Henry Ford Health System

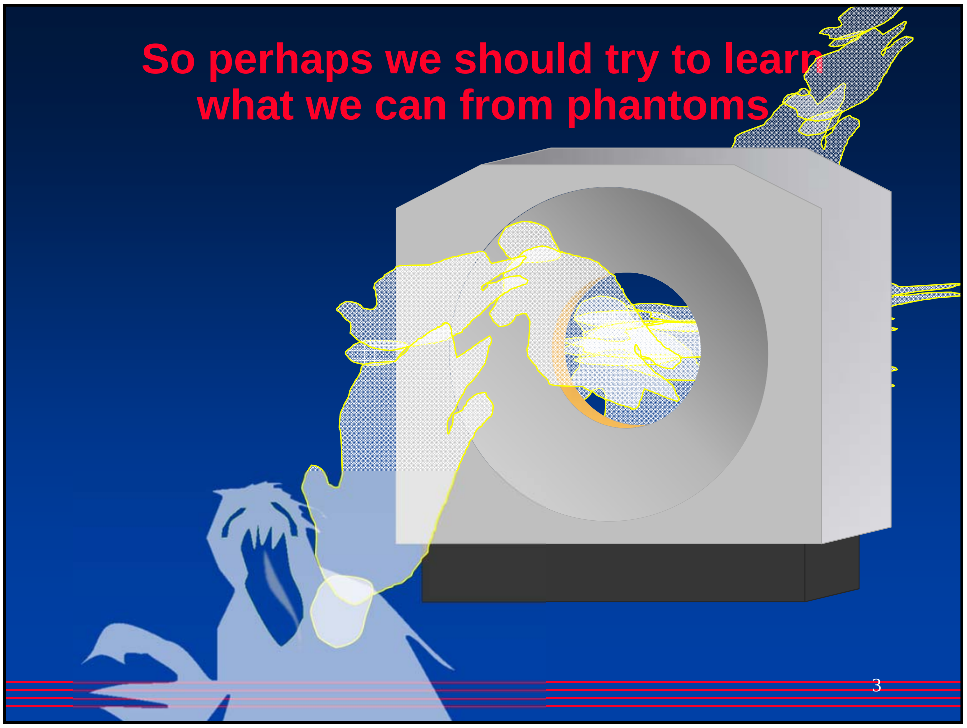
CT Dosimetry Basics (and What We Can Learn From Making Measurements On Cylinders)

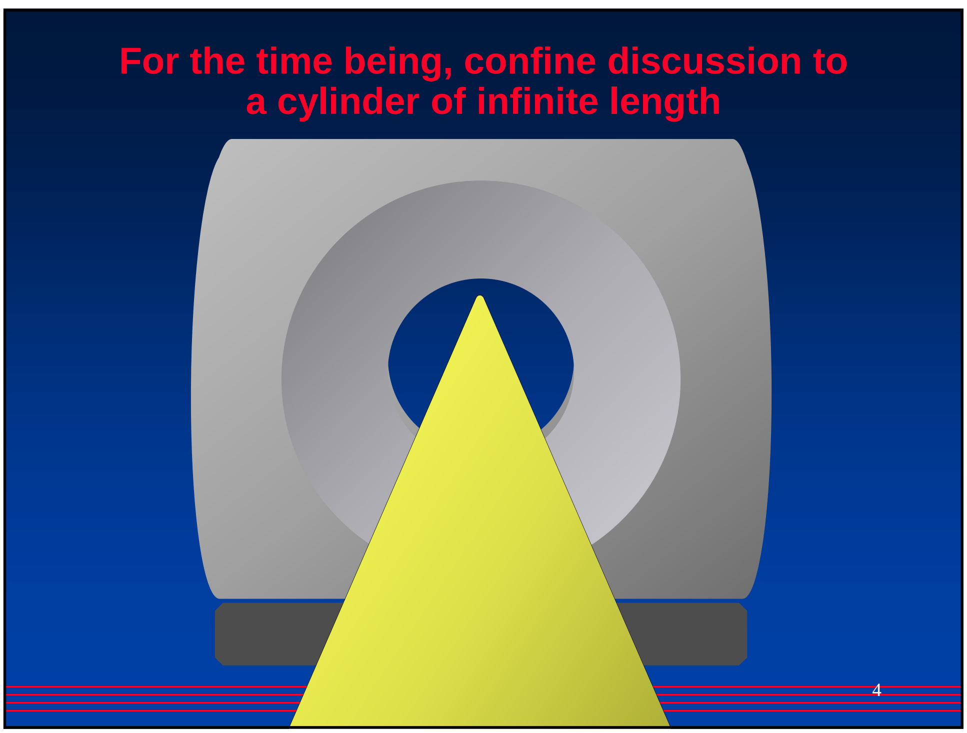
Donovan M. Bakalyar, Ph.D., FACR
Diagnostic Radiology

1

It may not always be a good idea to investigate radiation dose clinically

2


So perhaps we should try to learn
what we can from phantoms
3


For the time being, confine discussion to
a cylinder of infinite length
4

Topics

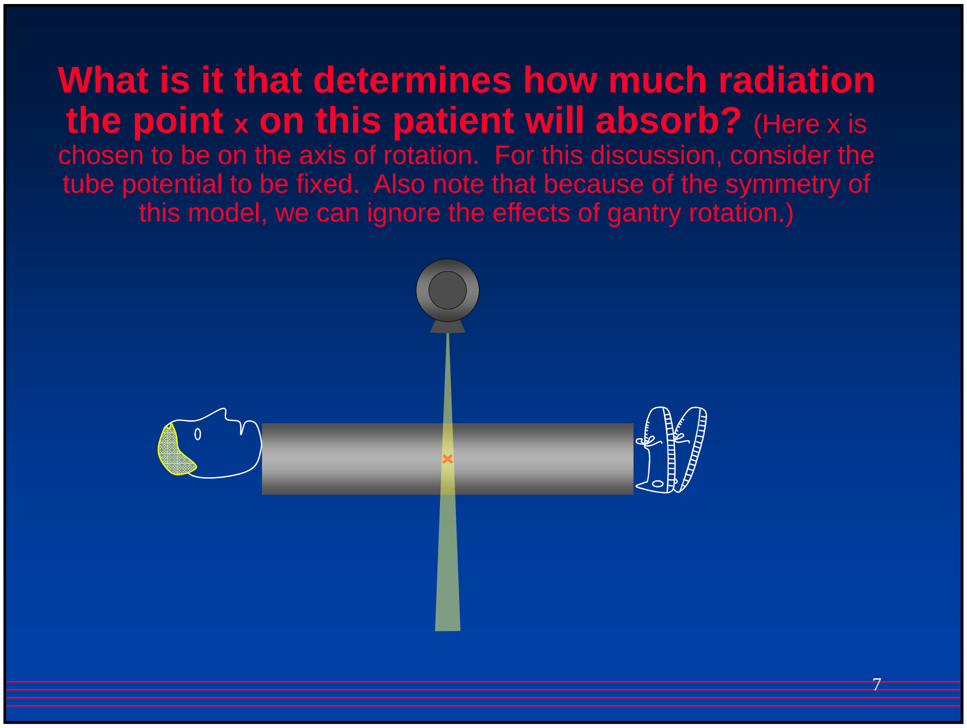
I. Operator controlled factors going into CT radiation dose
II. The infinite cylinder, its parameters and TG111
III. Details in the development of D_{eq} and its relatives
IV. The finite scan, $h(L)$ and CTDI
V. SSDE and accounting for patient size

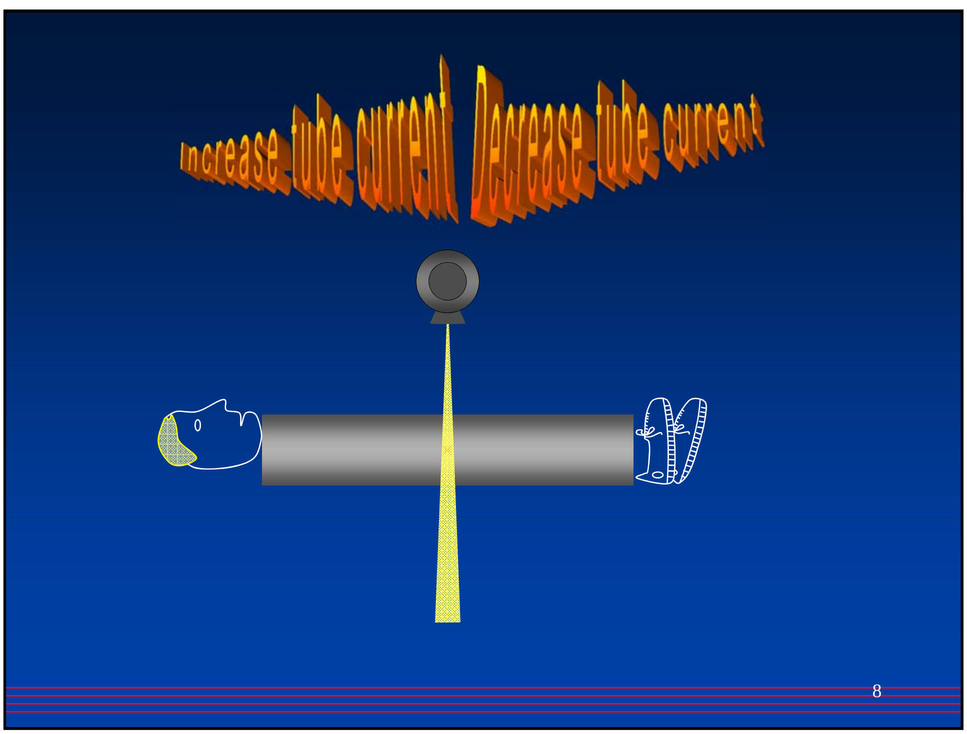
5

CT Dose

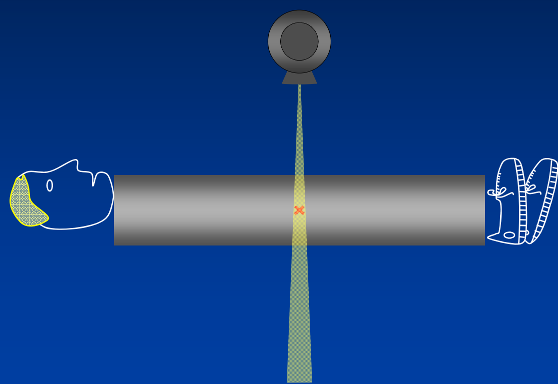
The radiation dose from CT is relatively high, often in the same ballpark as interventional radiology and cardiology. There is some difference in that ordinarily, the radiation in CT exams is applied more uniformly and so there may be less risk of radiation burns than for interventional procedures involving the same total absorbed energy.

6


What is it that determines how much radiation the point x on this patient will absorb? (Here x is chosen to be on the axis of rotation. For this discussion, consider the tube potential to be fixed. Also note that because of the symmetry of this model, we can ignore the effects of gantry rotation.)
x
7


Increase tube current! Decrease tube current!
8

Control the amount of time x-ray beam is on.

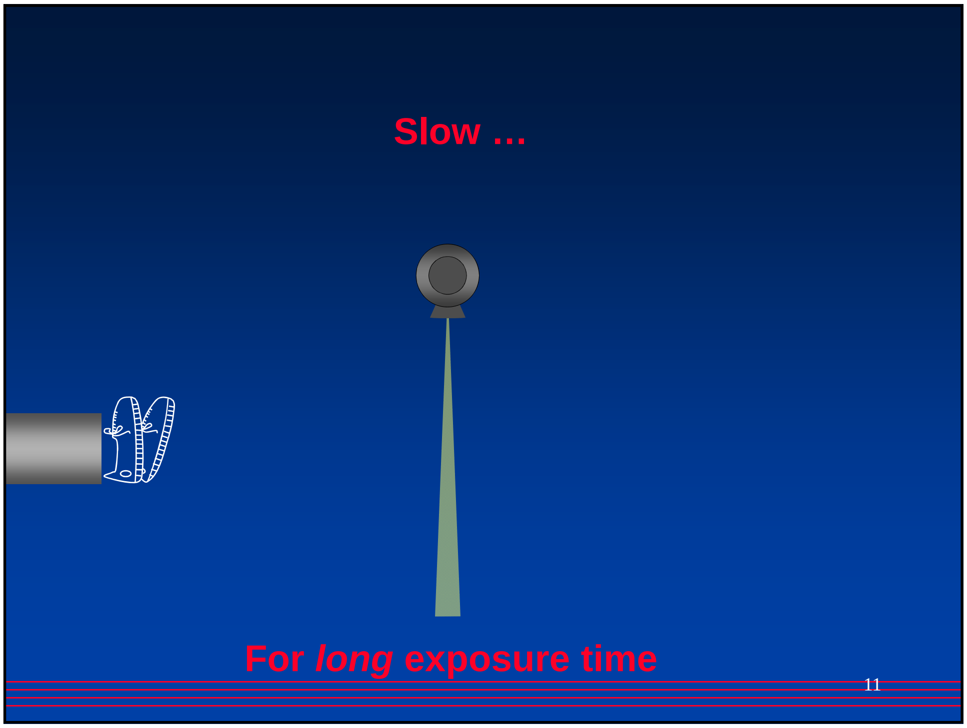


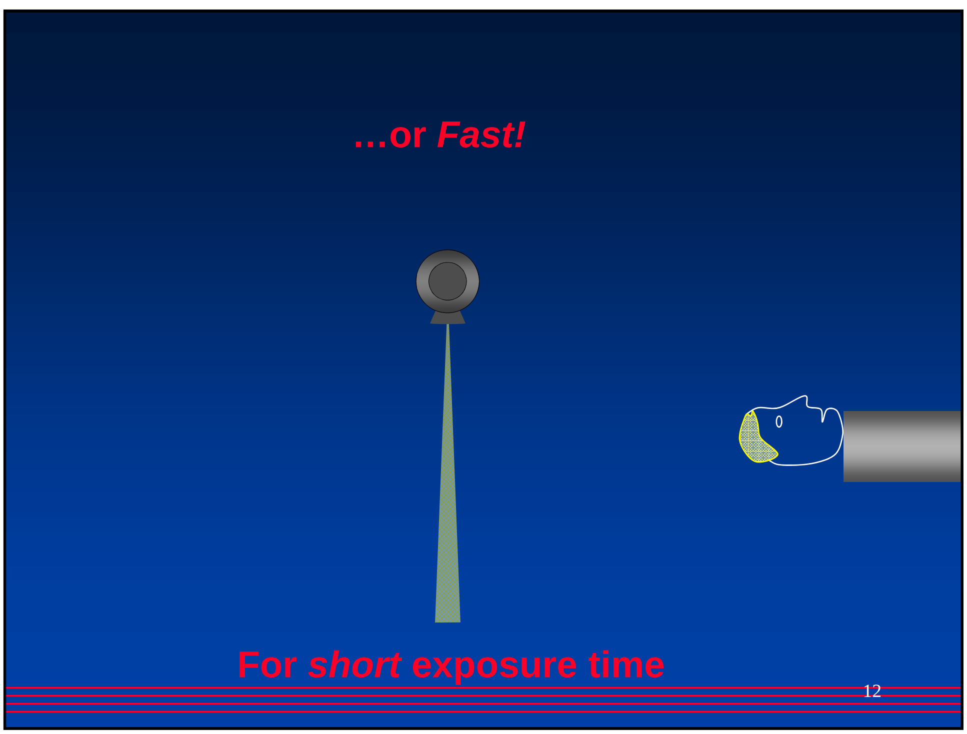
9

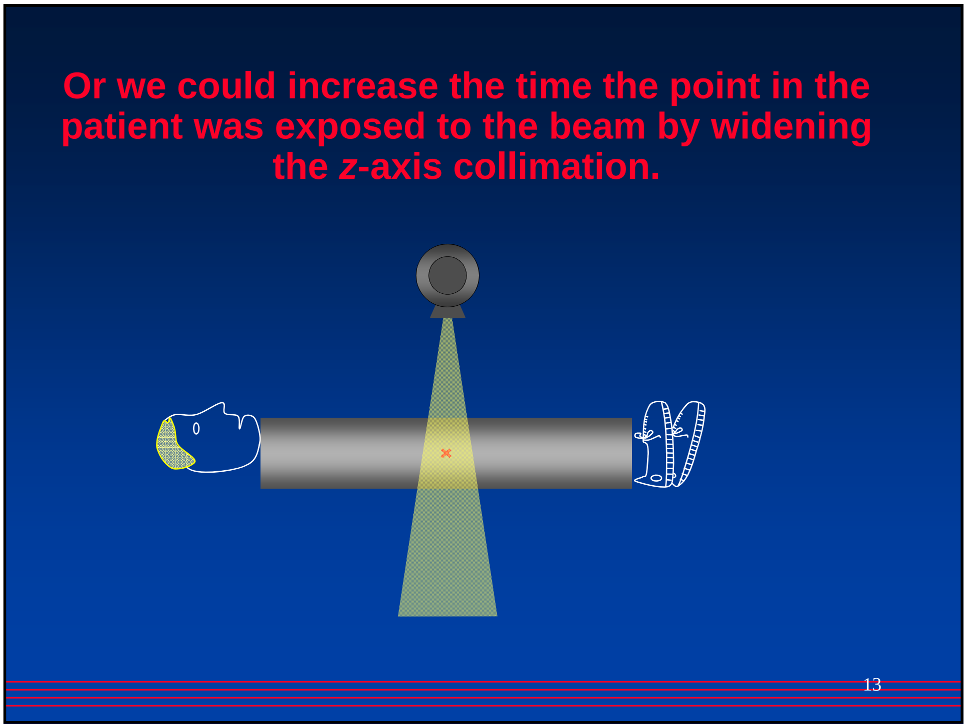
We have been looking at one data acquisition cycle of what is officially called an *axial* scan (or sequential scan—Siemens or scan and scan—Toshiba).

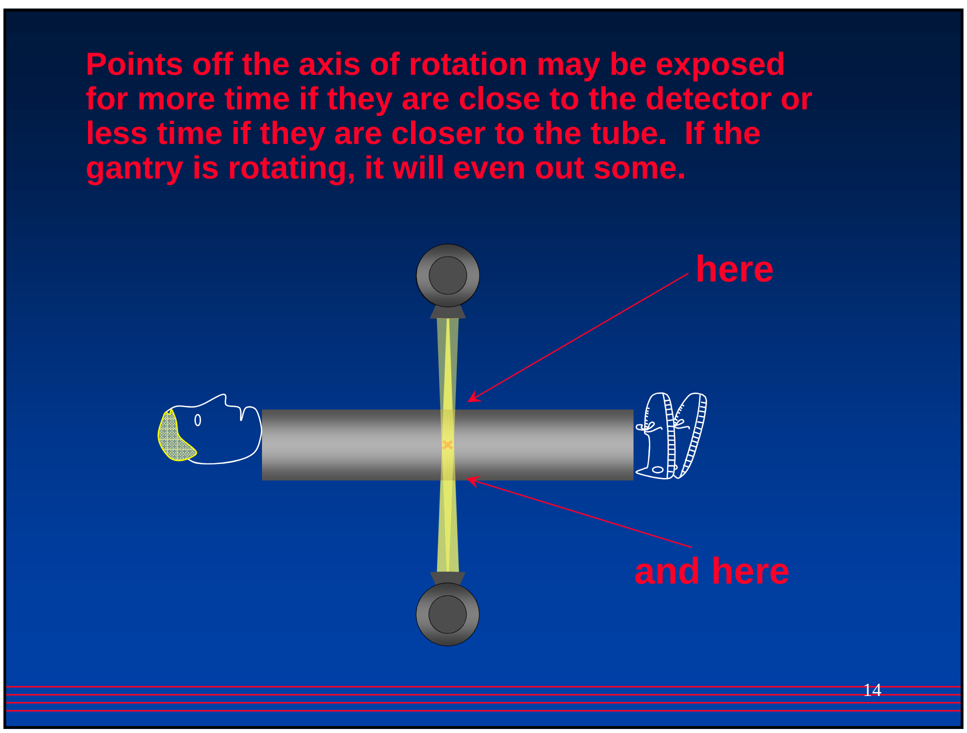
An alternative method of scanning is called a *helical* (or spiral) scan where the tube is left on and the patient is moved through the beam continuously.

10


Slow …
For *long* exposure time
11


…or *Fast!*
For *short* exposure time
12


Or we could increase the time the point in the patient was exposed to the beam by widening the *z*-axis collimation.
13


Points off the axis of rotation may be exposed for more time if they are close to the detector or less time if they are closer to the tube. If the gantry is rotating, it will even out some.
here
and here
14

mAs Confusion

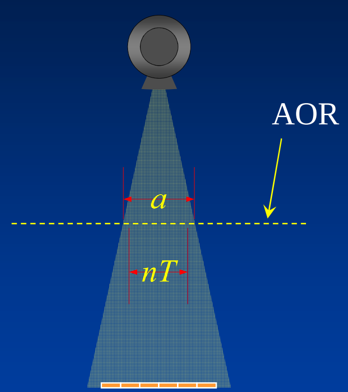
Though the meaning of *mA* is clear, *s* can be ambiguous in CT. When *s* means, as shown in the previous slides, the time, t_d, that a point on the axis is exposed to *detected* radiation, *mA* x *s* is called “effective mAs” or “mAs/slice” (Philips). Thus “effective mAs” or “mAs/slice” in CT has essentially the same meaning as “mAs” does in other x-ray applications.

Not really a word, *mAs* is more of a mix of an acronym and algebraic symbolism and when used by itself in CT, *mAs* often, but not always, means *mA* x *s*/rotation which is not the same thing and does not tell as complete a story as “effective mAs.”

Now semi-official: *Tube current time product* and *effective tube current time product*. (AAPM CT Lexicon version 1.1, 08/31/2011)

15

Width of Detected Beam vs Actual Beam Width



a = Geometric projection of the *z*-collimator aperture onto the axis of rotation (AOR) by a “point” focal spot. For MDCT $a>nT$ to keep the overbeaming (“penumbra”) beyond the active detector length nT. This is called “overbeaming.”[1]

Which is correct? *a* or *NT*?

[1]R. L. Dixon and J. M. Boone, Med. Phys. **37**, 2703-2718 (2010).

16

Pitch (IEC)

$$Pitch = \frac{\underline{Table\ movement/rotation}}{(detected)\ beam\ width}$$

$$p = \frac{b}{nT}$$

where b is the table increment per rotation and p is the pitch. (The word *pitch* has a related meaning for screw threads. A course-threaded screw has a higher pitch than a fine-threaded screw. Usually pitch is used only for helical scans but it could be applied to axial scans as well.)

17

Pitch = 1

← One row of detectors →

Pitch = 2

Pitch = 1

← Two rows of detectors →

Pitch = 2

(Another definition for pitch, table movement per detector channel, is sometimes used. This second definition was used by Siemens on its first multislice machines. Toshiba calls this second definition the *helical pitch* and uses *pitch factor* for the standard definition.)

18

Pitch and Effective mAs

$$mAs/slice = effective\ mAs$$

$$= \frac{mA \times t_r}{p}$$

where t_r is the time per rotation. The final expression is the one most often used as the definition of effective mAs. Effective mAs (or mAs/slice—Philips) is generally used with helical scans though the generalized definition $mA \times t_d$ could be applied to axial CT as well as other x-ray modalities.

19

Proof:

The time t_d is the detected beam width nT divided by the table speed v and so the effective mAs is

$$mAs_{eff} = mA \times t_d = mA \times \frac{nT}{v}$$

Note that the effective mAs is independent of rotation time t_r

20

Proof (2):

Now, if we multiply and divide by the rotation time t_r , and do some algebra, the effective mAs becomes

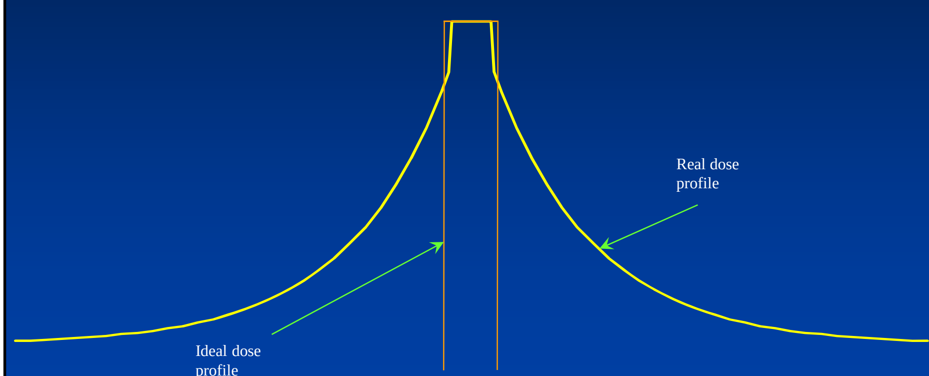
$$mAs_{eff} = \frac{mA}{v/(nT)} = \frac{mA \times t_r}{vt_r/(nT)} = \frac{mA \times t_r}{b/(nT)} = \frac{mA \times t_r}{p}$$

where $b = vt_r$ is the table increment per rotation and p is the pitch. The final expression is the one most often used as the definition of effective mAs.

21

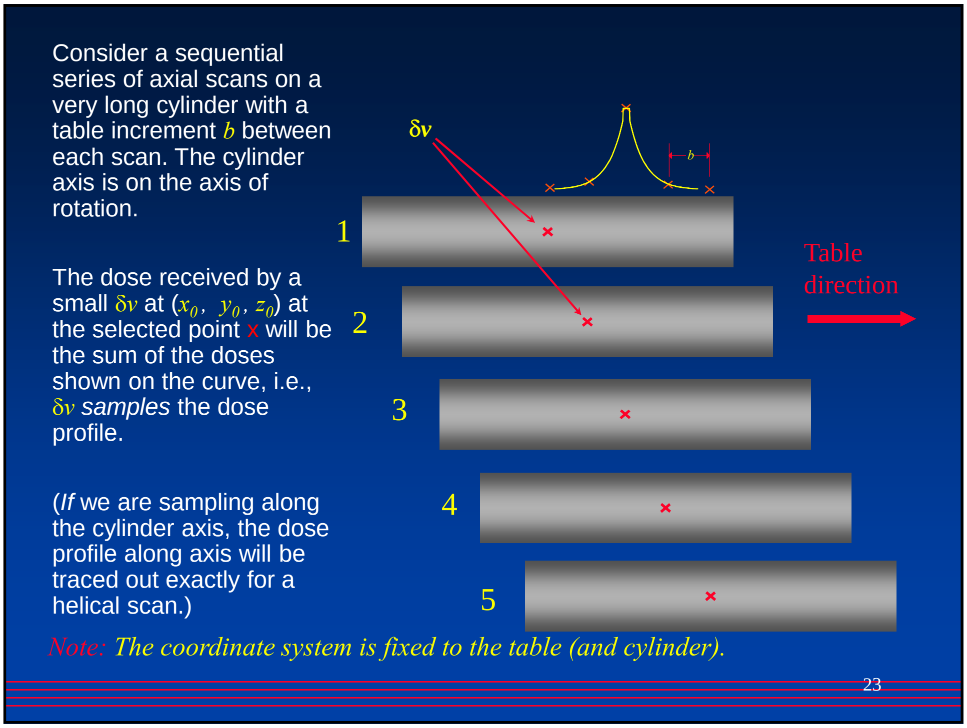
Dose Profile

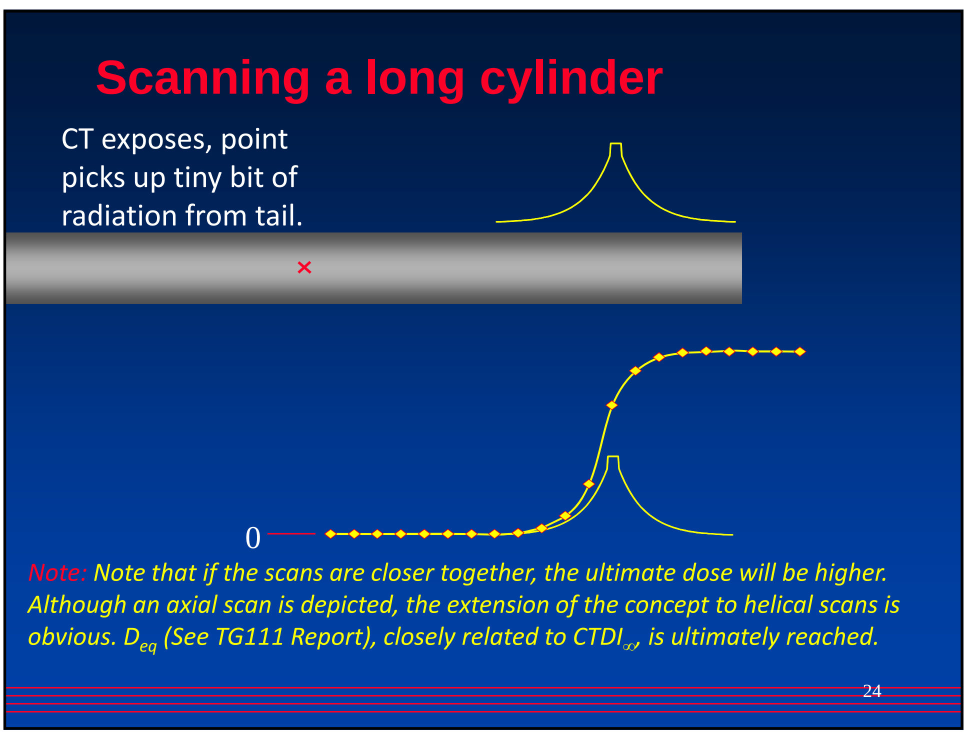
Because of scatter, the dose profile can extend well beyond the nominal collimated beam width a.



Based on data from TG200 very long phantom measurements.

22


Consider a sequential series of axial scans on a very long cylinder with a table increment b between each scan. The cylinder axis is on the axis of rotation.
The dose received by a small δv at (x_0, y_0, z_0) at the selected point x will be the sum of the doses shown on the curve, i.e., δv *samples* the dose profile.
(*If* we are sampling along the cylinder axis, the dose profile along axis will be traced out exactly for a helical scan.)
δv
b
1
2
3
4
5
Table direction
Note: *The coordinate system is fixed to the table (and cylinder).*
23


Scanning a long cylinder
CT exposes, point picks up tiny bit of radiation from tail.
0
Note: *Note that if the scans are closer together, the ultimate dose will be higher. Although an axial scan is depicted, the extension of the concept to helical scans is obvious. D_{eq} (See TG111 Report), closely related to $CTDI_{\infty}$, is ultimately reached.*
24

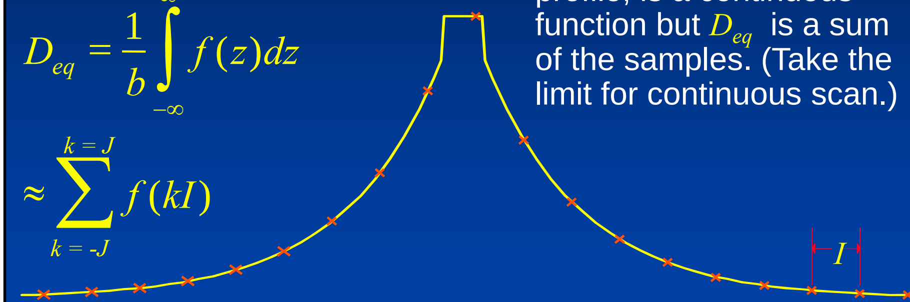
Dose from scan

(More detail, smaller b.) The total dose to the δv which samples the dose profile is

$$D_{eq} = \frac{1}{b}\int_{-\infty}^{\infty} f(z)dz$$

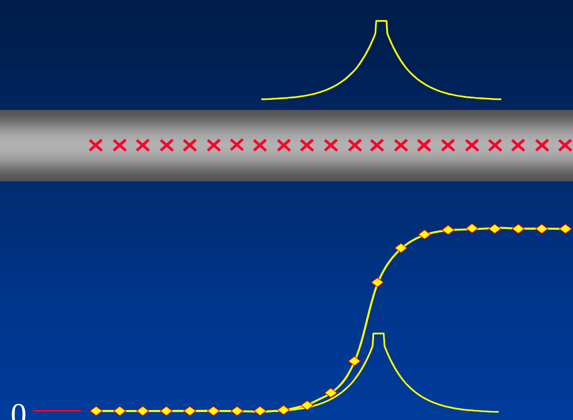
$$\approx \sum_{k=-J}^{k=J} f(kI)$$

where k is an integer, and Jb is far enough away so that the Nth contribution to the dose at the ROI is negligible. $f(z)$, the dose profile, is a continuous function but D_{eq} is a sum of the samples. (Take the limit for continuous scan.)



25

Alternatively,



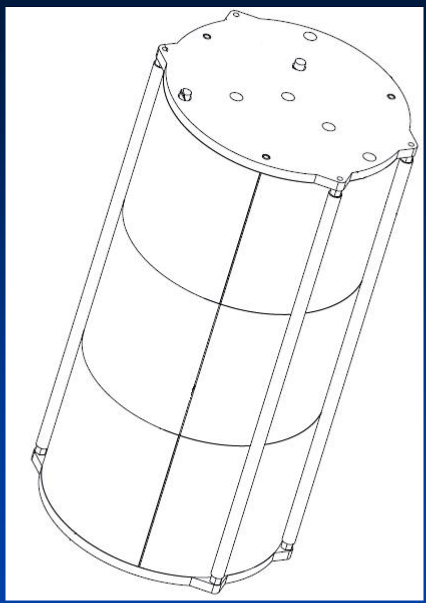
one could perform only one axial exposure with the phantom not moving and measure the dose at points all along the dose profile. Then add all the numbers together. (In order to make the number come out the same, the dose would be measured at points the same distance apart as the table movement per rotation of the previous slide.)

26

Infinity is impractical. How long does the cylinder need to be to effectively reach D_{eq}?

It is convenient to define L_{eq} as the value of L where D has reached 98% of its limiting value. (This is around 45 cm for 32 cm diameter acrylic.) For this scan of finite length, we no longer have translational symmetry. However, the central region of the scan has the same dose distribution as there *would* be for an infinite scan.

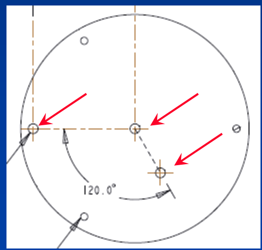
27



TG200 Phantom

(built by John Boone and his team at UC Davis).

30 cm in diameter by 60 cm in length and is made of high density polyethylene. There are three sections, 29.3 lb (13.3 kg mass) each. (The 32 cm CTDI phantom has a mass of 14.4 kg and weighs 31.7 lb). Three holes are bored deep into the phantom for probes.

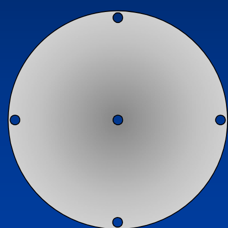


28

$\overline{D_{eq}}$

D_{eq} changes with radial depth. $\overline{D_{eq}}$ is the spatial average value of D_{eq} over the cross section of the cylinder.

Off axis measurements of D_{eq} are modulated by the asymmetric attenuation from the cylinder seen by the detector when the above method is used. Choosing a pitch < 1 nearly eliminates the effect of this modulation on the integral[1].



Note: Even with this ripple, the average dose over *any* cross section is still $\overline{D_{eq}}$ for an infinite scan.

[1]Dixon & Ballard, Med. Phys. 34, 3399 (2007).

29

E_{tot}

Dose (energy/mass) integrated over entire volume of cylinder.

$$E_{tot} = N\rho \boxed{\int_{-\infty}^{\infty}\int_{0}^{R} f(r,z)2\pi r\,dr\,dz} = \rho Nb\int_{0}^{R} 2\pi r\,dr\left\{\frac{1}{b}\int_{-\infty}^{\infty} f(r,z)dz\right\} = \rho L\int_{0}^{R} D_{eq}(r)\,2\pi r\,dr$$

$$E_{tot} = \rho\pi R^2 L\left\{\frac{1}{\pi R^2}\int_{0}^{R} D_{eq}(r)2\pi r\,dr\right\} = \rho\pi R^2 L\overline{D_{eq}}$$

(Note the resemblance of $\overline{D_{eq}}$ and E_{tot} to $CTDI_{vol}$ and DLP.) Though E_{tot} divided by the scanned volume is equal to $\overline{D_{eq}}$, the energy spills out of the scanned volume and so $\overline{D_{eq}}$ is *not* equal to the average energy within the scanned volume (for a finite scan).

30

Finite Scans on an Infinite Cylinder: *h*(*L*)

Consider a scan of length L along a line (in the z direction) within the cylinder. The dose at the center of the scan is $D(L)$ and

$$\lim_{L\to\infty} D(L) = D_{eq}$$

Define $h(L) = D(L)/D_{eq}$.

- h(L) has the limiting upper value of 1 as L increases.
- $h(L)$ is nearly independent of kV and collimation width.
- The scatter component is significantly greater in the center than at the periphery.

31

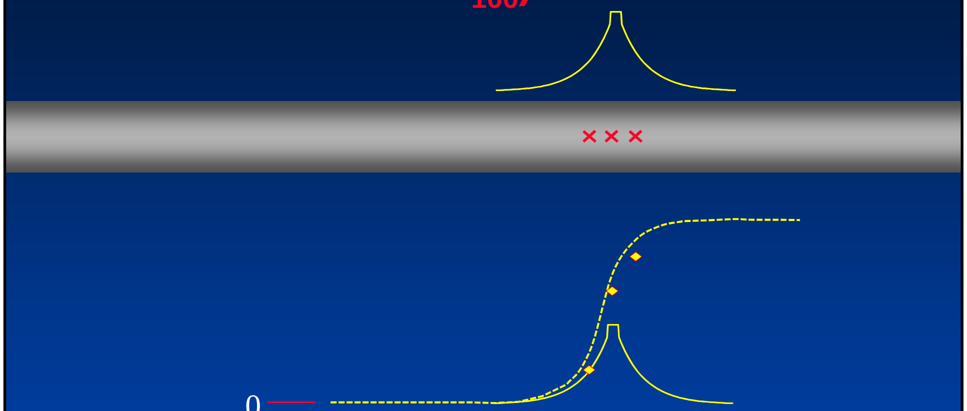
$\mathbf{CTDI_L}$ and D(L)

$$CTDI_L = \frac{1}{nT}\int_{-L/2}^{L/2} f(z)dz \qquad D(L) = \frac{1}{b}\int_{-L/2}^{L/2} f(z)dz$$

b is the table increment per rotation and nT is the *detected* beam width, measured at the axis of rotation. nT is less than the collimated beam width a. (For a pitch of 1, the actual beam overlaps the detectors.) In the limit as L grows very large, these parameters go to $CTDI_\infty$ and D_{eq}. (In order to confuse matters further, $CTDI_{vol}$ effectively has the divisor b instead of nT. More on that later.

32

What about a finite scan (e.g. $CTDI_{100}$)?



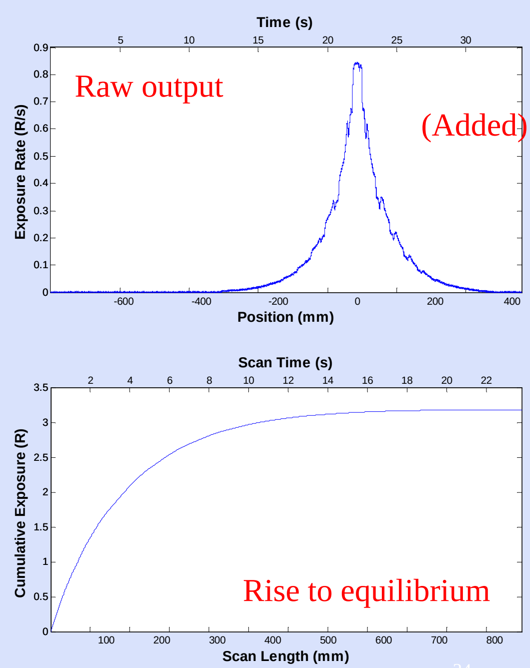
An ion chamber 100 mm long is not long enough to capture the entire dose profile and leaves off the tails. Thus, as you can see, the cumulative value is less than would be for an infinite scan.

33

- **D_{eq} and scan length**

- **120 kV helical scan w/ the probe positioned at the CENTER of the phantom**
  - Integral of the dose profile with increasingly longer scan lengths. *H(L)* only weakly dependent on kV and collimation

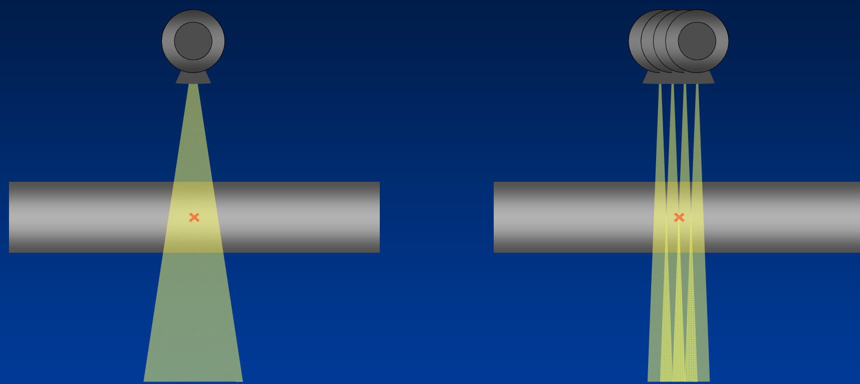
Slide from Sarah McKenney
(SSM20-05 • 03:40 PM, Wednesday)



34

Irradiated length

(Added)

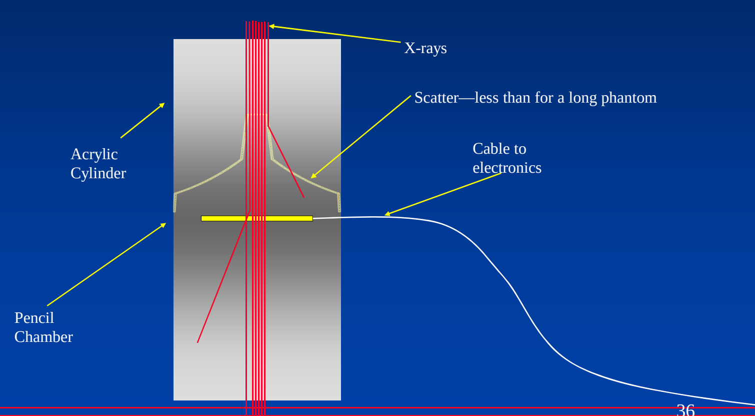


Equivalence between $f(0)$ and $D_L(0)$ described in detail by Dixon and Boone (Med. Phys. 37 (6) pp. 2703-2718, 2010 and summarized in AAPM TG111 report. (Note that *some* discrepancy is expected.)

35

CTDI_{100} using Pencil Chamber

For CTDI_{100}, scanning through 100 mm is replaced by integrating the dose from a single gantry rotation with a stationary table using a 100 mm long pencil chamber.

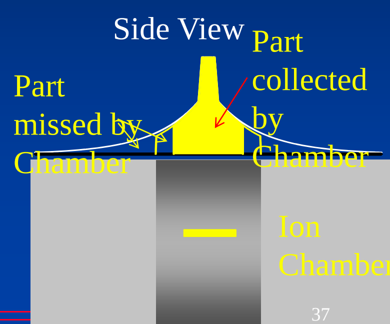


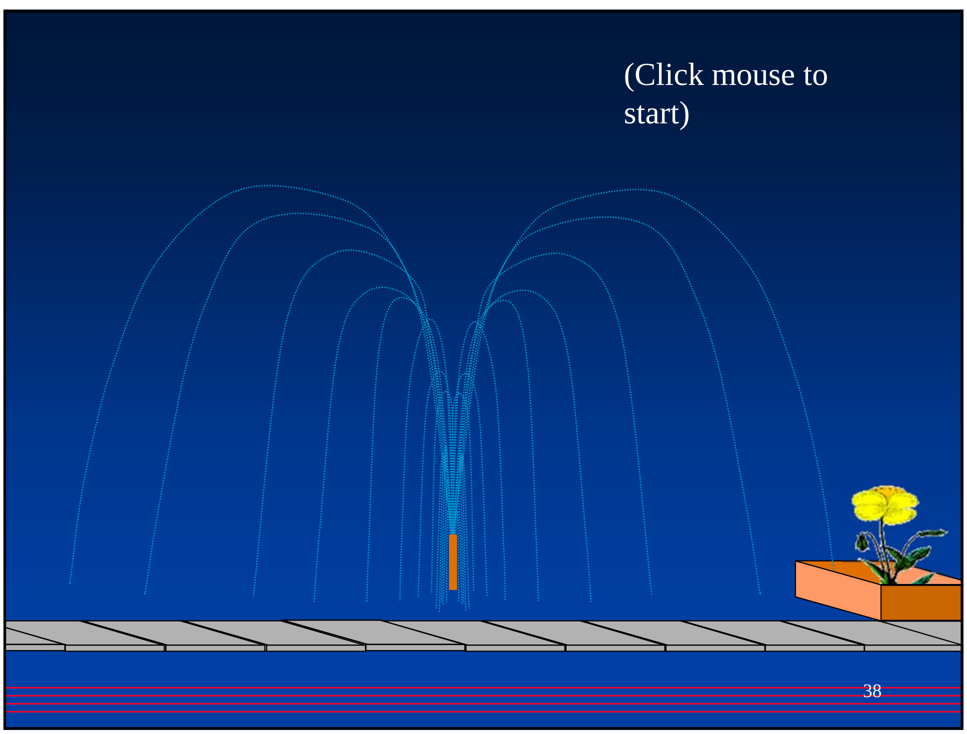
36

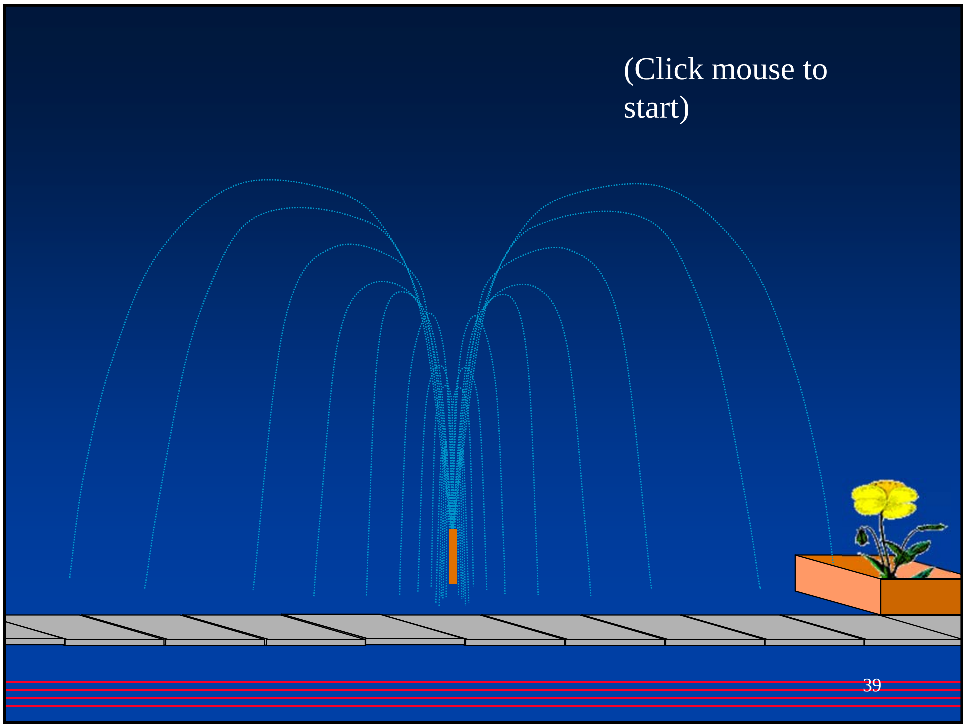
$CTDI_{100}$ using Pencil Chamber

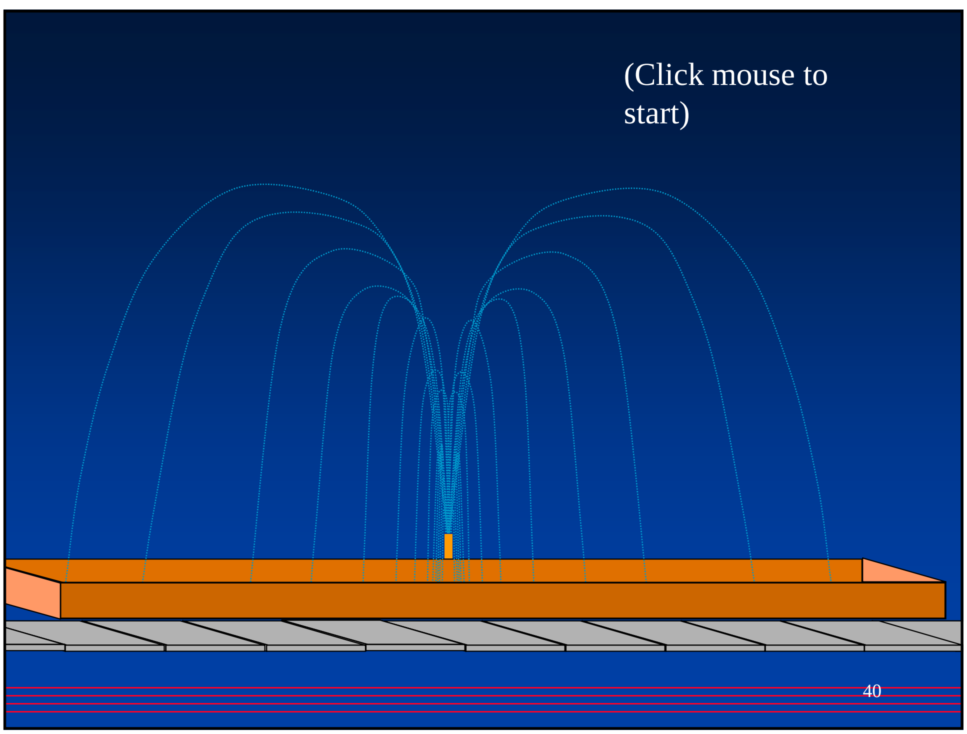
- CTDI is currently measured with 100mm ionization chamber
- Acrylic cylinder that is 160mm long and either 160 mm or 320 mm diameter

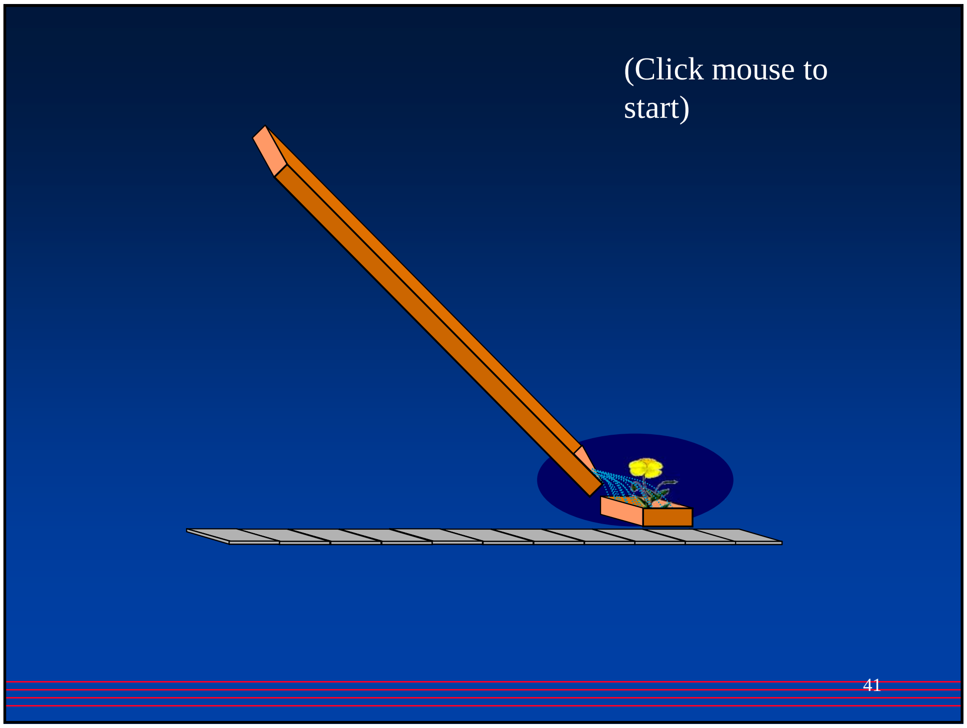
- The bottom figure shows that the 100 mm long pencil chamber (solid yellow) only collects a portion of the radiation profile.
- The white line is the profile that would occur for an infinitely long phantom

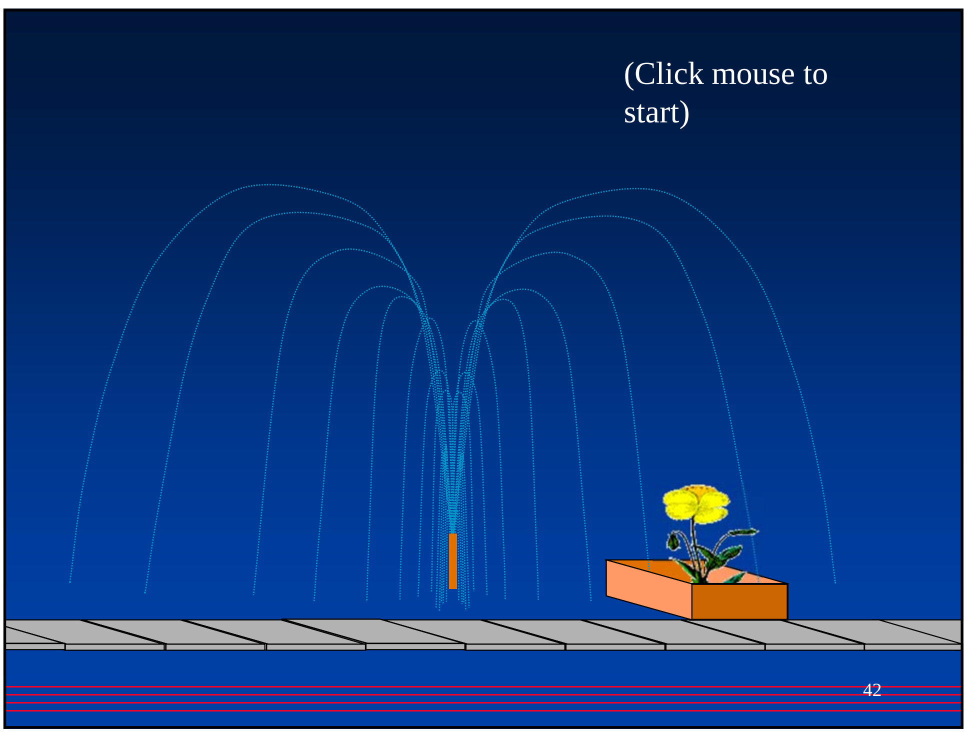


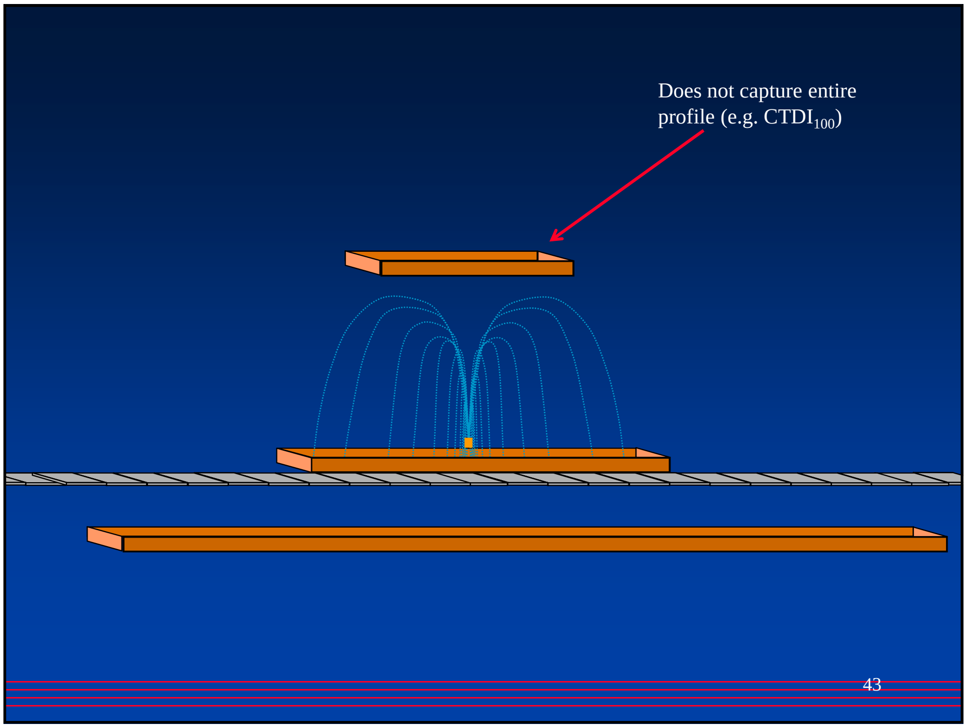


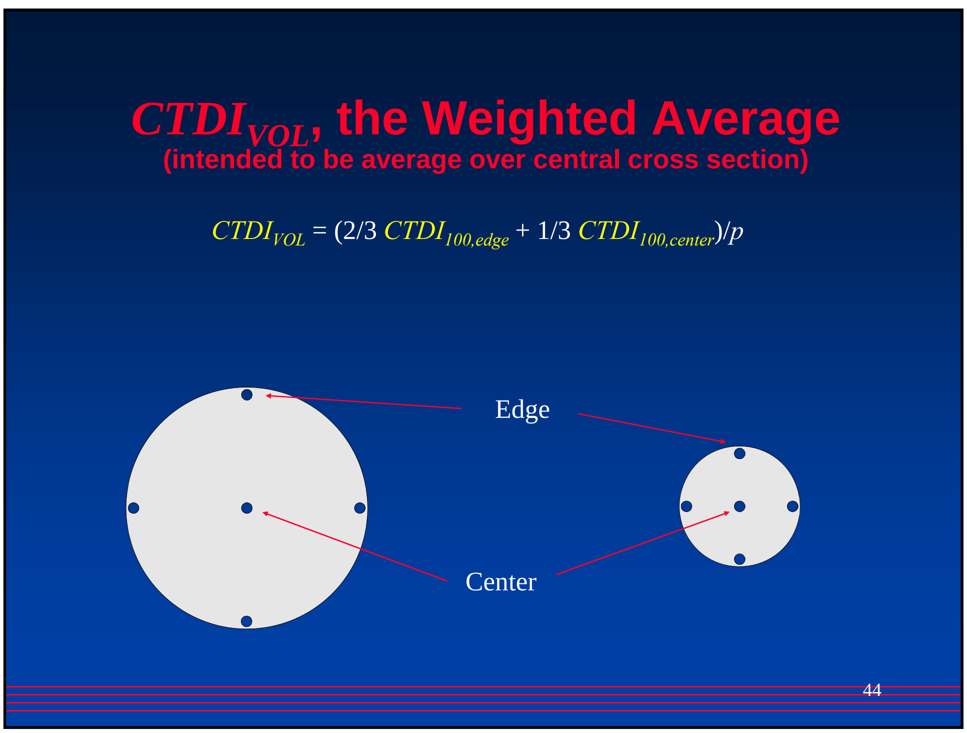

(Click mouse to
start)
39


(Click mouse to
start)
40


(Click mouse to
start)
41


(Click mouse to
start)
42


Does not capture entire profile (e.g. CTDI100)
43


CTDIVOL, the Weighted Average
(intended to be average over central cross section)
CTDIVOL = (2/3 CTDI100,edge + 1/3 CTDI100,center)/p
Edge
Center
44

$CTDI_{VOL}$ is a useful index of scanner output[1]

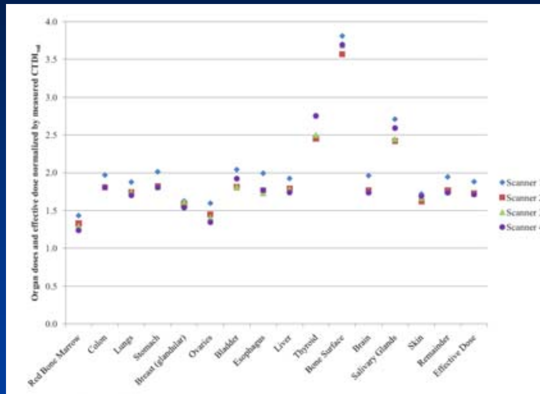
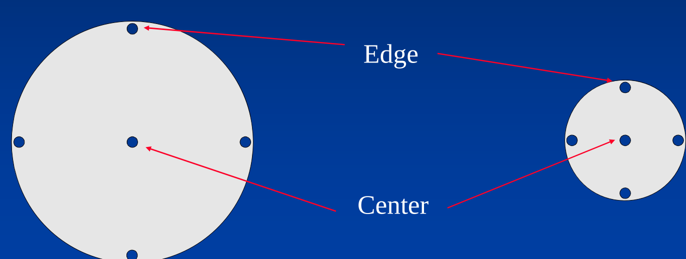


FIG. 2. $CTDI_{vol}$, S normalized organ ($nD_{S,O}$), and effective ($nD_{S,ED}$) doses for scanners 1–4.

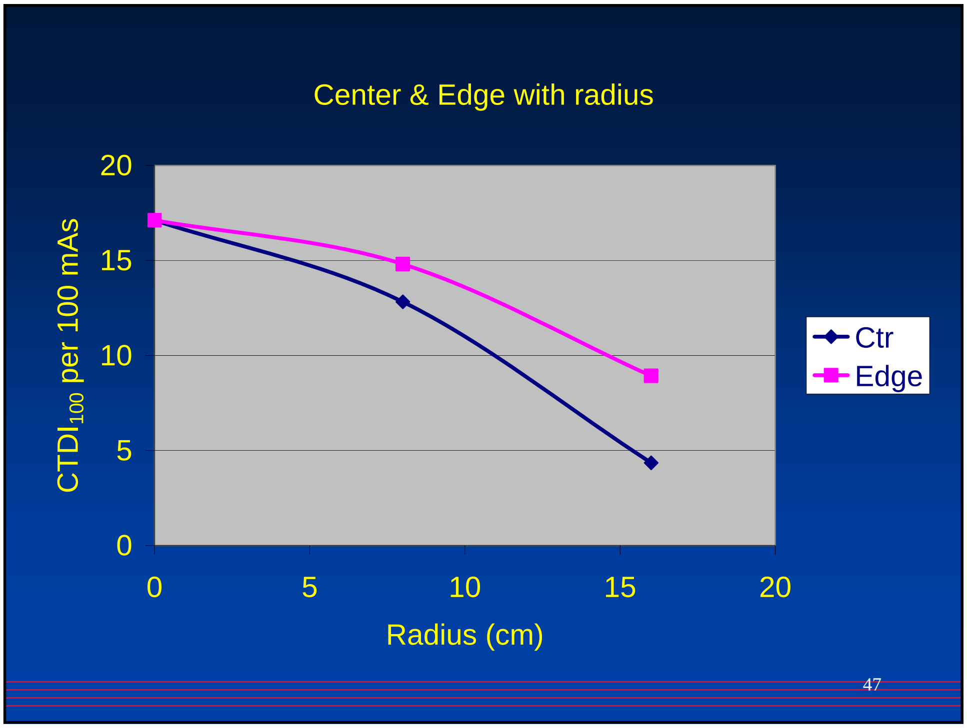
[1]A. Turner et al.,, "The feasibility of a scanner-independent technique to estimate organ dose from MDCT scans: Using CTDIvol to account for differences between scanners," Med. Phys. 37(4), 1816-1825 (2010).

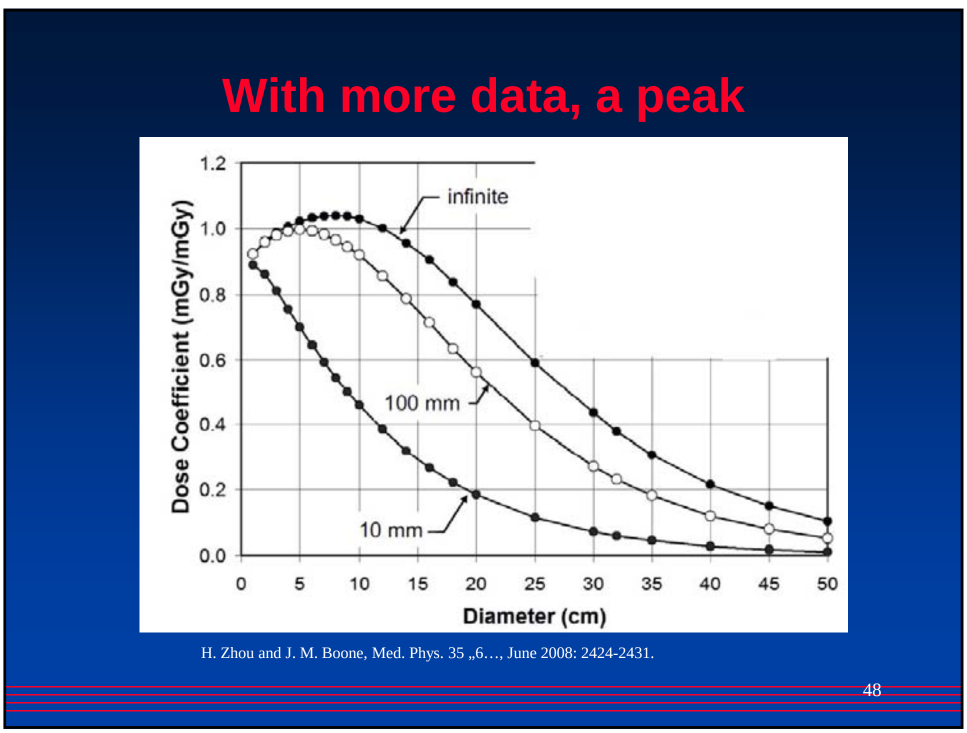
45

Patient Size: $CTDI_{100}$ for center and edge for two sizes and for air

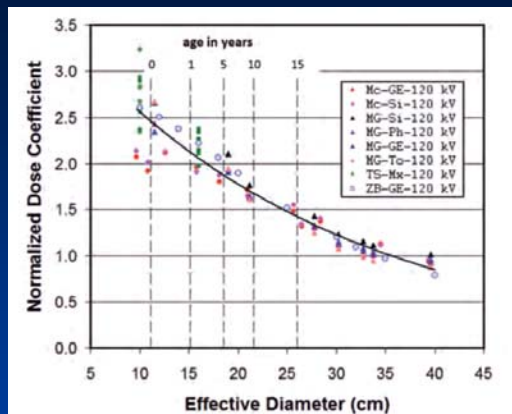


46


Center & Edge with radius
20
15
10
5
0
CTDI100 per 100 mAs
0
5
10
15
20
Radius (cm)
Ctr
Edge
47


With more data, a peak
Dose Coefficient (mGy/mGy)
1.2
1.0
0.8
0.6
0.4
0.2
0.0
0
5
10
15
20
25
30
35
40
45
50
Diameter (cm)
infinite
100 mm
10 mm
H. Zhou and J. M. Boone, Med. Phys. 35 „6…, June 2008: 2424-2431.
48

TG204



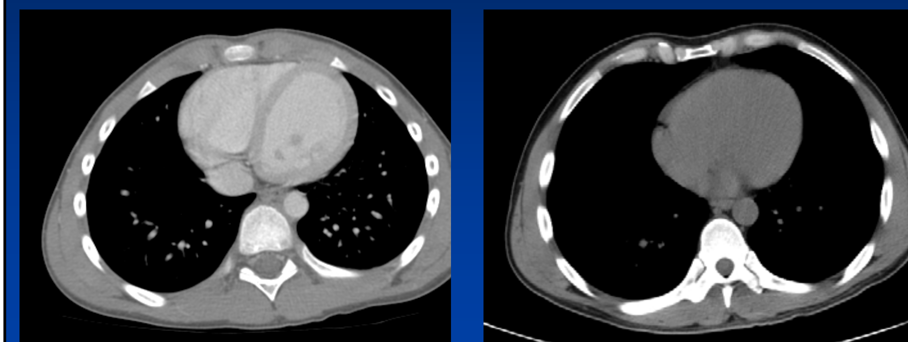
The normalized dose coefficient for the 32 cm PMMA $CTDI_{vol}$ phantom is shown as a function of effective diameter. The individual data points correspond to four independent research groups, as indicated in the key. (Mc=McCollough, MG=McNitt-Gray, TS=Toth/Strauss, ZB=Zhou/Boone). Scanners represented are also indicated in the key (GE=General Electric, Si=Siemens, Ph=Phillips, To=Toshiba, Mx=Mixed Scanner manufacturers).

49

Is a child a small adult?

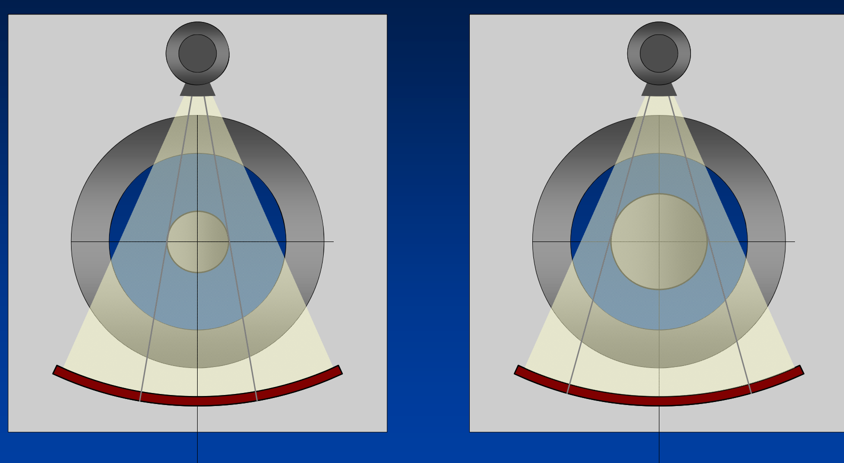
50

Child and adult: Which is which?



51

What needs to happen here?



52

So, how does this affect patient dose in a child?

- Magnification requires a higher detector dose.
- But lower attenuation allows us to reduce dose.
- How does this balance out? Along with size considerations, we also need to account for the diagnostic needs of the child.
- Goske, Strauss et al., suggest that SSDE can be used to adjust an adult protocol to fit a child (or a small adult or a very large adult).
- (What about slice width?)

M. J. Goske, K. J. Strauss et al., "Diagnostic Reference Ranges for Pediatric Abdominal CT," Radiology 268(1), 208-218 (2013).

53

$$SSDE\ ratio = 0.14 + 0.025 \times BW/cm$$

where BW is body width (lateral patient dimension).
14.4 cm (small child) for a ratio of 0.5
34.4 cm ("standard" adult) for a ratio of 1

M. J. Goske, K. J. Strauss et al., "Diagnostic Reference Ranges for Pediatric Abdominal CT," Radiology 268(1), 208-218 (2013).

54

Summary & Conclusions

- Effective mAs is mA x direct exposure time.
- Differences between physical and detected (used for imaging) beam can be significant.
- Considering infinite scans (translational symmetry) leads to simpler (and cleaner) physics.
- For a given (long) cylinder, the rise to equilibrium function $h(L)$ is very robust.
- $CTDI_{vol}$, though more complex than D_{eq}, is a very useful indicator of scanner output.
- SSDE gives us a simple way to scale dose indices to patient size.

55

Thank you

Henry Ford Health System

56