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

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**Diagnostic Radiology**

## Introduction

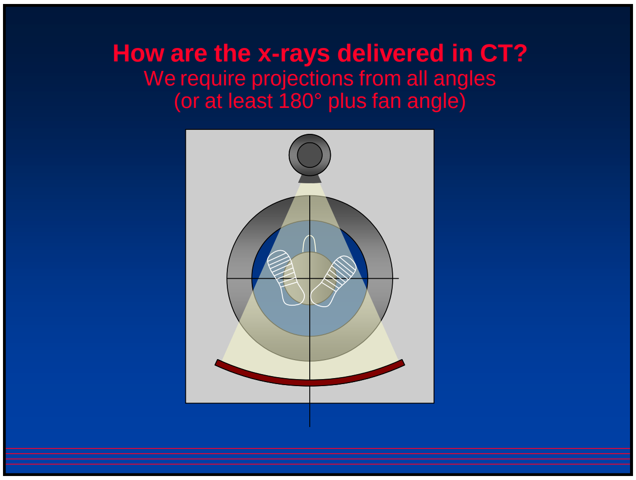
To be completed later

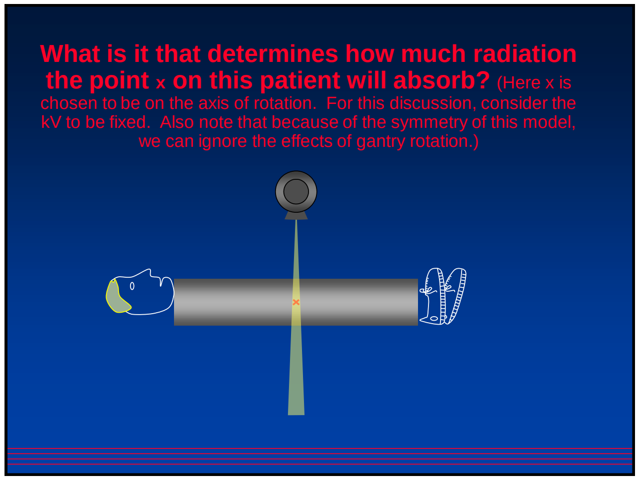
## Topics

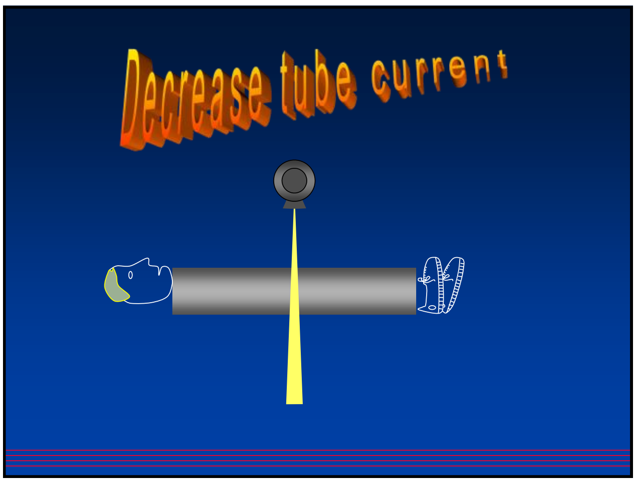
I. Operator controlled factors going into CT radiation dose
II. The infinite cylinder, its parameters and TG111
III. Details in the development of $D_{eq}$ and its derivatives
IV. The finite scan, $h(L)$ and CTDI
V. SSDE and accounting for patient size

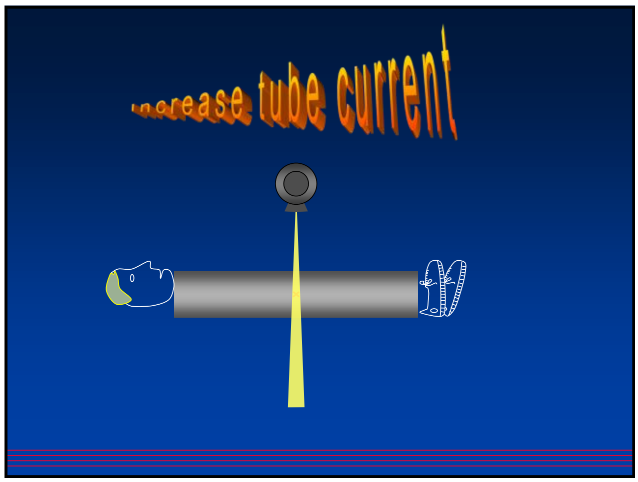
## CT Dose

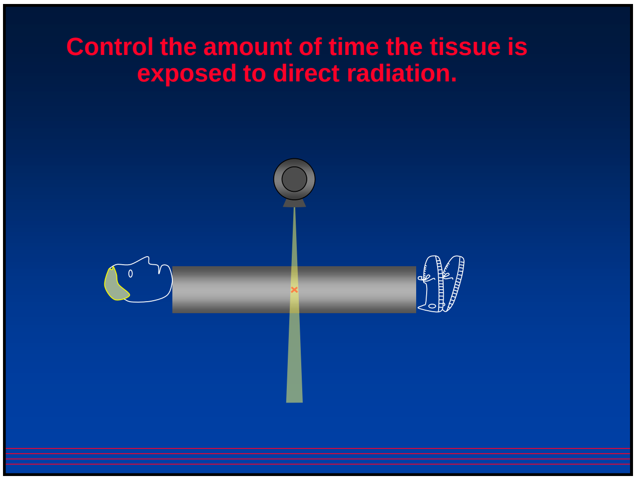
The radiation dose from CT is relatively high compared to most (but not all) x-ray exams. It is often in the same ballpark as interventional radiology and cardiology. There is some difference in that ordinarily, the radiation in CT exams is applied more uniformly and so radiation burns are generally less of a risk than for those modalities for the same total absorbed energy.


How are the x-rays delivered in CT?
We require projections from all angles
(or at least 180° plus fan angle)


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 kV to be fixed. Also note that because of the symmetry of this model, we can ignore the effects of gantry rotation.)

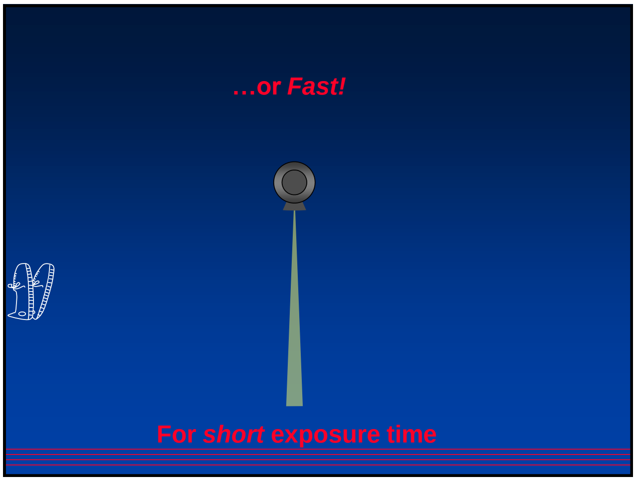

Decrease tube current


Increase tube current

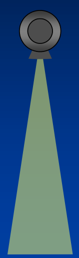



We have been looking at one data acquisition cycle of what is officially called an *axial* scan (or sequential scan—Siemens or step and shoot—Toshiba).

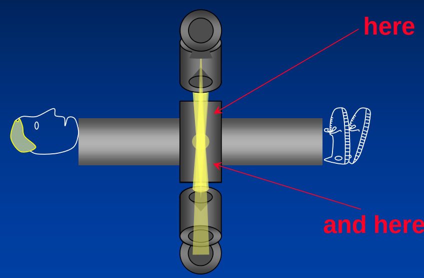
An alternative method of scanning is called a *helical* (or spiral) scan where the tube is left on and the patient is moved through the beam continuously.

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



**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.**



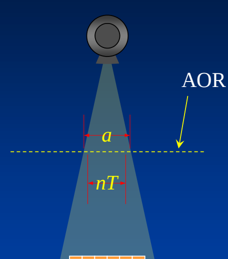

## mAs Confusion

Though the meaning of *mA* is clear, *s* can be ambiguous in CT. When *s* means, as shown in the previous slides, the time that a point on the axis is exposed to radiation, *mA* x *s* is called “effective mAs” or (unfortunately) “mAs/slice.” 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 quite 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)

## 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 penumbra beyond the active detector length *nT*. This is called “overbeaming.”[1]

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

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

Discuss & explain with one slide. Explain how NT (detected beam width) is important for image statistics while physical beam width *a* is appropriate in determining dose.

## Pitch (IEC)

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

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

where $I$ 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.)

← One row of detectors →

Pitch = 1

Pitch = 2

← Two rows of detectors →

Pitch = 1

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.)

## 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 definition above *could* be applied to axial scans (or any other *x*-ray exam).

## 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 $t_r$

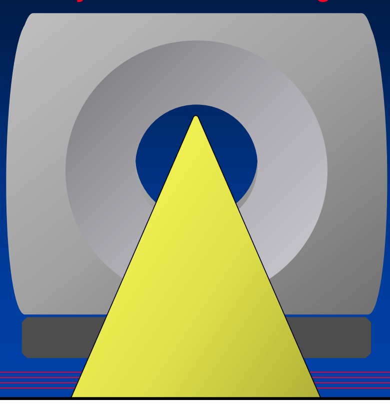
## Proof (2):

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

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

where $I$ 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.

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



## Consider two infinite scans with same mA, kV, table speed (m/s) and collimation. The first has double the rotation time of the second. Here *dose* means $D_{eq}$

| | |
|---|---|
| 0% | 1. Dose from first scan is twice that of the second. |
| 0% | 2. Dose from first scan is half that of the second. |
| 0% | 3. At high rotation time, penumbra begins to dominate so the dose from first scan is a little more than double that of the second. |
| 0% | 4. Dose is the same from both scans |
| 0% | 5. We need to know that the pitch is not much greater than 1 to know the dose ratio. |

Answer: 4-they are the same.

Ref: AAPM TG111 report

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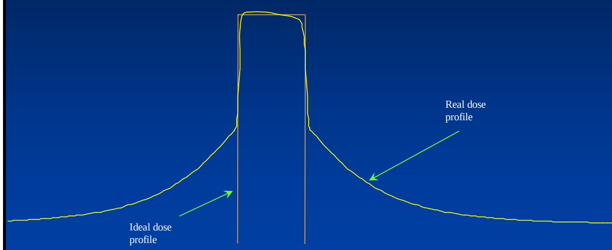
# Introduce $D_{eq}$, $\overline{D_{eq}}$ and $E_{tot}$ from TG111.

This will take a couple of slides. At this point just introduce the concept of $\overline{D}_{eq}$, as the spatially averaged dose in the center of the scan. A more formal description will occur later. Similarly introduce *Deq* and $E_{tot}$ informally.

# Dose Profile

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

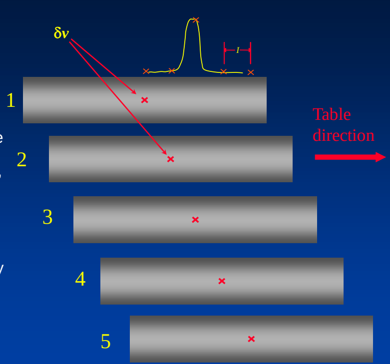



[1]Deduced from Fig. 4 in J. M. Boone, Med. Phys. 34(4), 1364-1371 (2007).

Consider a series of sequential (axial) scans on a very long cylinder with a table increment *b* between each scan.

The dose received by a small $\delta v$ at $(x_0, y_0, z_0)$ at the selected point × will be the sum of the doses shown on the curve $x_0$ i.e., $\delta v$ *samples* the dose profile.

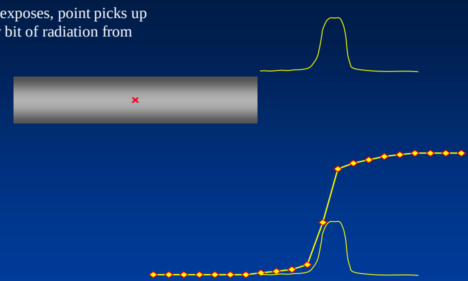
(For a spiral scan along the center of a cylindrically symmetric rod, the curve will be traced out exactly rather than sampled.)




*Note: The coordinate system is fixed to the table (and cylinder).*

# Consider a really long cylinder

CT exposes, point picks up tiny bit of radiation from tail.



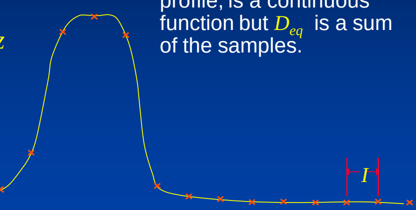
*Note: The coordinate system is fixed to the table (and cylinder) $D_{eq}$ (See TG111 Report) is ultimately reached. Note that if the scans are closer together, the ultimate dose will be higher. Although an axial scan is depicted, the extension of the concept to helical scans is obvious.*

## Dose from Axial Scan Series

(More detail, smaller $b$.) The total dose to the $\delta v$ which samples the dose profile is

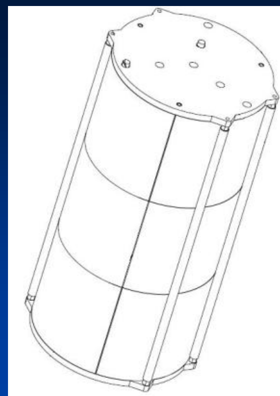
$$D_{eq} = \frac{1}{b}\int_{-\infty}^{\infty} f(z)dz \approx \sum_{k=-J}^{k=J} f(kI)$$

where $k$ is an integer, and $Jb$ is far enough away so that the $N$th 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.

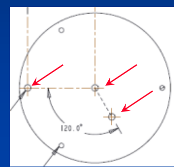



## TG200 Phantom

### (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.




## $E_{tot}$

We will show how the total energy absorbed by the cylinder in a single rotation is simply related to $\overline{D_{eq}}$ . It follows that $E_{tot}$ (the total energy absorbed in the scan) is simply $N$ times this value where $N$ is the # of rotations in the scan. This in turn leads to

$$E_{tot} = LA\overline{D_{eq}}$$

where $A$ is the area and $L$ is the scan length (suggesting a relationship to DLP). Thus $E_{tot}$ is the volume of the scan times the average value of $\overline{D_{eq}}$ . One has to be careful to interpret this properly. The energy peaks at the center of the scan and spills out beyond the edges of the scan. The next question emphasizes this:

## The following is true about $E_{tot}$

| | |
|---|---|
| 0% | 1. It is completely unrelated to DLP |
| 0% | 2. It is exactly equal to DLP |
| 0% | 3. Since it equals scan volume x $\overline{D}_{eq}$ , it is the average energy delivered to the cylinder over scan length $L$. |
| 0% | 4. It is greater than the energy delivered throughout the scan length. |
| 0% | 5. The relationship $E_{tot}$ = scan volume x $\overline{D}_{eq}$ is strictly true only when the scan length is at least $L_{eq}$, the scan length needed to reach $\overline{D}_{eq}$ . |

Answer: 3-It is greater (since some of the energy is deposited outside the scanned region).

Ref: TG111 report

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

(This is just a draft. Further development with illustrations is needed.) For an infinite cylinder, 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

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

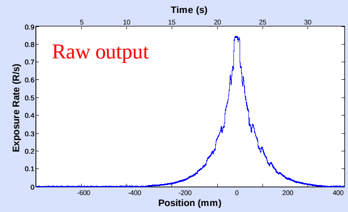
It is convenient to define a function $h(L) = D(L)/D_{eq}$. Thus $h$(L) has the limiting upper value of 1 as $L$ increases. Empirically, $h(L)$ has a weak dependence on both tube potential and collimation width. Also the scatter component is significantly greater in the center than at the periphery.

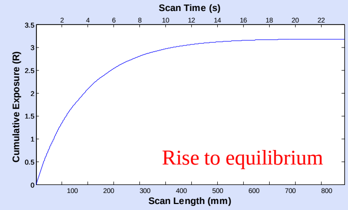
It is also convenient to define $L_{eq}$ as the value of $L$ where $D$ has reached 98% of its limiting value. (This is around 45 cm for 32 cm diameter acrylic.)

## • Rise to $D_{eq}$

- **120 kV helical scan w/ the probe positioned at the CENTER of the phantom**
  - The notches in the dose profile correspond to the attenuation from the patient table
- Integral of the dose profile with increasingly longer scan lengths. Integrals are centered around 0 mm

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




## The rise-to-equilibrium function *h*(*L*)

| | | |
|---|---|---|
| 0% | 1. | depends strongly on the collimation width. |
| 0% | 2. | depends strongly on tube potential. |
| 0% | 3. | approaches 1 (unity) more quickly at the periphery than at the center. |
| 0% | 4. | does not, surprisingly, appear to have any exponential dependency. |
| 0% | 5. | cannot be used to calculate the peak dose for values of $L$ less than $L_{eq}$. |

Answer: 3-Scatter diminishes earlier at the periphery.

Reference: TG111 report

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## CTDI

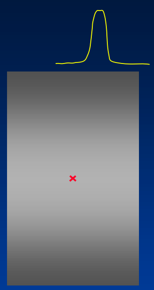
- Computed Tomography Dose Index
- $\mathrm{CTDI_{FDA}}$, $\mathrm{CTDI_{\infty}}$, $\mathrm{CTDI_{100}}$, $\mathrm{CTDI_{W}}$, $\mathrm{CTDI_{air}}$, $\mathrm{CTDI_{Vol}}$
- Measurement made from *single axial scan* (one revolution) using the cylindrical chamber
- Measurements are made on standard plastic cylinders to determine $\mathrm{CTDI_{FDA}}$, $\mathrm{CTDI_{100}}$, $\mathrm{CTDI_{W}}$, and $\mathrm{CTDI_{Vol}}$
- Measurement made in air (no phantom) for $\mathrm{CTDI_{air}}$

## CTDI

Here we will describe how CTDI is very similar in definition except that instead of the divisor being the table increment *b*, it is, instead, the *detected* beam width *NI*.

For $CTDI_{100}$, the phantom is 150 mm (~6") in length and the dose integral is measured using the 100 mm long pencil chamber and a single rotation in a fixed phantom. The measurement ***could*** be made ***directly*** by using a small chamber embedded in the phantom at point x and scanning over the central 100 mm.

In this scan, the dose at point x does not build up as much as it could for two reasons: 1) the abbreviated scan length and 2) lack of scatter from an extended cylinder. Extending the scan and cylinder lengths could add ~ 30% to the dose accumulating at the center.



## $CTDI_{100}$ measured at collimations of 40 mm and 1 mm

| | | |
|---|---|---|
| 0% | 1. | At 40 mm, CTDI is 40 times higher |
| 0% | 2. | At 40 mm, CTDI noticeably less than 1 mm due to penumbra |
| 0% | 3. | If values not within 5%, service should be called |
| 0% | 4. | $CTDI_{100}$ is same for each case |
| 0% | 5. | Because table speed is not given, we have incomplete information for the comparison |

Answer: 2-CTDI smaller at 40 mm than at 1 mm, often by a factor of two.

Ref: TG111 report and McCollough et al., Radiology (259), pp 311-316.

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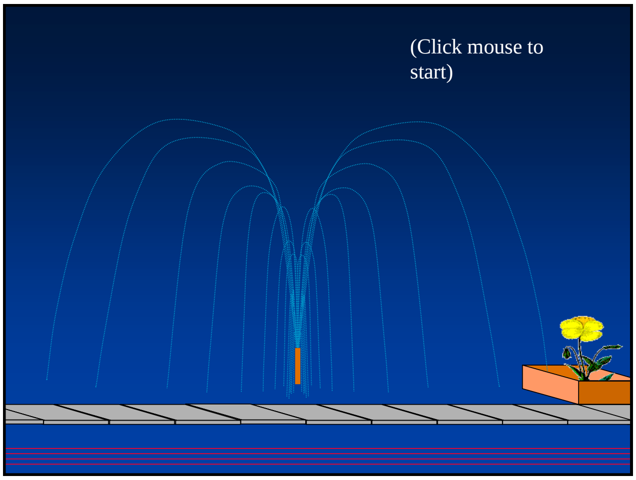
## Alternative

As an alternative, the CTDI method of measurement of the integral does so by using a long chamber and integrating all at once. (Add simple drawing to this slide. Decide about picture below.)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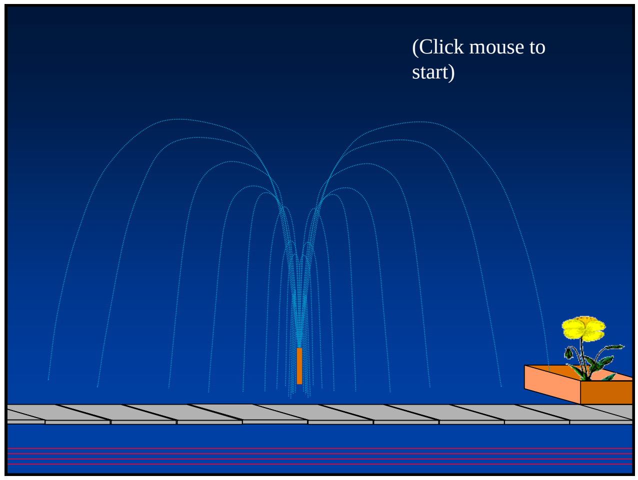
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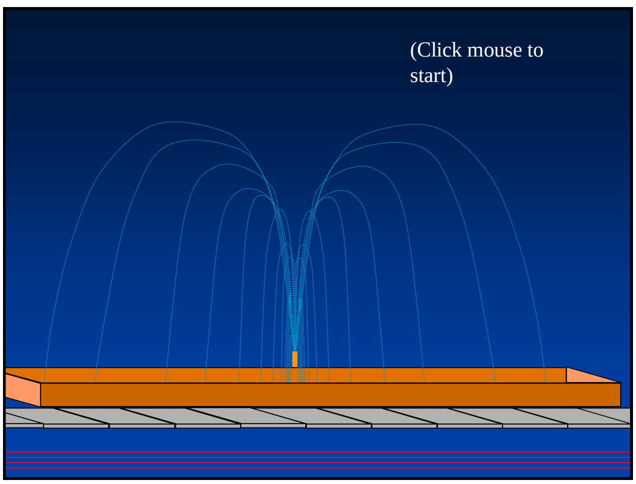
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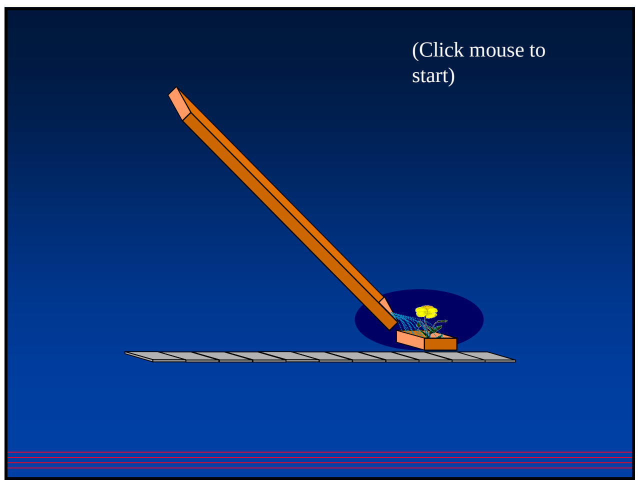
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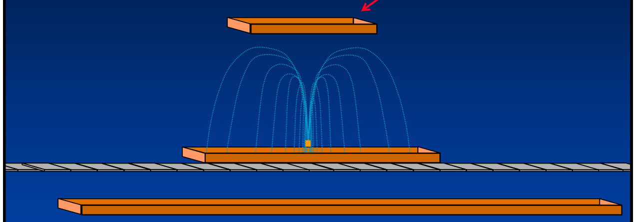
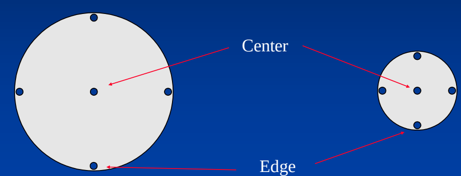


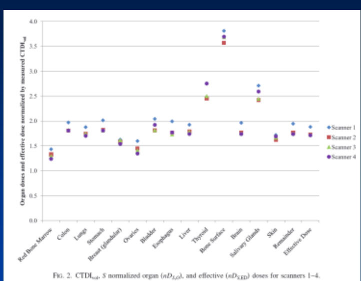

## $CTDI_{VOL}$, the Weighted Average

(intended to be average over central cross section)

$CTDI_{VOL}$ = (2/3 $CTDI_{100,edge}$ + 1/3 $CTDI_{100,center}$)/$p$




## With all of its warts, there is a new appreciation[1] for $CTDI_{VOL}$



[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).

## $CTDI_{vol}$

0% 1. describes the skin dose to an "average European male" abdomen.

0% 2. cannot be used to compare radiation outputs of different scanners.

0% 3. can be used to compare different scanners but only if accompanied by values of tube current, pitch and rotation time.

0% 4. at same kV is a good indicator of CT radiation output regardless of vendor, model, rotation time, pitch, tube current or collimation

0% 5. when multiplied by scan length, accounts for patient size.

Answer: 4

Ref: Turner et al., Med. Phys. 37(4), 1816-1825 (2010).

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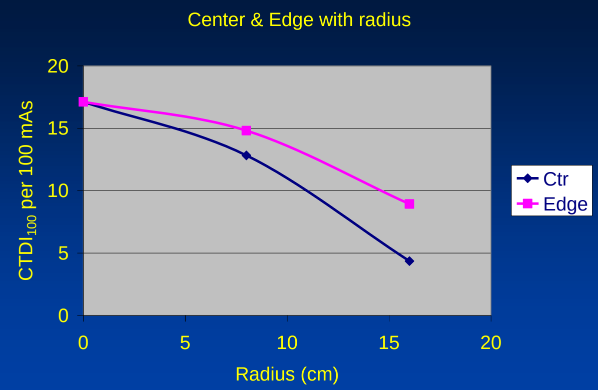
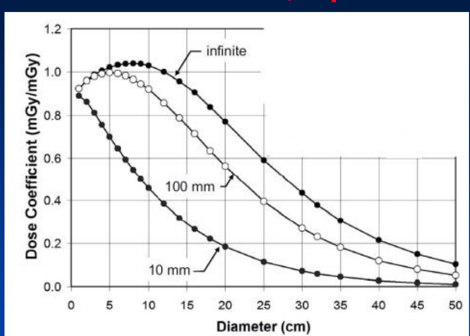
## Patient Size

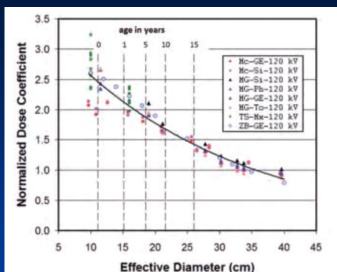



## With more data, a peak




H. Zhou and J. M. Boone, Med. Phys. 35 „6…, June 2008: 2424-2431.

## 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).

## Regarding TG204's Size Specific Dose Index (SSDE),

| | | |
|---|---|---|
| 0% | 1. | Dose is Energy/mass, already normalized and thus does not depend on patient size. |
| 0% | 2. | SSDE is determined from a linear extrapolation based solely on patient weight. |
| 0% | 3. | It incorporates a lookup table to account for vendor and model. |
| 0% | 4. | It is calculated using BMI and patient age. |
| 0% | 5. | It is independent of vendor, model and kV but uses measures of patient girth in patient scan region. |

Answer: 5-SSDE based solely on measurements relating to patient thickness or cross sectional area in the scan region.

Ref: AAPM TG204 report

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## The End