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

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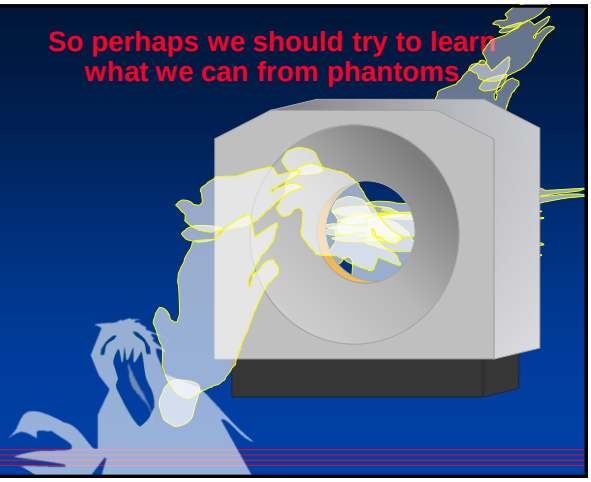
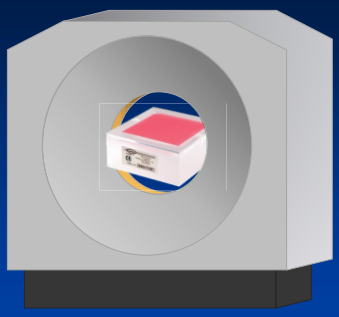
Diagnostic Radiology

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

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





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

Topics

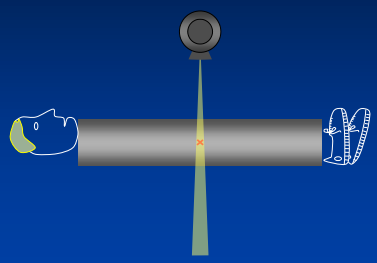
I. Operator controlled factors going into CT radiation dose
II. The infinite cylinder, its parameters and TG111
III. Details in the development of D_{eq} and its relatives
IV. The finite scan, $h(L)$ and CTDI
V. SSDE and accounting for patient size

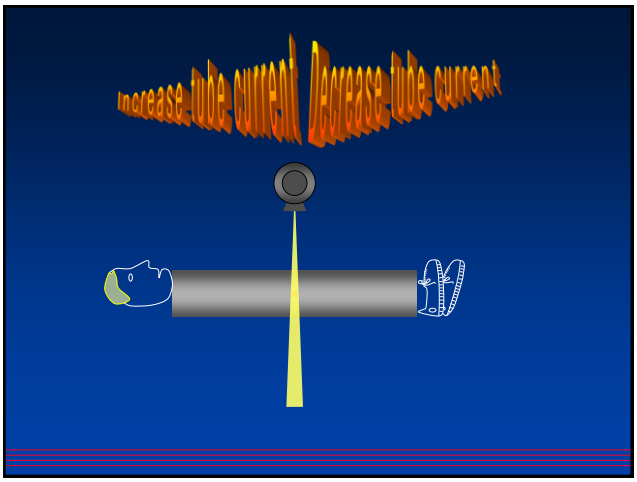
CT Dose

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.

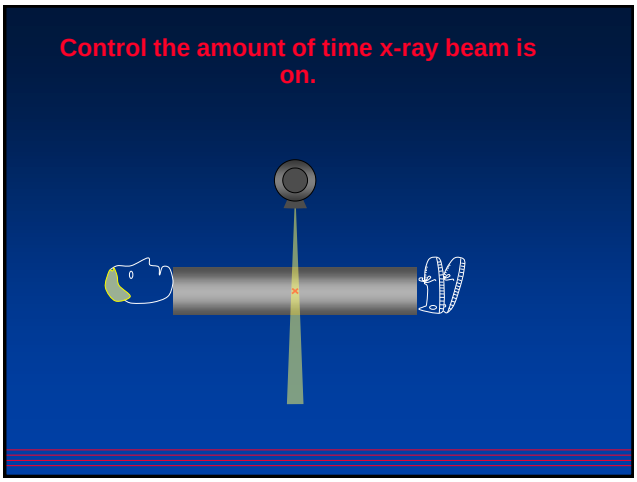
What is it that determines how much radiation the point x on this patient will absorb?

(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.)





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



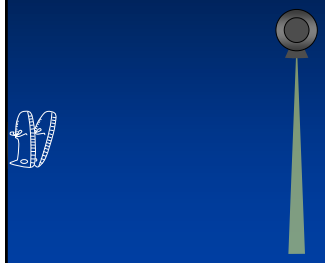
We have been looking at one data acquisition cycle of what is officially called an *axial* scan (or sequential scan—Siemens or scan and scan—Toshiba).

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.

Slow …

For *long* exposure time

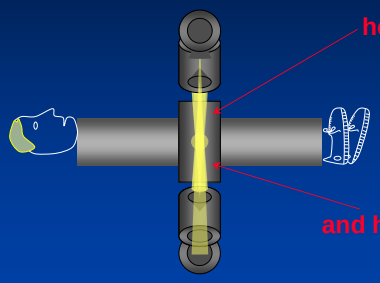
…or *Fast!*



For *short* exposure time

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.



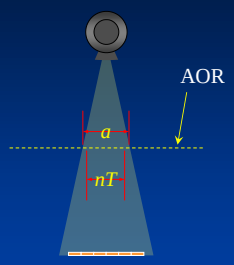
mAs Confusion

Though the meaning of *mA* is clear, *s* can be ambiguous in CT. When *s* means, as shown in the previous slides, the time, t_d, that a point on the axis is exposed to 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)

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]

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

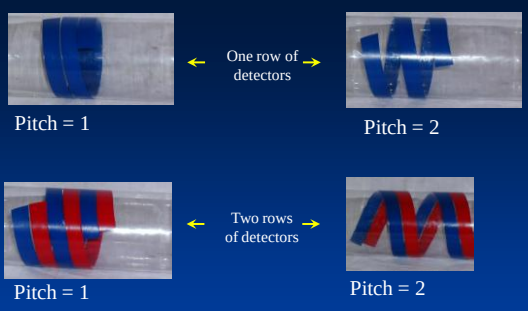
[1]R. L. Dixon and J. M. Boone, Med. Phys. **37**, 2703-2718 (2010).

Pitch (IEC)

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

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

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.)



(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 generalized definition $mA \times t_d$ could be applied to axial CT as well as other x-ray modalities.

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):

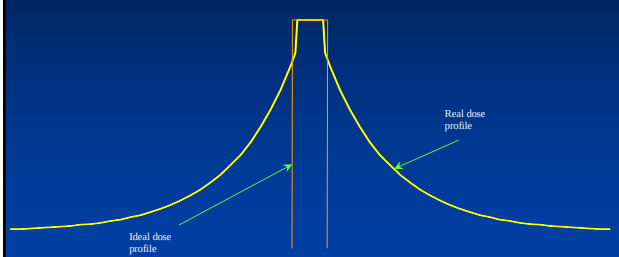
Now, if we multiply and divide by the rotation time t_r , and do some algebra, the effective mAs becomes

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

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.

Dose Profile

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

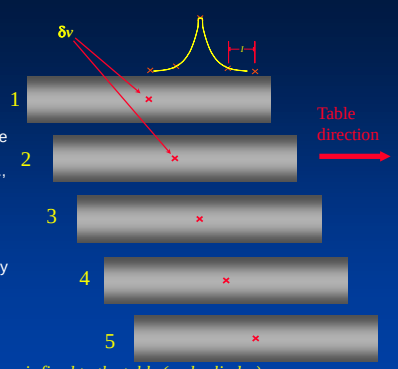


Based on data from TG200 very long phantom measurements.

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 δv at (x_0, y_0, z_0) at the selected point x will be the sum of the doses shown on the curve x_0 i.e., δv *samples* the dose profile.

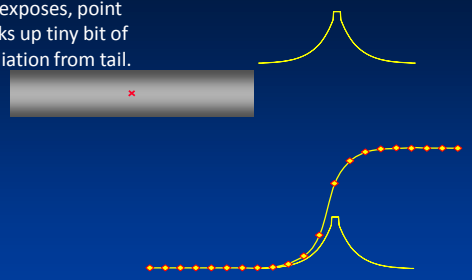
(For a spiral scan along the center of a cylindrically symmetric rod, the curve will be traced out exactly rather than sampled.)



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

For this 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.

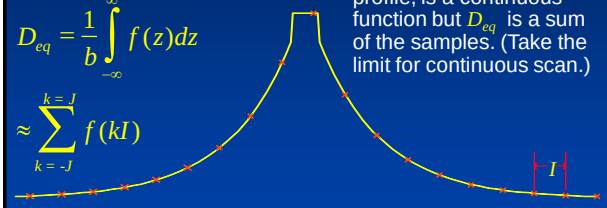
Dose from scan

(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.)



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. The dose from first scan is twice that of the second.
0%	2. The 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.
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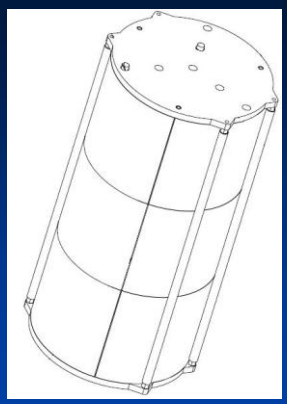
Answer: 4-they are the same.
Ref: AAPM TG111 report

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

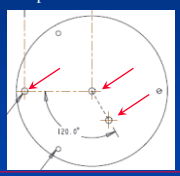
It is convenient to define L_{eq} as the value of L where D has reached 98% of its limiting value. (This is around 45 cm for 32 cm diameter acrylic.) For this scan of finite length, we no longer have translational symmetry. However, the central region of the scan has the same dose distribution as there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



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.



$\overline{D_{eq}}$

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

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

E_{tot}

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

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

(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).

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 $\overline{D_{eq}}$.

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Answer: 4-It is greater (since some of the energy is deposited outside the scanned region).

Ref: TG111 report

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

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

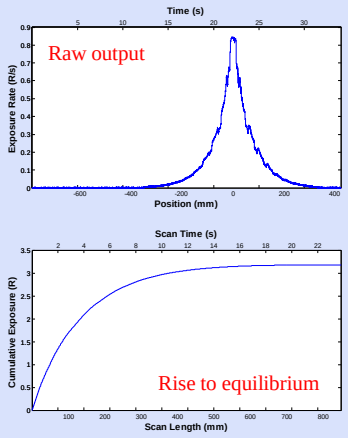
Define $h(L) = D(L)/D_{eq}$.

- 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.

• 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.
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Answer: 3-Scatter diminishes earlier at the periphery.

Reference: TG111 report

CTDI

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

CTDI

$CTDI_{100}$ is very similar in definition to $D(100)$ except that instead of the divisor being the table increment b, it is the *detected* beam width NI, i.e. it is a dose line integral divided by the beam width.

$CTDI_{vol}$ has much in common with $\overline{D_{eq}}$ except that it is for a finite length scan of 100 mm on a finite (150 mm) cylinder. In converting from $CTDI_{100}$ to $CTDI_{vol}$, a weighting formula (1/3 center, 2/3 periphery) is used to approximate the area average. The weighted average is divided by b/NI (pitch) and so $CTDI_{vol}$ has the same dependency on table increment as $\overline{D_{eq}}$ and is quite similar in definition to $\overline{D(100)}$, the central cross section dose average for a 100 mm scan.

Consider $CTDI_{100}$ measured at collimations (NI) of 40 mm and 1 mm

0% 1. At 40 mm, $CTDI_{100}$ is 40 times higher.

0% 2. At 40 mm, $CTDI_{100}$ is noticeably less than for 1 mm due to penumbra.

0% 3. If the values do not agree to within 5%, service should be called.

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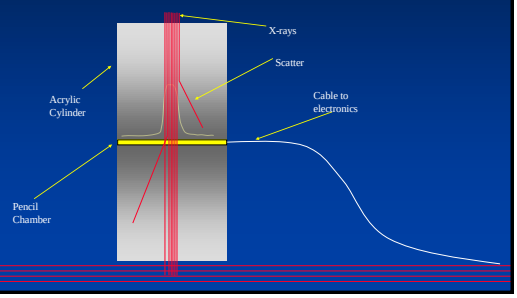
Answer: 2-CTDI smaller at 40 mm than at 1 mm, often by a factor of two.

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

Alternative: $CTDI_{100}$ 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.



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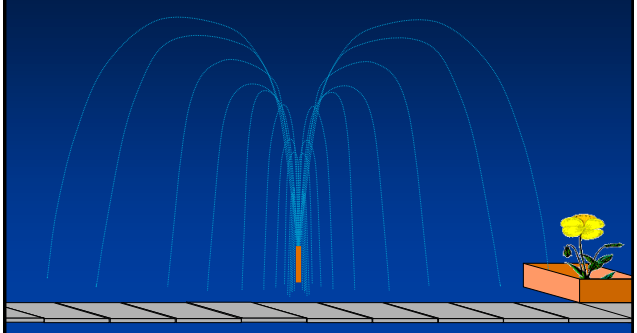

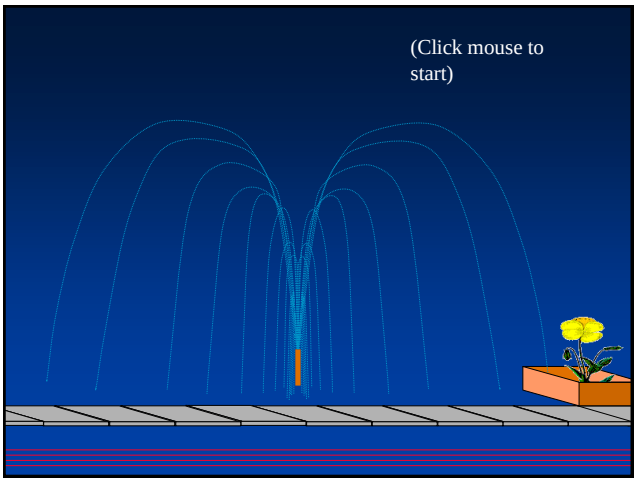
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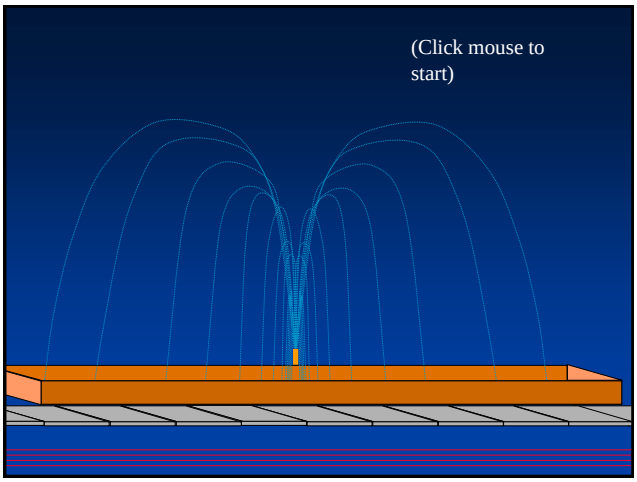
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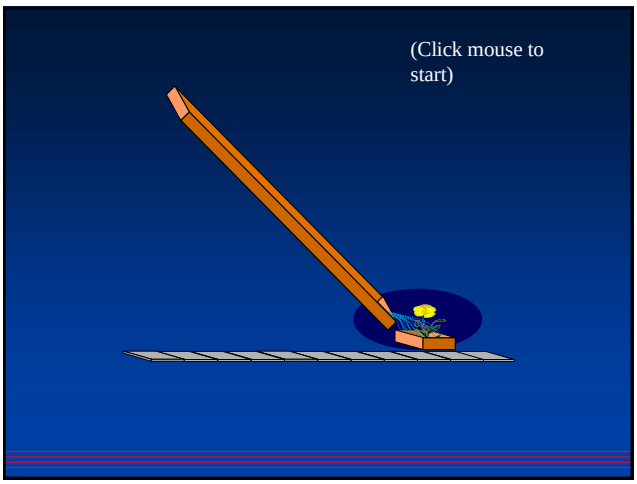
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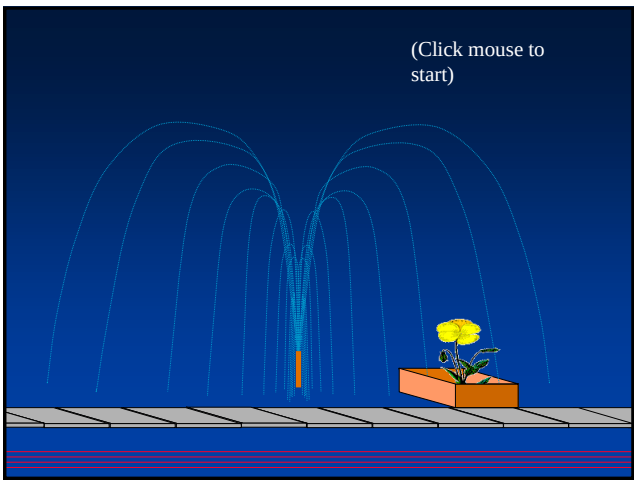
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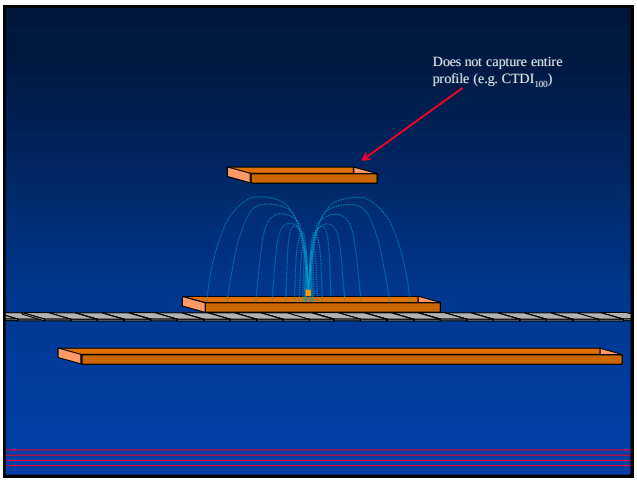
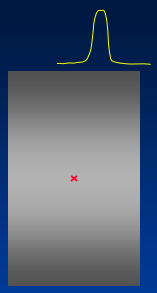


For $CTDI_{100}$, the phantom is 150 mm (~6") in length and the dose integral is measured using the 100 mm long pencil chamber and a single rotation in a fixed phantom. Of course, the measurement ***could*** be made ***directly*** by using a small chamber embedded in the phantom at point × and scanning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

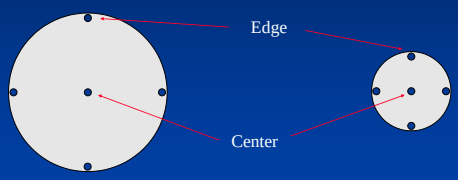
In this scan, the dose at point × does not build up as much as before for two reasons: 1) the abbreviated scan length and 2) lack of scatter from an extended cylinder. Extending the scan and cylinder lengths could add ~ 30% to the dose accumulating at the center.



$CTDI_{VOL}$, the Weighted Average

(intended to be average over central cross section)

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



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

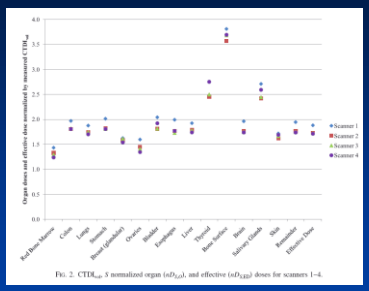


FIG. 2. $CTDI_{vol}$, S normalized organ ($nD_{S,O}$), and effective (nD_{EED}) doses for scanners 1–4.

[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).

$CTDI_{vol}$

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 the same kV is a good indicator of CT radiation output regardless of vendor, model, rotation time, pitch, tube current or collimation

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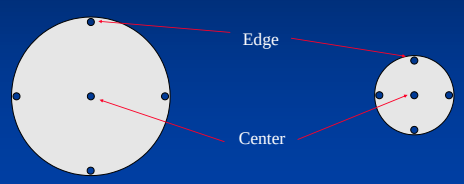
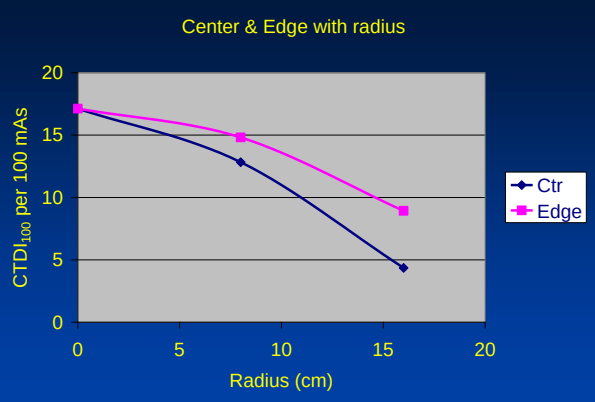
Answer: 4

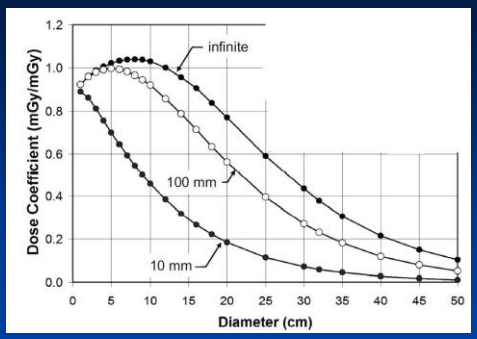
Ref: Turner et al., Med. Phys. 37(4), 1816-1825 (2010).

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



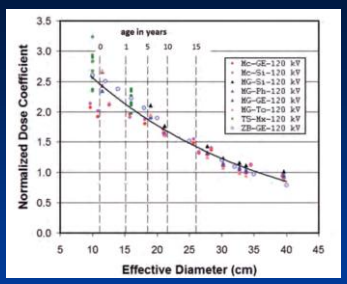


With more data, a peak



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.

10

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

Ref: AAPM TG204 report

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.

Thank you