Estimating Patient Radiation Dose from Computed Tomography

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Advances in Technology ...

- Helical CT: High output X-ray tubes, continuous gantry rotation/table motion
- MDCT: Over past 15 years total detector rows and beam width have increased
  - 2, 4, 8, 10, 16, 20, 32, 40, 64, 128, 256, and now 320 slice scanners
- Beam width up to 16cm
- Speed – gantry rotation time down to 1/3 sec

→ ...

Increasing scan speed & anatomical coverage

- Improved temporal resolution “freezes” physiologic motion
  - Scan entire chest in a breath hold (10 sec)
  - Scan entire heart in a heart beat (1 rotation)
- Growing number of clinical applications
  - Emergency room trauma scanning
  - Cardiac applications
  - Perfusion studies (repeated scans in one location)
  - Oncology treatment planning
- (… and billing opportunities)

→ …

More pediatric applications

- Fast scanners allow for pediatric CT imaging that previously may have required anesthesia, etc.
- 200% increase in pediatric CT over last few years

D.Frush

- Concerns
  - ↑ radio-sensitivity
  - ↑ organ and effective doses, particularly when technical factors are not adjusted
- 600,000 annual CT scans on children under 15 might result in 500 additional cancer deaths from IR

D.Brenner- AJR

CT dose: A concern?

- As of 2006, 62 million CT procedures performed annually

AAPM96/IMV Report in CT

- In U.S., CT comprises only 15% of all exams but generates 70% of delivered dose dose

Mettler 2003

- Diagnostic X-ray continues to increase in proportion of the population's total exposure to I.R.
  - 11% of total burden in 1980 to 17% in 2000
  - Up to 60% of manmade exposure
- Proportion of that burden due to CT has dramatically increased in last two decades
  - Growing at rate of 10-15% annually

NEXT – CT
Shrimpton & Edyvean
Mettler

CT as a Screening Tool?

- Risk vs. benefit
  - when scanning a symptomatic patient to render a diagnosis

    vs.

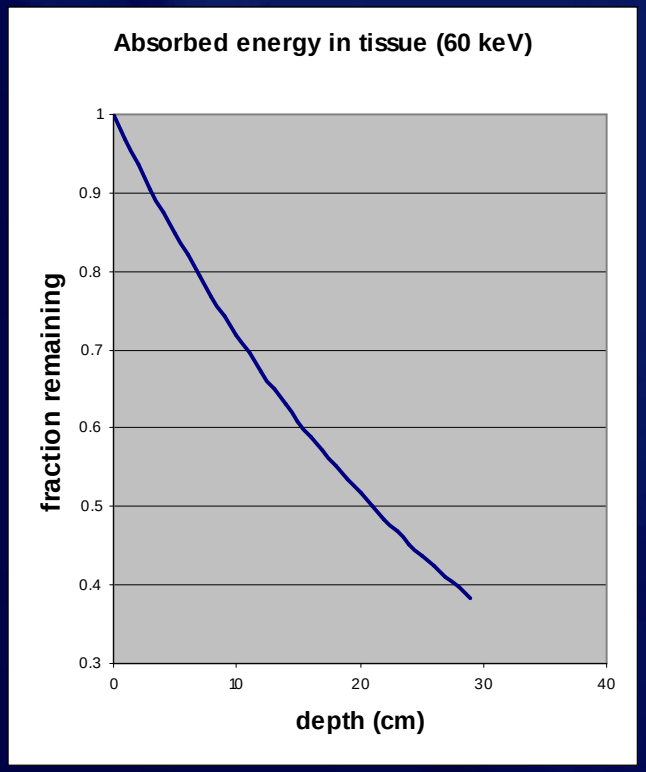
  - When used as a screening tool on an asymptomatic population

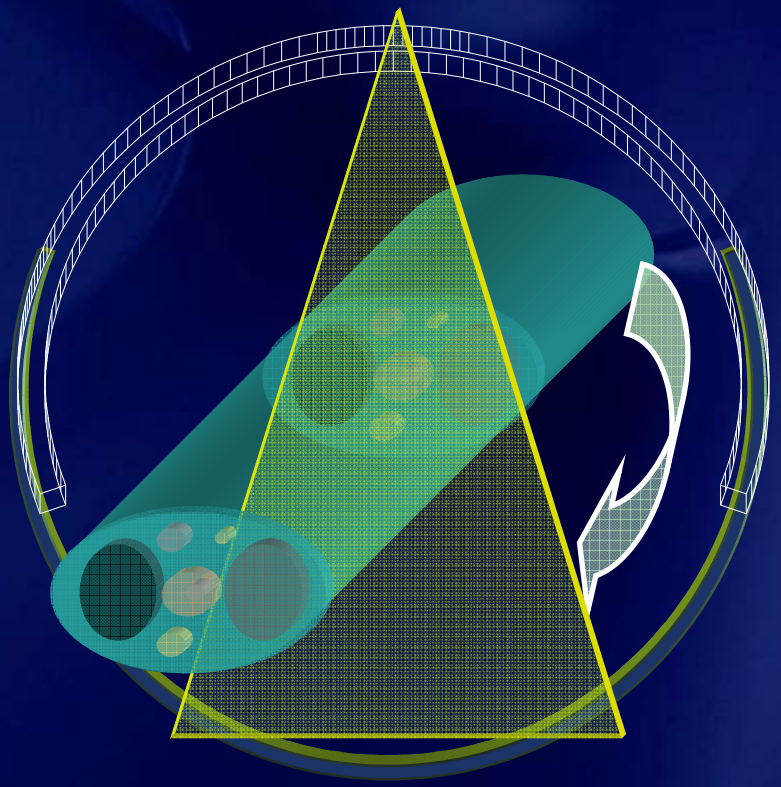
Federal Stance *(2005)*

- X-rays officially inducted to FDA list of carcinogens
- Sanction of the no-threshold model
- No safe dose of radiation
- Any increase in dose increases risk

CT Dose distribution differs from projectional

Unlike conventional projectional radiography where beam exit energy is a fraction of entrance, in CT rotating source encircles the body so that PA and AP entrance dose are nearly identical leading to a more uniform dose distribution and *generally higher organ dose*





Typical organ doses from diagnostic x ray examinations

Examination	Relevant organ	Relevant organ dose (mGy)
Dental x ray	Brain	0.005
PA Chest x ray	Lung	0.01
Lateral chest x ray	Lung	0.15
Screening mammogram	Breast	3
Adult abdominal CT*	Stomach	11
Adult head CT*	Brain	13
Neonate abdominal CT*	Stomach	25
Neonate head CT*	Brain	65

* *On average, there are two CT exams per treatment course*

* *CT doses estimated based on a 200 mAs setting*

CT dose paradigm

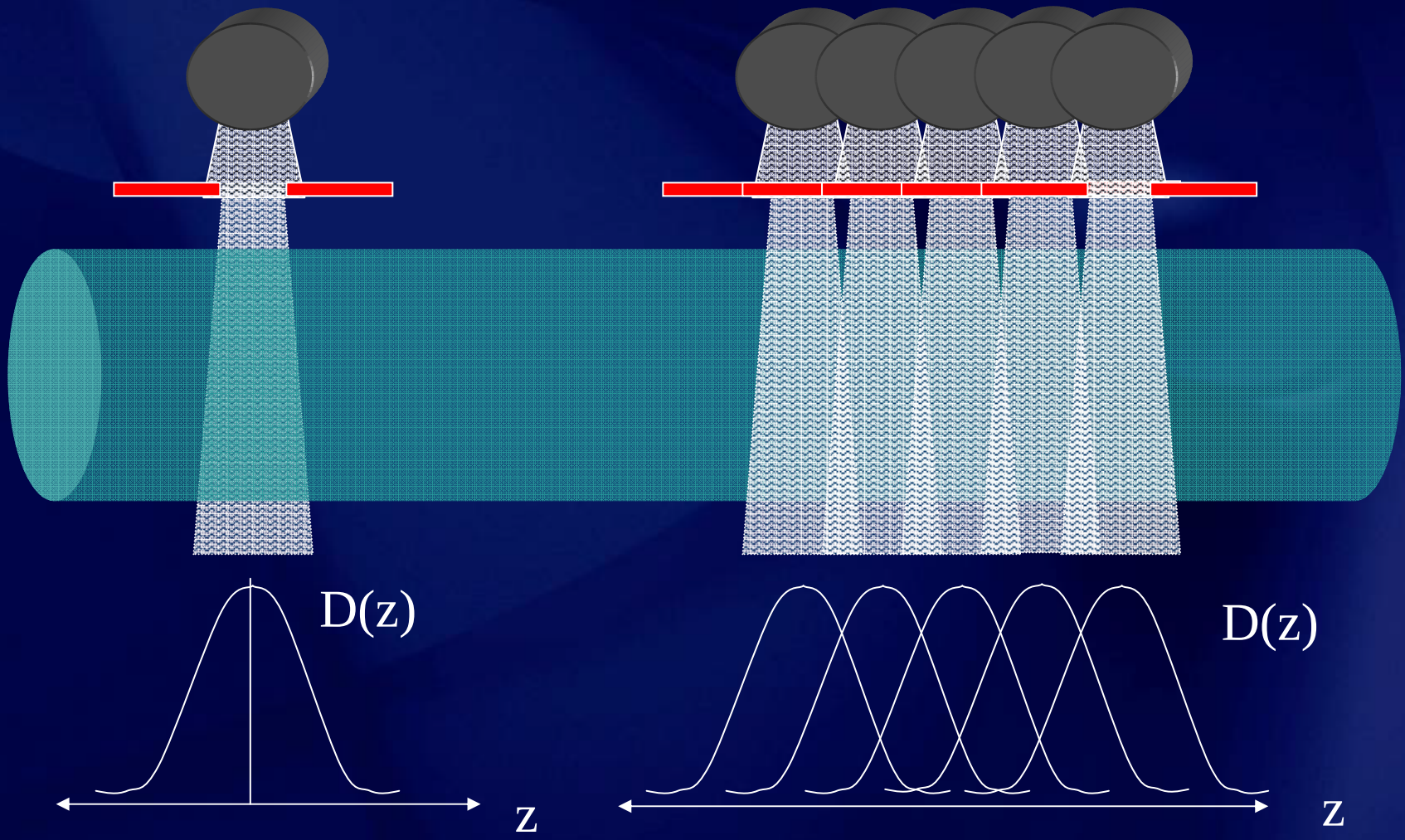
The local tissue dose from a *single* slice is *not* the same as the dose in the very same tissue when additional adjacent slices are made.

... because each additional slice scatters radiation into adjacent slices.

- D(z) = dose profile along z-axis from:

Single slice acquisition

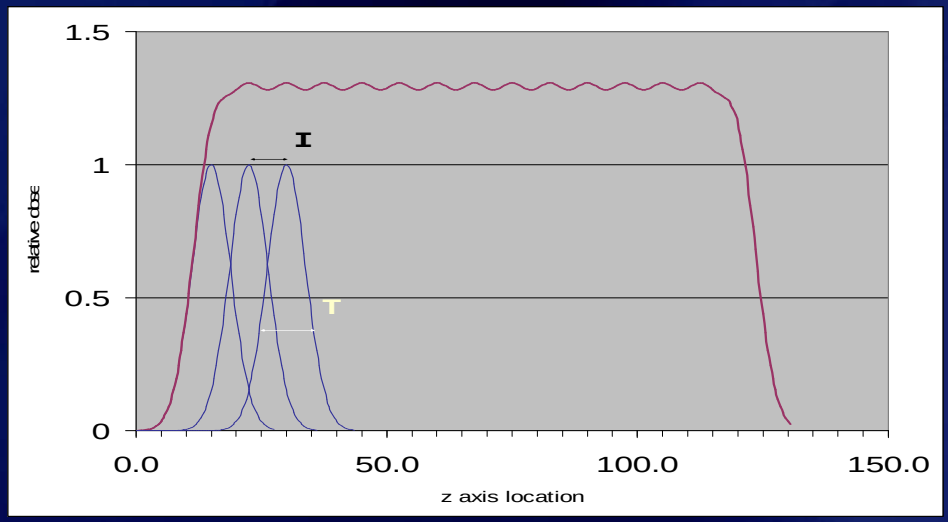
Multiple slice acquisition



Dose from multiple slices

- Even if non-overlapping slices (and ignoring beam penumbra) scatter tails of multiple contiguous scans overlap and contribute to an increased integral dose profile
- Function of:
  - Single Scan Profile Width (T)
  - Number of scans (N)
  - Spacing (I)

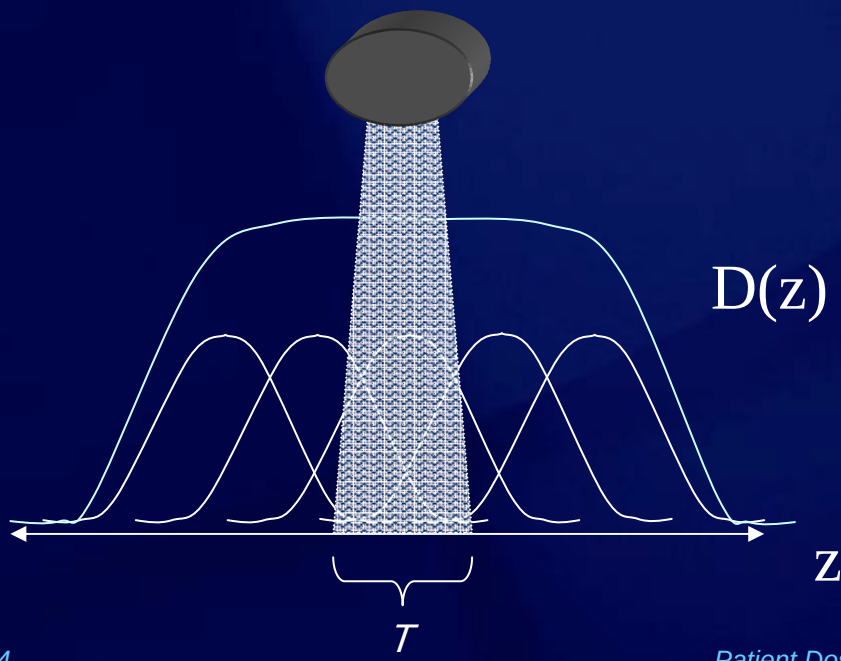
Case for I = 0.5 T →

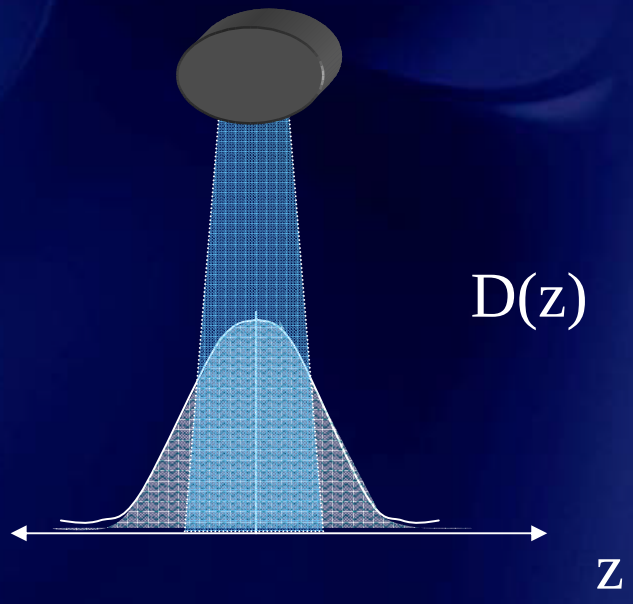


Computed Tomography Dose Index (CTDI) – defined

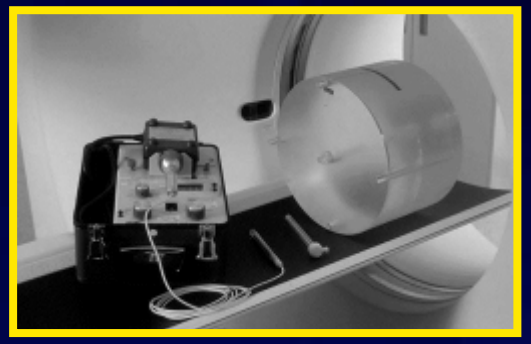
$$CTDI = (1/T)\int_{-\infty}^{\infty} D(z)dz$$

Total area of D(z) *under width T* of central scan of multiple scan profile = total area of single scan dose profile (including scatter tails)

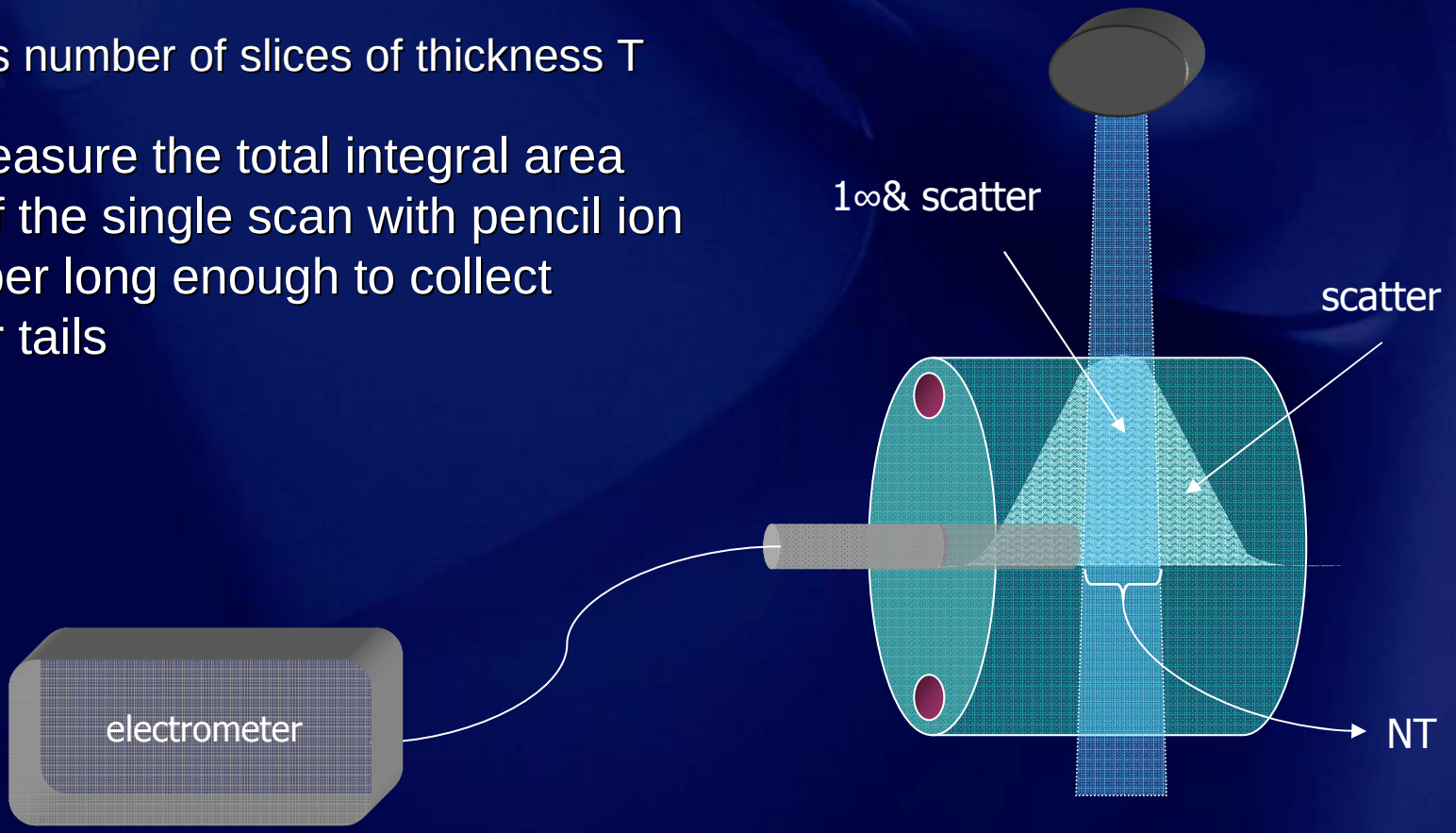




Measuring CTDI



- Single axial scan (in phantom to emulate patient scatter) of nominal *beam* thickness NT where:
  - N is number of slices of thickness T
- We measure the total integral area D(z) of the single scan with pencil ion chamber long enough to collect scatter tails



CTDI – FDA

- FDA defined CTDI over 14 slices (n is the number of slices/acquisition)

- **$CTDI = (1/nT) \int_{-7T}^{7T} D(z)\, dz$**

- This assumed that you either had:
  - TLDs or film to measure D(z) profile ... OR
  - A 100 mm chamber covering 14 - 7mm slices
  - Can overestimate dose for thin slices

$CTDI_{100}$

- *Standardizes* integration limits across 100 mm chamber
  (rather than 14 slices of varying size)

$$CTDI_{100} = \left(\frac{1}{NT}\right)\int_{-5cm}^{5cm} D(z)dz \qquad \rightarrow \quad = (ELCf)/(NT)$$

where

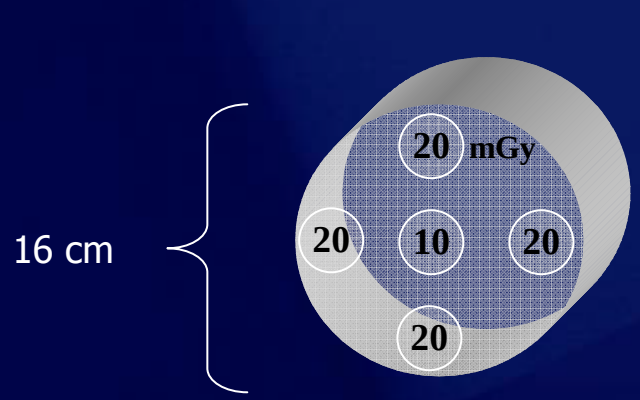
- E = measured value of integrated exposure
- L = active length of pencil ion chamber, typically 100 mm
- f = conversion factor from exposure to dose
- C = electrometer calibration factor - typically close to1.0

- N = *actual* number of data channels used during one axial scan
- T = nominal slice width of one axial image (scan collimation)

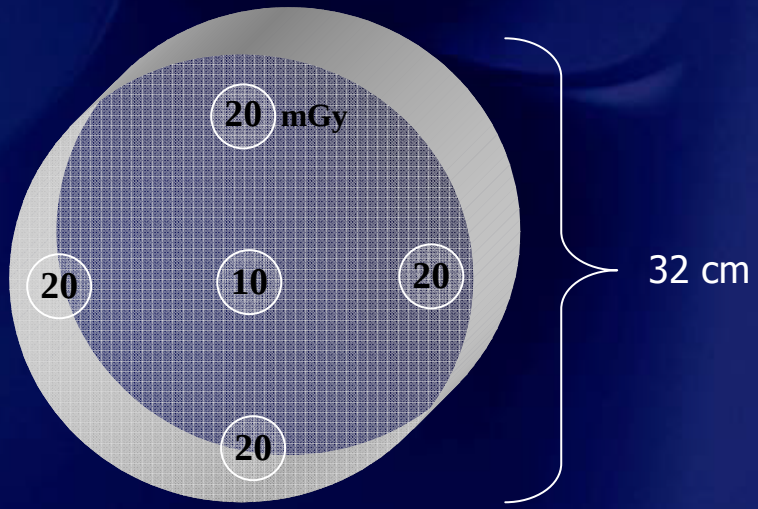
} *beam collimation*

$CTDI_{100}$

- $CTDI_{100}$ Measurements are done:
  - In Both Head and Body Phantoms
  - Using ONLY AXIAL scan techniques
    (CTDI = Area under the single scan dose profile)
  - At isocenter and at least one peripheral position in each phantom



Head



Body

Question: Dose to what?

When determining $CTDI_{100}$, one is calculating dose to …?

33% 1. air

20% 2. tissue

40% 3. acrylic

0% 4. none of the above

7% 5. $CTDI_{100}$ is not a function of material

What f factor to use?

- When determining $CTDI_{100}$, one is calculating dose to → *air*

→ Air - *f factor of 0.87 rad/R*

Tissue - *f factor of 0.94 rad/R*

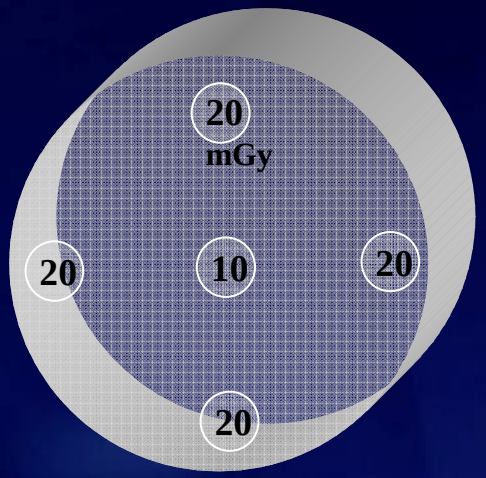
Acrylic - *f factor of 0.78 rad/R*

$CTDI_{FDA}$ and often what vendors report

AAPM Report 96: Measurement, Reporting, & Management of Radiation Dose in CT (Jan-08)

$CTDI_w$

- Due to attenuation CTDI is not homogeneous across the FOV – particularly in 32 cm body phantom where gradient between periphery and center is on order of 2:1
- To arrive at a single descriptive value we use a *weighted average* of center and peripheral $CTDI_{100}$



$$CTDIw = (1/3)\ CTDI_{100,\ center} + (2/3)\ CTDI_{100,\ peripheral}$$

$CTDI_w$ Limitations

$CTDI_w$ does describe dose within a *single x,y scan plane*

- measured for a single *axial* scan
- does reflect scatter contributions from adjacent scans
- assumes contiguous, non-overlapping scans

$CTDI_w$ *does not* ...

- describe dose for helical scans *(though close for pitch =1)*
- do well for non-adjacent slices frequently seen in helical scanning where X-ray beam may overlap (common with MDCT) or where extended pitch leaves gaps between rotations.

Volume CTDI ($CTDI_{vol}$)

- Calculated from CTDIw
- Represents the *average dose* in the central region of a multiple scan exam
- Averages over x, y *and z*
- Accounts for helical pitch

$$CTDI_{vol} = \frac{1}{pitch} \bullet CTDI_{w}$$

PITCH = table index per rotation (I) / *total* nominal scan width (NT)

Question: *Which of the following CT scan protocols - all made on different scanners with similar measured $CTDI_{100}$ values - would you expect to result in the highest $CTDI_{vol}$?*

(All are done at 120 kVp and identical beam collimations)

0% 1. mA = 400, rotation time = 0.5 sec, pitch = 1

17% 2. mA = 100, rotation time = 1 sec, pitch = 0.5

17% 3. Effective mAs = 200, pitch = 0.5

17% 4. Effective mAs = 200, pitch = 1

50% 5. There is no difference in $CTDI_{vol}$ for these protocols

Not all scanners provide same information to user: Siemens & Philips use concept of “effective mAs” which already takes pitch into account

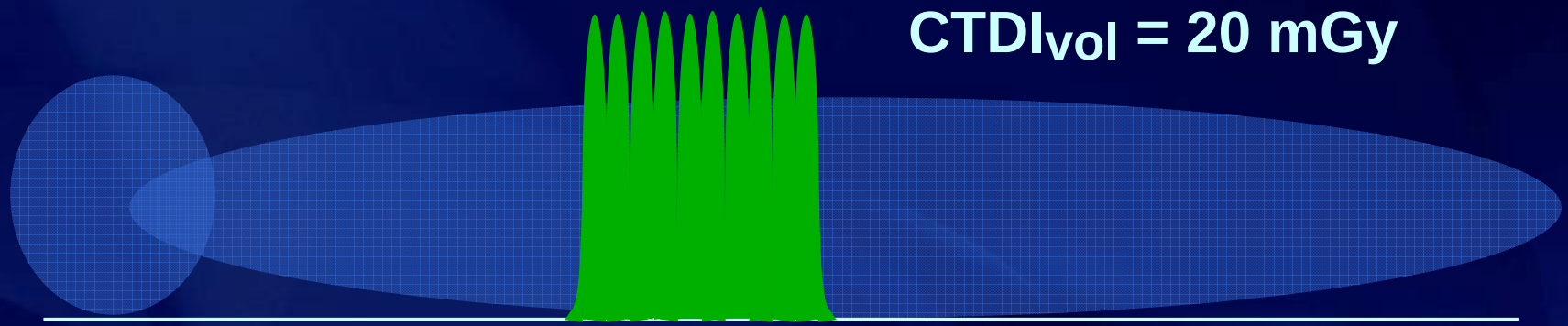
Effective mAs = (mA sec)/pitch*

1. mA = 400, rot. time = 0.5 sec, pitch = 1: *eff. mAs = (400*0.5)/1 = 200*
2. mA = 100, rot. time = 1 sec, pitch = 0.5: *eff.mAs = (100*1)/0.5 = 200*
3. Effective mAs = 200, pitch = 0.5: *eff.mAs = 200*
4. Effective mAs = 200, pitch = 1 *eff.mAs = 200*

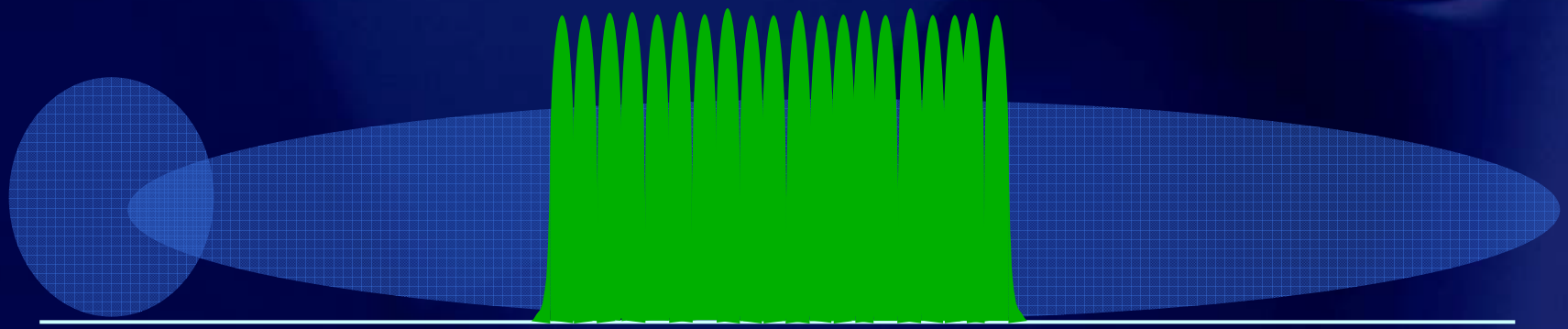
5. There is no difference in CTDIvol for these protocols – *all have the same effective mAs*

CTDI Limitations

- Says nothing about *length* of scan ...



$CTDI_{vol}$ = 20 mGy



$CTDI_{vol}$ *still* = 20 mGy

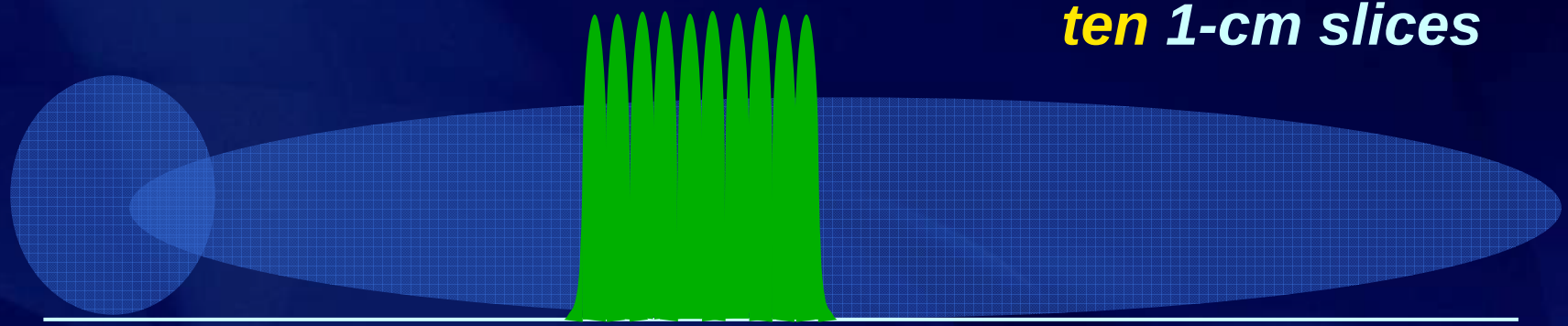
How do we represent the greater biologic risk?

Dose Length Product (DLP)

- Represents *integrated dose* in terms of total scan length (# slices • slice width)
- DLP = $CTDI_{vol}$ (mGy) • scan length (cm)
- DLP reflects total energy absorbed

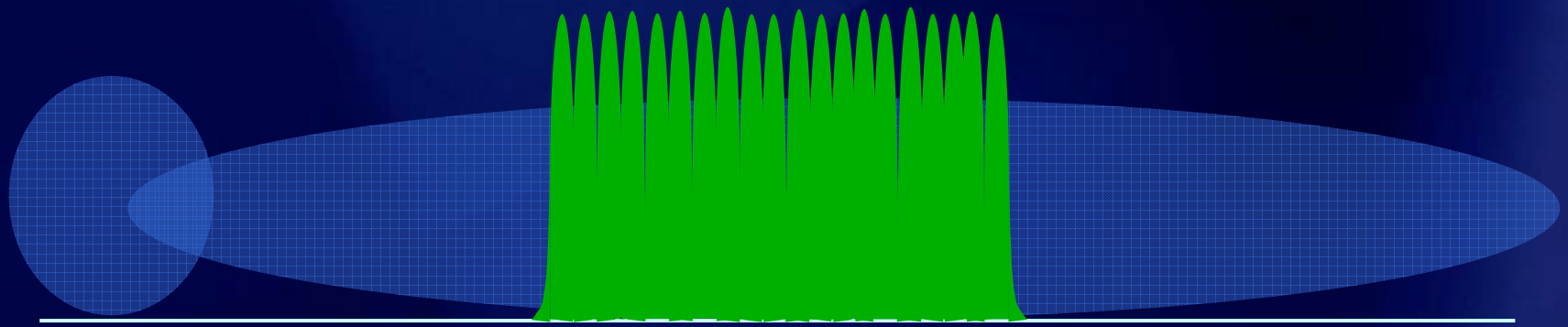
DLP = 200 mGy•cm

$CTDI_{vol}$ = 20 mGy
ten 1-cm slices



DLP = *400* mGy•cm

$CTDI_{vol}$ *still* = 20 mGy
twenty 1-cm slices



DLP from $CTDI_{vol}$

Table 1. Illustrative values for $CTDI_{vol}$ and DLP for common CT exams for (a) 4-channel MDCT and (B) 16-channel MDCT

Table 1a: 4-channel MDCT (120 kVp)

Exam	*Beam Collimation*	*Pitch*	*mAs per Rotation*	*Scan Length (cm)*	*$CTDI_{vol}$ (mGy)*	*DLP (mGy-cm)*
Head	4 x 2.5	Axial	250	15	55.0	825
Chest	4 x 5	0.75	100	40	12.0	480
Abdomen	4 x 5	0.75	150	20	19.1	382
Abdomen ≈ & Pelvis	4 x 5	0.75	150	40	19.1	764

Table 1b: 16-channel MDCT (120 kVp)

Exam	*Beam Collimation*	*Pitch*	*mAs per Rotation*	*Scan Length (cm)*	*$CTDI_{vol}$ (mGy)*	*DLP (mGy-cm)*
Chest	16 x 1.25	0.938	150	35	13.3	466
Abdomen	16 x 1.25	0.938	212	28	18.8	526
Pelvis	16 x 1.25	0.938	212	25	18.8	470

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Problems with CTDI methodology

- Assumes entire scatter tails are captured in their entirety by the chamber
- In body phantom, $CTDI_{100}$ underestimates MSAD (for pitch 1) by approximately 30% *(Boone)*
- The wider the collimation the worse the problem *(MDCT 25 & 30 mm beam widths now common, 160 mm now being introduced)*

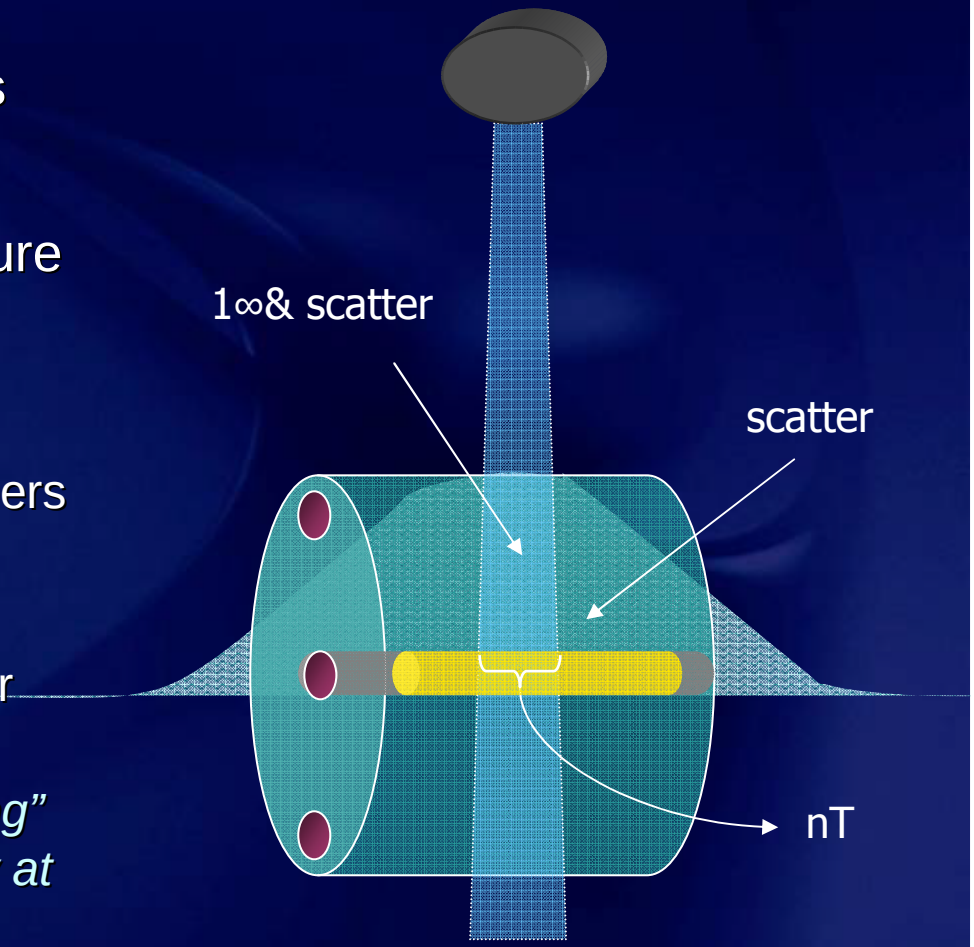
Problems with CTDI methodology

- Chamber is 100 mm & phantom is 150 mm length
- Chamber under-reports: can capture primary, but scatter tails are lost
- How to measure wider beams?
  - Longer phantoms & pencil chambers

    Expensive & impractical

  - Multiple scans with small chamber

    May be sensitive to dose inhomogeneities or "dose striping" from diverging beam (especially at periphery) and tube position



CTDI Limitations: Increasing beam width …

Courtesy K. Geleijns

(Leidin University, The Netherlands)

Courtesy K. Geleijns

(Leidin University, The Netherlands)

- 256 Channels
- CT Body Phantom
- 150 mm Long Phantom is too small ….
- 100 mm Pencil Chamber is too small

$CTDI_{100}$

$CTDI_{300}$

- 256 Channels
- CT Body Phantom
- 350 mm Long Phantom
- 300 mm Pencil Chamber

Correction factors …

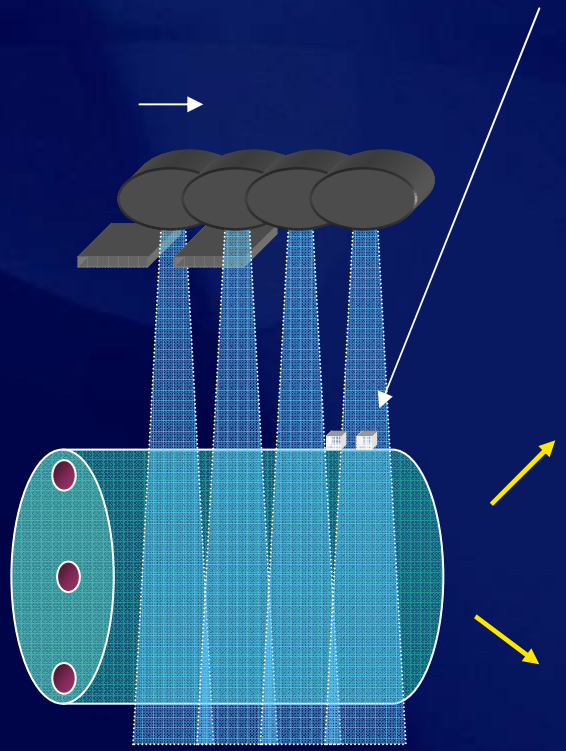
Courtesy K. Geleijns

(Leidin University, The Netherlands)

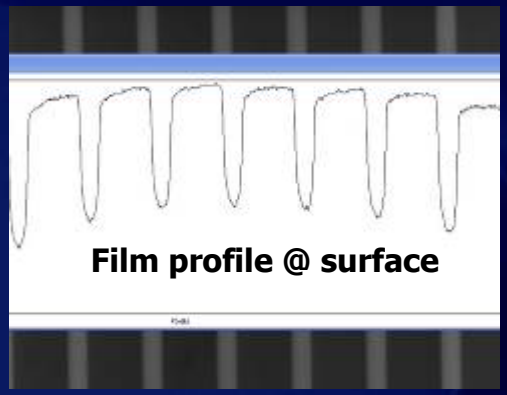
Average dose (D) in a 150 mm long body phantom measured with a 100 mm CT chamber relative to $CTDI_{300}$ (CTDI)

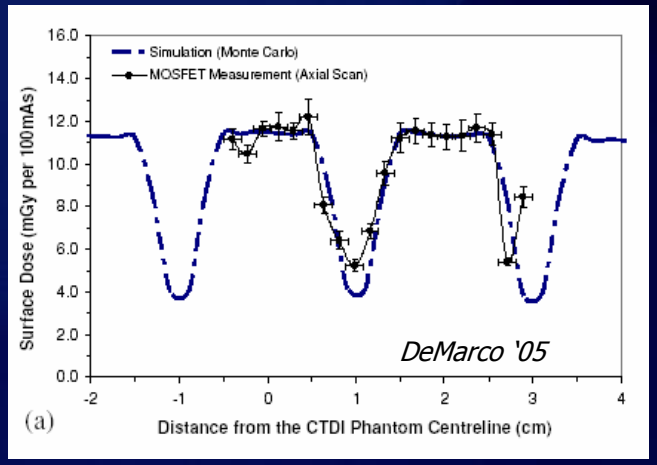
kV	Wedge	Phantom	Center	Periphery	Weighted
			D/CTDI	D/CTDI	**D/CTDIw**
80	**L**	**body**	69	104%	**97%**
80	**M**	**body**	69	103%	**96%**
120	**L**	**body**	67	99%	**91%**
120	**M**	**body**	67	99%	**91%**

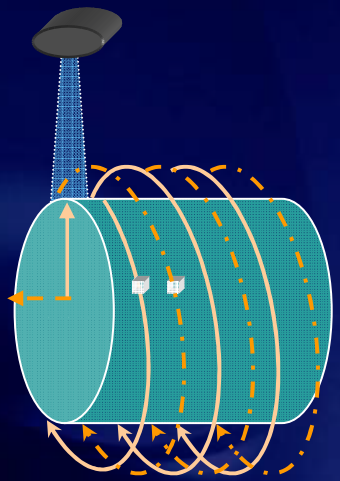
Small volume measurement devices can be sensitive to non-uniform CT radiation pattern – particularly at surface



32 cm CTDI phantom
contiguous axials







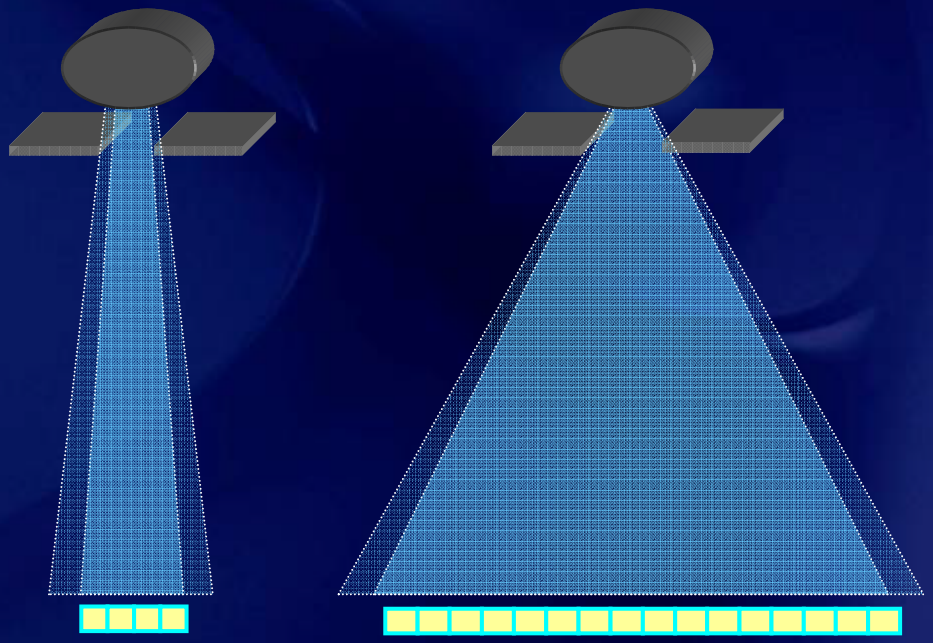
As well as *phase* in helical scans *(position of tube at start of scan)*

MDCT dose efficiency: Over-beaming

For each detector row in MDCT to see same radiation intensity, all must be located in the umbra region of beam, the penumbra (which could be integrated into the signal of a single detector) is not utilized and adds to patient dose

Percentage loss greatest with relatively few detector rows

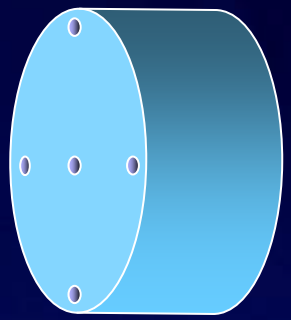
Cody, et al. Report of CT dose for MDCT scanners used in NLST - suggests improved dose efficiency in more complex scanners.



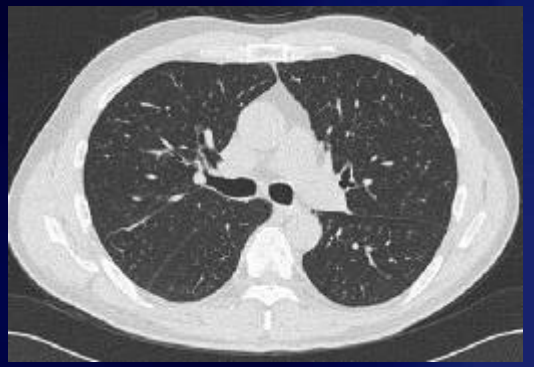
4 x 1mm

16 x 1.25 mm

Phantoms vs. Patients



- Phantoms easy to work with - symmetric, homogeneous, and standardized. Patients exhibit none of these features
- Not a good estimate for objects that vary in size and shape from reference cylinders
  - CTDI tends to overestimate dose for large patients and underestimate for small/pediatric patients
  - Without dose modulation, entrance dose increased in the lateral relative to the AP projection ($1/r^2$ source –skin distance)



What is the question being asked?

- CTDI may provide a useful benchmark for comparing scanners and protocols
- By itself CTDI is not a good estimate of organ dose or radiation risk
- To estimate risk we need to investigate dose to organs and Effective dose

Effective Dose (H_E)

- Effort to equate the partial body exposure of diagnostic X-ray to a whole body equivalent stochastic risk

$$H_E(Sv) = S\, w_T \text{¥} H_T\,(Sv)$$

- w_T values are tissue weighting factors that assign stochastic risk relative to radiating whole body

ICRP60 w_T values

gonads	0.2	stomach	0.12
colon	0.12	lung	0.12
bone marrow	0.12	breast	0.05
esophagus	0.05	bladder	0.05
liver	0.05	thyroid	0.05
bone surfaces	0.01	skin	0.01
remainder	0.05		

Determining Tissue/Organ Dose?

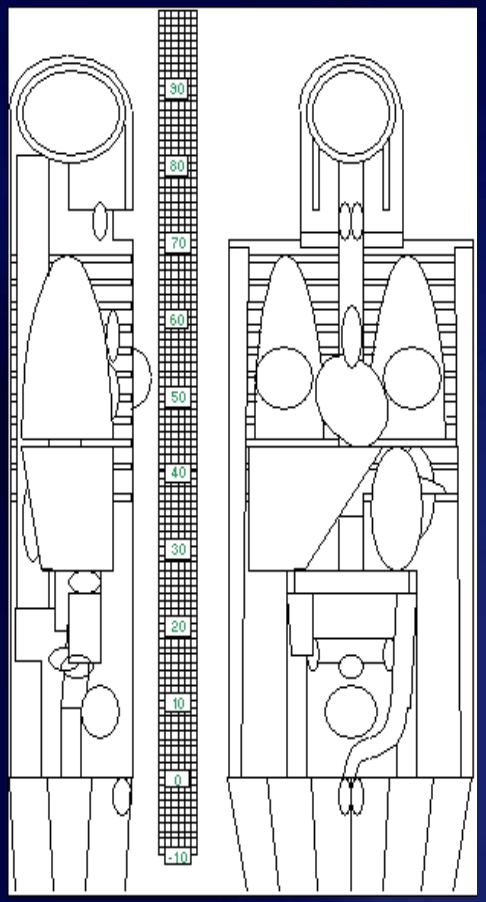
- TLDs and MOSFETs on anthropomorphic phantoms an improvement over homogeneous plastic cylinders, but ...
  - Absolute dose measurements require reference calibration of solid state detectors and appropriate energy corrections
  - “Point” measurements must be extrapolated to organs, whole body.
  - Measurements specific to a given phantom size, habitus, and composition not necessarily generalizable to actual patients

Organ dose with Monte Carlo

- Computation intensive method that tracks large numbers of individual photons through mathematical models of phantoms and patients and calculates dose deposition from individual photon/tissue interaction processes

Monte Carlo dose estimates of standard phantom model

- NRPB: Monte Carlo transport of CT spectra through MIRD phantom
- Based on mathematically described organ models of a standard, hermaphroditic, adult of given tissue composition.
- Dose estimates based on original, contiguous axial scan data from earlier scanners but can accept input for helical, modern MDCT protocols with scanner “matching” factors.
- Input CT model, scan protocol, and scan extent and database generates table of organ doses and an effective dose.
- Multiple methods have evolved or are derived from this method *(Huda, Atherton, ImPACT, LeHeron, Kalendar)*



Normalized effective dose (k) coefficients

- European Working Group - CT Quality guidelines
- Set of factors relating Monte Carlo based organ dose to DLP values
- Standardizes CT dose reporting for typical scans
- Thus effective dose can be estimated (within 10%) directly from DLP values (which *can be measured* in phantom and/or obtained from scanner console)

Effective dose [a] (k) * DLP

Table 3. Normalized effective dose per dose-length product (DLP) for adults (standard physique) and pediatric patients of various ages over various body regions. Conversion factor for adult head and neck and pediatric patients assume use of the head CT dose phantom (16 cm). All other conversion factors assume use of the 32-cm diameter CT body phantom[78,79]

Region of Body	*k (mSv mGy^{-1} cm^{-1})*				
	0 year old	*1 year old*	*5 year old*	*10 year old*	*Adult*
Head and neck	0.013	0.0085	0.0057	0.0042	0.0031
Head	0.011	0.0067	0.0040	0.0032	0.0021
Neck	0.017	0.012	0.011	0.0079	0.0059
Chest	0.039	0.026	0.018	0.013	0.014
Abdomen ≈& pelvis	0.049	0.030	0.020	0.015	0.015
Trunk	0.044	0.028	0.019	0.014	0.015

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Example: "*Low dose*" Chest protocol

- Protocol: 120 kVp, 80 mA, 0.5 sec. rot., 4 x 2.5 beam width
- 100mm pencil chamber at center position of 32cm phantom reading: 26.7 mR
- Center $CTDI_{100}$ = (ELCf)/(NT) = 2.3 mGy

$$\frac{(26.8mR)(100mm)(1.0)(0.87rad/R)}{(4)(2.5mm)}\sum\left(\frac{1R}{1000mR}\sum\frac{10mGy}{rad}\right)=2.3mGy$$

- Similarly, $CTDI_{100}$ at 12:00 = 4.7 mGy
- $CTDI_w$ = (2/3 center + 1/3 periphery) = 3.9 mGy

Example: *Low dose* Chest protocol

- Applying the measured axial CTDI values to a helical protocol where the table incrementation is 15mm/rotation
- $CTDI_{vol} = CTDI_w/pitch = CTDI_w \times (NT)/I = CTDI_w\ (4 \times 2.5/15)$

  $= (3.9\ mGy)\ (0.667) =$ 2.6 mGy
- Assuming a 35 cm long chest:

  DLP = (2.6 mGy)(35) = 91 mGy-cm
- And using the k factor for an adult chest of 0.014:

  Effective dose = (91)(0.014) = 1.3 mSv

Limitations of current dose determination methods

- No specific modeling of MDCT
  - *Schmidt & Kalendar recently modeled Siemens Vol. Zoom*
- MDCT have broader beam widths, different shape & composition of bow-tie filters, shorter focal to isocenter distances
- Helical scanners have variable pitch (0.5-2) and introduce non-contiguous slices
- Recent advent of tube current modulation - both in plane and longitudinally along gantry axis - requires specific modeling techniques to simulate

Limitations

- Extrapolating calculated organ dose estimates for mathematical models of a standard man to actual patients is problematic
- Actual patient size & morphology can have significant impact on dose *(Huda, Cody)* as can age, gender, and ethnicity

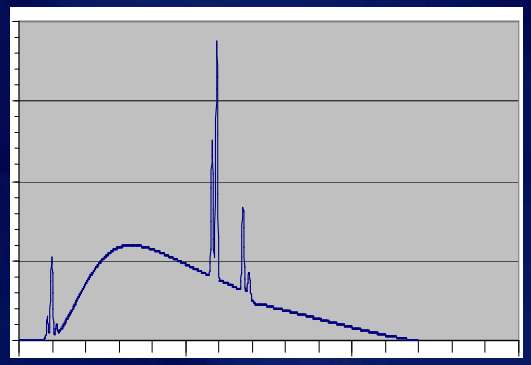
Next step →

Specific scanner, scan protocol, and patient modeling

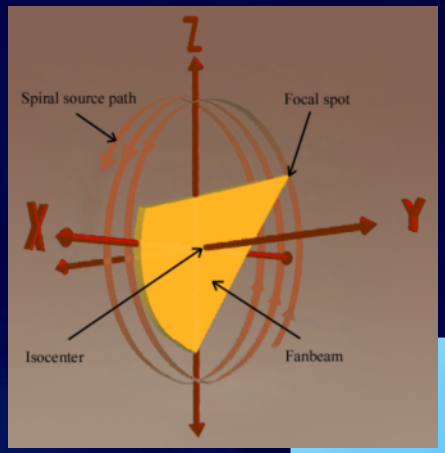
Monte Carlo simulations that model current MDCT scanners in helical (or axial) mode transporting photons through voxelized models of real patients

Models that explicitly specify scanner & scan geometry including:

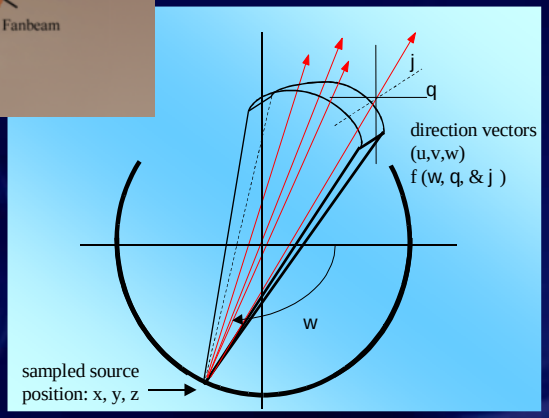
- Source to isocenter
- Beam collimation
- Beam path (helical, non contiguous, etc.
- Beam spectrum (vendor supplied)
- Bow-tie equalization filter (modeled on proprietary vendor information of shape and composition)

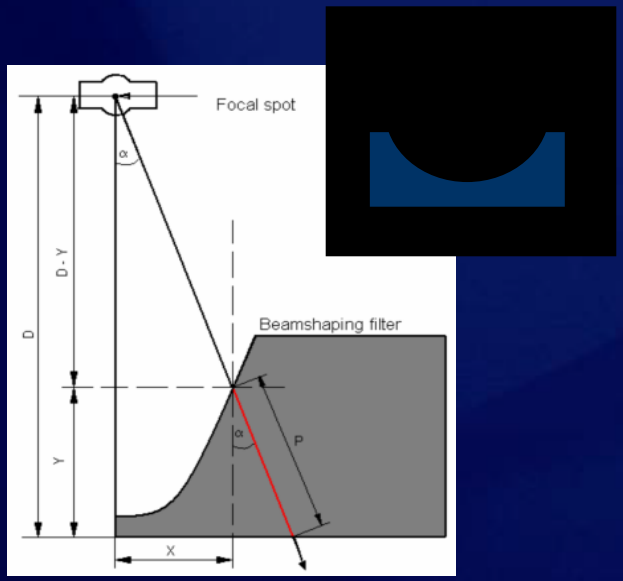


CT X-ray spectra



Collimation & scanner geometry





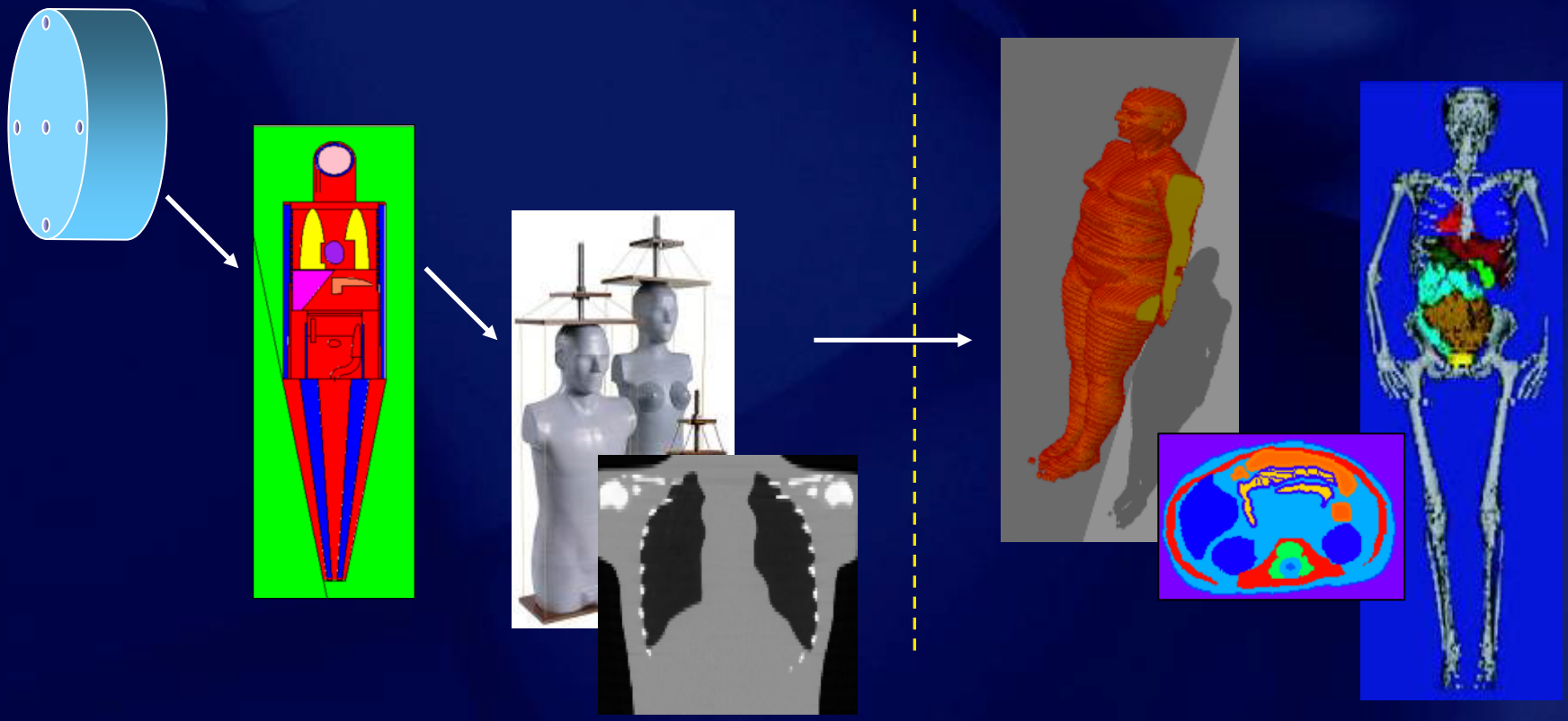
Bow-tie filter

CT source:
User defined,
Scanner & Protocol Specific

Monte Carlo Simulation Application: Evolution of Phantom Dosimetry

Model and benchmark against "conventional" dosimetry phantoms

Calculate irradiation patterns, organ dose, whole-body effective dose in patient-specific voxelized phantoms



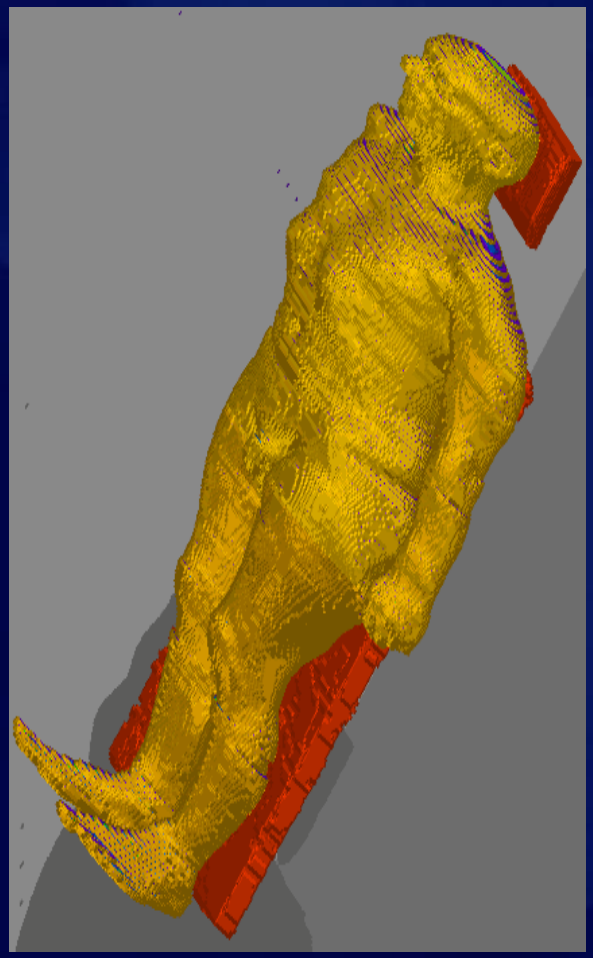
Patient Based Phantoms

Voxelized Patient Models

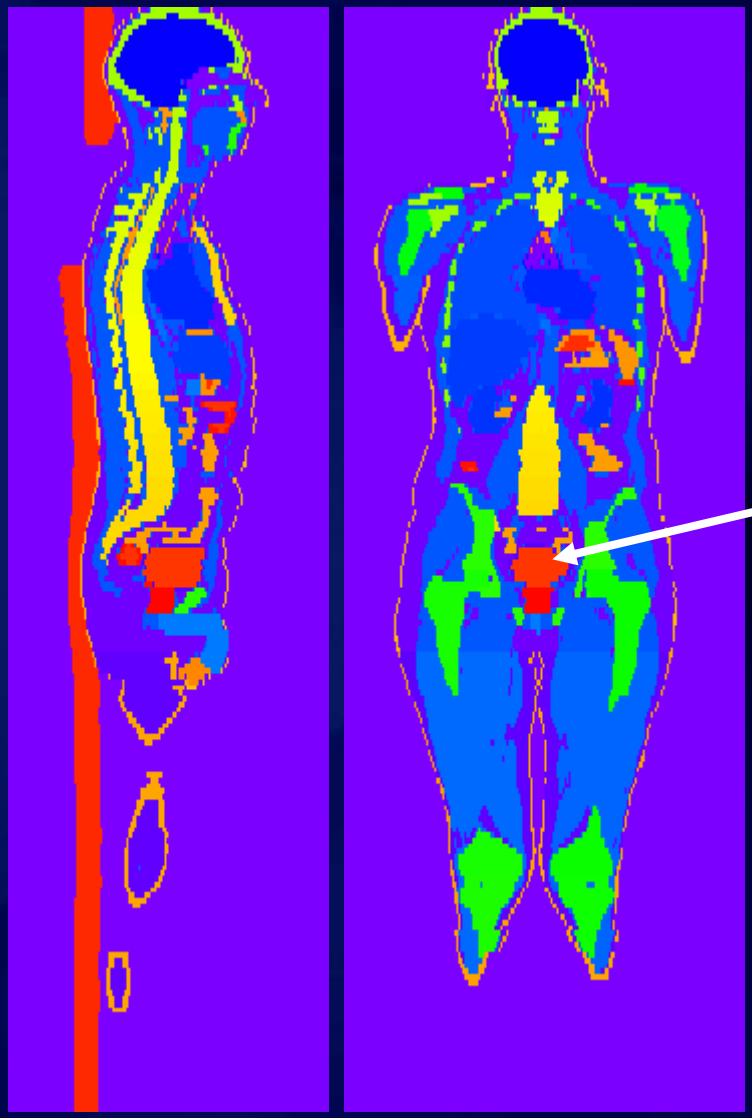
- From GSF (Petoussi-Henss, Zankl et al, PMB, 2002)
- Visible Human based upon the CT data from the Visible Human Project of the National Library of Medicine

GSF Model	Age (yr)	Gender	Weight (kg)	Height (cm)	Body-Mass Index (kg/m^2)
Baby	8 wks	Female	4.2	57	12.9
Child	7	Female	22	115	16.4
Irene	32	Female	51	163	19.2
Golem	38	Male	69	176	22.2
Donna	40	Female	79	170	27.3
Frank	48	Male	65*	97*	***
Helga	26	Female	81	170	28.0
Visible Human	38	Male	103	180	30.9

Patient Based Phantoms



GSF - Golem



The GSF data is based upon *organ segmentation* of the original CT scan sets.

ICRU 44 elemental composition and mass density

Patient based: Simulation Scan Protocols

- For each size patient model, simulate whole-body scan *(MDCT scanner - GE Lightspeed 16)*, mapping HUs of tissues into Monte Carlo materials
- Evaluate:
  - organ dose
  - whole-body effective dose
- Scan Protocol:
  - Top of head to mid-thigh
  - Helical scan, pitch=1, 120 kVp
  - Body bowtie for Adults, head bowtie for Baby and Child
  - 16 x 1.25 mm nominal beam collimation
  - On a per 100 mAs basis

Question: *On a normalized - per mAs basis (e.g. mSv/mAs) - how would you expect effective dose to vary with increasing patient size/weight?*

20% 1. Increase

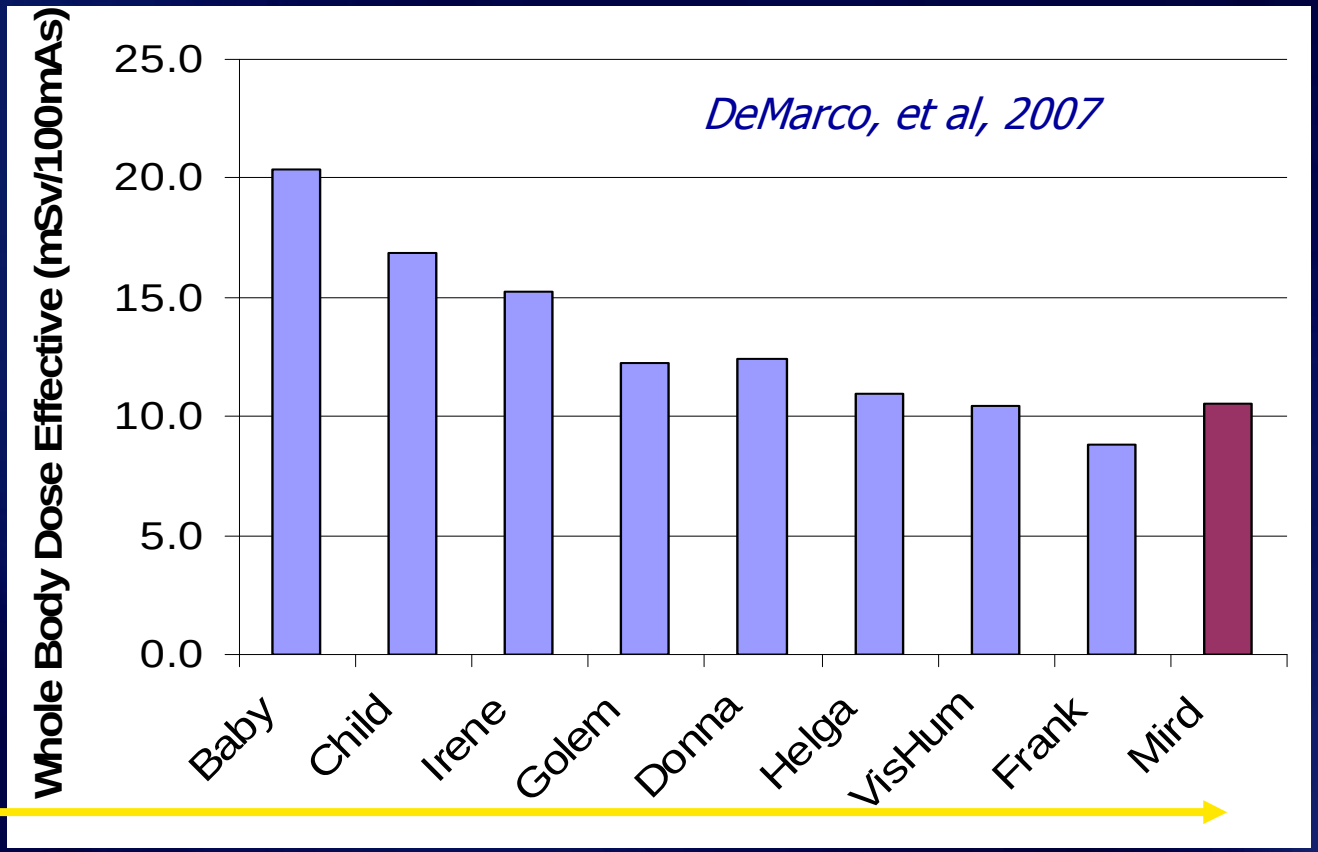
20% 2. Decrease

20% 3. Remain the same

20% 4. Cannot be determined

20% 5. Effective dose is independent of patient size

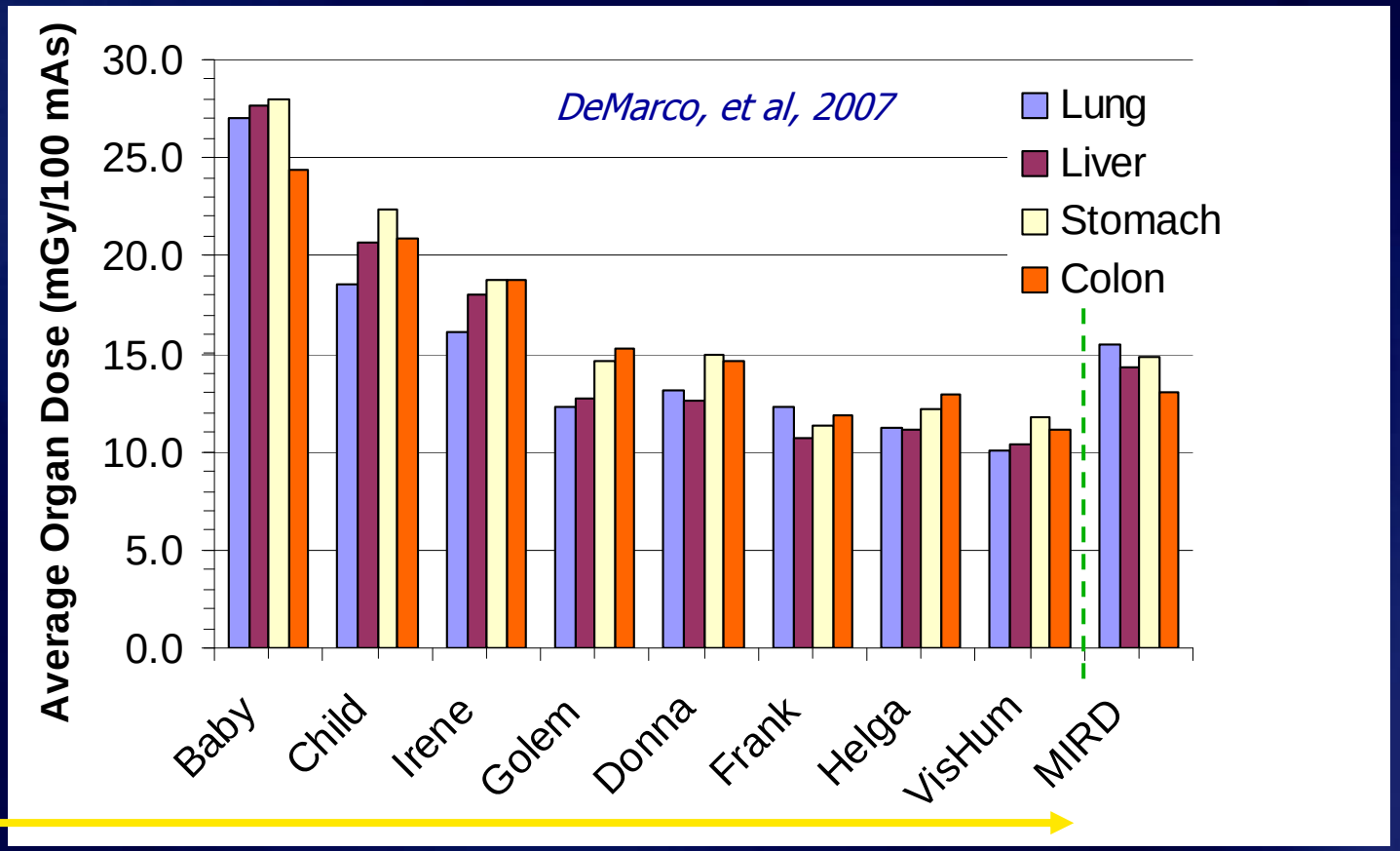
Patient based – Whole Body Scan



Increasing Body Size

Whole-Body Scan, Whole-Body Effective Dose (mSv per 100 mAs)

Results – Whole Body Scan

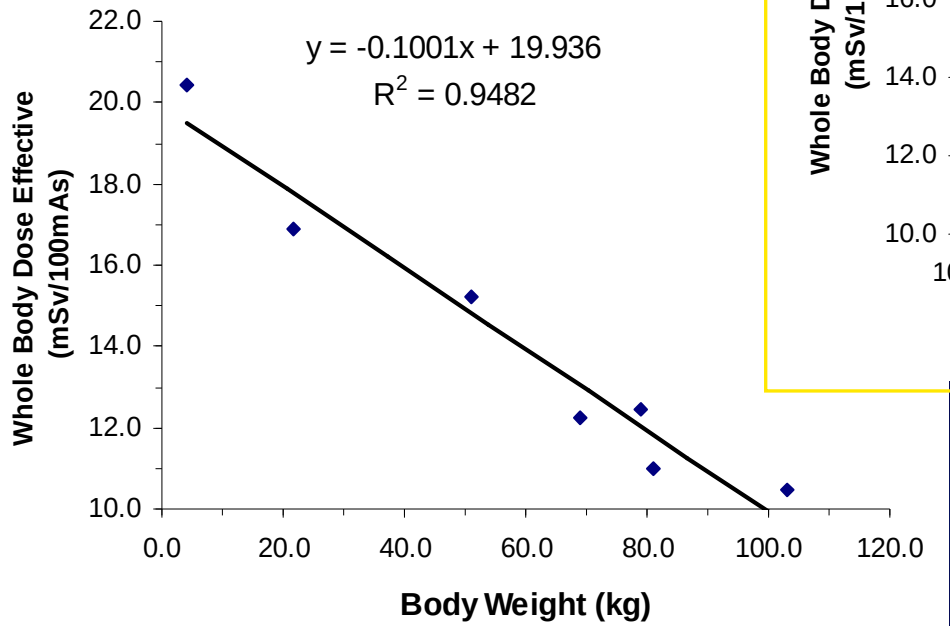


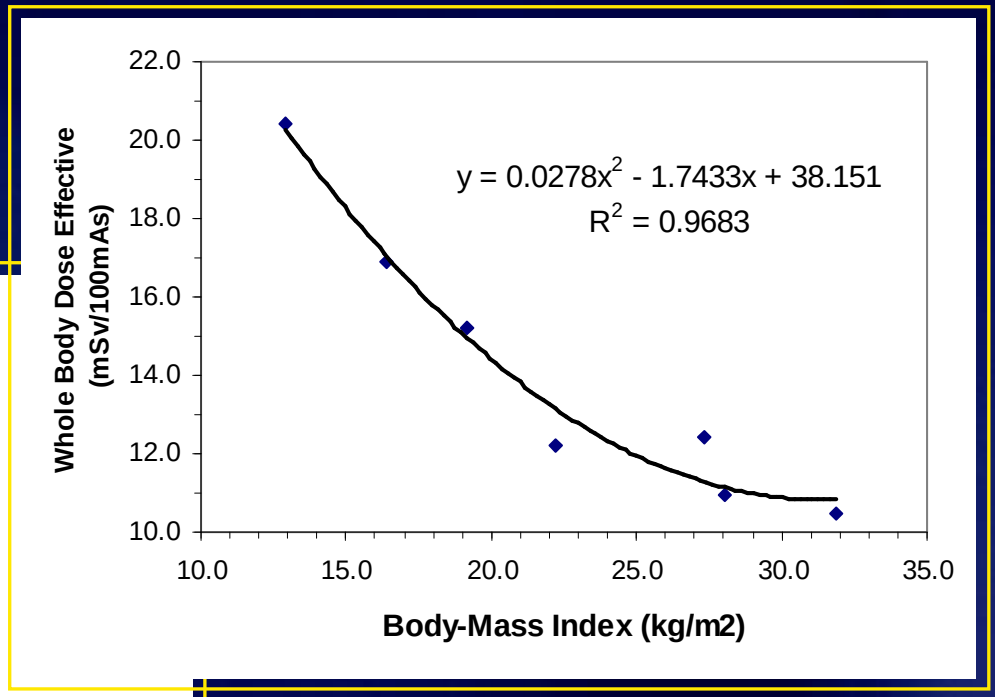
Increasing Body Size

Whole-Body Scan, Average Organ Dose

Results – Whole Body Scan

Variation of Whole-Body Dose with *Weight …*



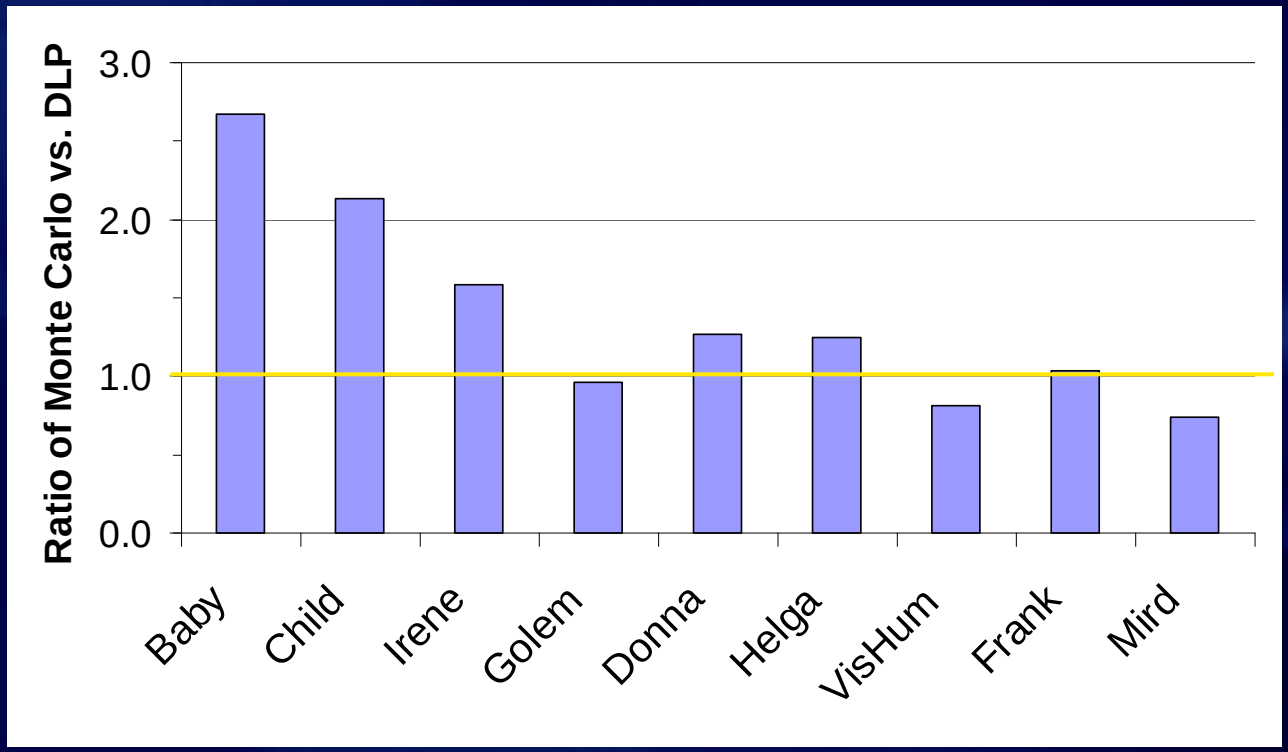


… and *Body Mass Index*

Lung Screening Protocol

- For each model, simulate low dose lung cancer screening protocol
- Evaluate
  - whole-body effective dose
  - organ dose to lung and thyroid
- Scan Protocol
  - **Thoracic inlet to base of lungs (into liver)**
  - **Helical scan, pitch=1.375, 120 kVp**
  - **16 x 1.25 mm nominal beam collimation**
  - **80 mAs (0.5 sec rotation time at 160 mA)**

Results – Low Dose Thoracic Scan

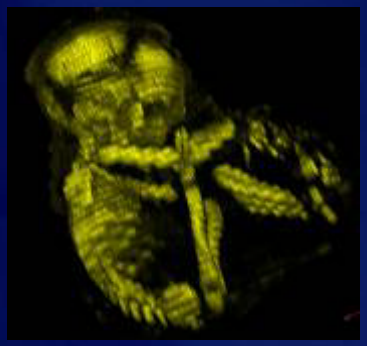


$$DLP = CTDI_{VOL} \times (scan\ length)$$

$$Effective\ Dose_{mSv} = DLP \times k$$

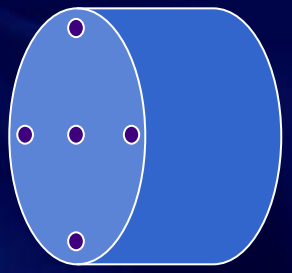
Region of body	k (mSv • mGy^{-1} • cm^{-1})
Head	0.0023
Neck	0.0054
Chest	0.017
Abdomen	0.015
Pelvis	0.019

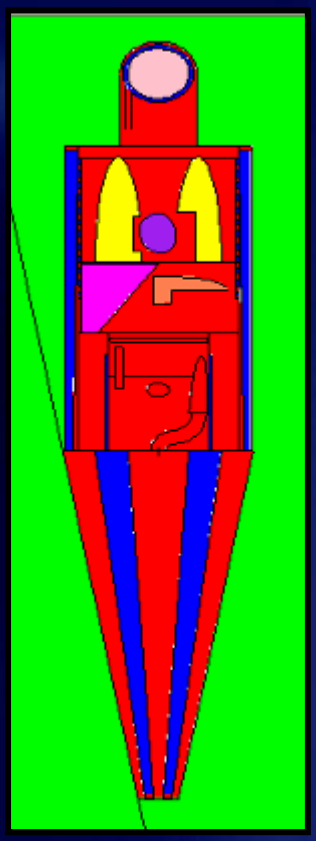
European Guidelines, Jessen, 1999



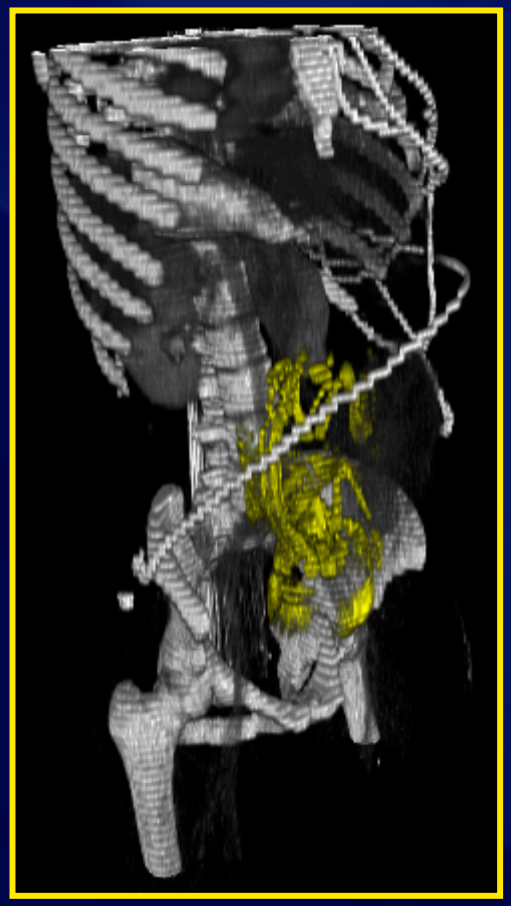
Fetal Dose with MDCT

- Existing methods:
  - Idealized geometric models
    - *Not pregnant*
  - Single dose estimate
    - Limited allowance for patient size
    - Single gestational age (<8 weeks)
    - Homogenous fetus
  - No standard of truth for comparison
- How well do these represent a range of patient anatomies & gestational age?



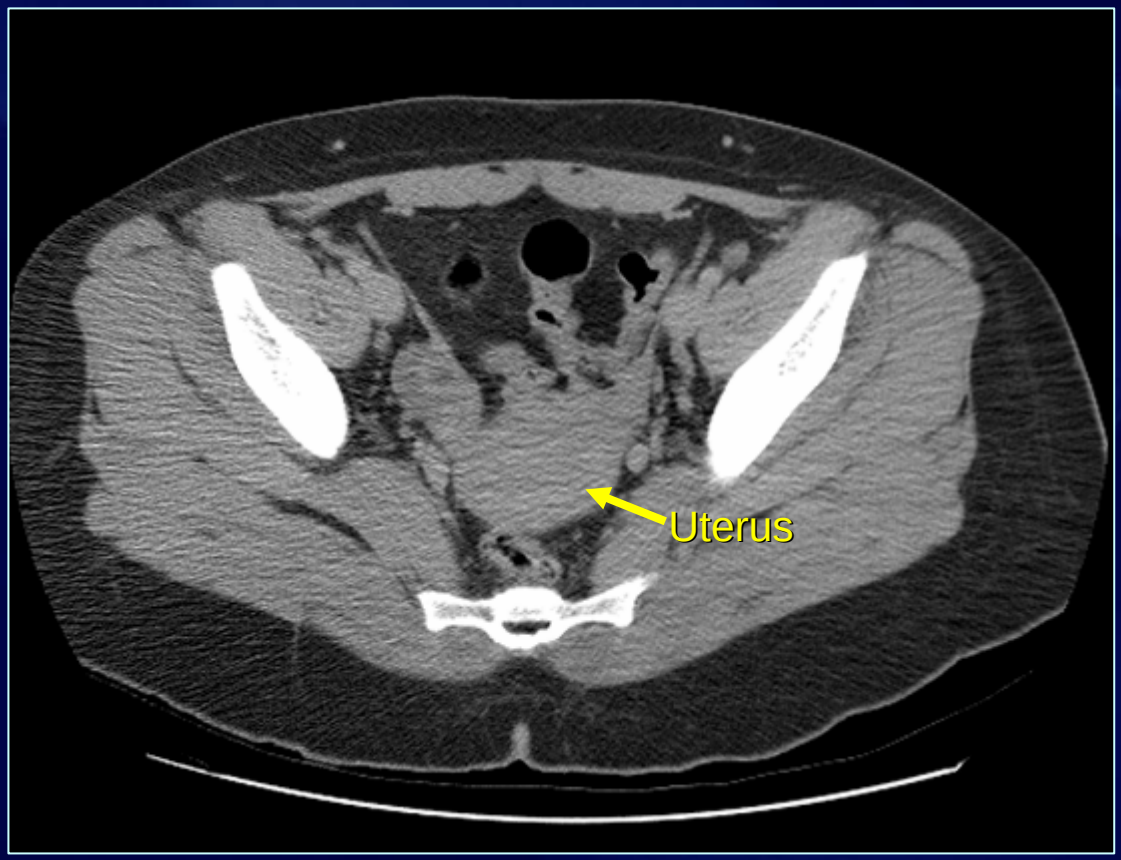


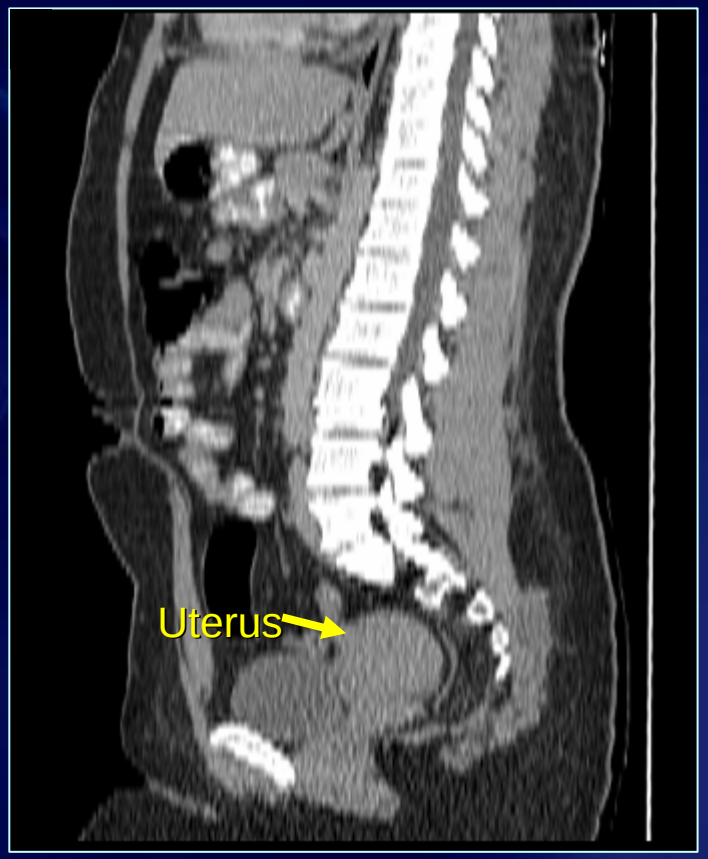
Monte Carlo Approach



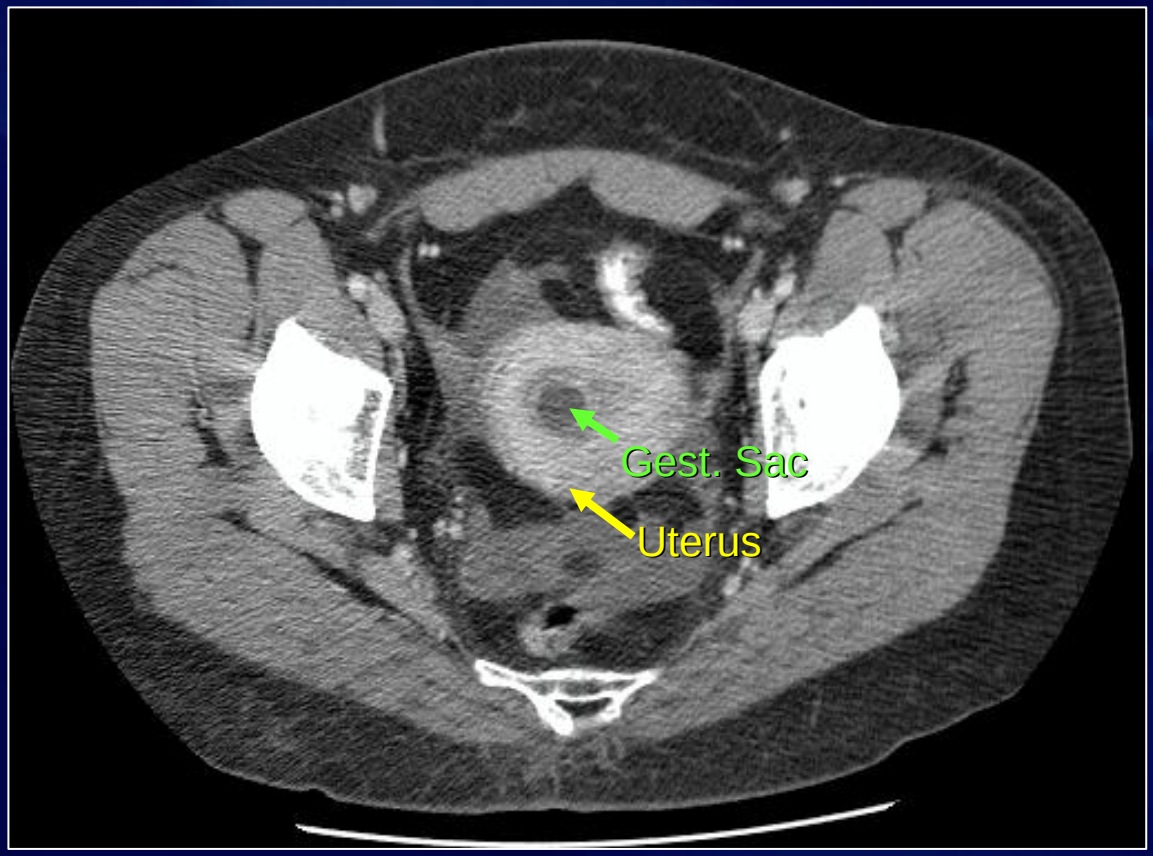
- Model CT scanner characteristics
  - GE LightSpeed 16, pitch 1
- Model 27 pregnant patients of gestational age of < 5 weeks to 37 weeks
  - Voxelized models created from actual patient image sets
  - including early and late term pregnancies
  - Mother's size
  - Fetal size
  - Gestational age
  - Fetal composition (bone, tissue)

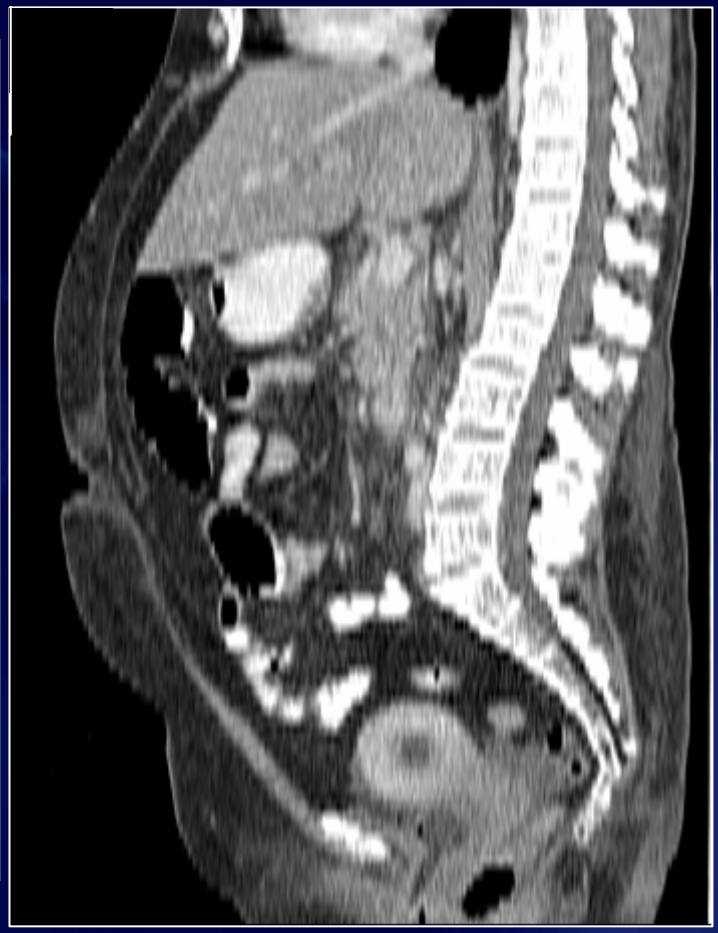
< 5 weeks (gestation sac not visible)





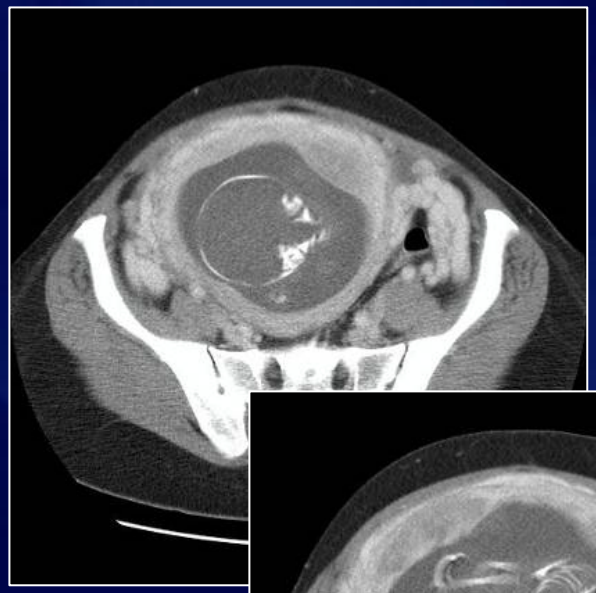
7 weeks (embryo not visible)

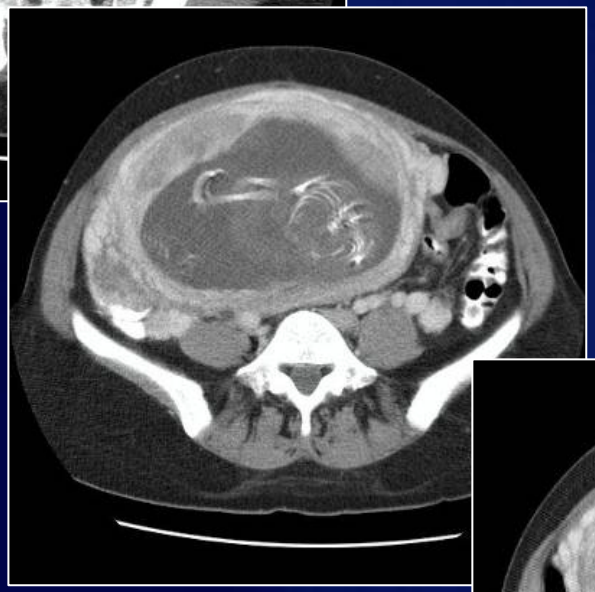


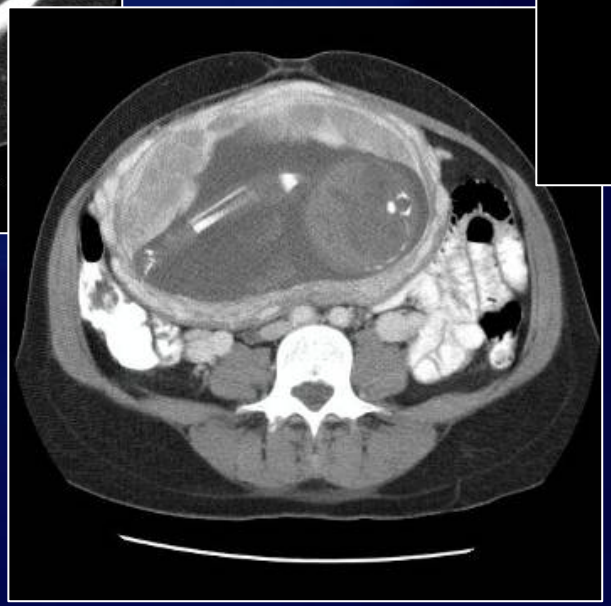


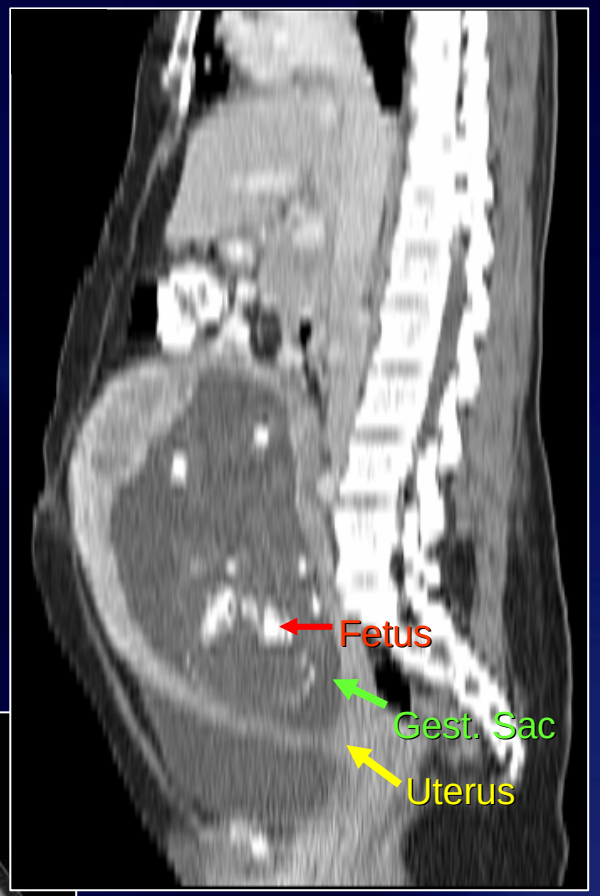
Average Maturity:

24 weeks



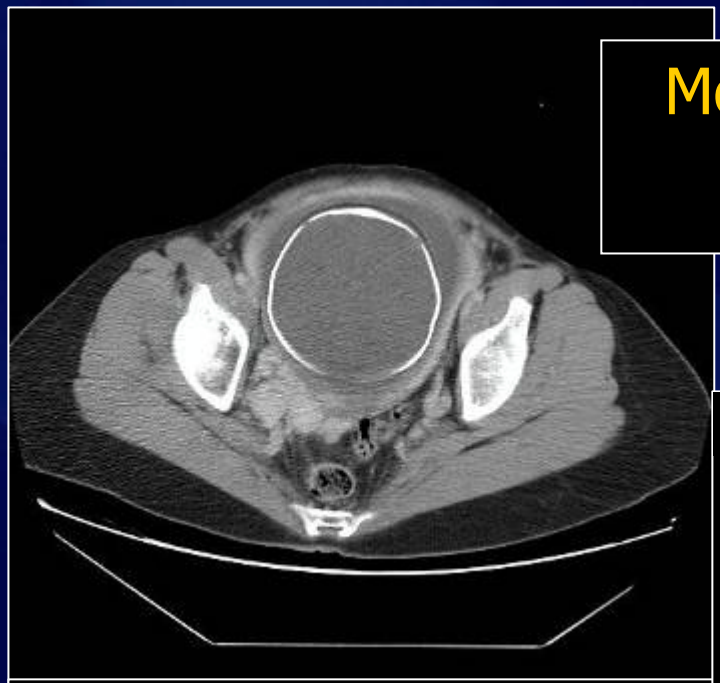


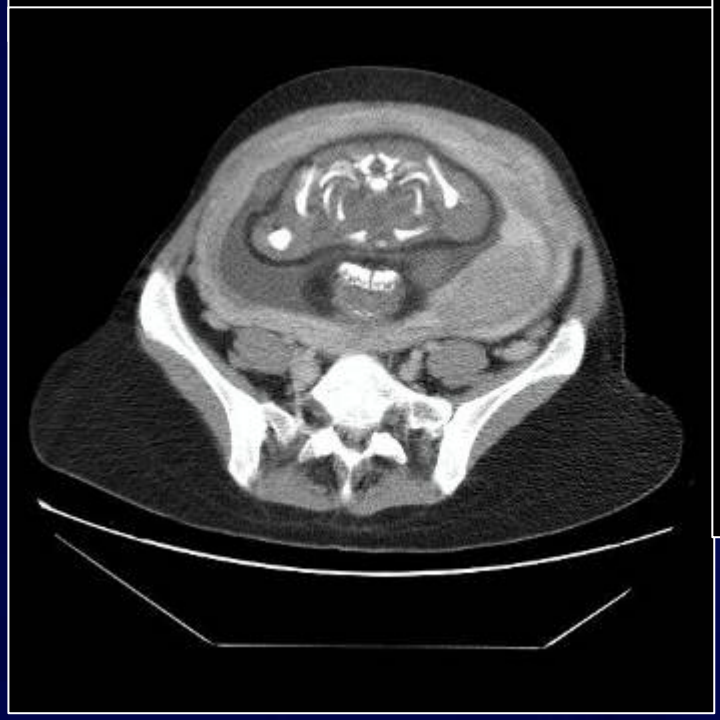


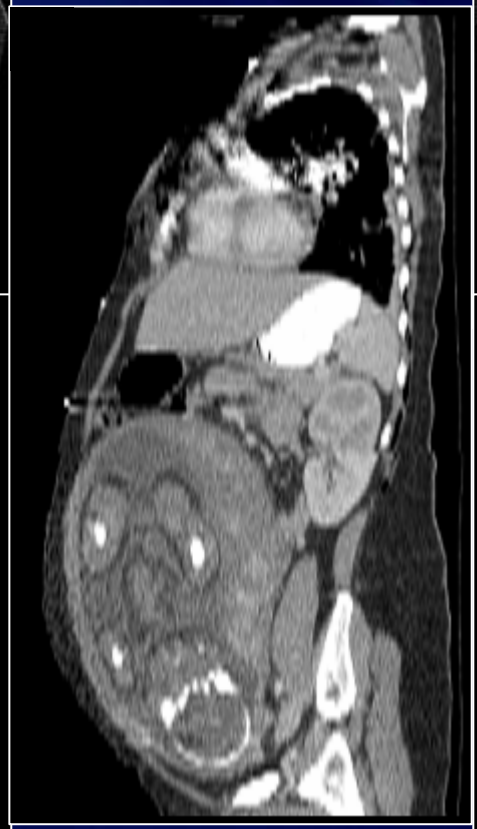


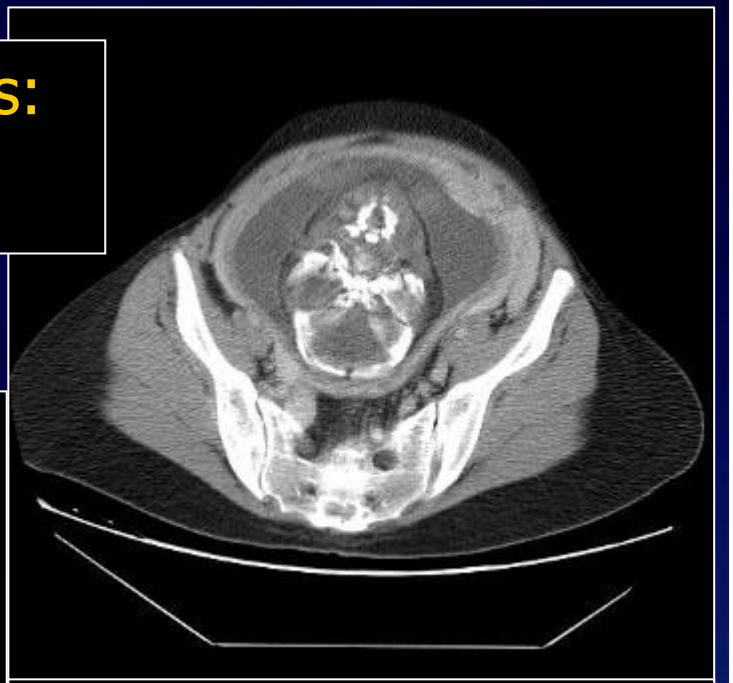
Most Mature Fetus:

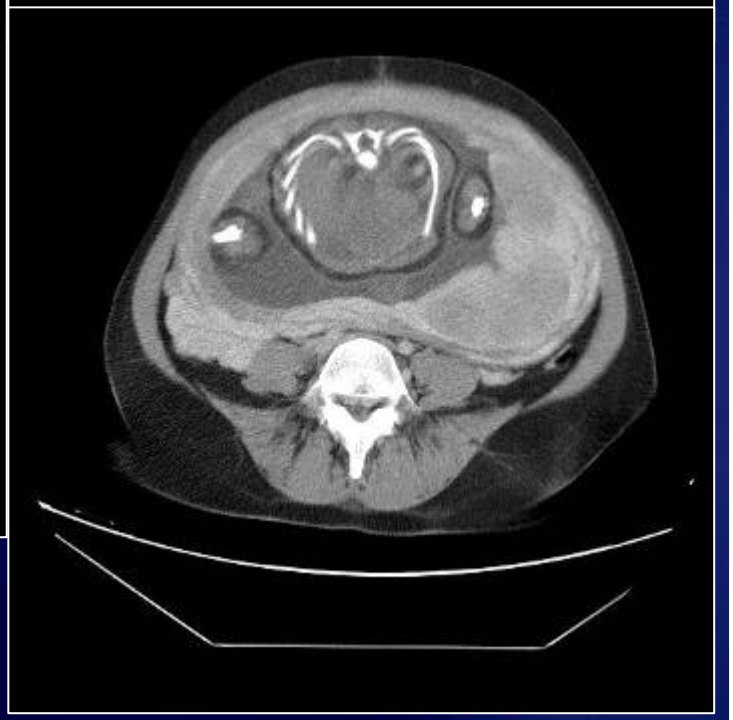
36 weeks





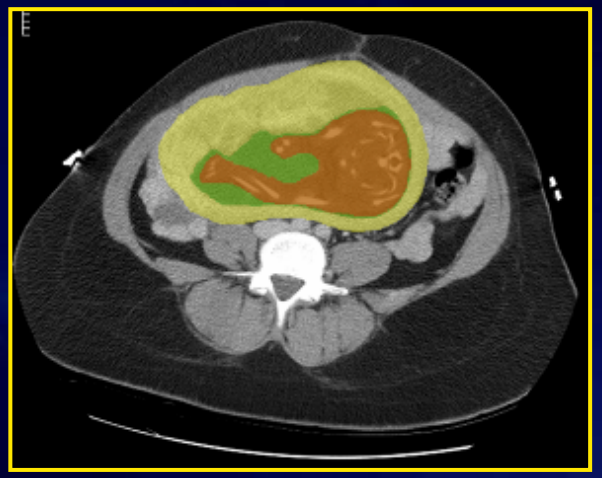


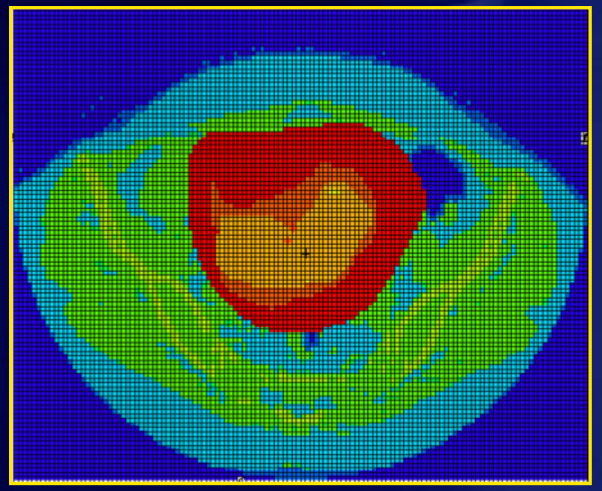


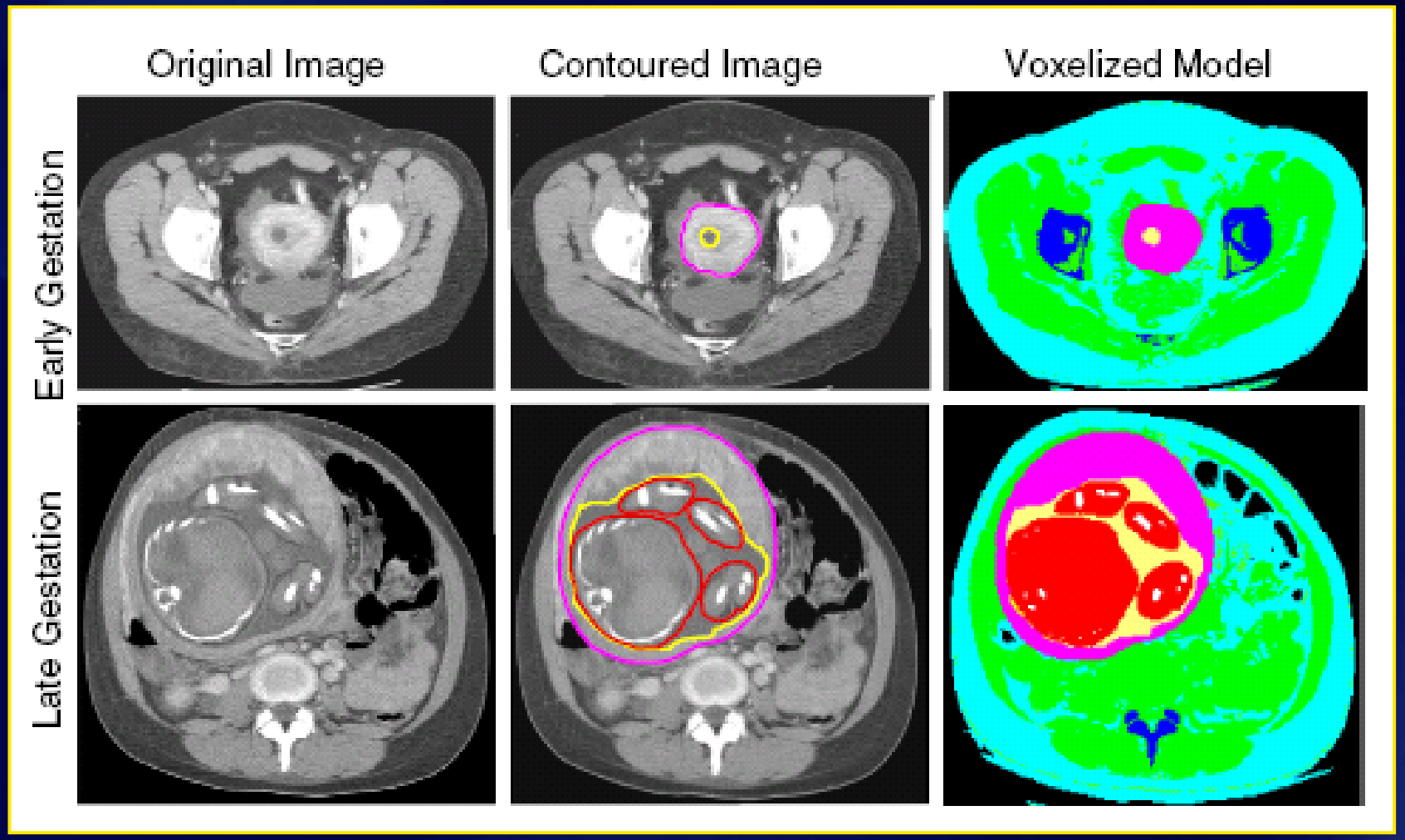


Voxelized Patient Models

- Radiologist contoured organs:
  - Fetus (if visible)
  - Gestational sac (if visible)
  - Uterus
  - Within the fetus, voxels assigned to bone or soft tissue
- Outside of uterus voxels assigned to one of 6 tissue types (based on HU):
  - Lung, fat, water, muscle, bone, air






Original Image
Contoured Image
Voxelized Model
Early Gestation
Late Gestation

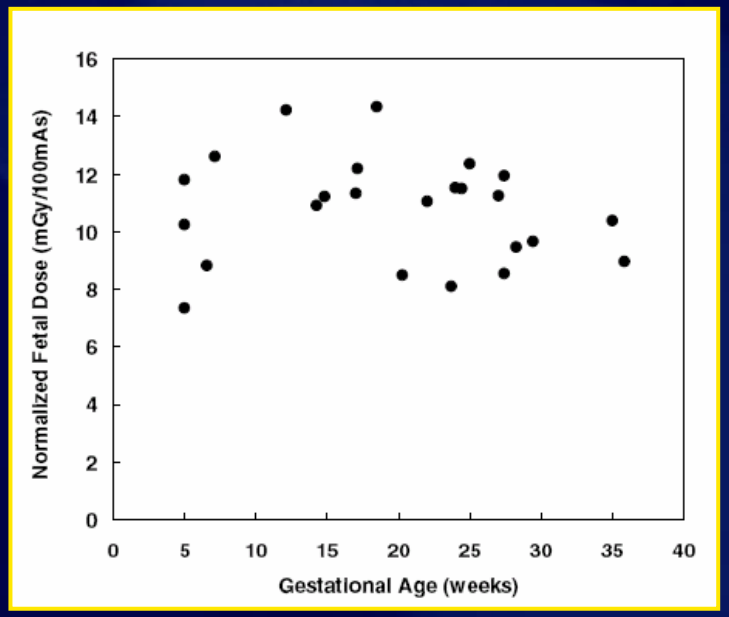
Fetal results

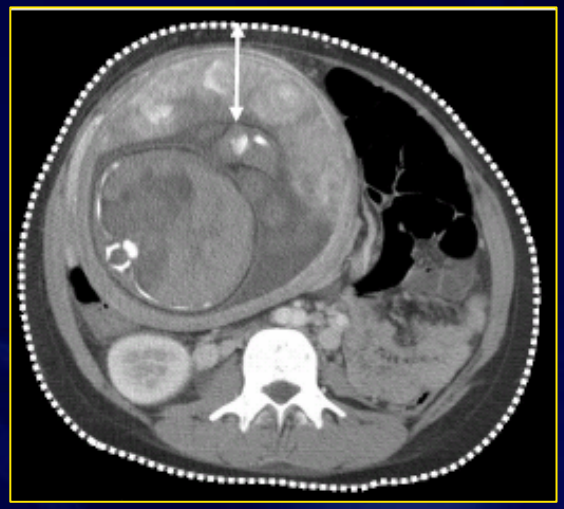
Angel, et al. 2008

	Fetal dose (mGy/100 mAs)	
METHOD	Average	Range
Patient Specific (n=27)	10.8	7.3-14.3
ImPACT (n=1) *(MIRD based uterine dose)*	12	***
Felmlee (n=1)	11.3	9-13.6 *(size allowance +/- 20%)*

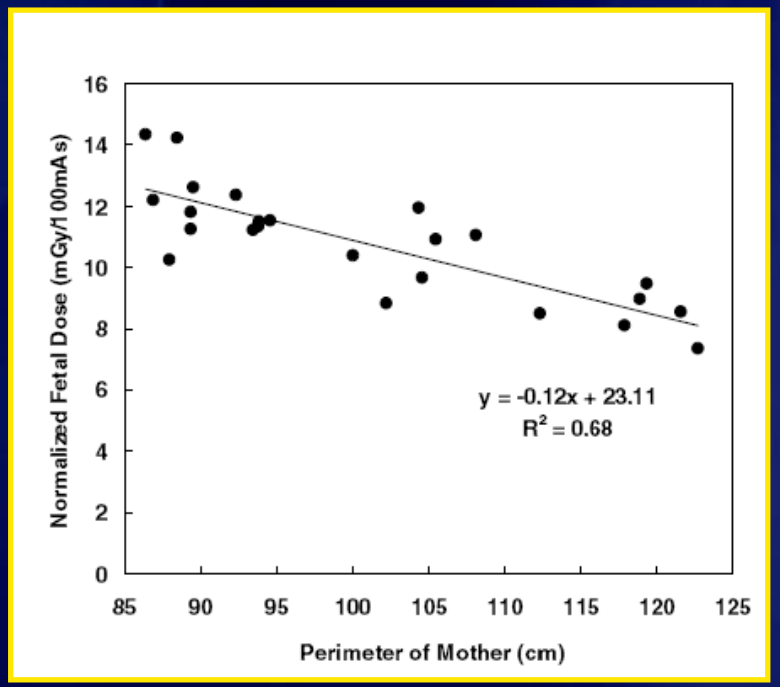
Comparison models (that don't account for gestational age) tend to overestimate fetal dose

Normalized radiation dose *does not* appear to correlate with gestational age …





but *does* correlate with mother's perimeter



Angel, et al. 2008

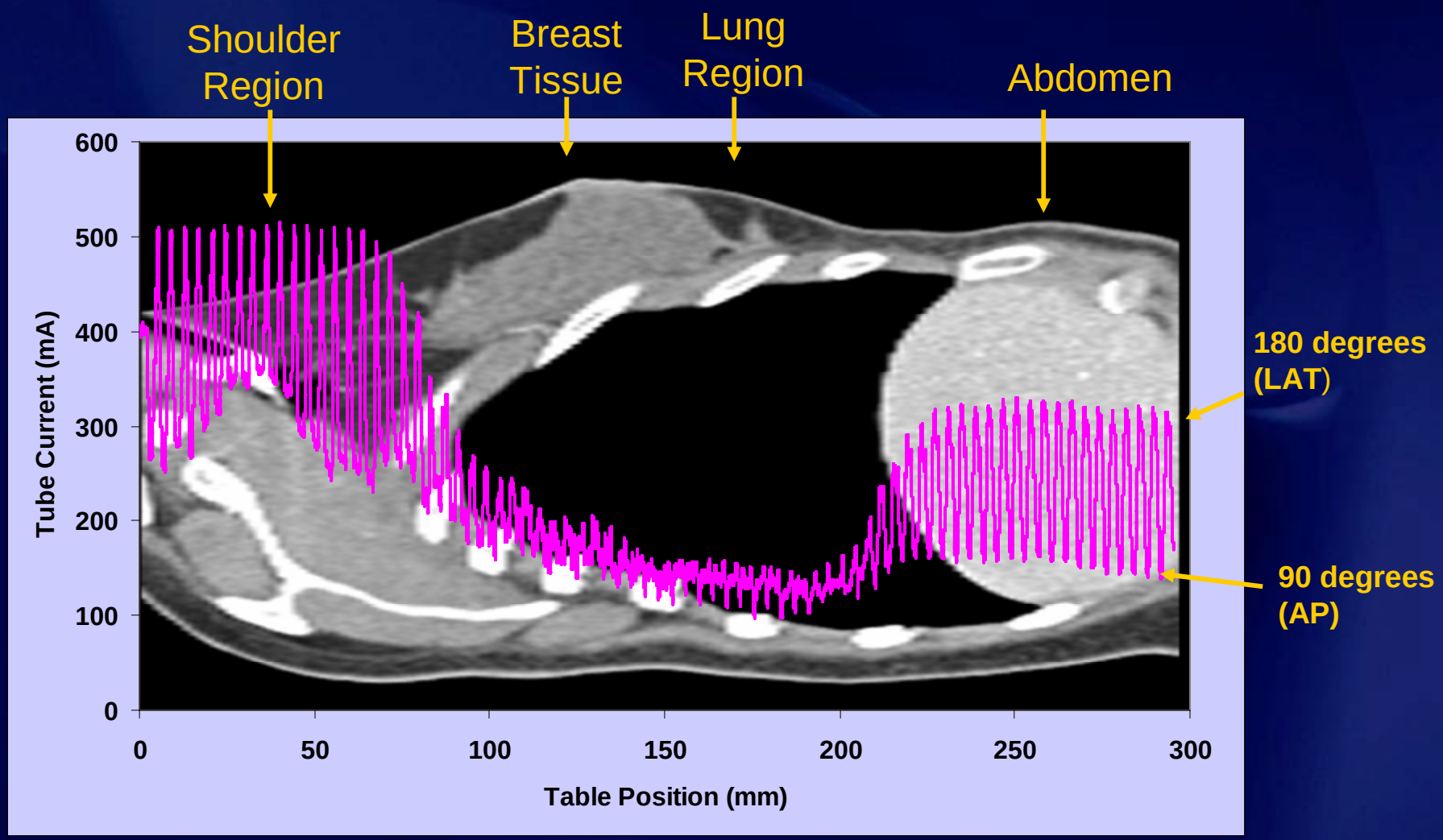
What about Dose/Tube Current Modulation?

- Use scanner/patient specific modeling to determine:
  - How much is overall dose reduced?
  - What happens to individual organ dose?

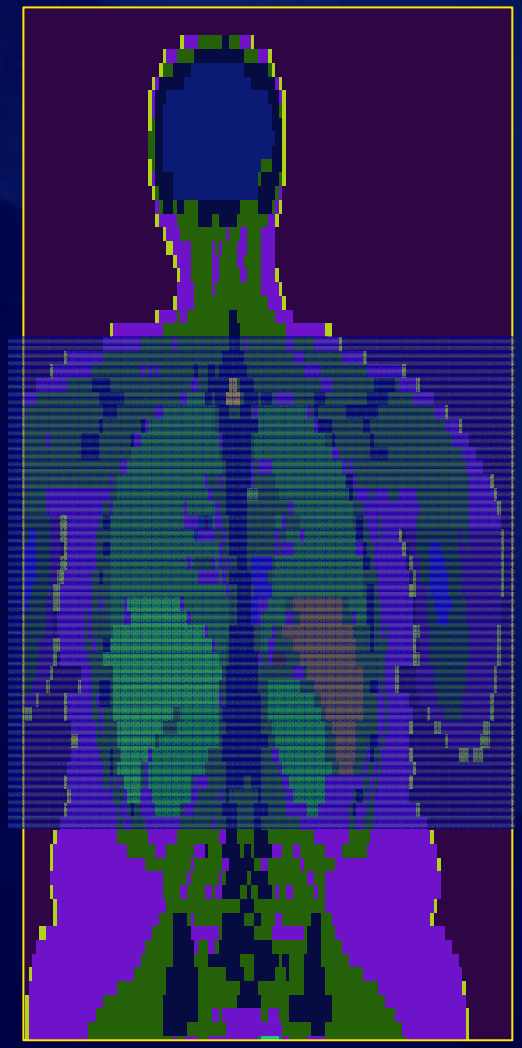
Variable mA Monte Carlo model

- For variable tube current, modify scanner X-ray source model to vary output as a function of gantry angle (in plane modulation) and z axis position along source path
- Use data from actual patient scan that provides:
  - tube current vs. tube angle vs. table position
  - (mA vs. θ vs. z)

Conventional Long Axis Modulation



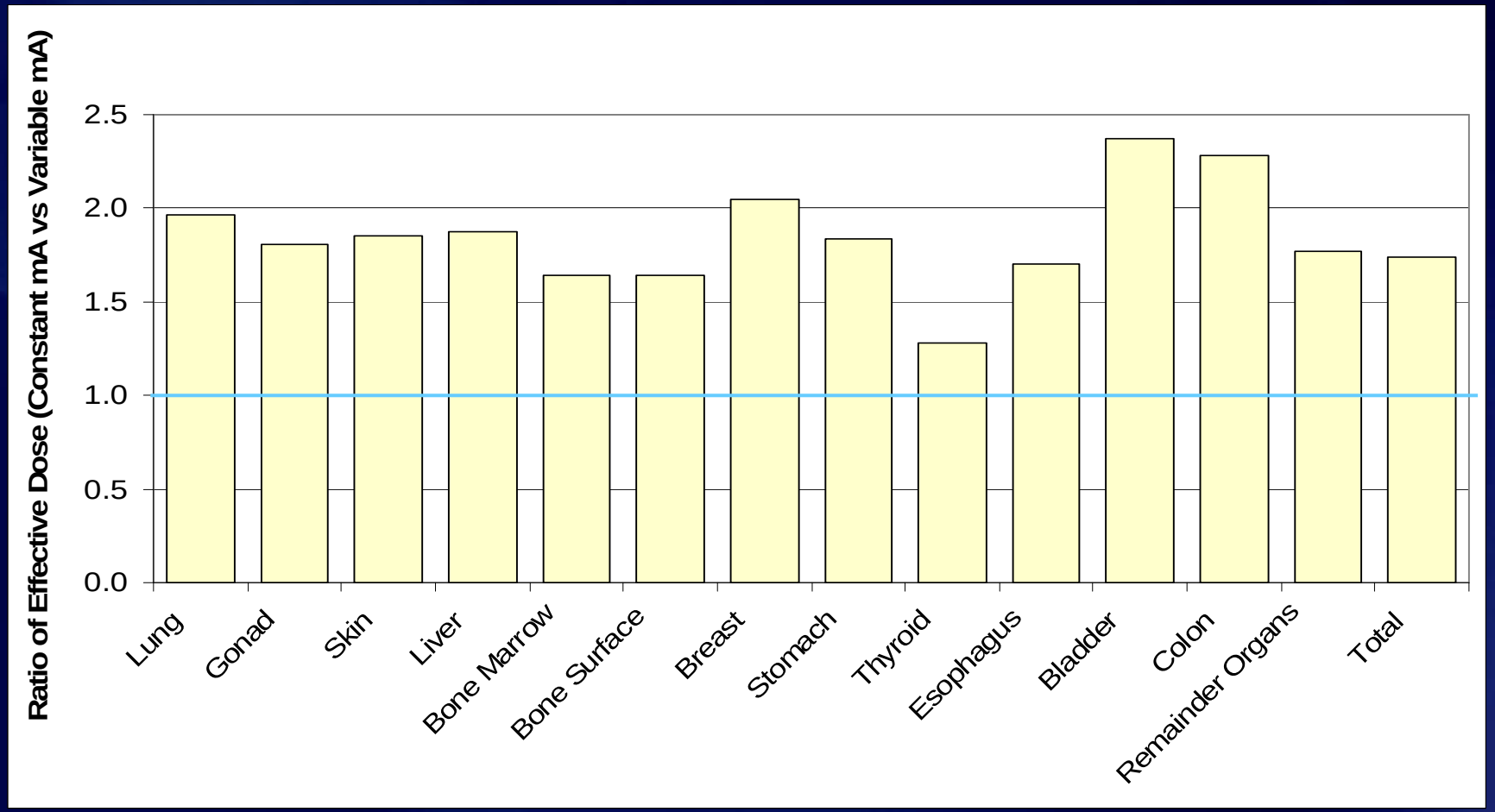
Siemens Sensation 16
applied to GSF "Donna"
(extrapolated)



Effective Dose Table *(w/ weight factors)*

	variable mA	constant mA	Ratio
Lung	0.750	1.473	1.96
Gonad	0.004	0.007	1.81
Skin	0.016	0.030	1.85
Liver	0.258	0.482	1.87
Bone Marrow	0.114	0.187	1.64
Bone Surface	0.040	0.065	1.64
Breast	0.284	0.581	2.04
Stomach	0.737	1.354	1.84
Thyroid	0.813	1.044	1.28
Esophagus	0.350	0.595	1.70
Bladder	0.000	0.001	2.37
Colon	0.073	0.166	2.28
Remainder Organs	0.082	0.146	1.77
Total	3.522	6.131	1.74

Ratio of maximum mA to average mA = 1.77



Reference mA ??

Dose to individual organs from tube current modulation?

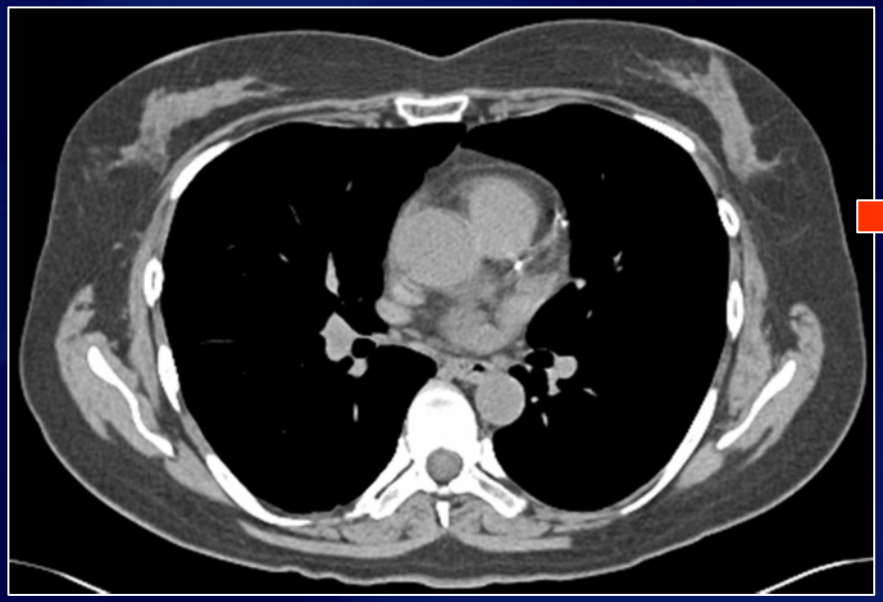
- Dose reduction is relative – what is comparison mA?
- Current work being done on effect of dose modulation schemes on breast dose

Voxelized Patient Model

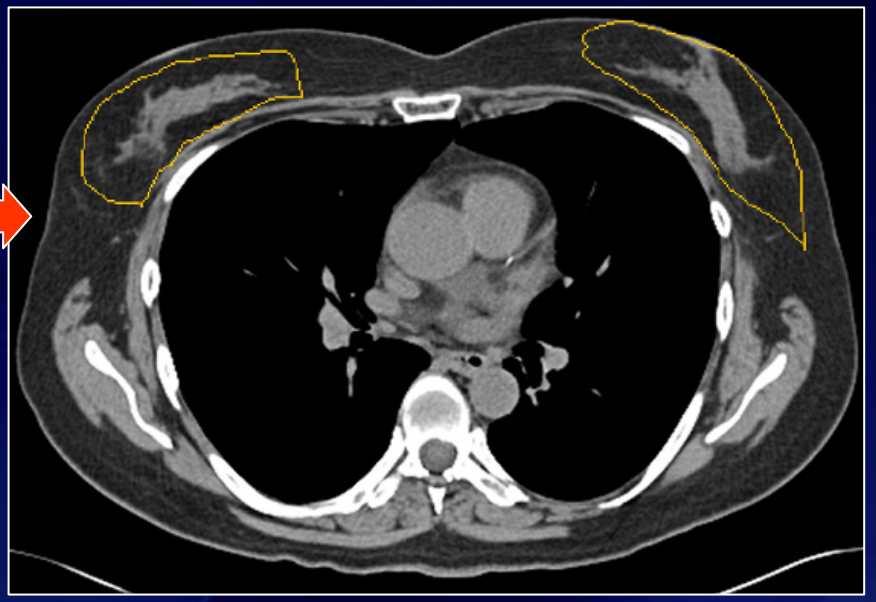
- **Voxelized models from** actual patient images
- Radiologist contoured breast tissue
  - (glandular +adipose)
- **Glandular tissue** automatically segmented
- **Lung tissue** semi-automatically segmented
  - 5 lung density categories, depending on HU
- Voxels outside the breast and lung regions automatically assigned to material types:
  - fat, water, muscle, bone, or air
  - further sub-divided into 17 density categories, depending on its HU value

***(as defined by ICRU 44)**

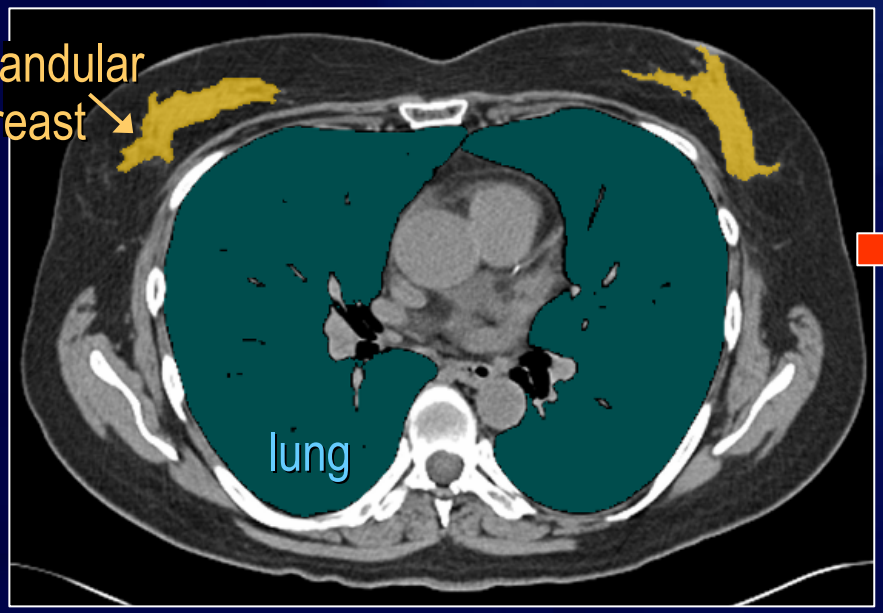
Original Image



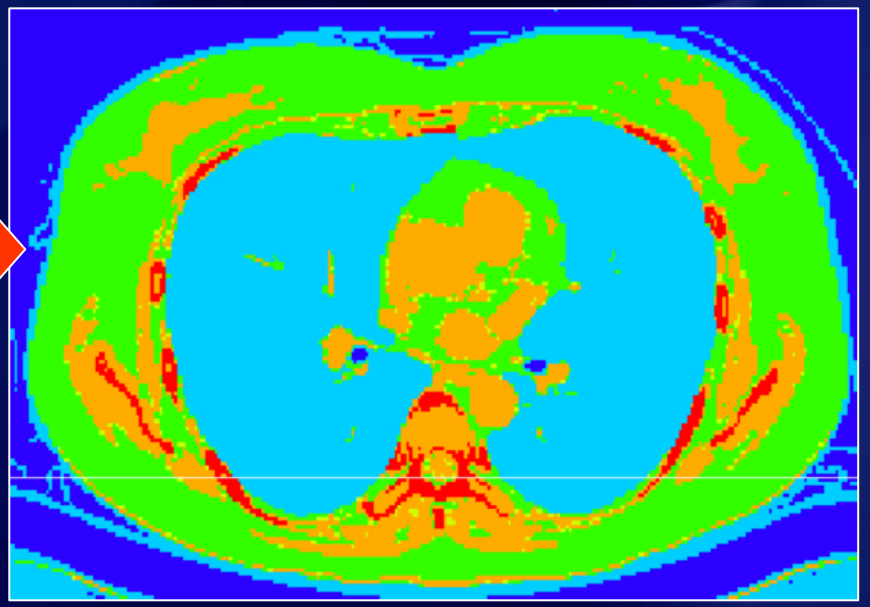
Radiologist's Contour



Segmented Image



Voxelized Model



Patient specific calculations in the clinic?

- Validated Monte Carlo modeling can be used to ask specific questions and could serve as a “gold” reference standard
- Not generally practical on an individual patient basis
- Patient and scanner specific Monte Carlo modeling probably better suited for characterizing dose - and identifying factors that effect dose- across a range of patient, scanners, and scan protocols
- Create data tables of scaling factors to allow estimation of patient and organ dose calculated from standardized measurements made on a CT scanner

Recommended reading

- AAPM Report 96: The Measurement, Reporting, and Management of Radiation Dose in CT (Jan-08)

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