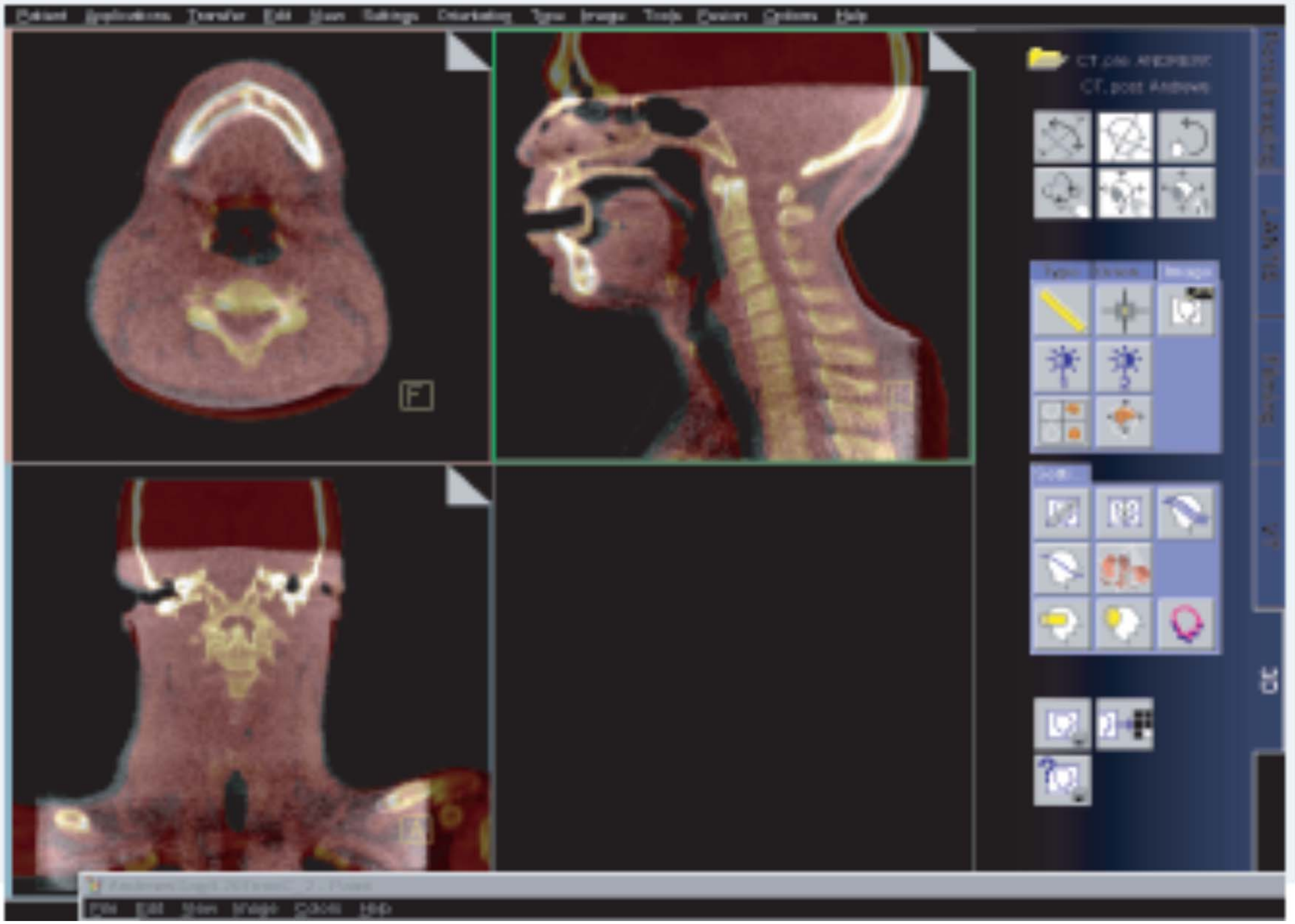


Thoracic duct embolization


Spin: 39
Tilt: -90
F


Spin: -0
Tilt: 0
A

RO alignment



Slide courtesy of Peter Balter Ph.D. and M Kara Bucci MD

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