Tutorial for CT Technologists: Dose, Image Quality and Best Practices

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next

What you will learn

Individual technique factors and their role in patient radiation dose and image quality

Principles of CT dosimetry, and the meaning of Dose Indices displayed by the scanner

Pathways to estimation of Effective Dose, and how to communicate risk

Image quality metrics

Major manufacturers' approaches to modulate tube current and reduce image noise

Performance standards and accrediting body requirements

Approaches to CT fluoroscopy

Patient shielding guidelines

Best practices in CT scanning

How to use this tutorial: The blue left and right arrows   at the bottom of each page navigate back and forth, one page at a time. The green 'home' button  at the bottom left of the page takes you directly to the contents page. From the contents page, you can jump directly to each individual section using the blue boxes next to each topic, with page numbers xx. Additionally, green back-and-froth boxes xx → ← xx with page numbers are provided at several locations, linking concepts as they appear in the text.

Contents

CT Technique Factors

Patient dose in CT depends of the following parameters:

- Tube kilo-voltage (kV)
- Tube current (mA)
- Tube rotation time (seconds)
- Detector configuration (radiation beam width)
- Pitch (also known as pitch factor, PF)
- Scan length

These are pre-programmed into the protocol

For mA-modulated protocols, mA is set by the scanner after the scout scan. Therefore, it is patient specific, based on patient girth, as well as noise reduction mechanism, if available and used

Depends on
how tall the patient is
prescribing physician (scan range prescribed)
scanner characteristics (amount of overranging),
technologist's skill and practice (amount of overscanning)

Historically, the focus of CT imaging was to produce the 'best possible' image, i.e., an image with the least amount of noise (graininess) and maximize spatial resolution (the ability to see fine details) and contrast resolution (the ability to distinguish areas with close CT numbers). This came at the cost of increased patient dose. With the rapid increase in the number CT scans over the past two decades, there has been a paradigm shift from 'the best possible' image to a 'diagnostically adequate' image. Recent advances in noise reduction technology have enabled us to maintain image quality with acceptable amounts of noise, while decreasing patient dose.

The next few slides consider the impact of each of the above parameters on dose.

kV

Kilo-voltage determines the number of photons produced by the x-ray tube, as well as their energy.

kV primarily controls image contrast, i.e., the difference between light and dark areas of the image. It also affects image noise.

If the kV is too low or too high, the image will have excessive contrast or lack contrast. Both extremes are undesirable.

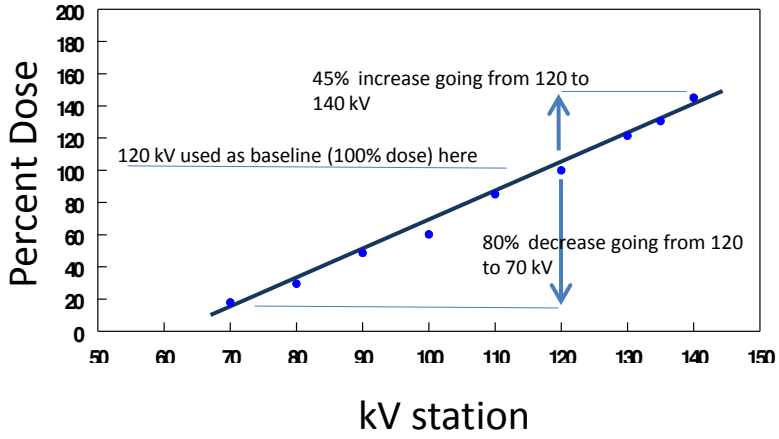
Only a few distinct kV stations are available: typically, 80, 100, 120, and 140. Few other scanners offer 70, 90, 110, 130 and 135 kV stations. The vast majority of protocols use 120 kV, or the closest available station.

As kV increases, patient dose increases linearly, but this is much steeper than a 1:1 increase.

Compared to 120 kV:

140 kV increases dose by about 45%
100 kV decreases dose by about 40%
80 kV decreases dose by about 70%
70 kV decreases dose by about 80%

However, the scanner must have the capability to compensate for the change in contrast, increased noise and possible artifacts caused by this these low kV stations. Thin adults and pediatric patients are good candidates for lower kV imaging.



For iodine contrast imaging, dropping the kV from 120 to:

100 kV increases iodine attenuation by 25%

80 kV increases iodine attenuation by 70%

Low kV imaging can therefore be beneficial for multi-phase scans where iodine visibility is important.

100 kV protocols for noninvasive coronary CTA significantly reduce dose, while not significantly affecting image quality in non-obese patients.

mA

mA primarily controls the number of photons, and therefore, image noise (or graininess).

Noise is measured as the Standard Deviation (SD or Std. Dev.) when an ROI (region of interest) is drawn.

The smoothest (least noisy) images require higher mA, but the noise benefits of higher mA taper off at a certain point.

Patient dose increases linearly with mA, in a 1:1 fashion. Doubling the mA doubles the dose. Halving the mA halves the dose.

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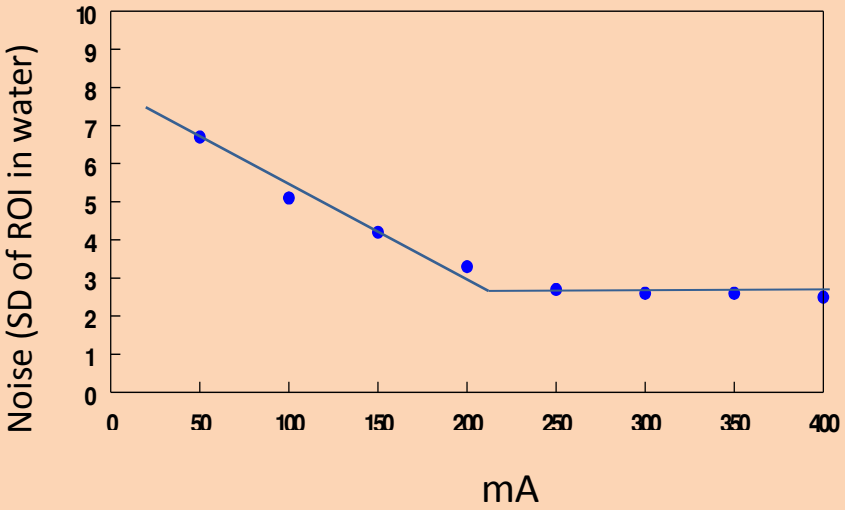
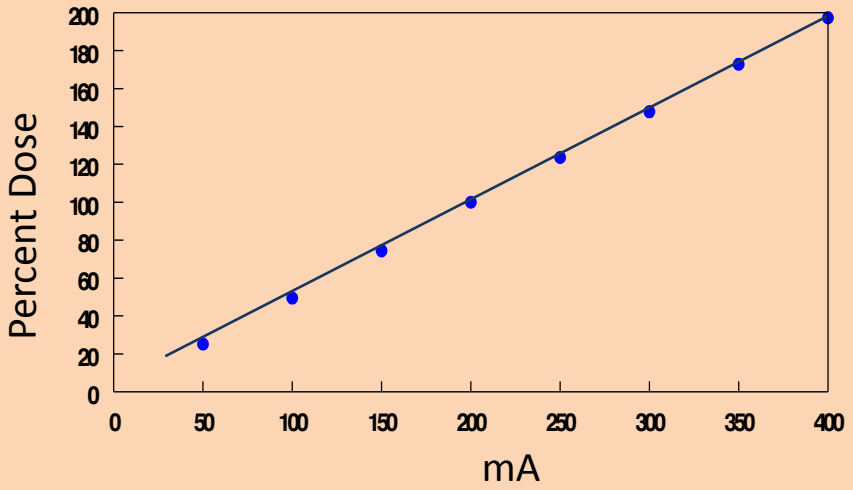
In the example below, noise decreases linearly with mA, up to 200 mA. Beyond that, the mA has no affect on noise. Unnecessary increase in mA therefore has no benefit. It only causes an unnecessary increase in patient dose, in this case, from 100% to 200%





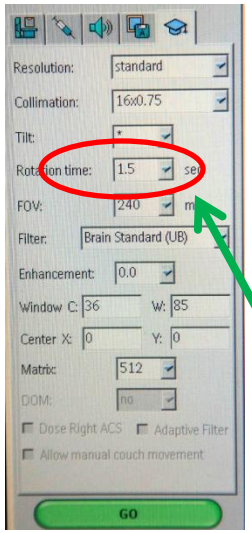
Time (s) and mAs

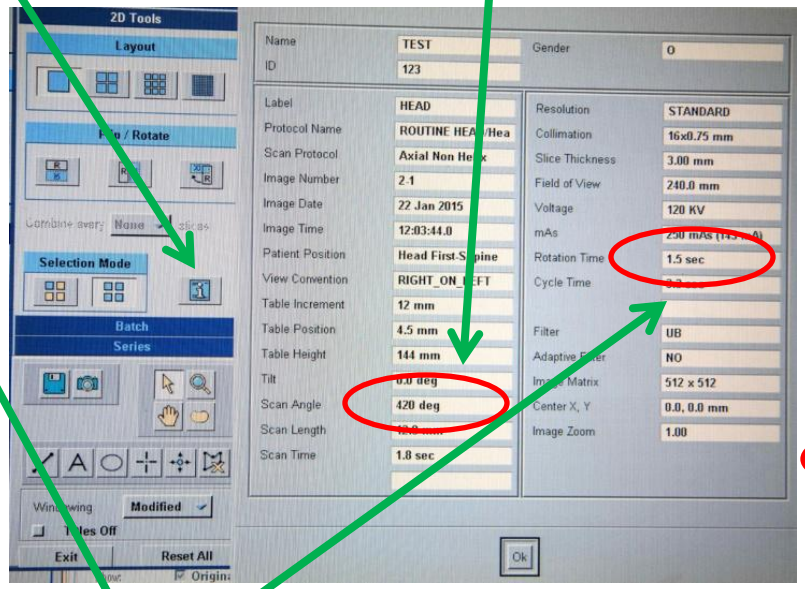
The effect of mA and tube rotation time (in seconds) are generally considered together, as mA multiplied by time (mAs). Increasing mAs has the same affect on dose and image noise as increasing mA; they work in tandem.

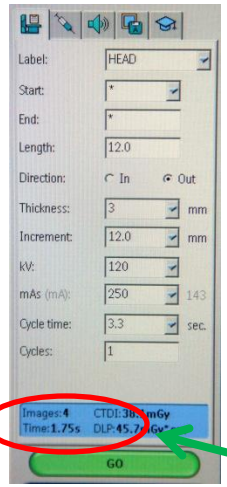
For a given mAs, increasing mA and decreasing time is preferable, because the longer the tube rotation time, the greater are the chances for patient motion artifacts.

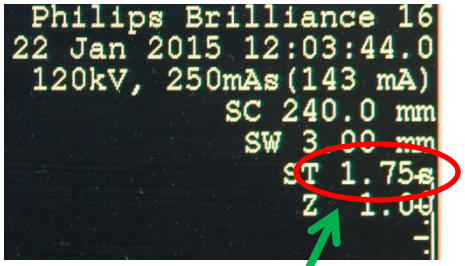
Example: a head scan is done at 200 mA, 2 sec (400 mAs)
Performing the same scan at 400 mA, 1 sec (400 mAs) is a much better technique, delivering the same dose.
The same protocol at 666 mA, 0.6 sec (still 400 mAs) is a vastly improved technique, if your scanner can handle the increased tube load

Axial sans on some Philips scanners are programmed for 420° rotation, as opposed to the traditional 360°. This can be checked using the 'I' (information) button under 'quick view'. The scan time set under the 'graduate cap' tab is NOT the correct rotation time. The correct tube rotation time can be found on the annotation on the image, or in the blue bar on the tech tab.









Wrong

Correct

Detector configuration: N×T

N is the number of detectors used in a particular protocol, and T is the thickness of the detectors. N multiplied by T gives radiation beam thickness (beam width), and is usually referred to as 'NxT' or simply 'NT', in millimeters. N and T are always quoted at the isocenter of the gantry, not at the detector surface.

NT is not the same as image (slice) thickness, although you may have slice thicknesses equaling NT. Examples of NT for various scanners are given on the next page. All these are axial and helical scan configurations. A few additional and unique NTs may be present exclusively for helical scans (not presented).

T Frequently, two or more detectors may be combined to form broader 'channels' , even though they are still called 'detectors'

GE: All individual detectors are 0.625 mm thick. Two are combined to form 1.5 mm, four are combined to form 2.5 mm, six are combined to form 3.75 mm, and eight are combined to form 5 mm channels. On the 64 slice VCT and Optima, none are combined

Philips: Individual detectors are 0.6 mm, 0.625 or 0.75 mm thick. Two of the 0.625 are combined to form 1.25 mm, and four are combined to form 2.5 mm channels. Two of the 0.75 are combined to form 1.5 mm, four are combined to form 3.0, and 6 are combined to form 4.5 mm channels

Toshiba: Individual detectors are 0.5 or 1.0 mm thick. They are combined to form 1, 2, 3, 4, 5, 6, 8 and 10 mm channels

Siemens: Individual detectors are 0.5 mm, 0.6 or 0.75 mm thick. Two of the 0.5 are combined 1.0 mm, five are combined to form 2.5 mm, 10 are combined to form 5 mm, and 16 are combined to form 8 mm channels. Two of the 0.6 mm are combined to form a 1.2 mm channel. Two of the 0.75 mm are combined to form a 1.5 mm channel.

N Either Individual detectors, or channels are formed into groups, constituting N (# of detectors)

GE: 2, 4, 8, 16, 32, or 64 detector rows or channels are used

Philips: 2, 4, 8, 12, 16, 32, 64 or 128 detector rows or channels are used

Toshiba: For axial scans, 2 or 4 detector rows or channels are used. For helical scans, selections are different. For example, Aquilion 16: 16 channels; Aquilion 32: 32 channels, Aquilion 64: 32 or 64 channels are used.

Siemens: 2, 4, 5, 10, 12, 16, 20, 32 or 64 detector rows or channels are used

Detector configurations for a range of scanners

GE

GE LightSpeed, 4 detectors

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
2	×	0.625	=	1.25	27.77
4	×	1.25	=	5	24.15
4	×	2.5	=	10	19.10
4	×	3.75	=	15	18.78
4	×	5	=	20	16.91

GE LightSpeed, 8 detectors

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
2	×	0.625	=	1.25	27.79
4	×	1.25	=	5	24.12
8	×	1.25	=	10	20.23
4	×	2.5	=	10	20.14
4	×	3.75	=	15	18.74
4	×	5	=	20	16.25
8	×	2.5	=	20	16.87

GE LightSpeed, 16 detectors

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
2	×	0.625	=	1.25	23.47
4	×	1.25	=	5	27.46
16	×	0.625	=	10	20.91
8	×	1.25	=	10	21.31
4	×	3.75	=	15	20.13
16	×	1.25	=	20	18.69
8	×	2.5	=	20	17.80

GE Optima CT660, 32 detectors (64 slice)

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
1	×	1.25	=	1.25	42.71
2	×	1.25	=	2.5	29.72
4	×	1.25	=	5	24.64
8	×	1.25	=	10	21.20
16	×	1.25	=	20	18.62
32	×	1.25	=	40	17.40

GE VCT, 64 detectors

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
2	×	0.625	=	1.25	37.40
4	×	0.625	=	2.5	26.62
8	×	0.625	=	5	23.45
16	×	0.625	=	10	19.00
32	×	0.625	=	20	16.94
64	×	0.625	=	40	15.83

Philips

Philips Brilliance Big Bore, 16 detectors

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
2	×	0.6	=	1.2	14.80
4	×	0.75	=	3	18.00
4	×	1.5	=	6	12.70
4	×	3	=	12	10.60
16	×	0.75	=	12	11.60
4	×	4.5	=	18	10.60
8	×	3	=	24	10.60
16	×	1.5	=	24	10.60

Philips Brilliance, 40 detectors

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
2	×	0.625	=	1.25	36.40
12	×	0.625	=	7.5	17.20
16	×	0.625	=	10	16.70
12	×	1.25	=	15	14.30
40	×	0.625	=	25	14.00
32	×	1.25	=	40	11.80
16	×	2.5	=	40	11.50

Philips Brilliance, 64 detectors

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
2	×	0.625	=	1.25	25.20
12	×	0.625	=	7.5	17.40
16	×	0.625	=	10	16.10
12	×	1.25	=	15	14.70
40	×	0.625	=	25	13.60
64	×	0.625	=	40	13.10
16	×	2.5	=	40	13.10

Philips iCT, 128 detectors (256 slice)

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
2	×	0.625	=	1.25	21.20
4	×	0.625	=	2.5	20.40
8	×	0.625	=	5	21.80
16	×	0.625	=	10	18.80
32	×	0.625	=	20	16.20
64	×	0.625	=	40	14.10
64	×	1.25	=	80	13.00
128	×	0.625	=	80	13.00

For all scanners, $CTDI_{vol}$ data is for 120 kV, 200 mAs, body phantom, for a single axial scan without table movement (and therefore, same as $CTDI_w$)

Siemens

Siemens Somatom Sensation, 4 detectors

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
2	×	0.5	=	1	32.00
2	×	1	=	2	18.00
4	×	1	=	4	18.00
2	×	2.5	=	5	15.20
4	×	2.5	=	10	15.20
2	×	8	=	16	13.40
4	×	5	=	20	13.80

Siemens Somatom Sensation, 16 detectors

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
2	×	1	=	2	16.80
12	×	0.75	=	9	16.80
2	×	5	=	10	12.60
12	×	1.5	=	18	14.40
16	×	1.5	=	24	14.00

Siemens Somatom Definition AS, 20 detectors

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
5	×	1	=	5	11.86
10	×	1	=	10	11.86
20	×	0.6	=	12	16.06
12	×	1.2	=	14.4	15.22
16	×	1.2	=	19.2	14.56

Siemens Somatom Definition AS, 32 detectors (64 slice)

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
5	×	1	=	5	11.86
10	×	1	=	10	11.86
12	×	1.2	=	14.4	15.22
16	×	1.2	=	19.2	14.56
32	×	0.6	=	19.2	15.20

Siemens Somatom Definition AS, 64 detectors (128 slice)

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
5	×	1	=	5	12.33
10	×	1	=	10	12.33
12	×	1.2	=	14.4	15.81
32	×	1.2	=	38.4	13.80
64	×	0.6	=	38.4	13.49

Toshiba

Toshiba Aquilion, 4 detectors

N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
4	×	0.5	=	2	53.11
4	×	1	=	4	36.95
4	×	2	=	8	28.02
4	×	3	=	12	26.21
4	×	5	=	20	24.12
4	×	5	=	20	23.57
2	×	10	=	20	29.19
4	×	8	=	32	27.98

Toshiba Aquilion, 16 detectors

	N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
	4	×	0.5	=	2	56.90
	4	×	1	=	4	38.20
	4	×	2	=	8	28.10
	4	×	3	=	12	24.00
	4	×	4	=	16	24.20
	4	×	6	=	24	23.20
	4	×	8	=	32	23.50
helical	16	×	0.5	=	8	28.10
helical	16	×	1	=	16	24.20

Toshiba Aquilion, 32 detectors

	N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
	1	×	1	=	1	91.20
	4	×	0.5	=	2	56.90
	4	×	1	=	4	38.20
	4	×	2	=	8	28.10
	4	×	3	=	12	24.00
	4	×	4	=	16	24.20
	4	×	6	=	24	23.20
	4	×	8	=	32	24.00
helical	32	×	0.5	=	16	24.20
helical	32	×	1	=	32	24.00

Toshiba Aquilion, 64 detectors

	N	×	T (mm)	=	NT (mm)	CTDIvol (mGy)
	4	×	0.5	=	2	55.60
	4	×	1	=	4	36.60
	4	×	2	=	8	26.50
	4	×	3	=	12	25.80
	4	×	4	=	16	23.50
	4	×	6	=	24	22.50
	4	×	8	=	32	21.80
helical	32	×	0.5	=	16	23.50
helical	64	×	0.5	=	32	21.80

$CTDI_{vol}$

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overbeaming

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NT The combination of N and T gives rise to a particular beam width. For example:

On a GE LightSpeed 4 detector CT, two (N) of 0.625 mm *detectors* (T) give rise to a 1.25 mm radiation beam (NT)
four (N) of 1.25 mm *channels* (T) give rise to a 5 mm beam width (NT) … and so on

In several cases there may be more than one NT combination resulting in the same beam width (e.g., 8x1.5 and 4x2.5 on the GE LightSpeed 8 detector CT; both result in a 10 mm beam)

Only a few, representative scanner NT combinations are shown on the previous page. Many more scanners, with additional detector thicknesses, channel thicknesses and NT combinations are in commonly in use.

The table also presents the dose index $CTDI_{vol}$ for each of the NT combinations, under identical conditions (more on this later). For now, notice that as a general rule, $CTDI_{vol}$ *decreases* as radiation beam width (NT) *increases*. 21 →

Image Configuration refers to the number of images (slices) and thickness of each, obtained from a given radiation beam thickness. In the vast majority of cases, these combinations are simple multiples. For example: A 40 mm radiation beam width may have the following image configuration options:

4 images of 10 mm each; 8 images of 5 mm each; 16 images of 2.5 mm each; or 32 images of 1.25 mm each

While $CTDI_{vol}$ depends on detector configuration (NT), it is *independent* of image configuration. For example, each of the image configurations for the 40mm beam shown above will have the *same* $CTDI_{vol}$. Think of this as baking bread. You may decrease or increase the number of calories (*dose*) by using less or more dough *(beam thickness).* But once the bread is baked and fully consumed, it does not matter how many pieces you slice it into *(image thickness):* the total number of calories will be the same.

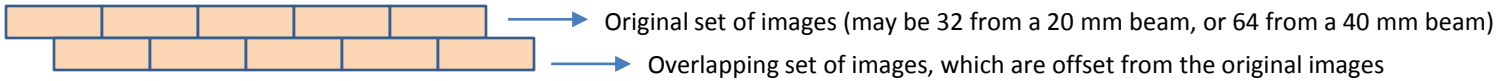
Flying focal spot technology: In many cases, the number of images generated exceeds the beam width. This is done by generating two sets of signals from each detector or channel, one from the radiation beam originating from the large focal spot on the anode, the other from the small focal spot. The beam 'flickers' between the two focal spots at a very rapid rate. Examples:

Siemens Somatom Sensation and Definition AS 64: Have 32 detector rows
Siemens Somatom Definition AS 128: Has 64 detector rows
Philips iCT 256: Has 128 detector rows
Toshiba Aquilion Prime 160: Has 80 detector rows

In some cases, a large volumetric image is 'parsed' into smaller 'slices' in post-processing. Here, a varying focal spot is not employed. Examples:

GE Optima CT660, 128 slice option: Has 64 detector rows, the additional 64 images can be generated by image processing software
GE Optima CT660, 64 slice option: Has 32 detector rows, the additional 32 images can be generated by image processing software
GE Discovery CTHD750 has similar options
Siemens Somatom Definition AS 64: Volumetric images generated for some detector configurations are 'sliced' as needed
Medtronic O-arm: Obtains a volumetric image using a single digital flat panel detector, and generates 192 slices of 1 mm each

'Overlapped Recon' is the technique employed by GE, and is available only for axial scans during retro recon.



Pitch (P)

Pitch refers to how the patient table is going to be moved during a scan. It is the relationship between table speed and beam width during one tube rotation.

Mathematically, pitch = mm of table movement during one tube rotation (mm/rot, also called *increment*, *interval* or 'I'), divided by beam thickness (NT in mm).

If the table moves 40 mm during one tube rotation (40 mm/rot), and the radiation beam width (NT) is also 40 mm, each scan will be next to the previous, with no overlaps or gaps. Here, the pitch is 1. You get this as: $\frac{40 \text{ mm/rot (I)}}{40 \text{ mm (NT)}} = \mathbf{1.0}$

If the table is moves slower (say 20 mm/rot) for the same beam width (NT=40 mm), the pitch will be $\frac{20 \text{ mm/rot (I)}}{40 \text{ mm (NT)}} = \mathbf{0.5}$
There will be overlaps between each scan. Compared to a pitch of 1, patient dose will be higher

If the table moves faster (say 60 mm/rot) for the same beam width (NT=40 mm), the pitch will be $\frac{60 \text{ mm/rot (I)}}{40 \text{ mm (NT)}} = \mathbf{1.5}$
There will be gaps between each scan. Compared to a pitch of 1, patient dose will be lower

Downsides of too high pitch settings: The ability to detect small lesions (e.g., pulmonary nodules) can be compromised. Low pitch values result in decreased noise, but at the cost of higher dose. 36 →

Toshiba scanners display two values of pitch: Pitch Factor (PF) is another name for pitch (P), as described above
Helical Pitch (HP) is pitch (or PF) times number of detectors (N)

Example: if N×T is 64 × 0.5mm = 32mm, and I is 20 mm/rot, then PF = 20/33 = 0.625, and HP = 0.625×64 = 40

$P = \frac{I}{NT}$	$I = P \times NT$	$NT = \frac{I}{P}$	$PF = \frac{I}{NT}$	$PF = \frac{HP}{N}$	$HP = PF \times N$

mAs_{eff} and mAs/slice

Effective mAs is mAs divided by pitch, and is a quantity used by Siemens and Toshiba scanners. mA_{seff} is displayed only for helical scans, not for axial (sequential) scans.

mAs/slice is identical to mAs_{eff}, but used by Philips scanners.

GE scanners have thus far stayed away from this descriptor.

Every once in a while, it becomes necessary to tease out individual parameters like mA and tube rotation time from the morass of displayed information, for instance to fill out ACR accreditation paperwork, or to make sense of the units to while making protocol adjustments. The following relationships apply:

$mAs_{eff} = \frac{mAs}{P}$	$mA = \frac{mAs_{eff} \times P}{s}$	$s = \frac{mAs_{eff} \times P}{mA}$	$mA = \frac{mAs}{s}$	$s = \frac{mAs}{mA}$

Reconstruction Algorithms (Kernels)

Every manufacturer has a different approach to the numbers of available algorithms (convolution filters) and their names, although their functionalities are similar. Smoother algorithms result in better low contrast resolution and less noise, while sharper algorithms favor better high contrast (spatial) resolution.

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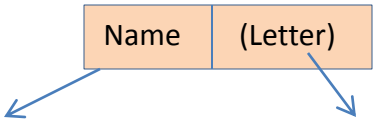
GE has the simplest approach: Typically, eight selections are available:*

Soft	for tissues with similar densities, but not useful for un-enhanced scans
Chest	for mediastinum and lung detail studies
Standard	for routine exams, e.g., chest, abdomens, and pelvis scans
Detail	for post myelograms, where hybrid tissue detail and bone edges are important
Bone	for High resolution exams (including Hi Res Chest) and sharp bone detail
Lung	for interstitial lung pathology
Bone+	for sub mm detailed head work
Edge	for small bone work in the head, as well as high resolution scans

* Source: GE Optima CT660 User manual

Philips uses an approach somewhat similar to GE
e.g., **Sharp(C)**

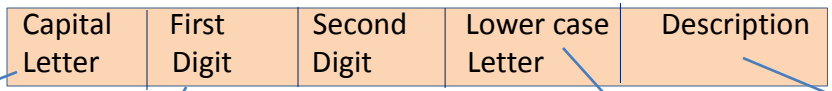


Standard
Sharp
Detail
Lung
Lung Enhanced

A = very smooth
B = smooth
C = sharper
D = sharp,
E = sharper, body only
L = sharpest, body only
YA = sharp, head only
YB = sharper
UA = head only, minimizes beam hardening artifacts
UB = head only, for small lesions
UC = head only, for small lesions

Reconstruction Algorithms (Kernels) cont'd

Siemens has perhaps the most detail-oriented kernel selection system. The following naming convention is used:
e.g., **B30f medium smooth**



Kernel Type
B = body
H = head
C = children (pediatric)
D = dual energy
U = ultra high resolution
T = topogram (scout)
S = special application
H and J: used with SAFIRE

Resolution
1 = smoothest
2
3
4
5
6
7
8
9 = sharpest

Version #
1
2
3
4
5
6
7
8
9

similar in sharpness, but different 'impression'.

Scan mode
f = fast
s = standard
h =-high resolution
u = ultrahigh resolution

Smooth
Medium smooth
Medium smooth +
Very smooth
Sharp
Very sharp
Ultra sharp, etc.

Not all 2 digit combinations are available

Toshiba's algorithms are use the suffix FC for main scans, and FL for scouts (scanograms). Increasing numbers within each group mean sharper algorithms

FC 1-5 = Body algorithms without beam hardening correction (BHC)
FC 7-9 = Body algorithms without BHC, providing increased contrast
FC 11-15 = Body algorithms with BHC
FC 17-19 = Body algorithms with BHC, providing increased contrast
FC 20-26 = Head algorithms with BHC, coarse grain size (FC 26 = increased contrast)
FC 30, 35, 81 = Bone algorithms; FC 30+ has edge enhancement
FC 41-44 = Head algorithms without BHC
FC 46-49 = Head algorithms without BHC, pediatric (FC 49 = increased contrast)

FC 50-53 = Standard lung algorithms
FC 55, 56 = Standard lung algorithms, decreased noise
FC 62-68 = Head algorithms with BHC
(FC 68 = increased contrast)
FC 83-86 = High Res lung algorithms
FC 81-81 = Temporal bone algorithms

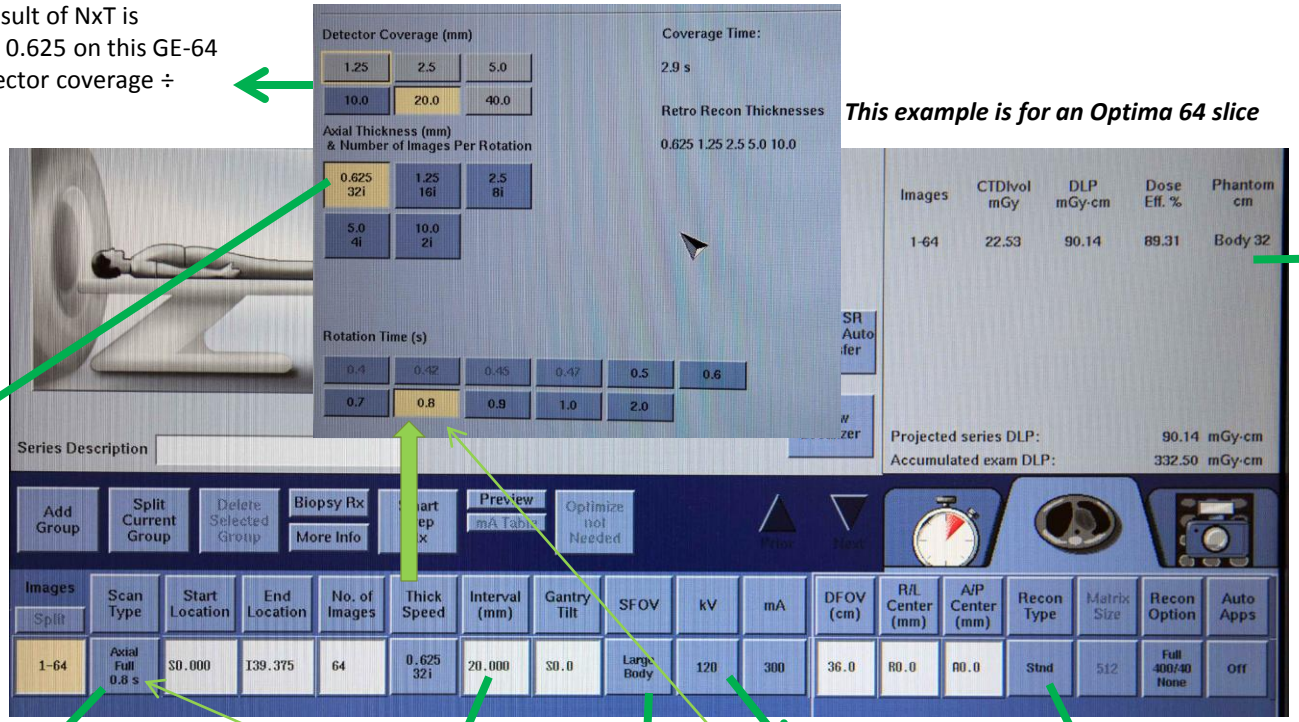
FL -4 = Scanogram algorithms

Screen Shots: GE (axial)

Detector configuration: Result of NxT is provided. Since T is always 0.625 on this GE-64 CT, you calculate N as Detector coverage ÷ 0.625. Thereby:

NxT ÷ T = N

1.25 ÷ 0.625 = 2
2.5 ÷ 0.625 = 4
5.0 ÷ 0.625 = 8
10 ÷ 0.625 = 16
20 ÷ 0.625 = 32 (selected)
40 ÷ 0.625 = 64



This example is for an Optima 64 slice

Phantom diameter for the selected SFOV is shown (32 cm or 16 cm), along with dose indices and z-axis efficiency for all detector configurations (not just those below 70%)

Image Configuration:
32 images of 0.625 mm thickness (selected).
Other options for this 20 mm beam are:
16 images of 1.25 mm;
8 images of 2.5 mm
4 images of 5 mm
2 images of 10 mm

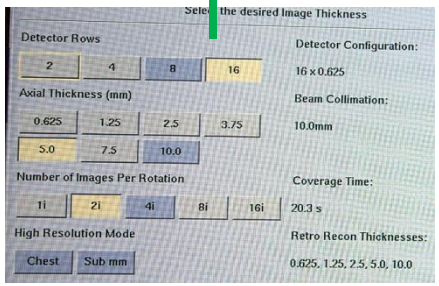
Axial scan technique factor layout: NxT breakups are not shown on GE-64 slice scanners; one has to infer them from beam thickness (20 mm in this case, quoted as 'detector coverage' in the 'Thick Speed' insert) and the fact that all detectors are 0.625 mm. Therefore, in this selection, NxT = 32 x 0.625 = 20 mm.

For axial scans, 'interval' refers to table increment between successive scans. In this case, a 20 mm travel ÷ 20 mm beam thickness = pitch of 1.0.

Only kV and mA are provided; GE does not display mAs (in this case, 300 mA x 0.8 sec = 240 mAs). The 0.8 sec tube rotation time is provided here and here.

Reconstruction algorithm

For 8, 16 and other slice GE scanners (aside from 64 slice), detector configuration displays are clearer. For the axial protocol selected, NxT = 16x0.625 =10 mm beam thickness; providing 2 images (2i) of 5 mm each.



This example is for a LightSpeed 16 slice

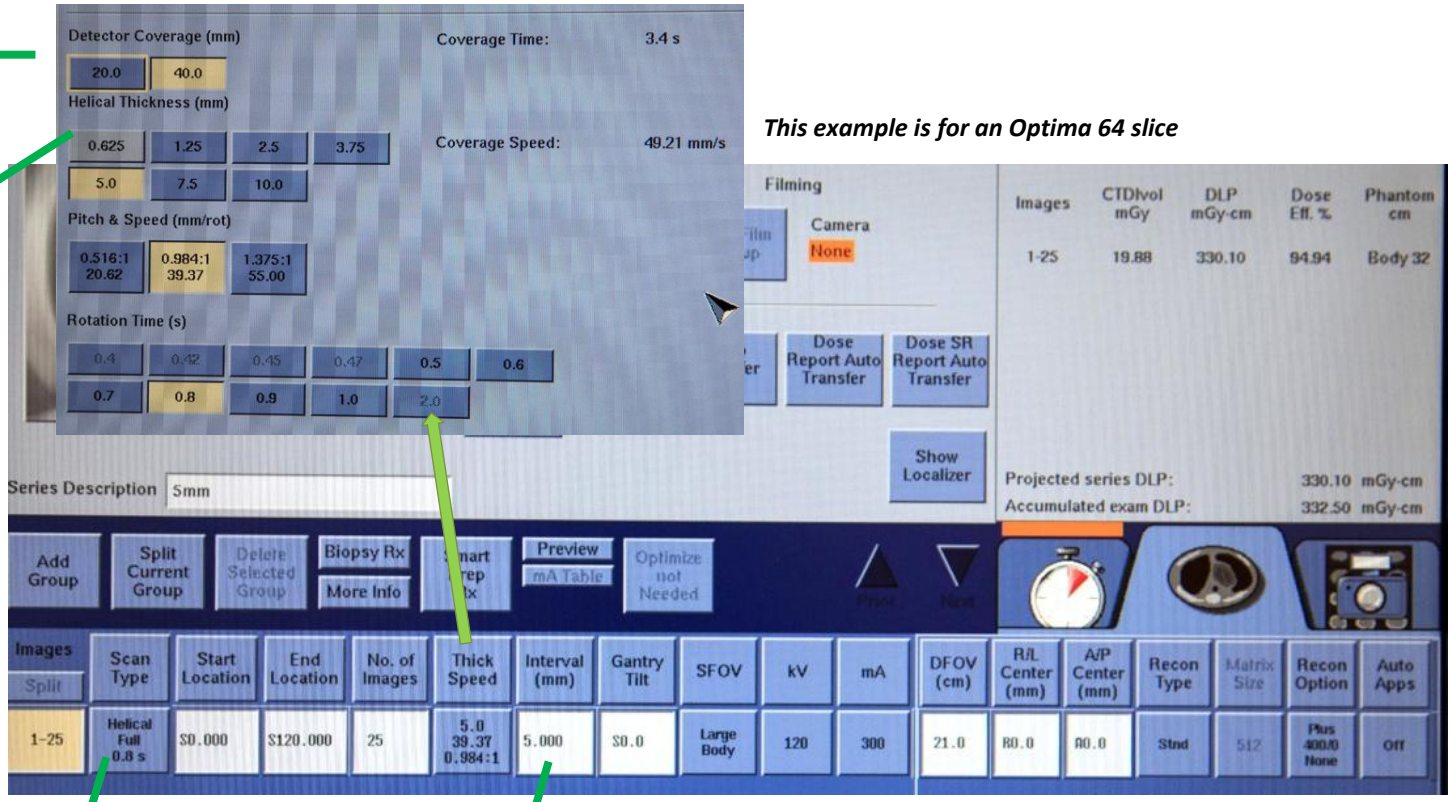
Scan field of view (SFOV): several options are provided: Large Body and Small Body options refer to a 32 cm phantom, with 50 cm and 32 cm maximum display fields of view (DFOV), respectively. Head, Ped Body and Ped Head refer to a 16 cm phantom, all providing 32 cm maximum DFOVs. There will be subtle differences in CTDIvol between the various 32 cm and 16 cm phantom selections.

Screen Shots: GE (helical)

Detector configurations available on this protocol:
32 x 0.625 = 20 mm
64 x 0.625 = 40 mm (selected)

Image configurations for the 40 mm beam selected:
64 images of 0.625 mm (not available, grayed out)
32 images of 1.25 mm
16 images of 2.5mm
8 images of 5 mm (selected)
etc.

This example is for an Optima 64 slice



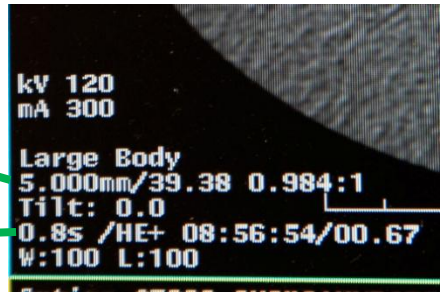
Helical scan technique factor layout: Image thickness, table increment and pitch are shown together; as well as in the inset under 'Thick Speed'. Pitch on GE scanners is always denoted as 'pitch:1'

For helical scans, 'interval' refers to distances between images (unlike table increment for axial scans)

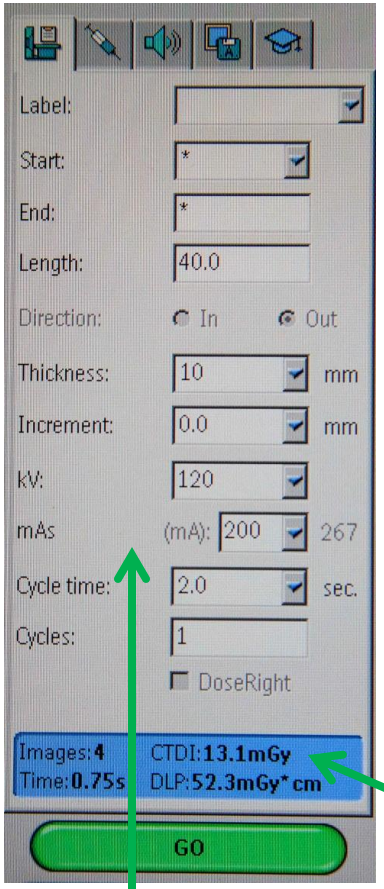
Helical scan image display annotations:

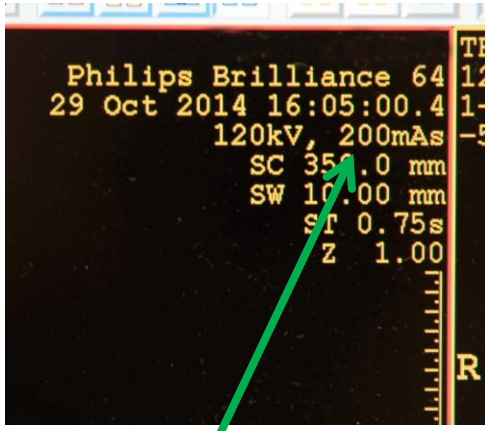
Image thickness, table increment and pitch

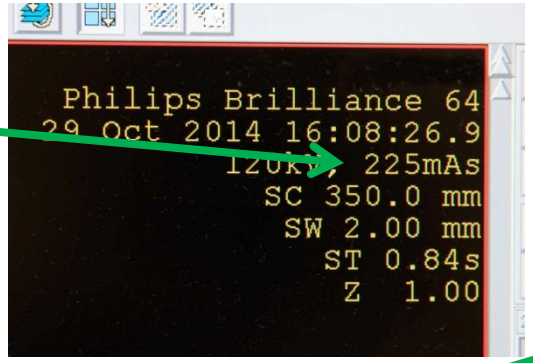
Tube rotation time

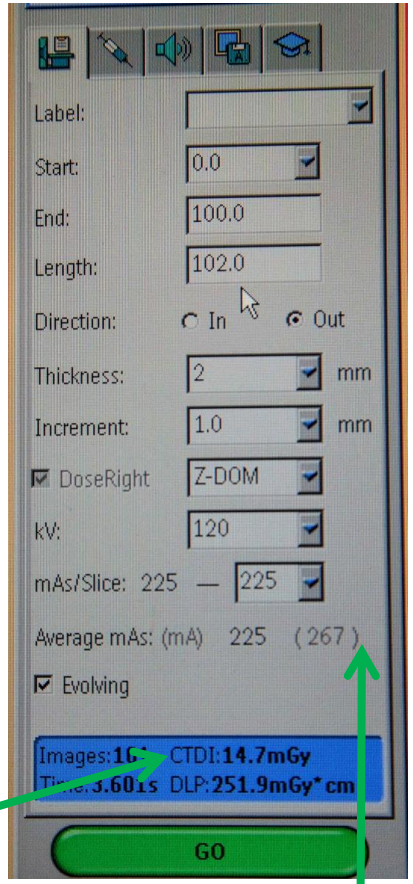


Screen Shots: Philips









For both axial (top) and helical scans (right), The annotation shows 'mAs'. Bear in mind that while the annotation is correct for axials, the mAs annotated for helical scans is actually mAs/slice (or mAs eff).

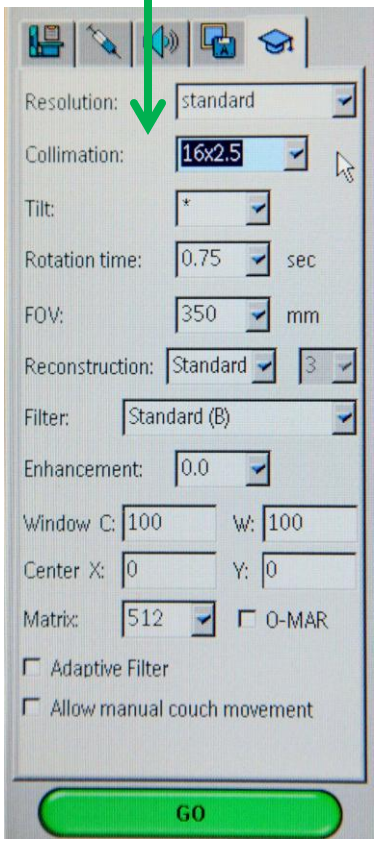
As with Toshiba, Philips scanners do not specify what phantom (32 or 16 cm) the dose index refers to. If a pediatric torso scan is performed under a 'body' protocol, you need to double the displayed CTDIvol to get the correct value.

For axial scans, the protocol tab displays mAs. In more recent Philips software revisions (as shown here), mA is also displayed in gray. Therefore, tube rotation time for this protocol is mAs / mA = 200 / 267 = 0.75 sec

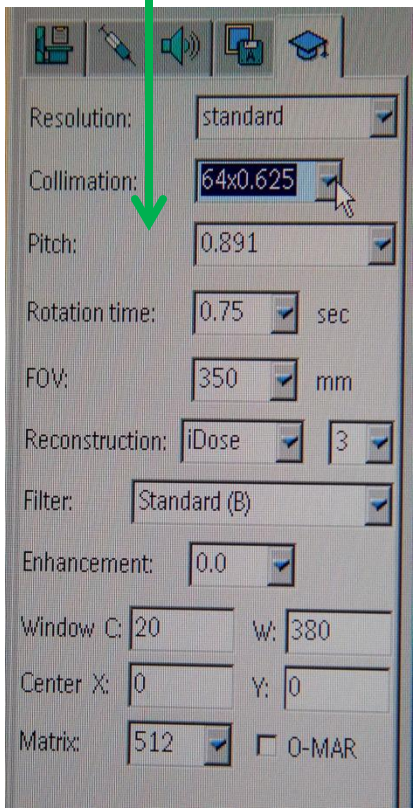
For helical scans, the protocol tab displays mAs/slice (which is the same as mAs effective, i.e., mAs / pitch) . In more recent Philips software revisions (as shown here), average mAs and mA are also displayed in gray. Average mAs is the estimated mAs to be applied across the scan; it is the same as mAs/slice in the above image because no scout image has been obtained. Tube rotation time for this protocol is mAs / mA = 225 / 267 = 0.84 sec. What other information can you glean from this screen? Since mAs_{eff} = mAs / pitch, pitch = mAs / mAs_{eff} = 225 / 225 = 1.0

Screen Shots: Philips

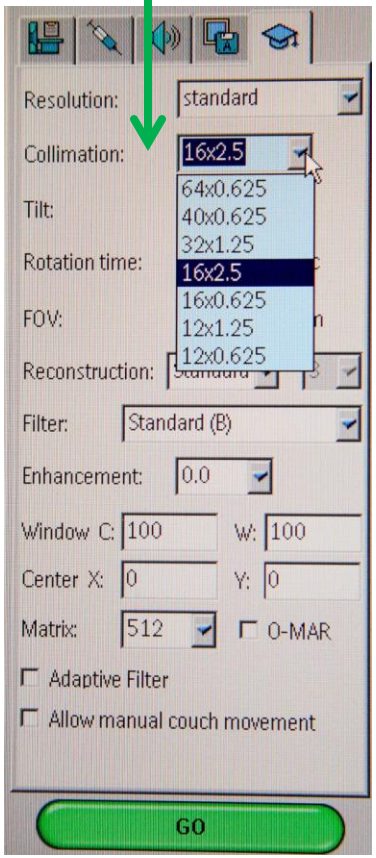
NxT combinations are shown in the 'collimation' drop-down menu under the 'graduate cap' tab. This is an axial protocol.



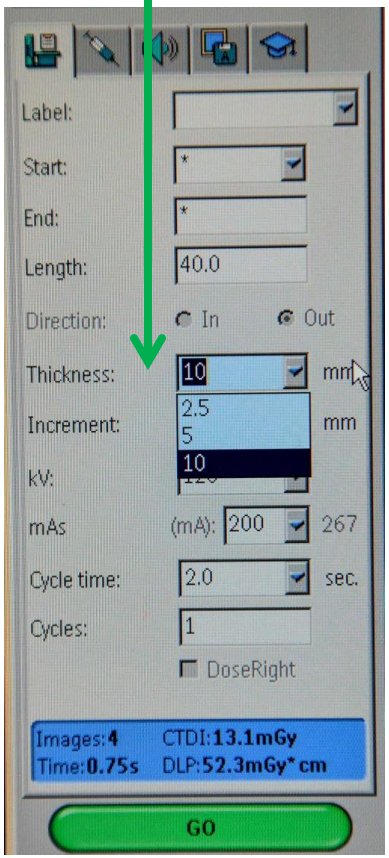
Pitch is displayed below the 'collimation' tab for helical protocols. For axial protocols, this space is occupied by 'tilt' (left image); while table incrementation is shown below the 'thickness' tab (rightmost image)



In this particular axial protocol, available beam thicknesses, from top to bottom (as you multiply the NxT values), are 40, 25, 40, 40, 10, 15 and 7.5 mm, respectively.



For the 16 x 2.5 = 40 mm NxT configuration selected (leftmost image), image thickness options from top to bottom are 16 images of 2.5 mm, 8 images of 5 mm, and 4 images of 10 mm (selected). The # of images is shown in the dark blue section at the bottom

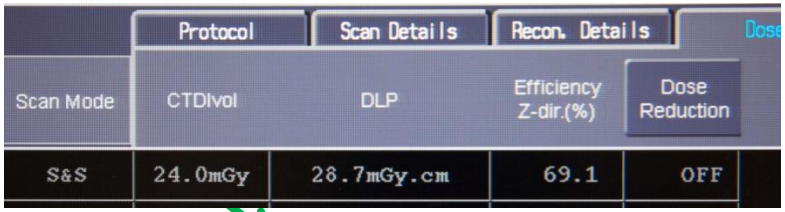


Screen Shots: Toshiba

S&S (Scan & Scan) and S&V (Scan & View) are Toshiba's axial (sequential) scan modes.

200 mA, 1.0 sec, 3.0x4 (TxN) = 12 mm, and 12 mm couch movement (table increment). Therefore, pitch = 12/12 = 1.0; mAs eff = 200/1 = 200.

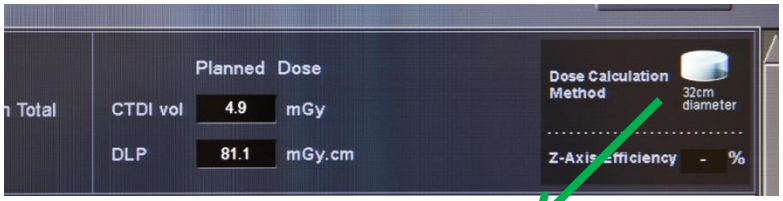
The 'Thickness' tab reveals NxT options available in axial mode on this 16-slice scanner. For Toshiba, the first entry is T, and the second is N. Resulting beam widths are 2, 4, 8, 12, 16, 24 and 32 mm, respectively.



On this Aquilion 16, dose indices do not specify what phantom they refer to. It is for the user to interpret that the values refer to a 32 cm phantom if SFOVs of M, L or LL are selected, and to a 16 cm phantom if SFOVs of SS or S are selected. Therefore, if a pediatric torso scan is done using M, L or LL SFOV, you need to double the displayed CTDIvol to get the correct value.

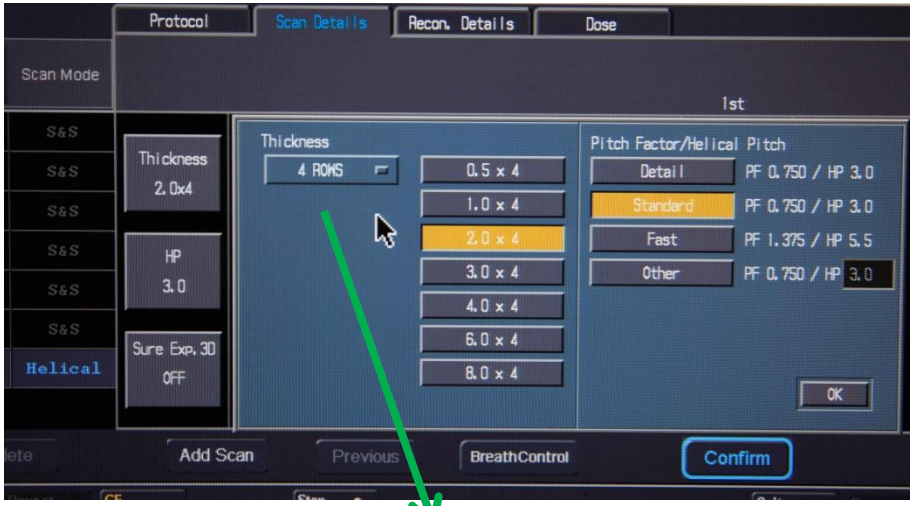
For TxN = 3.0x4 = 12 mm beam thickness, the image thicknesses available are 3, 6 and 12 mm. Therefore, image configurations possible are 4 images of 3 mm; 2 images of 6 mm, and one image of 12 mm. All 3 options result in 12 mm 'worth' of images.

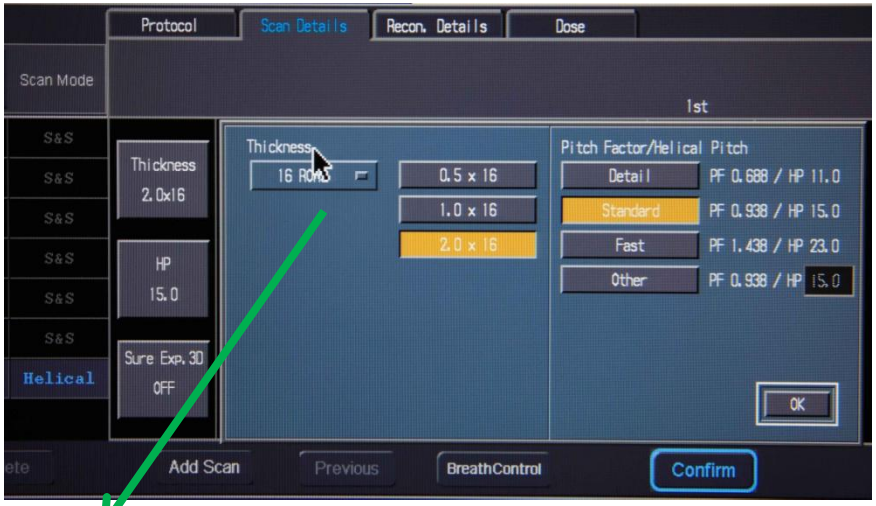
Display field of view (DFOV) is specified in numbers (mm), and the corresponding scan field of view (SFOV) in letters. SFOV options available are SS, S, M, L and LL.



Newer models, such as this Aquilion Prime, display phantom information with CTDI

Screen Shots: Toshiba





In the Helical scan mode on an Aquilion-16, NxT options are presented as 4 ROW sets or 16 ROW sets. Default Pitch (PF in Toshiba's parlance) and HP combinations have different names: Detail, Standard and Fast. The 'Other' option lets the user change HP. On the left image, TxN selected = 2.0 x 4 = 8 mm. Since the selected P = 0.750, table increment should be 8 x 0.750 = 6 mm per rotation (since I = P x NT). HP = PF x N = 0.75 x 4 = 3.0, as displayed.

On the right image, TxN = 2.0 x 16 = 32 mm; P = 0.938, therefore I will be 0.938 x 32 = 30.016 mm per rotation. HP = 0.938 x 16 = 15.008 (rounded down to 15, as displayed).

<u>Can you decipher this screen?</u>

200 mA with 1.0 sec selected. Therefore, mAs = 200 x 1 = 200

TxN selected = 2.0 x 16 = 32 mm

HP selected = 15.0. Therefore, Pitch = HP / N = 15 / 16 = 0.9375, rounded us as 0.938

I = P x NT = 0.938 x 32 = 30.016 mm per rotation

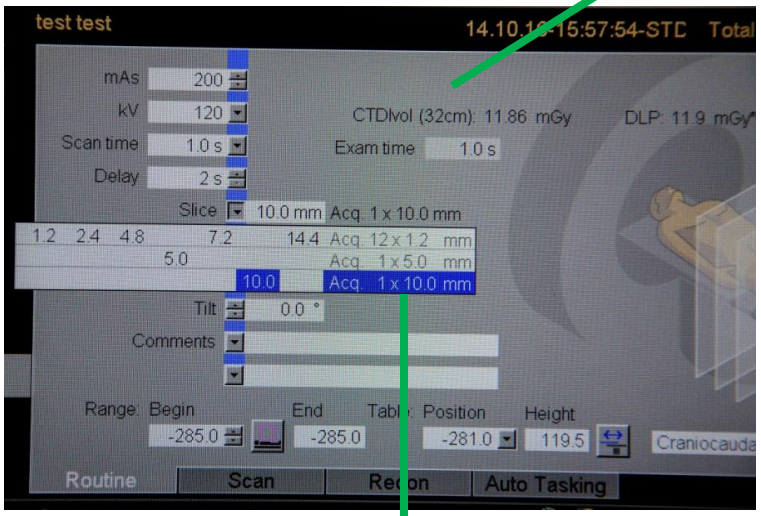
mAs effective = mAs / pitch = 200 / 0.938 = 213.219, rounded to 214 (displayed)

Since Small SFOV is selected, the displayed dose index (CTDIvol) will be for a 16 cm phantom, appropriate for a adult head, pediatric head, and pediatric abdomen.

The images will be 10 mm thick, with a recon interval of 10 mm, therefore contiguous (no overlaps or gaps in images)

Screen Shots: Siemens

Dose index display specifies which phantom (32 or 16 cm) it refers to.



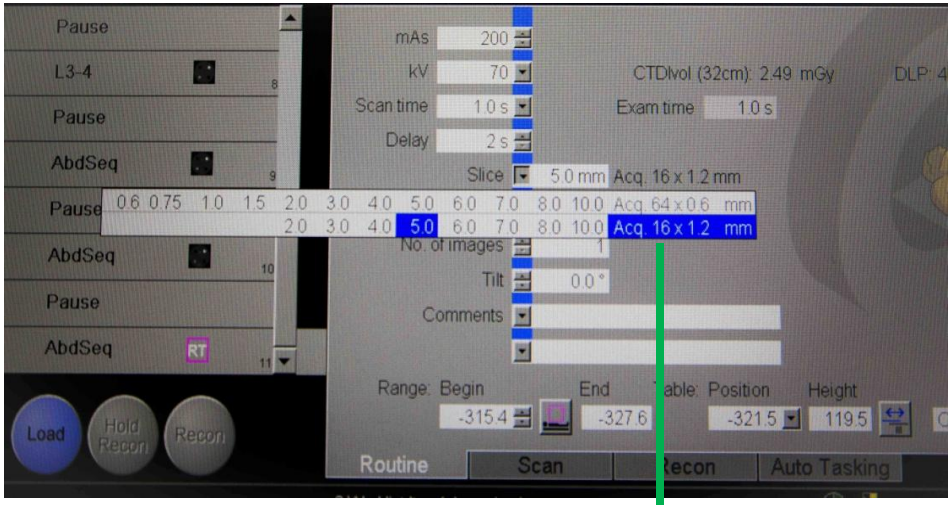


Image thicknesses as well as NxT combinations can be pulled up on the 'slice' tab. Image configurations possible can be inferred by dividing NxT by image thicknesses on the left side. The 3 'acq' combinations available on this protocol refer to:

(a) 12 x 1.2 = 14.4 mm beam thickness. This beam can generate 12 images of 1.2 mm, 6 images of 2.4 mm, 3 images of 4.8 mm, 2 images of 7.2 mm, and 1 image of 14.4 mm. To derive # of images, divide NxT by image thickness, i.e., 14.4 ÷ 1.2, 2.4, 4.8, 7.2 or 14.4 give 12, 6, 3, 2 and 1 image, respectively

(b) 1 x 5.0 = 5 mm beam thickness, providing 1 image of 5 mm

(c) 1 x 10.0 = 10 mm beam thickness, providing 1 image of 10 mm

You will notice that all possible image configurations 'add up' to the same beam thickness for a given NxT.

This helical protocol, on a Definition AS-64 slice canner, has a choice of two NxT combinations: 32 x 0.6 = 19.2 mm beam thickness, and 16 x 1.2 mm = 19.2 mm. Note: This scanner is not a true 64 slice, but a 32 slice with a flying focal spot. Therefore, any mention of '64' on the protocol page is actually a '32'. Thus, both NxT combinations result in the same beam thickness. Image thickness combinations for these two configurations are more complicated than on the example on the left; the numbers do not 'add up'. For instance, working your way from left to right on the top row, you get 32 images of 0.6 mm (19.2 ÷ 0.6 = 32); 25.6 images of 0.75 mm; 19.2 images of 1 mm; 12.8 images of 1.5 mm, and so on. While the concept of fractions of # of images does not make sense, this is an example where a large volumetric image produced from multiple helical tube rotations is 'parsed' into thinner slices; some images at the ends of the volume are not processed.

Screen Shots: Siemens

<u>Can you decipher this screen?</u>

mAs eff displayed = 200, tube rotation time = 1 sec

NxT = 32 x 0.6 = 19.2 mm (the 64 in the display is actually 32, with a flying spot)

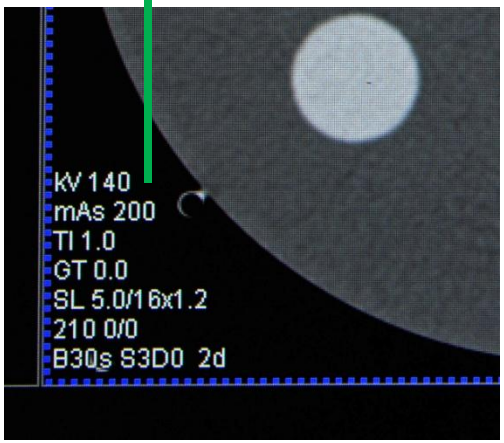
Pitch = 0.6. Therefore, I = P x NT = 0.6 x 19.2 = 11.52 mm per rotation

mA = (mAs eff x P) / sec = (200 x 0.6) / 1 = 120

mAs = mA x sec = 120 x 1 = 120

For helical scans, mAs-eff is displayed.

For axial scans, mAs is displayed.



CT Dosimetry

This section will consolidate the technique factor information presented so far, and look at how they fit into dosimetry.

CT dose is measured using acrylic cylinders of two different diameters: a 32 cm 'body' phantom, and a 16 cm diameter 'head' phantom (which is also used for pediatric body dose measurements). Exposure measurements are made at the center and the periphery (entrance) locations of the phantoms, which are positioned at isocenter.

It is well recognized that this system of measurements is far from perfect, because the phantoms do not resemble human shapes and sizes. The most significant drawback is that the method tends to underestimate dose to very thin patients, and overestimate dose to obese patients.

Dosimetry is performed only on single, axial scans without any table movement. The effect of multiple axial scans (i.e., sequential scans) and helical scans is incorporated mathematically.

Exposure measurements (typically in milliroentgen, mR) are normalized to the 100 mm length of the measurement chamber, and converted to $CTDI_{100}$ (mGy). This quantity is converted to weighted-CTDI ($CTDI_{w,}$ mGy) by giving different weights to the entrance and center exposure measurements. $CTDI_w$ divided by pitch is known as volume-CTDI ($CTDI_{vol,}$ mGy). $CTDI_{vol}$ multiplied by scan length provides Dose Length Product (DLP, mGycm). A detailed treatment of the above steps in dosimetry is beyond the scope of this tutorial.

For single axial scans where the table is not moved, $CTDI_w = CTDI_{vol}$.

CTDI refers to CT dose index. It is **not** patient dose. It is a measurement of radiation output of the scanner, and therefore an *indicator* of patient dose, much like the Dow Jones Index is an indicator of the stock market, or ERA is an indicator of baseball pitcher's performance. CTDI is quoted in the units of milligray (mGy)

CT Dosimetry cont'd

$CTDI_{vol}$ (mGy) is one of the two dose indices displayed by all recent CT scanners (the other being DLP, see later). It incorporates the effect of all the technique parameters programmed into the scan protocol (or determined by the scanner during mA modulation): kV, mA, tube rotation time, detector configuration (NxT) and pitch. Therefore, all the above parameters affect $CTDI_{vol.}$

$CTDI_{vol}$ is a useful quantity to compare different protocols, study the effects of changing a particular technique factor (such as kV, mA, s, N×T or P), and compare different makes and models of CT scanners.

$CTDI_{vol}$ (=$CTDI_w$) presented on page ← 6 are for identical techniques (120 kV, 200 mA, 1 sec, body phantom, for most of the axial NxT combinations available) on these scanners.

← 7

As a general rule, $CTDI_{vol}$ values decrease as beam width increases (with a few exceptions here and there). This means, larger radiation beam thicknesses are more efficient than thinner beams. The thinnest beams are used only in a few instances, such as for high-resolution axial chest scans, where large gaps (intervals) are specified between successive tube rotations.

All the factors going into $CTDI_{vol}$ are known before the scan. $CTDI_{vol}$ displayed on the protocol page even before the scout scan, and then corrected after the scout scan determines the mA (and sometimes, kV) to be used, for mA modulated (or mA and kV modulated) scans.

For this reason, ACR reference levels and pass/fail levels, as well as NEMA Notification Values (NVs) and Alert Values (AVs) are quoted for $CTDI_{vol.}$

$CTDI_{vol}$ for the head phantom is approximately *double* that of the body phantom (for identical parameters), because the phantom is half the size of the body phantom (16 cm vs. 32 cm diameter). This makes sense conceptually for head scans, because the head is more radio-resistant than the body (two layers of bone, protecting neurons that are not as susceptible to radiation damage than soft tissues). Therefore the head can 'tolerate' higher doses than the body, as reflected in their respective $CTDI_{vol.}$

CT Dosimetry cont'd

This also makes sense conceptually for pediatric body scans (which are evaluated using the 16 cm head phantom). The higher dose, relative to the body phantom, reflects the fact that radiation risks are much higher in children than adults. Therefore, technique factors need to be toned down for pediatric patients. Suppose both adult and pedi abdomen scans were evaluated using the 32 cm phantom: the $CTDI_{vol}$ would be the same, but the radiation risk would be far higher for pedi patient than the adult (but this would not be reflected in the numbers). If the pedi scan parameters were cut in half, the $CDTI_{vol}$ would be halved, and the radiation risk would begin to be at par with the adult scan. Since the 16 cm phantom is used for pedi bodies, the new $CTDI_{vol}$ would be almost identical to the 32 cm phantom, thereby reflecting the fact that the risks are at par with the reduced technique. To further reduce the risks for the pedi scan with respect to adults (which is one of the goals of dose reduction efforts), technique parameters need to be reduced even more: the younger the age group, the greater is the reduction required.

The above rationale is presented again in table below:

Technique factors (theoretical)	**$CTDI_{vol}$ if same phantom is used**		**$CTDI_{vol}$ if different phantoms are used**	
	Adult (32 cm)	**Pediatric (32 cm)**	**Adult (32 cm)**	**Pediatric (16 cm)**
120 kV, 200 mAs, 40 mm beam, pitch = 1	16 mGy	16 mGy	16 mGy	32 mGy
120 kV, 100 mAs, 40 mm beam, pitch = 1	8 mGy	8 mGy	8 mGy	16 mGy
120 kV, 50 mAs, 40 mm beam, pitch = 1	4 mGy	4 mGy	4 mGy	8 mGy
	Increased risk to the pedi patient is *not* reflected in the numbers		Increased risk to the pedi patient is reflected in the numbers	

If you take the time and effort to comprehend the above rationale, the concepts of Image Gently will become intuitive, and easy to implement

CT Dosimetry cont'd

DLP (**Dose Length Product**, in mGycm) is the second dose index displayed by CT scanners. It is merely $CTDI_{vol}$ multiplied by scan length in cm. Example: for a $CTDI_{vol}$ of 25 mGy and scan length of 30 cm, DLP = 25×30 = 750 mGycm

This parameter is patient specific. For the same protocol with identical $CTDI_{vol}$, a taller patient will result in higher DLP because his/her scan length will be longer. However, these differences will be readily apparent only if patient heights are substantially different (e.g., a 4.5 foot vs. a 6 foot patient). For the vast majority of patients, scan lengths will be the same ballpark for identical regions. Scan lengths differ more often because of overranging, and overscanning (individual physician's preferences in prescribing scan ranges, and with differences in how a technologist sets the scan boundaries). To a smaller extent, scan lengths may also differ for different scanner makes and models, because of non-identical technology.

DLP is an important step in determining Effective Dose (more on this later). Different combinations of $CTDI_{vol}$ and scan lengths of the same organs producing the same DLP can have identical Effective Doses. Conversely, scans for patients of approximately the same height, with identical $CTDI_{vol,}$ *passing* ACR trigger levels, but having considerably different scan lengths for the same organs, will result in very different Effective Doses (and therefore, risk). **Example:**

Theoretical abdomen / pelvis scans on three patients:

Patient Girth	Height	CTDIvol (mGy)	Sc. Lgth. (cm)	DLP (mGycm)	Eff. Dose (mSv)
Thin	Med.	16	30	480	7.2
Avg.	Med.	22	30	660	9.9
Large	Med.	28	30	840	12.6
Expected outcome. Larger patients with same height get more Effective Dose					

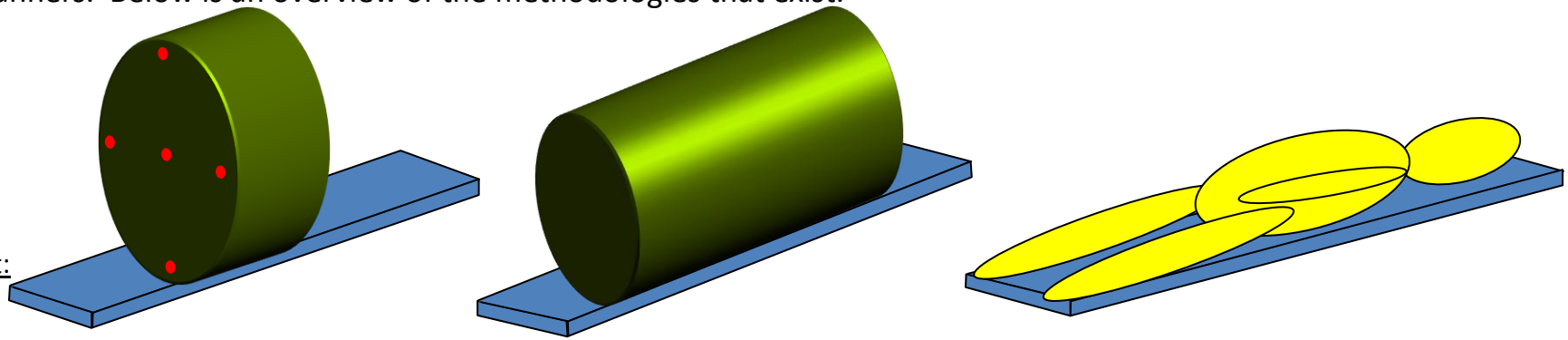
Patient Girth	Height	CTDIvol (mGy)	Sc. Lgth. (cm)	DLP (mGycm)	Eff. Dose (mSv)
Thin	Tall	16	41.25	660	9.9
Avg.	Avg.	22	30	660	9.9
Large	Short	28	23.57	660	9.9
Possible outcome. Patients with different body weights & heights can get equal Effective Dose					

Patient Girth	Height	CTDIvol (mGy)	Sc. Lgth. (cm)	DLP (mGycm)	Eff. Dose (mSv)
Avg.	Med.	22	25	550	8.25
Avg.	Med.	22	30	660	9.9
Avg.	Med.	22	35	770	11.55
Undesirable outcome. Why were scan lengths different? Merits investigation					

If you obtain the scan length from the images produced (or from the scout scan), and multiply by the CTDIvol, you *ought* to be able to calculate the DLP. However, *this may not be the same* as the DLP displayed by the scanner. Why? Actual scan lengths are generally longer than the span of images displayed, because of overranging by the scanner (an engineering aspect). Therefore, it is important to record both the displayed $CTDI_{vol}$ and DLP rather than trying to calculate the DLP from the length of images produced

CT Dosimetry cont'd

Approaches to patient dose. Further steps in dosimetry are not currently incorporated into CT scanners. Below is an overview of the methodologies that exist.



Input:
kV
mA
time
N×T
Pitch

Phantom measurement → $CTDI_{vol}$ (mGy) → Input: cm → DLP (mGycm) → Input: *K* factor → ***K*-factor method (current standard method)** → Estimate **Effective Dose (mSv)**

$CTDI_{vol}$ (mGy) → **SSDE method (new, but incomplete)** → ROAD BLOCK → Estimate **Effective Dose (mSv)**

Input: patient diameter
Size Specific Dose Estimate (mGy)

Input:
Patient metrics
Computational phantom
Computer code

Virtual Phantom Method (primarily for research) → Individual organ doses → Input: Tissue Weighting Factors → Estimate **Effective Dose (mSv)**

Estimate **Effective Dose (mSv)** → Evaluate and Communicate Risk

CT Dosimetry cont'd

Effective Dose. The objective of any dosimetry scheme for a modality involving x-rays is Effective dose, in units of millisievert (**mSv**), traditionally designated millrem (**mrem**).

Effective dose incorporates the long term adverse effects of radiation (primarily cancer), and converts doses to different organs (or groups of organs, in the case of CT scans) into 'whole body' doses. It provides a convenient, uniform currency for comparing doses across modalities and age groups.

Effective dose therefore serves as a surrogate for **risk**. Since risk is difficult to quantify, the numerical value of mSv or mrem is a good alternative.

Once Effective Dose is estimated, it has to be communicated in a meaningful manner. Perhaps the best method is to compare it to natural background radiation, which everyone on the planet is exposed to. For instance, most of us in the continental USA will be exposed to about 24 abdomen CT scans - worth of natural background radiation during our lifetimes.

The concept of Effective Dose, in its current form, was born in 1990. Its original intent was to quantify occupational risk in radiation workers, not doses from medical procedures. It is strictly a concept applicable to large groups of people, and not to individuals. Nevertheless, in the absence of any other metric for quantifying medical radiation dose, it is widely used, with several caveats attached.

One caveat is that Effective Dose can only be *estimated*, because it originates from measurements made on cylindrical phantoms. Despite its limitations, it serves to adequately answer the fundamental dosimetry question from a patient who has undergone a CT scan, "*What is the likelihood that I will be harmed from this exam?*" *

*AAPM report #96, 2008

CT Dosimetry cont'd

_k_-factor method. The current method of estimating Effective Dose from DLP is to multiply it by a '*k*' factor, first described in 2000. The *k*-factor has units of mSv per mGycm. These factors have been worked out for various CT-scan anatomies (head, neck, chest, abdomen and pelvis), for combinations (head + neck; abdomen + pelvis; chest + abdomen + pelvis) for adults as well as pediatric patients of various age groups.

Anatomy	**Adult**	**Pediatric**			
		0-6 mo	6 mo - 4 yr	4 - 9 yr	9 - 18 yr
	k factor				
Head	0.0021*	0.011	0.0067*	0.004	0.0032
Neck	0.0054	0.017	0.012	0.011	0.0079
Head + Neck	0.0031	0.013	0.0085	0.0057	0.0042
Chest	0.017	0.039	0.026	0.018	0.013
Abdomen	0.015*			0.020*	
Pelvis	0.019				
Abdomen + Pelvis	0.015	0.049	0.03	0.02	0.015
Trunk	0.015	0.44	0.028	0.019	0.014

* Used by ACR to calculate Effective Dose in their accreditation program

Example: The estimated Effective Dose for an abdomen scan on a 5 year old, with a DLP of 160 mGycm is

$$160 \text{ mGycm} \times 0.020 \ \frac{\text{mSv}}{\text{mGycm}} = 3.2 \text{ mSv}$$

CT Dosimetry cont'd

SSDE method. In 2011, a new method was developed that takes into account *patient diameter* into dosimetry calculations, thus overcoming one of the biggest shortcomings of the cylindrical phantom measurement method.

The SSDE method involves measuring the AP and lateral dimensions of the axial scan slice from the patient's image, and applying a correction factor from the SSDE table*. Multiplying the $CTDI_{vol}$ by correction factor generates SSDE, in mGy. There are other methods of determining SSDE, as well.

Example: Given the image below and a $CTDI_{vol}$ of 16 mGy for the 32 cm body phantom, what is the SSDE?

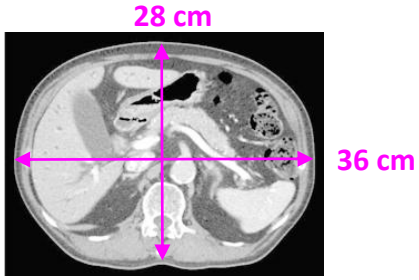


Image source: radiologypics.com

AP diameter = 28 cm
Lateral diameter = 35 cm
AP + Lateral = 64 cm
Conversion factor from SSDE table = 1.16
SSDE = 16 mGy × 1.16 = <u>18.56 mGy</u>

Despite this advancement, there is a major roadblock on the SSDE dosimetry method: The remaining steps have not yet been worked out.

- You cannot multiply SSDE (mGy) by scan length (cm) to obtain a corrected DLP
- You cannot apply the *k*-factor to obtain Effective Dose

Unless these steps are worked out, SSDE methods remains an incomplete approach. Once these steps are established, the SSDE method will probably become the standard method to estimate Effective Dose

Nevertheless, the SSDE method is now recommended by Image Gently as the preferred approach for setting pediatric protocols because it is an improvement over the cylindrical phantom model.

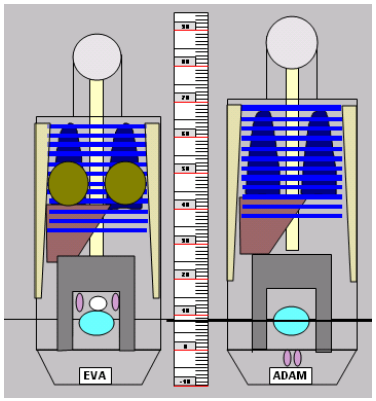
70 →

*AAPM report # 204, 2011

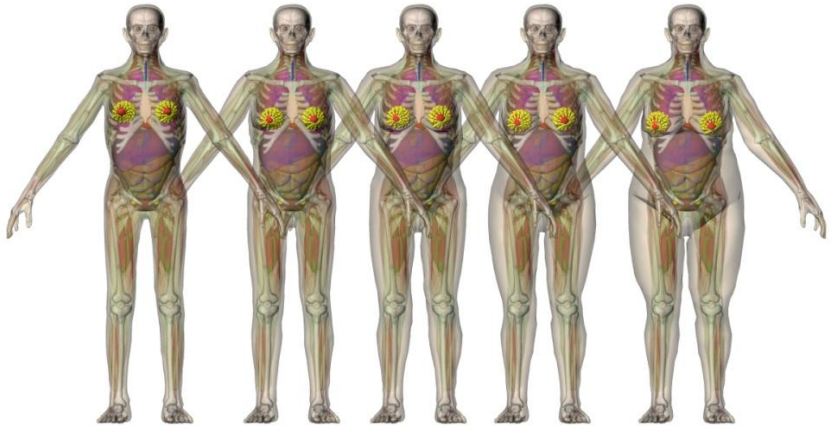
CT Dosimetry cont'd

Virtual Phantom method. Since the 1980s, mathematical phantoms have been used to calculate organ does for radiopharmaceuticals, by the MIRD (Medical Internal Radiation Dose) method. These methods were then extended to CT scans.

Geometric phantoms of the earlier days have given way to *voxelized* phantoms. Mathematical phantoms have been created for the average adult male and female; children (newborn, 1, 5, 10 and 15 year olds), as well as overweight and obese patients.



Eva and Adam geometric phantoms, from CT-Expo


Voxel phantoms (skinny to obese) from Virtual Phantoms, Inc.*

CT scan parameters and patient biometric data are applied to the suitable phantom, and Monte Carlo simulations (a concept that originated from the gambling industry, hence the name) are performed. Significant computing power is required to run these computer codes. The programs calculate doses to individual organs, and using published tissue weighting factors, estimate Effective Dose.

Virtual phantoms have traditionally found utility in research, although they are starting to make inroads into hospital based CT scanners. The website *impactscan.org* is a popular tool that uses the virtual phantom method. Software solutions are available from groups such as Virtual Phantoms, Inc., and Bayer Healthcare (Radimetrics).

*Image source: virtualphantoms.com

Risk and Risk Communication

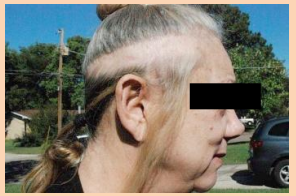
Risks from ionizing radiation exposure can be classified into two:

Stochastic (or probabilistic risk) is the long term risk of radiation-induced cancer. As accumulated dose increases chronically over the lifetime, the greater the probability of this outcome. Exposures early in life therefore pose greater risk than exposures in adulthood. The prevailing assumption, primarily for policy making purposes, is that there is no threshold dose below which stochastic risk is absent: this is the basis on the linear, no threshold (LNT) model. Prevalence of this risk from low doses (such as from diagnostic imaging, including from CT scans) is controversial, and will perhaps never be conclusively proven or disproven. The dosimetry unit to address stochastic risk is the **mSv** (or **mrem**).

Deterministic risk is the short-to-medium term possibility of tissue injury from acute radiation exposures. This includes skin reactions such as skin reddening, temporary or permanent hair loss, desquamation, ulceration and necrosis; induction of cataracts, sterility, damage to the unborn fetus, etc. It is well established that there are threshold doses before each of these effects become prevalent, and increasing doses increase the severity of the effect (rather than just the probability that they may occur). The dosimetry unit to address deterministic risk is the **mGy** (or **mrad**).

Of the two risks, stochastic risks are of greater concern for CT exposures. As a greater percent of the population gets exposed to ever increasing amounts of radiation, the concern is that the overall prevalence of cancer may increase. Moreover, radiation induced cancer takes 10-30 years to manifest. Age at exposure is therefore an important determinant of stochastic risk, with children being more susceptible than adults.

Deterministic risks are relatively rare for diagnostic procedures. However, in an episode at Cedar-Sinai Medical Center during 2008-2009, over 200 patients undergoing brain perfusion studies were exposed to doses of 3000-4000 mGy to the skin, resulting in hair loss such as shown in the photograph. Many of the recent regulatory and accreditation initiatives on CT scanners can be traced back to this incident.



Source: latimes.com

Note: this was not a Cedar-Sinai patient

The actions taken to decrease patient dose are aimed at minimizing both stochastic and deterministic risks.

Risk and Risk Communication cont'd

When patients want to know their dose from a CT scan, they are essentially asking "what is the risk from this scan?" For the most part, they are knowingly or unknowingly referring to stochastic risk.

Informing the patent about their estimated effective dose in mSv hardly does the patient any good, because the lay public cannot relate to this, or any other dosimetry unit. It is imperative to communicate dose information in a way that patients and members of the general public can understand it.

The most practical means to achieve this is to compare effective dose to **natural background radiation**. Although comparisons can also made to other radiographic procedures (such as number of chest or dental x-rays), this is tantamount to chasing a moving target: technologies for all radiographic procedures are evolving rapidly, and these do not present a stable baseline to make a comparison against.

Natural background over much of the continental United States is a stable quantity, not expected to change in the foreseeable future. While the most recent reports estimate it at 3.11 mSv (311 mrem) per year, the general approach is to round it off to **3 mSv (300 mrem) per year**. One of the few exceptions is natural background in the high Colorado plateau, where the increased elevation and Rocky Mountains cause it to be 4.5 mSv (450 mrem) per year. Both these values provide useful cornerstones to compare radiation doses from CT scans, as well as all other diagnostic procedures involving ionization radiation.

Examples:

Procedure	Effective Dose (mSv)*	Equivalence in natural background radiation (years & months)	
		Continental USA	Colorado Plateau
Head	2	7 months	5 months
Chest	7	2 yrs & 3 months	1 yr & 6 months
Abdomen and Pelvis	10	3 yrs & 2 months	2 yrs & 2 months
A lifetime (80 years) worth of natural background in the continental USA (3 x 80 = 240 mSv) is roughly equivalent to 24 abdomen and pelvis scans at 10 mSv per scan.			

*Source: NCRP Report # 160, 2009

Image Quality Metrics

Noise, Contrast-to-noise ratio (**CNR**), Low Contrast Resolution (**LCR**) and High Contrast Resolution (**HCR**) are four important image quality metrics.

Noise (the amount of *graininess* in the image) is quoted in Hounsfield Units (HU) as the Standard Deviation (SD) value that is displayed when an ROI is drawn. Larger the SD, greater the noise.

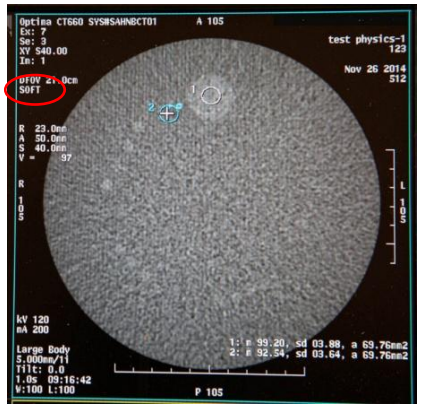
Parameters		Implications
Decreasing kV increases noise	33 →	Lower kVs can only be used for small patients or body parts Lower kVs cannot be used for dense materials (such as bone) Lower kVs can used for improved iodine contrast visualization Lower kVs can be used for CT Angiography
Decreasing mAs increases noise	← 3	Increasing mAs decreases noise, but only up to a certain point. Lower mAs results in lower dose, in a 1:1 relationship
Increasing the pitch can increase noise	36 →	Excessively high pitch values, especially above 1.5, can cause noise levels to be intolerable However, decreasing the pitch is an expensive way to decrease noise.
Thinner images have increased noise	37 →	Increasing the thickness of diagnostic images ('slice thickness') is the most economical way of improving image quality: there is no expense in terms of increased dose.
More aggressive algorithms increase noise	32 →	For most applications, the 'standard' (or equivalent, depending on manufacturer) is suitable for the majority of applications. Siemens and Toshiba offer more choices to fine-tune algorithms (or kernels) than Philips or GE. A more aggressive algorithm is needed for visualizing high contrast objects (e.g., GE requires the use of bone algorithm for Hi Res Chest scans)

Image Quality Metrics cont'd

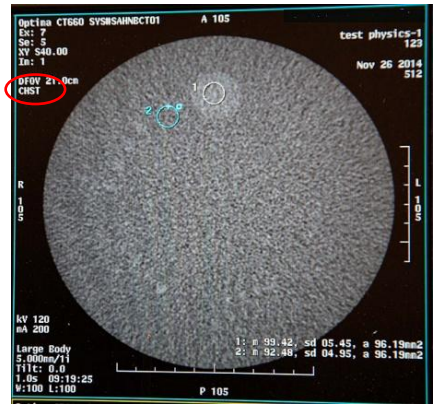
Reconstruction Algorithm vs. Noise

This image panel shows the effect of algorithm on noise, for a GE scanner which typically has 8 selections.
Images are of the LCR module of the Gammex phantom, and are arranged in the order of increasing noise. All exposures were made at identical techniques (120 kV, 200 mAs, axial scans). Notice that the lung algorithm visually looks the noisiest, but numerically only the third-most noisy. The graph on the next page shows noise (measured as SD of the ROI drawn in the background area next to the 25 mm cylinder; blue circles) as a function of algorithm.

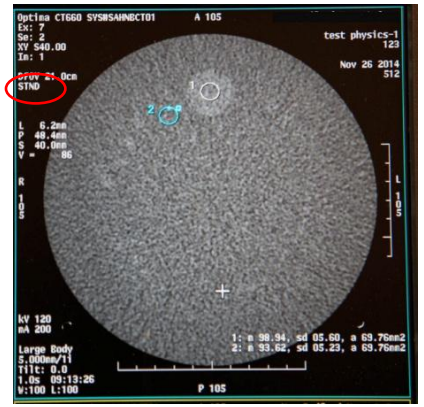
← 11 ← 31



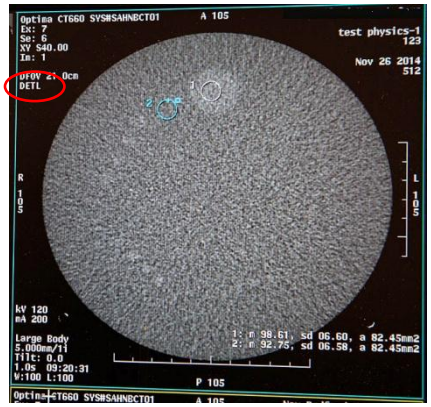
Soft



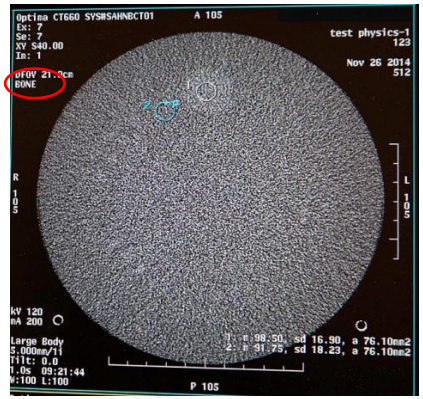
Chest



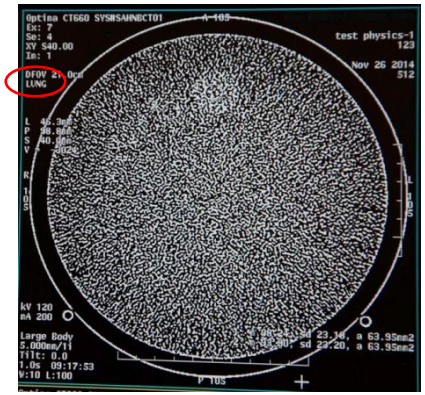
Standard



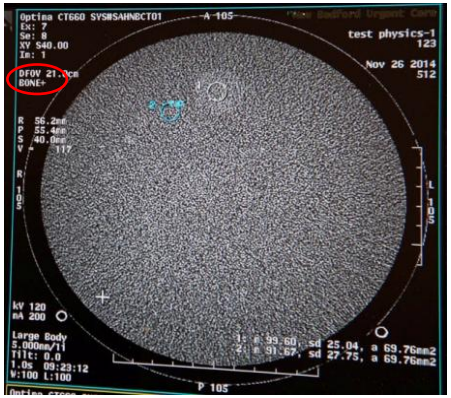
Detail



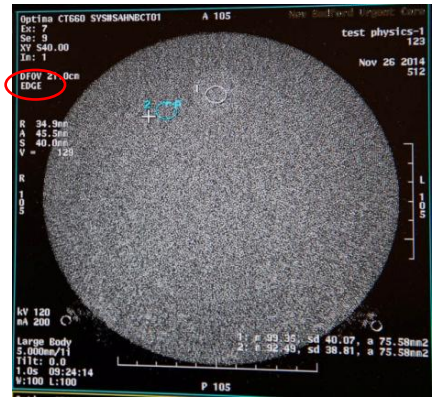
Bone



Lung



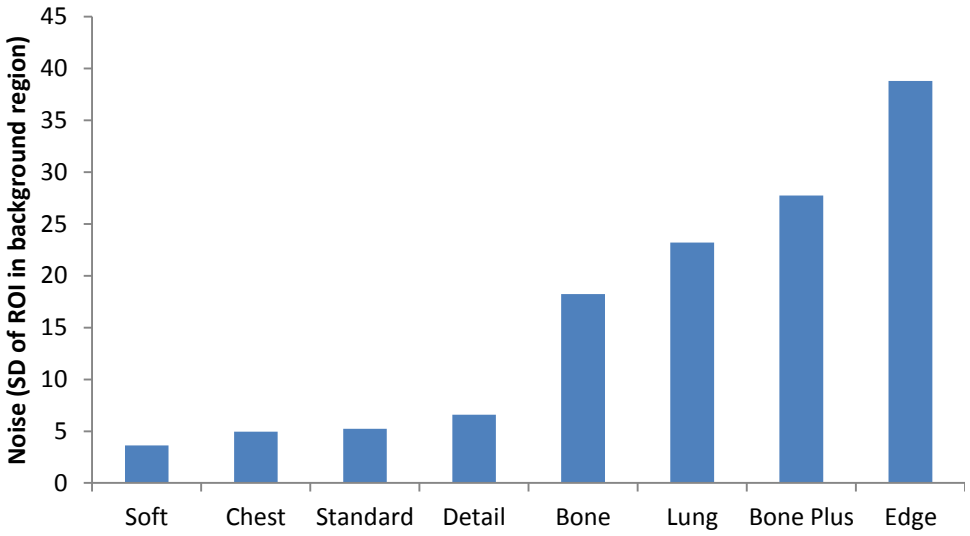
Bone+

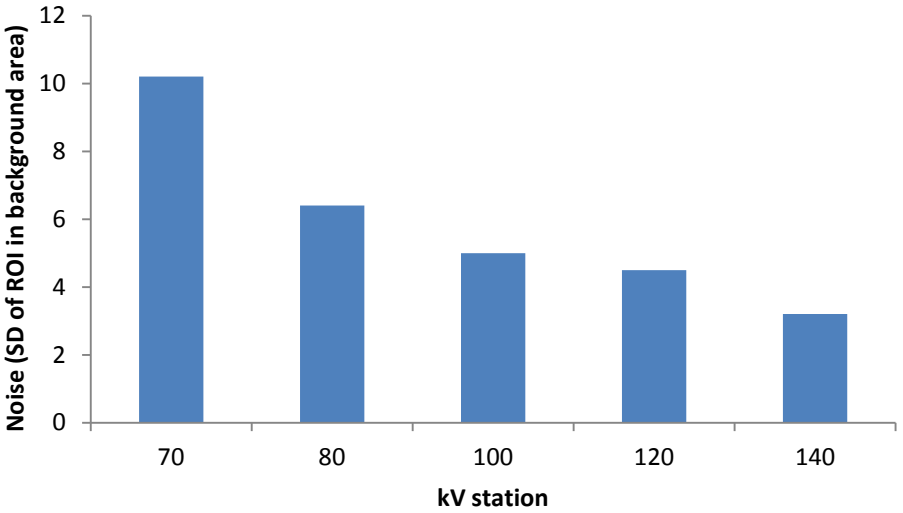


Edge

Image Quality Metrics cont'd

Variation of noise with algorithm, on a GE scanner





Decrease in noise with increasing kV, on a Siemens Somatom Definition scanner (one of the very few scanners that has a 70 kV station). All exposures were made at 200 mAs, axial scans, on the LCR module of a Gammex phantom. ROI was measured as shown on the images on the previous page.

← 31

Image Quality Metrics cont'd

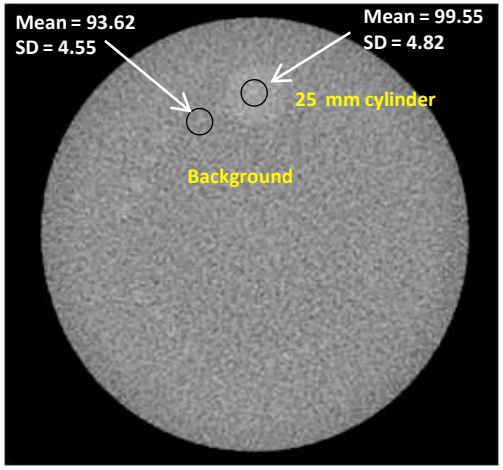
Contrast is the numerical difference between HUs in ROIs drawn over different materials (typically, object and background). Conceptually, it is the ability of the human eye to distinguish between two objects (or an object against a background) that do not differ much in terms of brightness, contrast or texture.

Contrast to Noise Ratio (**CNR**) is Contrast divided by Noise .

Example: The ACR uses module #2 of the Gammex phantom for this measurement. An ROI is drawn over the 25 mm cylinder (the 'object'), and one beside it to the left (the 'background' material). Suppose the mean and SD of the ROI over the object are 99.55 and 4.82; and over the background are 93.62 and 4.55, respectively,

Contrast = 99.55 – 93.62 = 5.93
CNR = contrast ÷ SD of Bkg = (98.94 – 93.62) / 4.55 = **1.3**

What if the contrast is a negative number? Suppose the two mean values above were reversed; you would get 93.62 – 99.55 = -5.93 for contrast. *In this case, simply ignore the negative sign proceed with the calculation.*



Gammex Phantom Module 2

A CNR of exactly 1.0 means that contrast is equal to noise (e.g., if the above numbers were 99.55 – 95.00 = 4.55; and 4.55 / 4.55 = 1.0). A CNR of less than 1.0 means that noise is greater than contrast (the image looks 'noisy'), and CNR of more than 1.0 means contrast is greater than noise (the image looks 'clean').

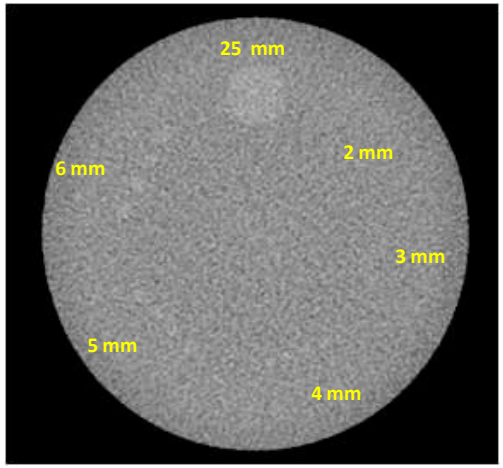
The ACR has established minimum criteria for CNR for the 4 phantom protocols required for accreditation

	Adult Head	Pediatric Head	Adult Abdomen	Pediatric Abdomen
Minimum **CNR** required	1.0	0.7	1.0	0.4

These CNR criteria indirectly decide the question "how low can I go on dose?" Setting protocols that have extremely low doses are counter-productive, because the images will lack adequate diagnostic quality. At the same time, by setting the minimum pediatric CNR requirements to 0.7 and 0.4, the ACR is sending the message that 'noisier' images are tolerable for children, thereby giving more weightage to dose reduction over image quality.

Image Quality Metrics cont'd

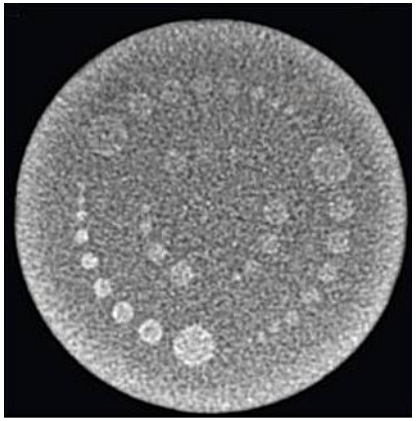
Low Contrast Resolution (**LCR**) is a more subjective quantification of the detectability of objects that do not differ a whole lot from background (i.e., objects that have very little contrast). Module 2 of the Gammex phantom, in addition to the 25 mm cylinder used for CNR measurement, has 5 sets of rods, of diameters 6, 5, 4, 3 and 2 mm. Counting which set of rods is visible under the scan parameters, at a window setting of 100 and level of 100 provides LCR. All 4 rods in a set must be visible to constitute a score. In the figure shown, the 4 mm rod set is visible.

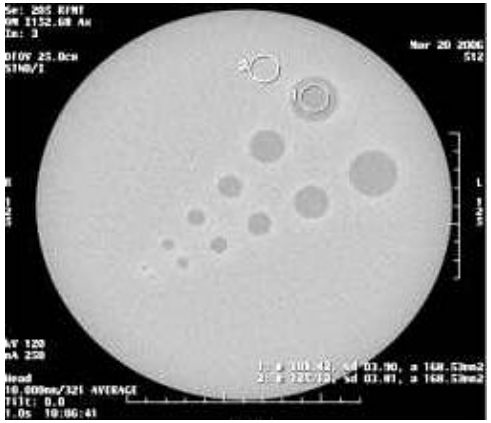


Gammex Phantom Module 2

Prior to 2008, the ACR used LCR as a pass/fail criteria for accreditation. Because of its subjectivity, the ACR abandoned LCR in favor of the more objective criteria of CNR. As a rule of thumb, at least the 6 mm set should be visible for a routine adult abdomen technique.

Every good CT quality control phantom will have an LCR module, although they may differ in the number and size of objects, and therefore, scoring and method of evaluation. Therefore, it is impossible to compare scores of different CTs obtained using different phantoms. A few LCR module examples are shown below.


LCR Module, Catphan 500 Phantom
Image source: health.siemens.com


LCR Module, Fluke AAPM CT Performance Phantom
Image source: openi.nlm.hih.gov

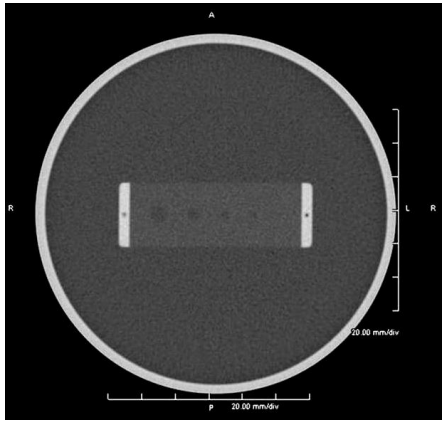
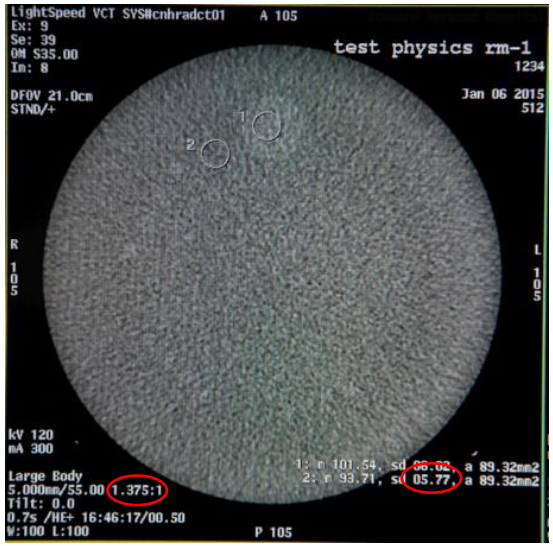
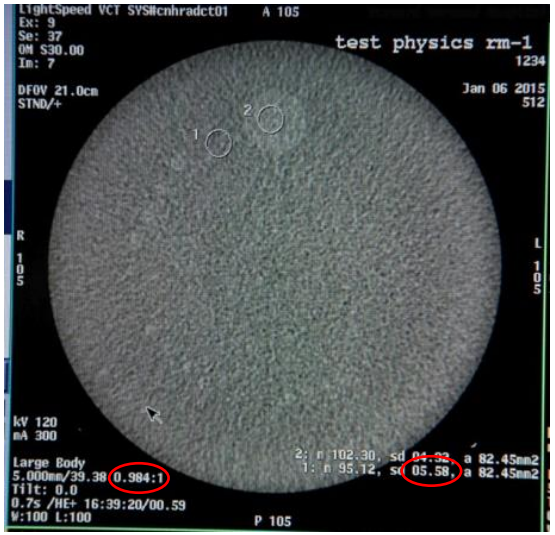

LCR Module, GE CT Phantom
Image source: ceessentials.net

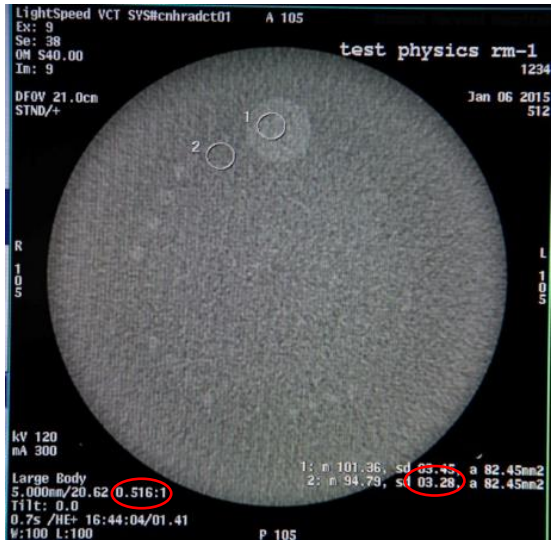
Image Quality Metrics cont'd

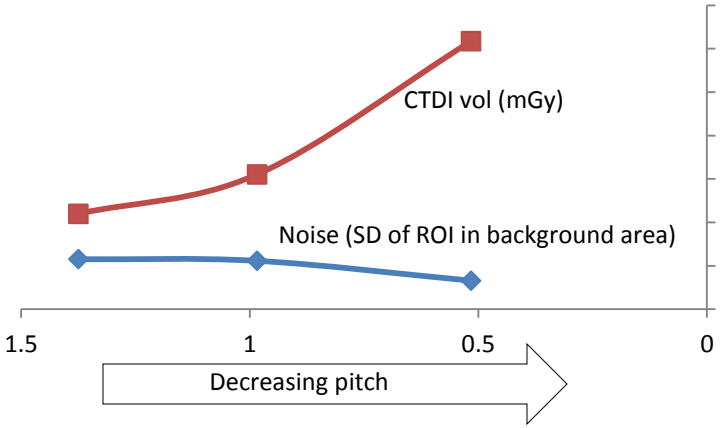
Pitch vs. Noise

Small pitch values are slightly less noisy (blue line in graph), have better low contrast resolution (notice improved cylinder visibility as pitch decreases), but come at the expense of steeply increased dose (red line in graph). All helical scans done at 120 kV, 300 mA, 0.7 sec (210 mAs), 5 mm images, standard algorithm, at decreasing pitch values of 1.375, 0.984 and 0.516, respectively.







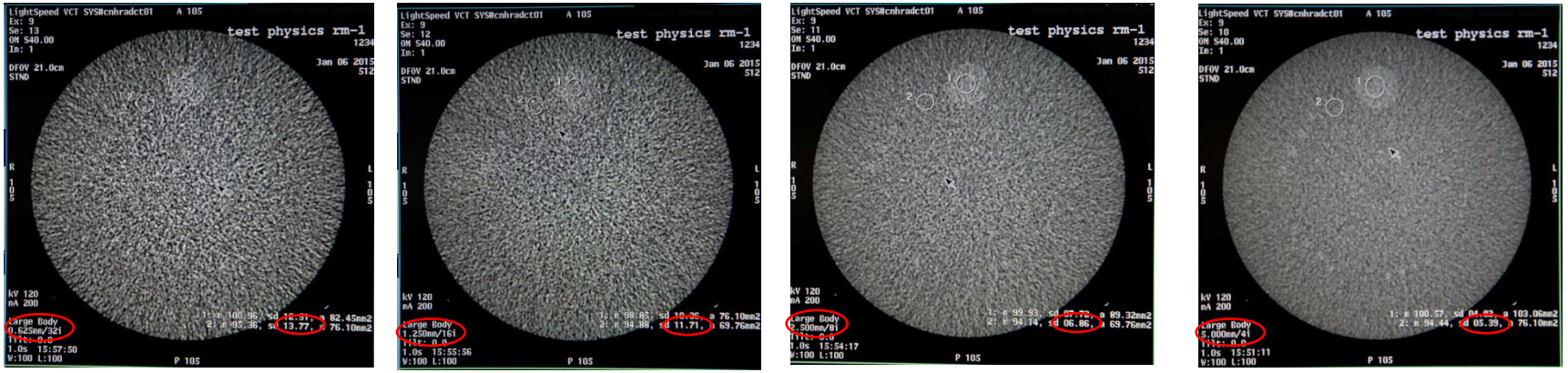


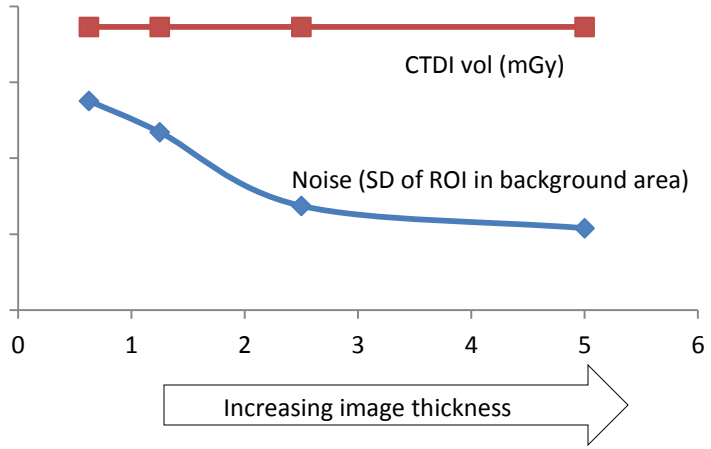
← 9
← 31
36

Image Quality Metrics cont'd

Image Thickness vs. Noise

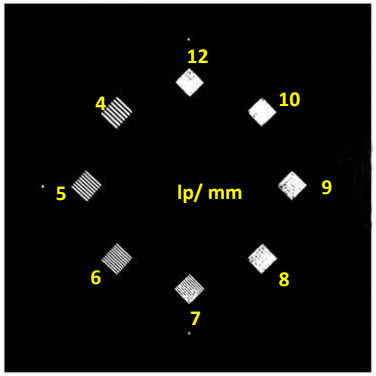
Unlike decreasing pitch, increasing image thickness is a very economical way to decrease noise (blue line in graph). There is no expense in terms of dose (red line in graph). All exposures at 120 kV, 200 mA, 1 sec, 200 mAs, 20 mm beam thickness, standard algorithm, providing images of 0.625 mm, 1.25 mm, 2.5 mm and 5 mm, respectively. Notice improvement in cylinder visibility as image thickness increases.





← 31

Image Quality Metrics cont'd



Gammex Phantom Module 4

High Contrast Resolution (**HCR**), is the subjective ability of the human eye to distinguish between objects that have very high contrast (e.g., bone vs. tissue. If ROIs were drawn over the two objects, the contrast would be a large number, for instance 925.55 – 95.23 = 830.32, a far cry from the CNR example on the page 34).

For most practical purposes, this boils down to detecting small or fine objects against a contrasting background, such as calcifications in tissue, or a crack in a bone. For this reason, HCR is also called **spatial resolution** or **line pair resolution.**

Module 4 of the Gammex phantom contains the HCR insert, consisting of line pair sets that go from 4 lp/cm (line pairs per cm) to 12 lp/cm. Although the ACR does not currently have pass/fail criteria on HCR, it was previously specified as at least 5 lp/cm for the routine abdomen protocol. All the lines in a particular block must be visible without interference (or aliasing) for a valid score. This evaluation is performed at a window width of 100 and level of approx. 1100.

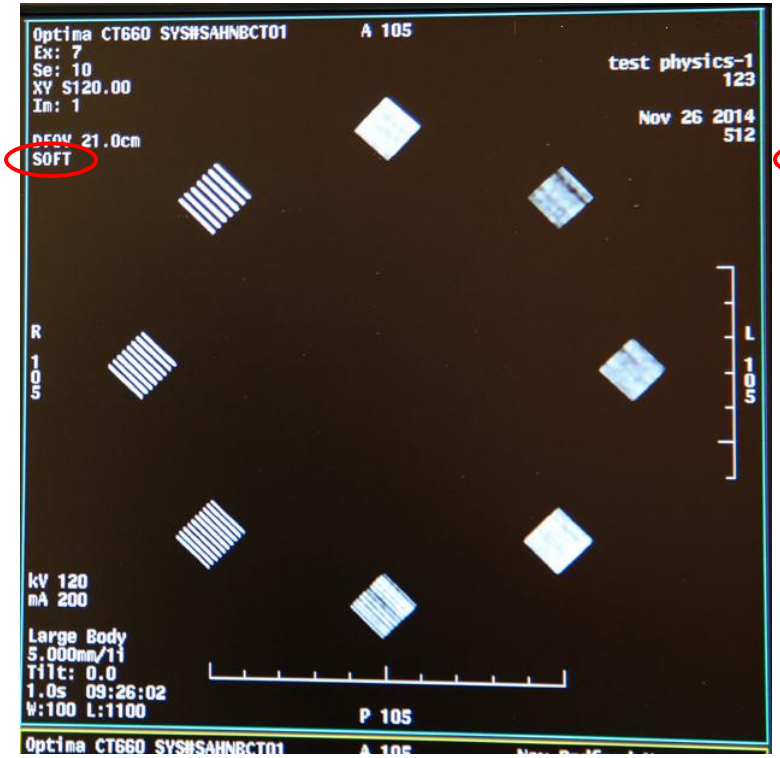
A line pair, as the name suggests, is two lines distinguishable by the naked eye. If the 7 lp/cm set is visible, it means that 14 lines are distinguishable in the space of one cm; or that 0.07 cm (calculated as 1 cm/14), or 7 mm, is the thinnest line visible, or the 'limiting resolution'. CT scanners do poorly in the HCR department than other imaging modalities, such as general radiography or mammography, where resolution is stated in lp/mm (as opposed to cm). For instance, a resolution of 9 lp/mm (0.06 mm limiting resolution) is commonly achieved in a Hologic Selenia digital mammography unit.

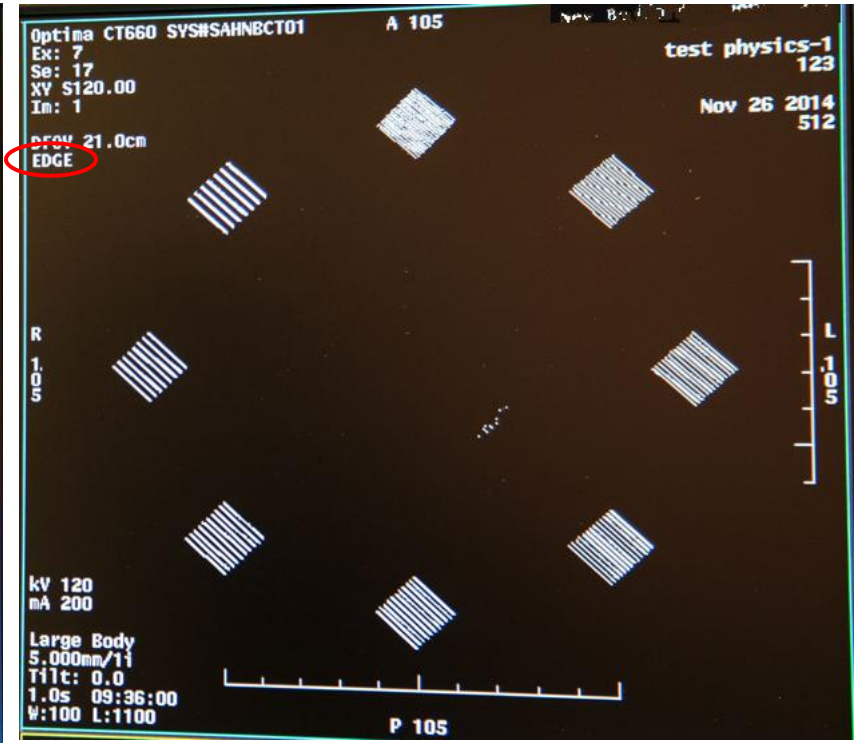
Just as CNR is the more objective method of quantifying LCR, HCR is quantified mathematically as **modulation transfer function** (**MTF**). A detailed treatment of MTF is beyond the scope of this tutorial. Automated QC programs of Siemens CT scanners calculate the MTF of the Siemens phantom HCR insert, and display a pass/fail based on their criteria.

Image Quality Metrics cont'd

Line pair visibility is greatly affected by the reconstruction algorithm. As a general rule, 'smoother' algorithms result in lower spatial resolution, while more 'aggressive' algorithms (also the noisiest images) are best suited for finer line pair visibility. Below are HCR images for the Soft and Edge algorithms (representing the opposite ends of the noise scale, as shown in page 32) for the GE scanner, with differences in lp/mm visibility evident.

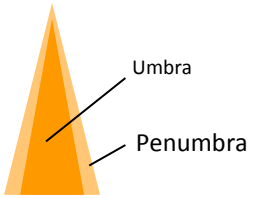
← 11





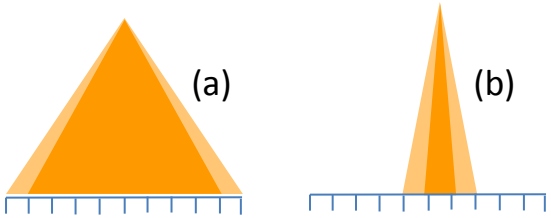
It follows that LCR and HCR are flip sides of visual image quality; and algorithm that fares best for LCR is the poorest for HCR, and vice versa. A 'middle of the road' algorithm such as Standard, is therefore a good choice for most protocols, providing a reasonable balance between LCR and HCR. Additional reconstructions can be performed for more specific imaging needs.

Overbeaming



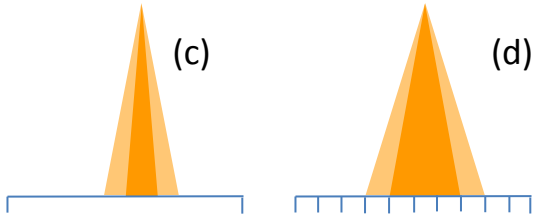
The 'useful' or 'primary' beam emerging from the x-ray tube has two components: the umbra region in the center, flanked by the penumbra on both ends.

For a given scanner, the size of the penumbra is constant, irrespective of beam thickness. Therefore, penumbra size is relatively larger for smaller beams, when compared to larger beams. Example: For the 10 mm beam (a), the umbra is 8 mm, and the penumbra is 2 mm (1 mm at each end), i.e., the penumbra is 20% of the total beam. For the 3 mm beam (b), the umbra is 1 mm, and the penumbra is still 2 mm, i.e., the penumbra is 67% of the total beam.



Let's say each mm of umbra is worth 2 'units' of dose, and each mm of penumbra, 1 unit *(these are arbitrary units and values, just for illustration)*. The beam in (a) is worth 18 units of and (b) is 4 units.

In the days of single detector CTs, the entire beam (umbra + penumbra) was captured by the detector; none of it was wasted. Example: a 3 mm beam used on a single 10 mm detector (c). With multi detector CTs, every detector row used to produce the image needs to be 'fed' the same beam intensity. Suppose three detectors, each 1 mm are used for the same 3 mm beam, as in figure (b). Detectors 1 and 3 are 'starved' of photons, while detector 2 is 'well fed'. This is not good for image production. The beam thickness now needs to be increased to 5 mm (d), to equally satisfy the three detectors.

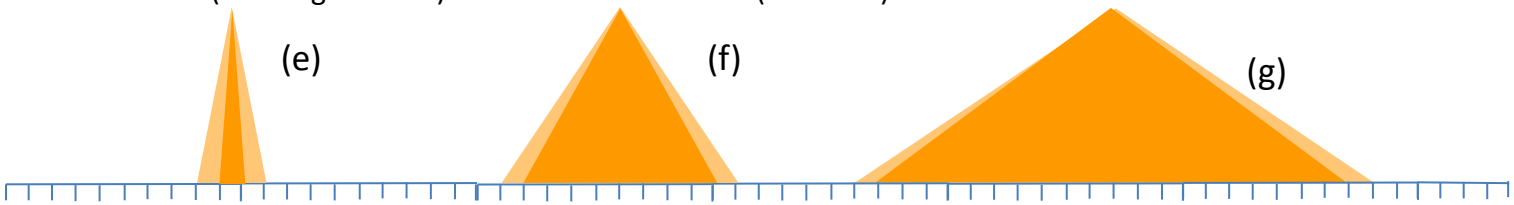


The single detector CT cost 4 units of dose, and the multi detector CT, 7 units: a 75% increase in dose. Thereby, switching from a single detector to multi detector CT represents an increase in dose. The need to increase beam thickness in this manner is called overbeaming, and is one of the dose tradeoffs of modern CTs.

Overbeaming cont'd

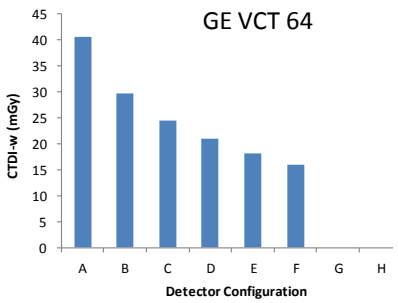
Following the same logic,

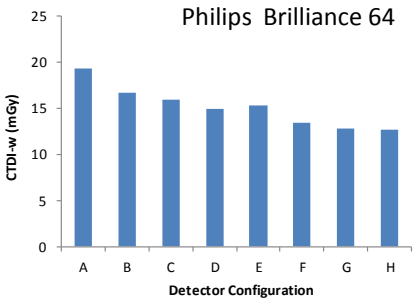
a 1 mm beam (e) would cause a 100% increase in dose (4 dose units for multi-detector vs. 2 units for a single detector);
a 10 mm beam (f) would cause a 10% increase in dose (22 vs. 20);
a 20 mm beam (g): 5% increase (42 vs. 40), assuming such a single detector scanner existed; and
a 40 mm beam (not diagrammed): 2.5% increase in dose (82 vs. 80)

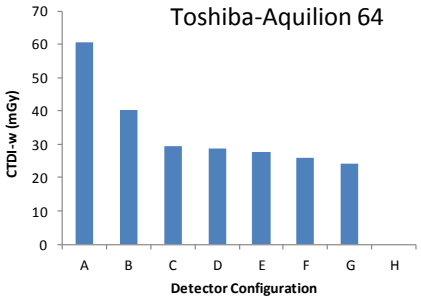


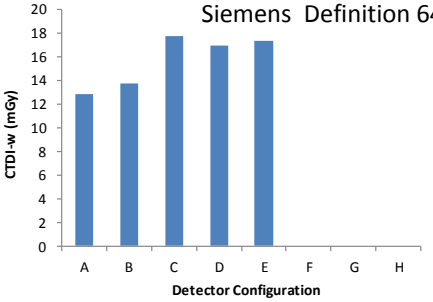
Notice that the *degree* of increase in dose falls as beam thickness increases. This is also a characteristic of overbeaming. As beam widths increase, the doses almost plateau out.

Overbeaming is the reason why **thinner radiation beams** have higher $CTDI_{vol}$ (and are therefore **less efficient**) when compared to **wider beams**, as was shown on the table on page ← 6. The spread of $CTDI_{vol}$ over the range of axial NxT is shown graphically in your annual CT testing report from F X Massé Assoc., as in the examples below:









Beam thickness in the above graphs increases from A to H. The downward trend in CTDI is evident. There are some exceptions, as in the Siemens Definition-64 (which is related to volumetric imaging, as was mentioned in page 8).

Overbeaming cont'd

The preferable method to minimize overbeaming is to *maximize* the number of detector rows (i.e., use the widest beam) possible.

There are only a very few instances where a narrow beam is preferred. **Example:** consider an axial protocol for high-resolution chest scans, where a 1.25 mm image slice is required every 10 mm. Suppose this is a GE-VCT 64, the technique is 120 kV, 200 mAs, the total scan length is 25 cm, and scans are to be contiguous. Obtain $CTDI_w$ from page 6.

← 6

Case 1: Use the **narrowest** beam: NxT = 2 x 0.625 = 1.25 mm. Table increment (I) = 10 mm. P = I/NT = 10 / 1.25 = 8.
CTDIvol = 37.4 / 8 = 4.675. DLP = 4.675 x 25 = 116.875. Using chest *k*-factor of 0.017, Effective Dose = 116.875 x 0.017 = **1.98 mSv**

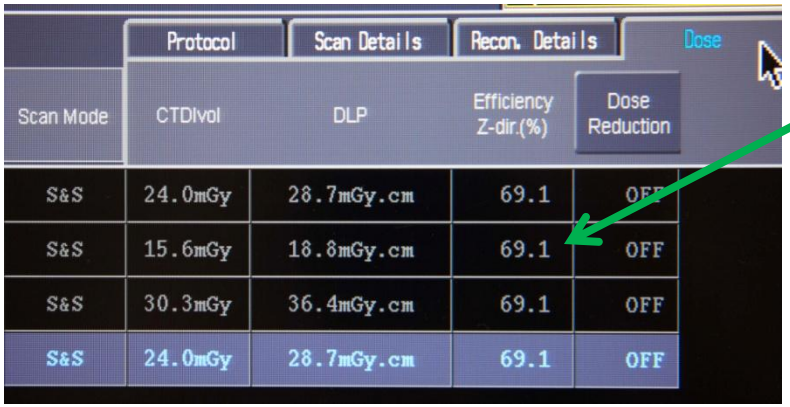
Case 2: Use the **widest** beam: NxT = 64 x 0.625 = 40 mm. Table increment (I) = 40 mm. P = I/NT = 40 / 40 = 1.
CTDIvol = 15.83 / 1 = 15.83. DLP = 15.83 x 25 = 395.75. Using chest *k*-factor of 0.017, Effective Dose = 395.75 x 0.017 = **6.72 mSv**

Therefore, even though the $CTDI_{vol}$ for the thin beam is 2.4 times higher than that of the wide beam, its use in this case results in 3.4 times less dose than using a wide beam.

Another instance when the widest beam is not preferable is for pediatric scans on very young / small patients. In this case, a larger beam width could easily cause overscanning of the tiny anatomy, and a slightly narrower beam is selected as an acceptable tradeoff. Another reason for selecting a narrower beam happens to be **overranging,** covered in the next section.

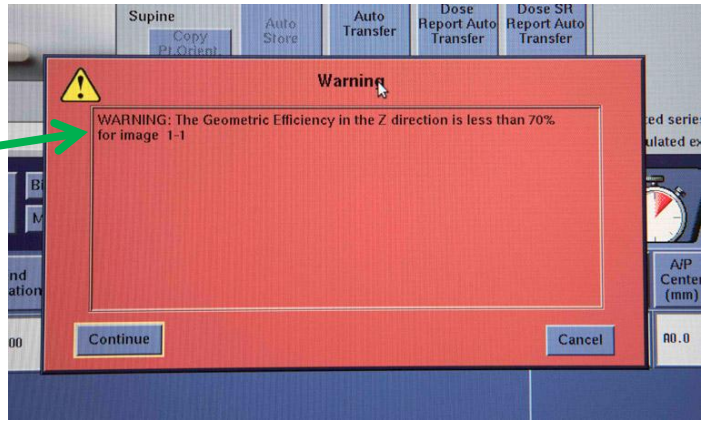
The degree of overbeaming is quantified by **geometric efficiency in the z-direction,** or **z-axis efficiency**. Regulations require that if the geometric efficiency of a detector configuration is less than 70%, a warning must be displayed to the operator. This is done via a pop up window, and requires operator acknowledgement.

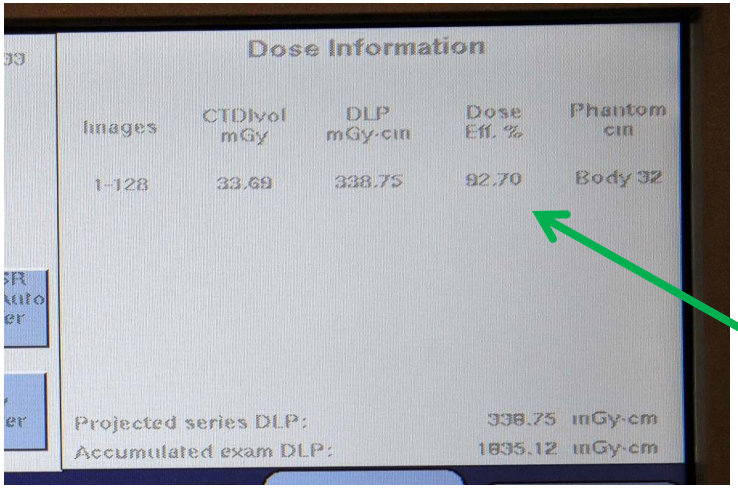
Overbeaming cont'd



Example of geometric efficiency display on a Toshiba 16 slice scanner. This display is required of the efficiency falls below 70%

Example of geometric efficiency display pop-up window on a GE scanner.





Some scanners, such as this GE Optima CT660, display geometric efficiency for all detector configurations.

Overranging

Overranging is the flip side of overbeaming, the 'dirty little secret' of helical scanning. It goes unnoticed and unreported because there are no readily available indicators to alert the user of this phenomenon. In mot cases, it is a 'necessary evil', but should be understood and controlled to the extent possible.

When a helical scan is performed, data from the first full turn (for scanners or protocols that used 360° interpolation) or half turn (for scanners or protocol that use 180 ° interpolation) of the gantry is used by interpolation algorithms. Only a small portion of the data collected is utilized. The rest is 'wasted' radiation.

This causes direct irradiation of organs beyond the planned scan length by a few mm to several cm.

There are two ways to detect and calculate the extent of overranging:

1. Carefully observe the beam-on indicator on the scanner gantry when a scan is in progress. Note down the table position at the instant the light comes on, and when it turns off. Compare this against the prescribed table positions.

2. Use the displayed DLP and $CDTI_{vol}$, as follows: Actual scan length in mm = $\frac{DLP}{CTDIvol}$ x 10
(The x10 is just to convert cm to mm)

There are currently no regulatory or accreditation standards that address overranging. For instance, ACR standards are build on $CTDI_{vol}$, which is oblivious of overranging. The effects of overranging can only be felt at the Effective Dose level. At the organ level, the effect of overranging should be taken into consideration when radiosensitive organs (eye, thyroid and testicles) are close to the edge of the prescribed scan length. Realize that even if these organs are placed 'just outside' the planned scan range, they can receive a significant radiation dose due to overranging. Therefore, it is important to compute the extent of overranging for such protocols, and set scan boundaries outside of the overrange area.

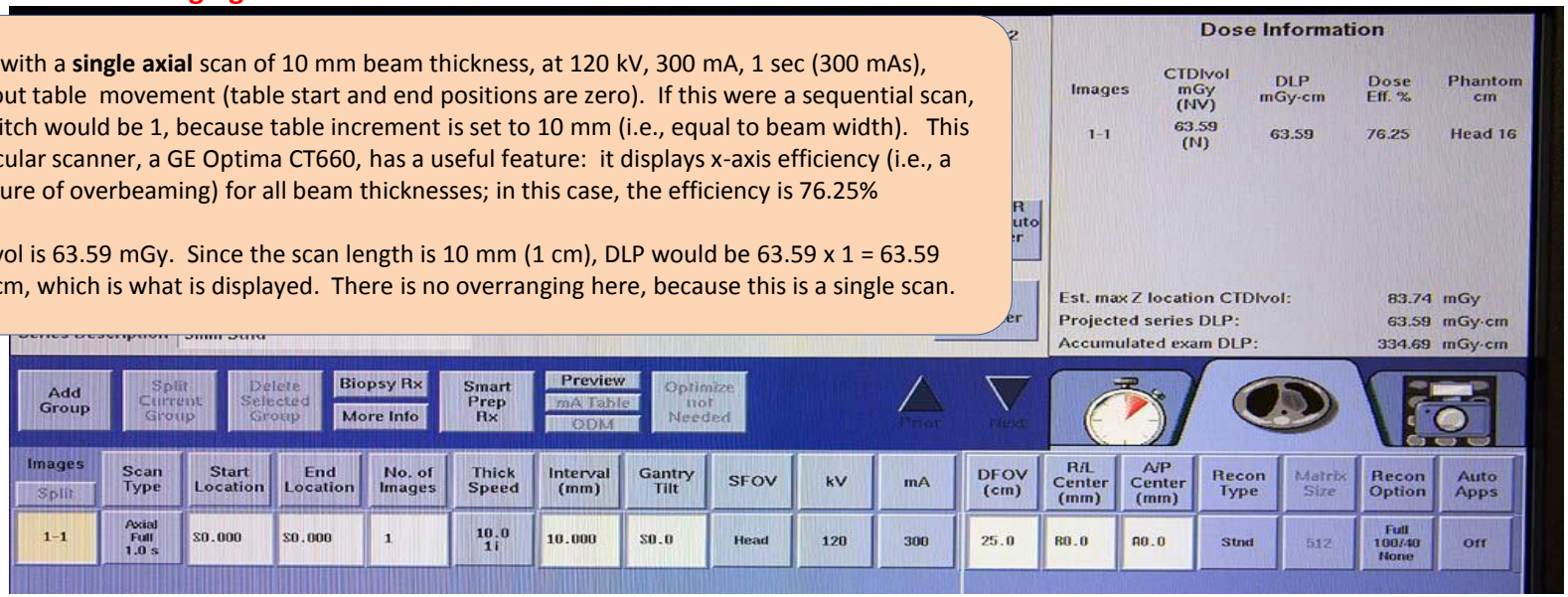
Factors that effect the degree of overranging are **scan type** (helical), **detector configuration** (beam thickness) and **pitch,** and **scan length**.

Overranging cont'd

Axial scan vs. Overranging

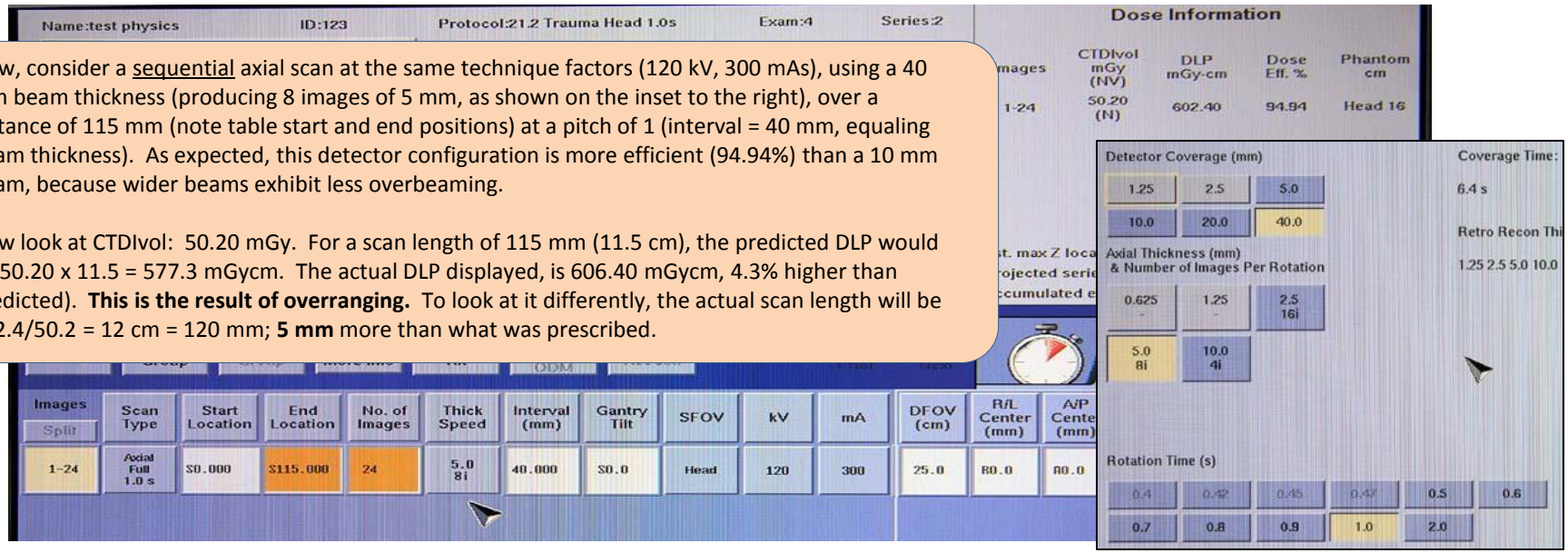
Start with a **single axial** scan of 10 mm beam thickness, at 120 kV, 300 mA, 1 sec (300 mAs), without table movement (table start and end positions are zero). If this were a sequential scan, the pitch would be 1, because table increment is set to 10 mm (i.e., equal to beam width). This particular scanner, a GE Optima CT660, has a useful feature: it displays x-axis efficiency (i.e., a measure of overbeaming) for all beam thicknesses; in this case, the efficiency is 76.25%

CTDIvol is 63.59 mGy. Since the scan length is 10 mm (1 cm), DLP would be 63.59 x 1 = 63.59 mGycm, which is what is displayed. There is no overranging here, because this is a single scan.



Now, consider a <u>sequential</u> axial scan at the same technique factors (120 kV, 300 mAs), using a 40 mm beam thickness (producing 8 images of 5 mm, as shown on the inset to the right), over a distance of 115 mm (note table start and end positions) at a pitch of 1 (interval = 40 mm, equaling beam thickness). As expected, this detector configuration is more efficient (94.94%) than a 10 mm beam, because wider beams exhibit less overbeaming.

Now look at CTDIvol: 50.20 mGy. For a scan length of 115 mm (11.5 cm), the predicted DLP would be 50.20 x 11.5 = 577.3 mGycm. The actual DLP displayed, is 606.40 mGycm, 4.3% higher than predicted). **This is the result of overranging.** To look at it differently, the actual scan length will be 602.4/50.2 = 12 cm = 120 mm; **5 mm** more than what was prescribed.



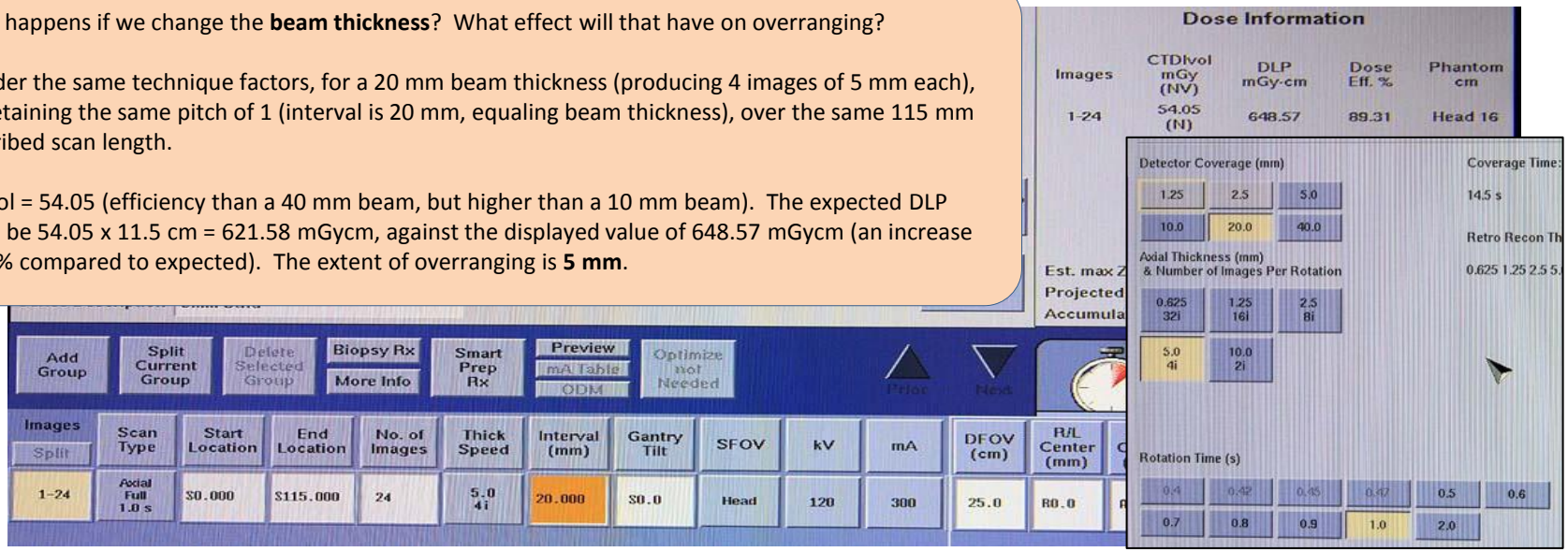
Overranging cont'd

Beam thickness vs. Overranging

What happens if we change the **beam thickness**? What effect will that have on overranging?

Consider the same technique factors, for a 20 mm beam thickness (producing 4 images of 5 mm each), and retaining the same pitch of 1 (interval is 20 mm, equaling beam thickness), over the same 115 mm prescribed scan length.

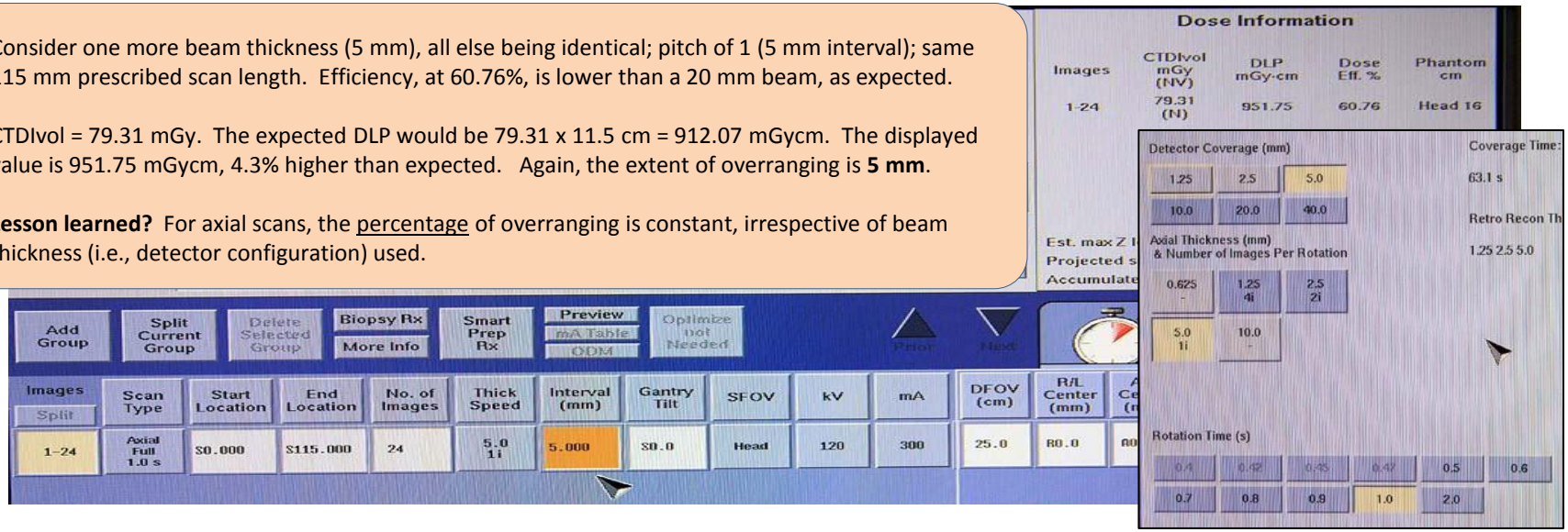
CTDIvol = 54.05 (efficiency than a 40 mm beam, but higher than a 10 mm beam). The expected DLP would be 54.05 x 11.5 cm = 621.58 mGycm, against the displayed value of 648.57 mGycm (an increase of 4.3% compared to expected). The extent of overranging is **5 mm**.



Consider one more beam thickness (5 mm), all else being identical; pitch of 1 (5 mm interval); same 115 mm prescribed scan length. Efficiency, at 60.76%, is lower than a 20 mm beam, as expected.

CTDIvol = 79.31 mGy. The expected DLP would be 79.31 x 11.5 cm = 912.07 mGycm. The displayed value is 951.75 mGycm, 4.3% higher than expected. Again, the extent of overranging is **5 mm**.

Lesson learned? For axial scans, the <u>percentage</u> of overranging is constant, irrespective of beam thickness (i.e., detector configuration) used.



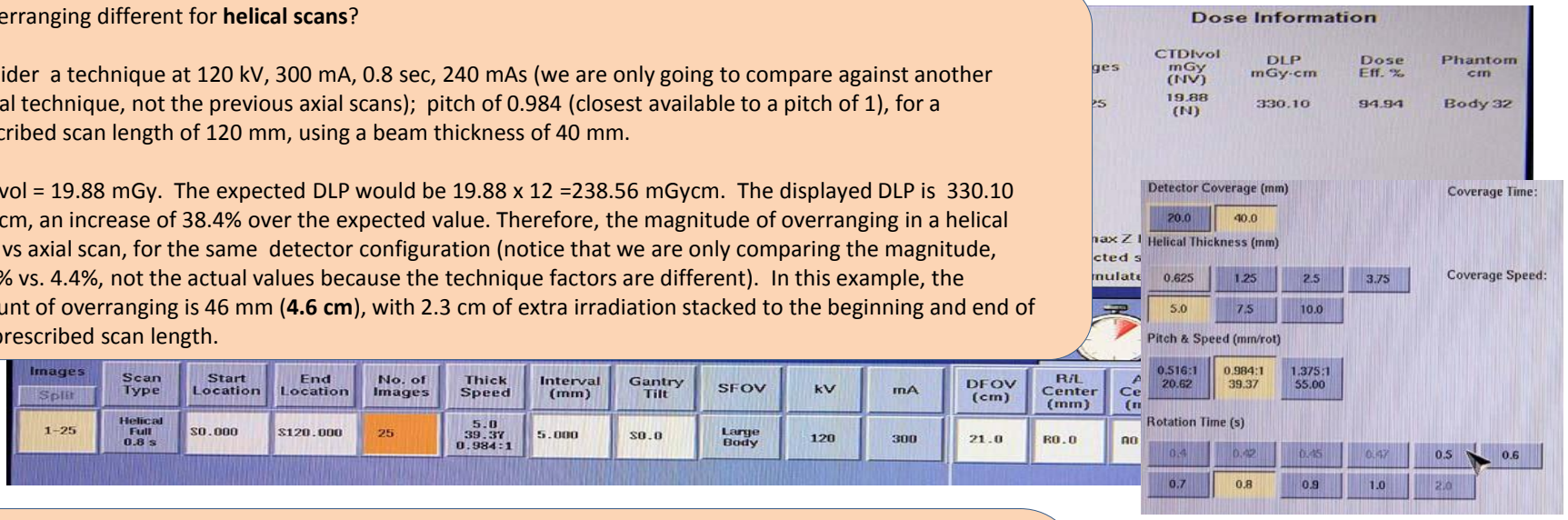
Overranging cont'd

Helical scan vs. Overranging

Is overranging different for **helical scans**?

Consider a technique at 120 kV, 300 mA, 0.8 sec, 240 mAs (we are only going to compare against another helical technique, not the previous axial scans); pitch of 0.984 (closest available to a pitch of 1), for a prescribed scan length of 120 mm, using a beam thickness of 40 mm.

CTDIvol = 19.88 mGy. The expected DLP would be 19.88 x 12 =238.56 mGycm. The displayed DLP is 330.10 mGycm, an increase of 38.4% over the expected value. Therefore, the magnitude of overranging in a helical scan vs axial scan, for the same detector configuration (notice that we are only comparing the magnitude, 38.4% vs. 4.4%, not the actual values because the technique factors are different). In this example, the amount of overranging is 46 mm (**4.6 cm**), with 2.3 cm of extra irradiation stacked to the beginning and end of the prescribed scan length.

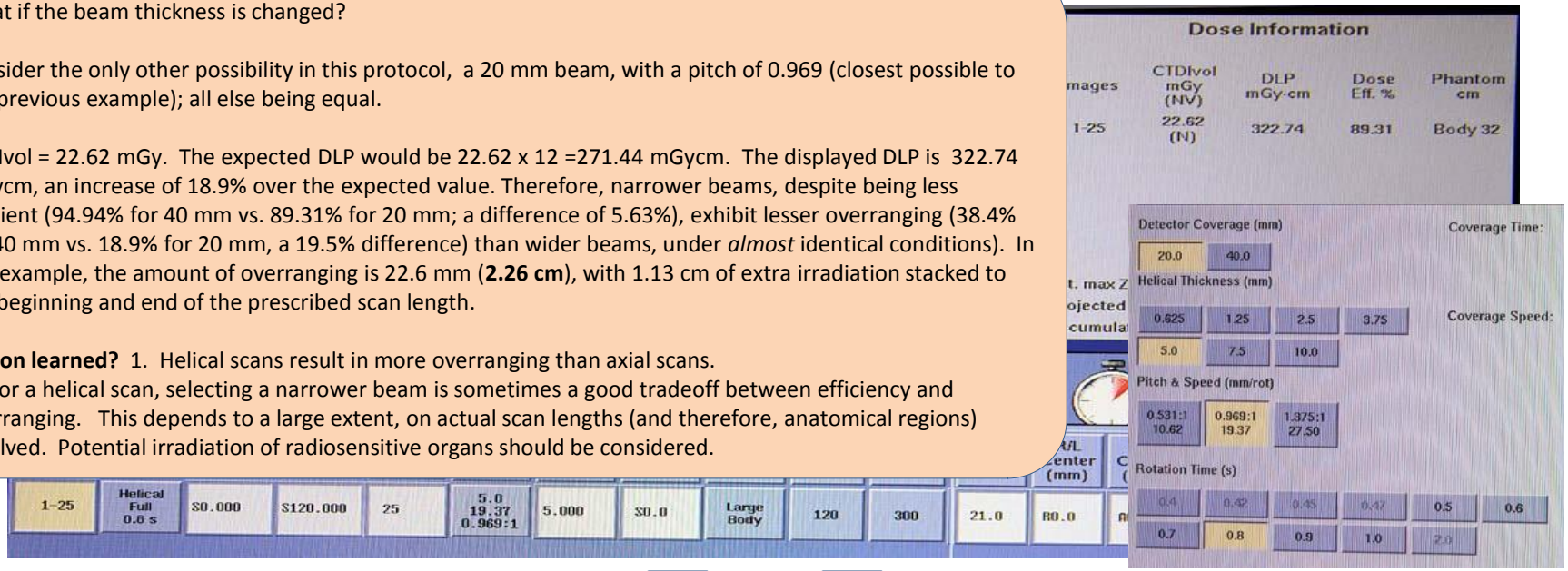


What if the beam thickness is changed?

Consider the only other possibility in this protocol, a 20 mm beam, with a pitch of 0.969 (closest possible to the previous example); all else being equal.

CTDIvol = 22.62 mGy. The expected DLP would be 22.62 x 12 =271.44 mGycm. The displayed DLP is 322.74 mGycm, an increase of 18.9% over the expected value. Therefore, narrower beams, despite being less efficient (94.94% for 40 mm vs. 89.31% for 20 mm; a difference of 5.63%), exhibit lesser overranging (38.4% for 40 mm vs. 18.9% for 20 mm, a 19.5% difference) than wider beams, under *almost* identical conditions). In this example, the amount of overranging is 22.6 mm (**2.26 cm**), with 1.13 cm of extra irradiation stacked to the beginning and end of the prescribed scan length.

Lesson learned? 1. Helical scans result in more overranging than axial scans.
2. For a helical scan, selecting a narrower beam is sometimes a good tradeoff between efficiency and overranging. This depends to a large extent, on actual scan lengths (and therefore, anatomical regions) involved. Potential irradiation of radiosensitive organs should be considered.



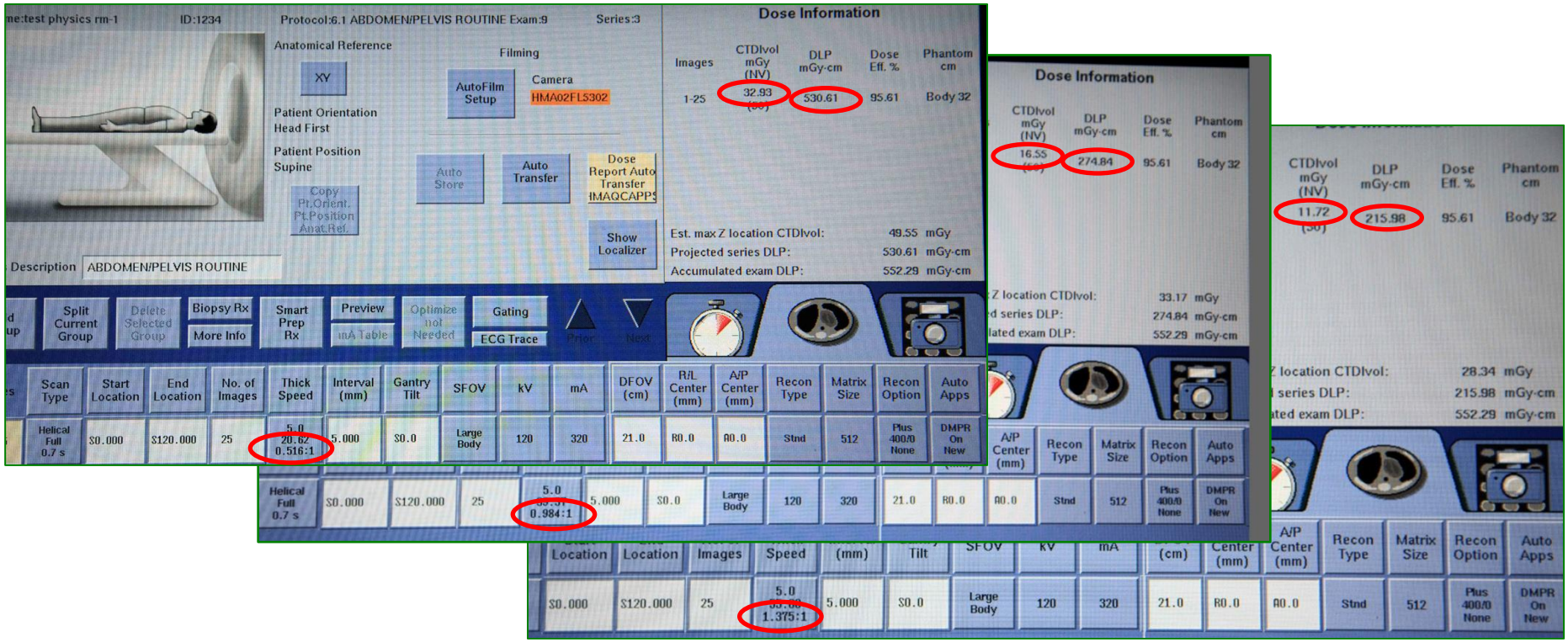
Overranging cont'd

Pitch vs. Overranging

Does **pitch** effect overranging?
Consider three helical exposures at 120 kV, 320 mA, 0.7 sec, 224 mAs, 40 mm beam thickness, 5 mm image thickness, 120 mm prescribed scan length, at the three available pitch options of 0.516, 0.984 and 1.375. Here is the analysis, following the same methods as in the previous pages:

Pitch	displayed CTDIvol (mGy)	Displayed DLP (mGycm) for the **12 cm** prescribed scan	Actual scan length (cm) (DLP ÷ CTDIvol)	Increased scan length due to overranging cm	Percentage
0.516	32.93	530.61	16.11	4.11 cm	34%
0.984	16.55	274.84	16.61	4.61 cm	38%
1.375	11.72	215.98	18.42	6.43 cm	53%

Lesson learned? High pitch values cause high overranging, making it necessary to be doubly careful in setting scan boundaries.



Overranging cont'd

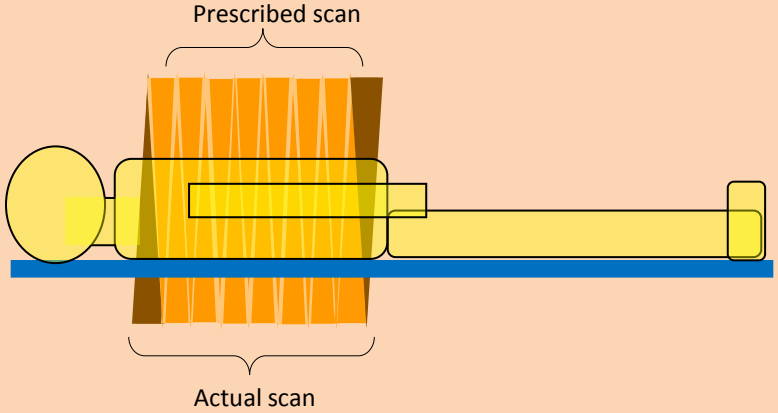
Scan length vs. Overranging

Finally, does **scan length** have an effect on overranging?
Consider helical exposures identical to those in the previous page, for different prescribed scan lengths, at pitch of 0.984

Prescribed Scan length (mm)	Actual scan length (mm)	Increased scan length due to overranging Percentage
30	76.1	153%
60	10.6	76%
120	16.6	38%
180	226.0	25%
220	266.1	21%

Since the amount of overranging is fixed for a given beam thickness and pitch, smaller scan lengths will have proportionately much higher overranging than longer scan lengths.

It also follows that one longer segment of a helical scan will result in lesser Effective Dose than two or more contiguous shorter scan segments.

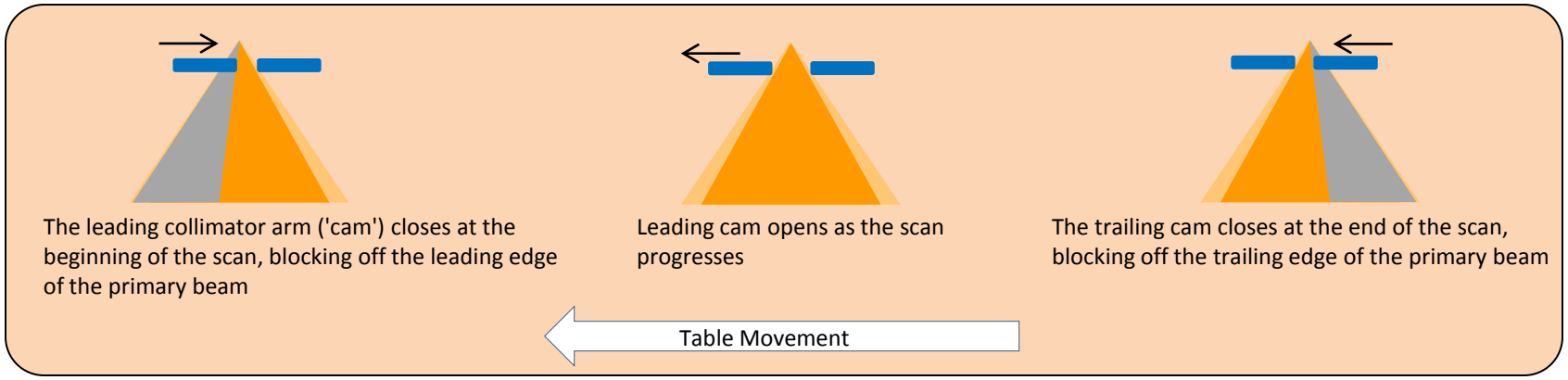


To summarize:
Degree of overranging <u>increases</u> with

- helical scans (vs. axial scans)
- wider beam thicknesses
- higher pitch
- shorter scan lengths
- fractionated scans

Overranging cont'd

Technologies. Manufacturers are beginning to address overranging in their widest-beam models. Generally, the method consists of collimators that open asymmetrically during the helical scan.



GE offers **dynamic z-axis tracking**, where collimators partially close at the beginning and end of the scan. **SmartTrack** is additional technology that constantly adjusts collimator openings during each tube rotation, in response to opposing forces such as gravity and centrifugal force.

Philips offers similar technology on its Brilliance iCT 256 slice scanner, which has a maximum beam thickness of 80 mm. Philips calls this technology the **eclipse collimator**.

Toshiba calls its technology **active collimator**.

Siemens calls its method the **adaptive dose shield**, which operates along the same lines as above.

These options available only on a few, select scanners.

Overscanning

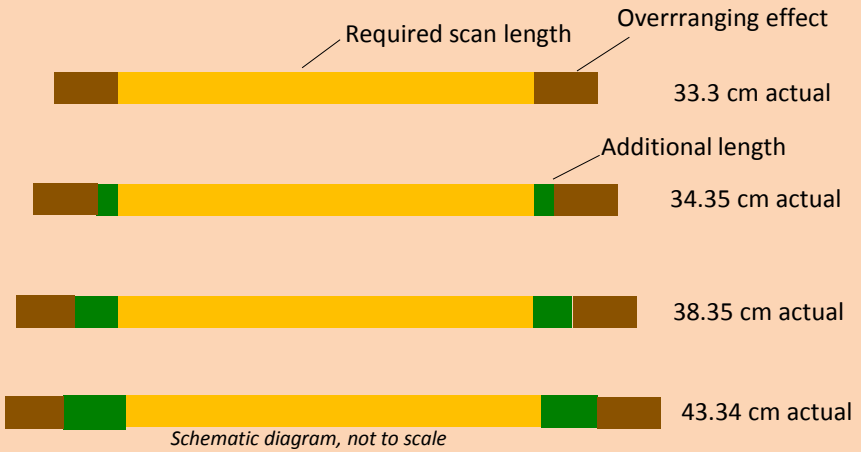
Although the terms overrranging and overscanning are used interchangeably in the literature, they are being treated as different in this tutorial. Overscanning is being considered as the practice of setting scan boundaries longer than prescribed, or the practice of prescribing longer scan lengths than diagnostically necessary.

Similar to overranging, the setting scan ranges longer than necessary increases patient effective dose, besides increasing the possibility of exposing radiosensitive organs to excess primary or scatter radiation.

In combination with overranging, effective doses can be 10-25% higher than intended, if care is not taken to strictly limit scan boundaries to even within 5 cm .

Example: Consider a GE LightSpeed 16. Scan parameters for routine abdomen scan at typical 'best practice' settings using the widest (16x 1.25 = 20 mm) beam, pitch of 1.375, 120 kV, 350 mA, 0.8 sec, 280 mAs, CTDI vol = 18.84 mGy (well under ACR reference level), for a required scan length of 30 cm.

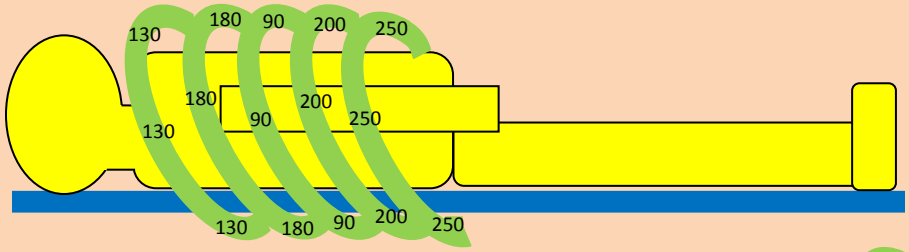
	Effective dose
a. Scan length set properly to 30 cm	9.4 mSv
b. 0.5 cm tacked on to both ends	9.7 mSv (**3% higher**)
c. 2.5 cm tacked on to both ends	10.8 mSv (**15% higher**)
d. 5 cm tacked on to both ends	12.3 mSv (**31% higher**)



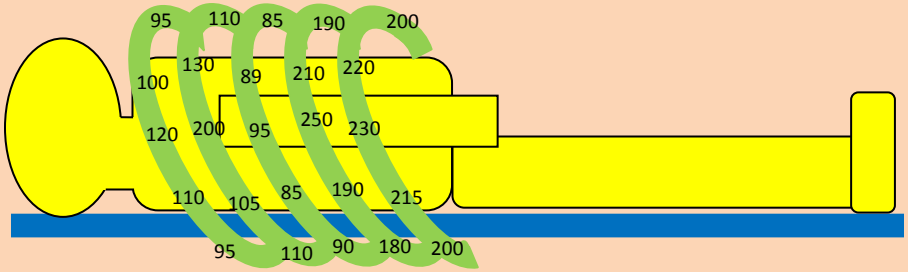
mA Modulation or AEC

All CT manufacturers now have some form of mA modulation technology (also known as Automatic Exposure Compensation, AEC) built into their scanners. mA modulation takes into account the individual patient's body habitus from scout scans (typically an AP or PA scout, and a lateral scout). Based on the size, shape and density of different body parts, mA is constantly changed from rotation to rotation, and within a rotation, during the main (image acquisition) scan. Dose from the scout scan is negligible, because it is done at very low mA (typically 30-50 mA)

Schematic diagrams (purely illustrative, the numbers are all made up) showing the difference between **Longitudinal** and **Angular** modulation, where mA changes depending on the thickness and composition of the anatomy scanned. Different manufactures use different terminologies to describe these modes. Longitudinal modulations requires one scout image, while angular modulation may require two scout scans (some systems do both using just one scout).



Longitudinal modulation, where each slice may have a different mA, but mA during each rotation is the same. Also called **z-axis modulation**.



Angular modulation, where mA changes constantly change during each rotation (and between rotations). Also called **x-y axis modulation**.

mA Modulation cont'd

Properly positioning the patient at isocenter is critical for mA modulation algorithms to work as intended. Being off centered by even a few cm can cause inordinately high doses, accompanied by sub-par image quality.

Phantom studies have shown that being off by even 10 cm above or below the isocenter can cause an increase of 40% or more in $CTDI_{vol}$.

Ensure that the coronal lasers on either side of the gantry overlap with each other. Mismatched laser lines are a clear indication that one, or both lasers are off, and neither can be relied upon to indicate isocenter height.

GE uses **Noise Index** (**NI**) as the parameter that decides the amount of mA modulation that will be applied. NI is a value the user selects, as the minimum level of noise (graininess) that can be tolerated at the center of the image. Recall that noise is the SD of a ROI. *The higher the Noise Index is set, greater is the tolerated noise, resulting in reduced mA, and therefore, reduced dose*. NI has to be increased in steps of at least 15-20% to see noticeable dose reduction. After the NI is set, if **Auto mA** is activated, the scanner will alter the mA from slice to slice, based on the patient's profile (i.e., each tube rotation will have the same mA). In addition, if **Smart mA** is selected, mA will vary during each tube rotation as well. Auto mA and Smart mA can be selected singly, or together.

Selecting a lower kV station while auto mA is active will not make a difference, because mA will increase proportionately to maintain the selected NI. The system will also adjust mA to account for various settings or pitch, to maintain a constant noise level

Parameters in Recon 1 are used for Auto mA calculations. If thicker slices are used for axial viewing and thinner slices are used for multiplanar reconstructions (in later Recons), set the thickest slices in Recon 1. As a general rule, image thickness is very important to noise: thinner the slices, noisier the images.

mA Modulation cont'd

Philips calls its mA modulation technology **Dose Right**, which has three components. (a) With **Automatic Current Selection** (**ACS**), the system has a stored set of pre-determined 'reference' mAs and noise values for different scans, based on a theoretical patient diameter. These reference values can be fine tuned manually. After a scout (Sureview) scan, the system determines the patient's maximum diameter. Based on the stored reference values and the patient's maximum diameter, the system sets *the maximum* mAs to be used in the scan, such that a constant noise level is achieved. The average and minimum mAs predicted for the scan are also displayed; the minimum is set to approx 40% of the maximum.

(b) Next, **Z- Axis Dose Modulation** (**Z-DOM**) uses the profile of the patient from the scout image to modulate mA along the longitudinal axis, with the maximum mA set by the ACS or manually. Each tube rotation can have a different mA. (c) If **Dynamic Angular Dose Modulation** (**D-DOM**) is selected, the system modulates mAs during each tube rotation. D-DOM applies its modulation only if at least 10% mAs reduction can be achieved. D-Dom is not recommended for head scans (because of symmetry)

ACS can be applied simultaneously with Z-DOM or D-DOM, but not both. Either Z-DOM or D-DOM (but not both together) can also be applied independently, without ACS

Toshiba uses a technology called **SURE Exposure**. mA is modulated both longitudinally and angularly, based on single or dual scout scans (scanograms), maintaining noise at a level that is set by the user. This level is selected from three presets: high quality, standard and low dose, each with its own mA thresholds and ceilings, which themselves are adjustable. Additional presets, such as ultra low dose, can be generated. **Boost 3D** is a data processing method that helps eliminate streak artifacts in images caused by highly attenuating objects such as bones in the shoulder and pelvis. **Quantum Denoising Software** (**QDS**) is designed to maintain high contrast resolution by edge detection, smoothening and edge enhancement.

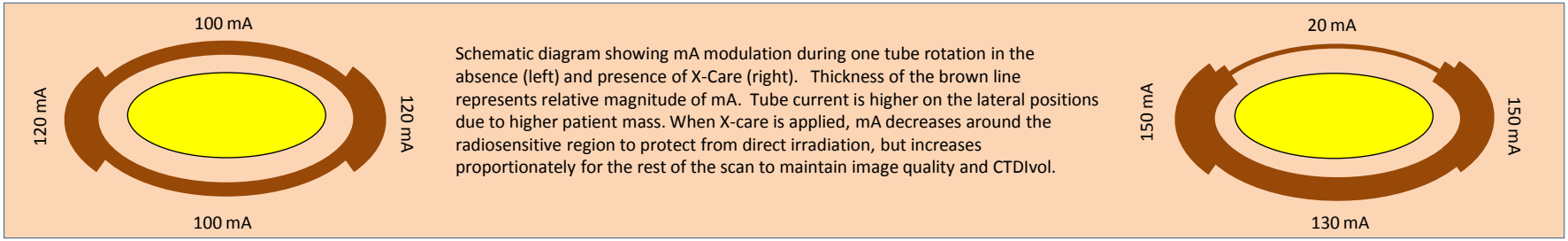
mA Modulation cont'd

Siemens uses a system called **CARE Dose4D** . The system has stored characteristics for a reference adult (75 kg) and pediatric (20 kg) patient. Initially, the user sets an mAs modulation strength (average, weak or strong) for both slim and obese patients or regions; these settings apply universally to all scans on the scanner. Siemens systems can work with one scout scan (Topogram), either in the AP or lateral direction; the other scout scan results are calculated.

If Care Dose4D is activated, an image **quality reference mAs (QRM)** is used to calculate the effective mAs for the entire scan, based on scout scan calculations and the stored modulation strength. Modulations takes place simultaneously between successive tube rotations and during a rotation; and take into account the pitch selected. Longitudinal modulation takes into account information from the scout scan. Angular modulation (which Siemens calls real-time angular dose modulation) measures the body's attenuation continuously during each tube rotation, and adjusts mA accordingly (hence, the requirement of just one scout scan). QRMs are input in units mAs_{eff}, but are a measure of the noise levels to be achieved. They must not be related to patient size.

CARE kV is another feature available on some Siemens models, where a suitable kV setting is selected by the system in addition to mA; the selected kV (70 kV being the lowest available) is used throughout the scan, so as to maintain a constant contrast to noise ratio. This feature is beneficial in exams that use contrast agents, as well as pediatric exams.

X-CARE is Siemens' organ-based dose modulation, where the mA is reduced when the x-ray tube is over the eyes or the female breast. In this method, tube current is reduced 80% over an 80° arc over the patient. To compensate and maintain the came $CTDI_{vol}$, there is a corresponding increase in mA at the underside of the patient. Thus, direct irradiation of the organs is avoided.



Schematic diagram showing mA modulation during one tube rotation in the absence (left) and presence of X-Care (right). Thickness of the brown line represents relative magnitude of mA. Tube current is higher on the lateral positions due to higher patient mass. When X-care is applied, mA decreases around the radiosensitive region to protect from direct irradiation, but increases proportionately for the rest of the scan to maintain image quality and CTDIvol.

Noise Reduction

In addition to mA modulation, manufacturers also have some type of noise reduction method built into their scanners (or sold as a separate package). The earliest scanners used algebraic reconstruction, which was replaced by filtered back projection. The newer methods involve some form of iterative reconstruction.

The basis of iterative reconstruction is to construct an 'ideal' (noiseless) image from the data acquired from a scan, compare it with the actual image from the same scan, and 'improve' the actual image over numerous cycles so that it matches the ideal image to the extent possible, with reduced noise.

GE uses **Adaptive Statistical Iterative Reconstruction** (**ASiR**). It allows the user to improve images that would have been unacceptably noisy. It also allows the user to supplement the work done by the noise index (by allowing higher NI values to be set), and thereby reduce dose. However, ASiR does not automatically cause mA reduction unless there is specific user input. For manual mA protocols (i.e., when mA modulation is not used), GE provides a table that relates ASiR to Noise Index values. ASiR is generally applied as a percentage level (or strength); around 40% seems fairly typical.

GE Scanners with Dose Reduction Guidance (**DRG**) feature have a more advanced method of entering ASiR levels, for both mA-modulated and non-modulated protocols. This is done by entering a percentage dose reduction desired, and automatically sets NI and ASiR %.

Veo is a new product by GE, which implements **Model Based Iterative Reconstruction** (**MBIR**). This is currently available as an option of the Discovery CT750HD model, and promises more noise and dose reduction than ASiR.

Noise Reduction cont'd

Philips uses an iterative reconstruction package called **iDose^4**. Different strengths (levels) of noise reduction can be set on a scale of 1-7, and iDose can be selected before the scan (prospectively), or during image reconstruction (retrospectively) if the raw scan data is available on the scanner.

Toshiba's noise reduction technology is called **Adaptive Iterative Dose Reduction 3D** (**AIDR 3D**). It works both on raw data and on reconstructed images, to produce final image that is low in noise, but preserves spatial resolution.

Siemens originally offered a product named **IRIS** (**Iterative Reconstruction in Image Space**) as its noise reduction solution. This involved the creation of a 'master volume image' that is cleaned up in 3-5 iterative steps, each step achieving greater noise reduction.

The newer offering from Siemens is **SAFIRE** (**Sinogram-Affirmed Iterative Reconstruction**), where the reconstruction starts from the raw image and scout scan image data rather from a 'master volume image'. Safire strength can be set from 1-5 by the user, a setting of 3 is fairly typical.

General methodology for use of noise reduction options:

1. Review image quality in the absence of the noise reduction (NR) features for selected clinical images
2. Perform additional reconstructions with varying strengths of ASIR, Safire, etc.
3. Aggressive settings will cause the images to look artificially smooth and 'plasticky', and may introduce new artifacts.
4. Select NR settings based on radiologist's preferences

There is no 'one size fits all' solution. This has to be a collaborative effort between the technologists and radiologists, with a lot of back-and-forth

ACR Action Levels

ACR specifies Reference Levels (RLs) and pass/fail (P/F) criteria based on **$CTDI_{vol}$**, for a particular weight/age class of patients, for a few selected protocols (selected presumably because the are the most commonly performed scans nationwide). These $CTDI_{vol}$ criteria serve to set the *upper boundary* of acceptable technique. What about the *lower boundary* of acceptable dose?

ACR criteria on **CNR** determine *how low you can go* with technique factors, before image quality becomes unacceptably poor. Most scanners have plenty lf leeway between these lower and upper boundaries

Maximum **$CTDI_{vol}$** allowed	Adult Head	Pediatric Head	Adult Abdomen	Pediatric Abdomen
Reference Level (mGy)	75	35	25	15
Pass/Fail Criteria (mGy)	80	40	30	20
Minimum **CNR** required	1.0	0.7	1.0	0.4

Each of these are for *routine* protocols, not specialized exams. An adult protocol refers to one typically used for a 20 year old, or a 154 lbs (70 kg) patient. The 70 kg is the international 'reference man' standard, and may not represent your typical patient. The pediatric abdomen protocol is to be typical of a 5 year old, or 40 lbs (18 kg) patient; who may also not be the typical pediatric patient you see at your facility. The pediatric head protocol is to be one typically used on a 12 month old. The adult abdomen $CTDI_{vol}$ is specified for the 32 cm phantom; the other three, for the 16 cm phantom.

It is strongly recommended that the protocols result in $CTDI_{vol}$ below reference levels for the above weight/age class of patients. If the scanner meets these criteria, it logically follows that larger patients will trigger higher $CTDI_{vol}$, and thinner patients will trigger trigger lower $CTDI_{vol}$ values.

The minimum CNR requirements for pediatric scans are lower than that of adults, reflecting their lower dose allowances, and setting the expectation that these scans can tolerate a certain amount of noise.

AAPM / NEMA Action Levels

The National Electrical Manufacturer's Association (NEMA), in collaboration with the American Association of Physicists in Medicine (AAPM), have recommended action levels for protocols that are more encompassing than ACR criteria. Two levels of criteria are specified as part of the NEMA XR-25 requirement. On scanners equipped with DoseCheck software, these criteria can be programmed in.

The first level is called a **Notification Value** (**NV**), which triggers a pop-up window notifying the technologist that the criteria has been exceeded; they can override the NV and continue the scan after specifying a valid reason. NVs apply to all protocols (unlike ACR criteria which are only for four routine protocols).

The second level is called the **Alert Value** (**AV**) which triggers a different pop-up window, holds up the exam in a password-protected lock, and requires the intervention of a supervisor (or assigned lead technologist) via a password. AVs apply for all individual parts in a study combined (example, all phases of a multiphase study), or for all exposures on a patient during a single exam.

	Adult Head	Pediatric Head	Adult Torso	Pediatric Torso
Suggested Notification Values (mGy)	80	50 for < 2 yr old	50	25 for 16 cm phantom
		60 for 2-5 yr old		10 for 32 cm phantom
	600 for brain perfusion		150 for retrospectively gated cardiac CT	
			50 for prospectively gated cardiac CT	
Suggested Alert Value (mGy): 1000 mGy for all scans				

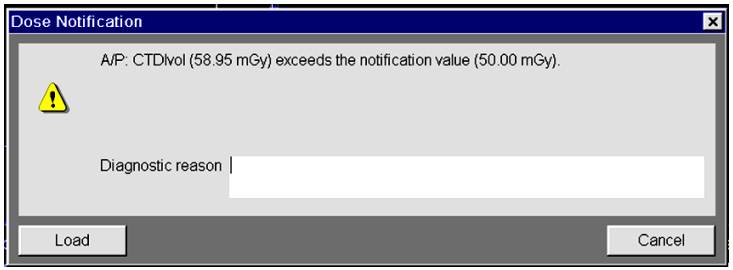
software logo

The purpose of the DoseCheck initiative is to act as a gatekeeper and prevent major scanning mistakes from happening. NVs are not set in stone; they are merely *starting points* that can be fine tuned over time for different protocols, once enough information exists about the range of $CTDI_{vol}$ triggered by your general patient population. If your patient population is generally overweight or obese, more liberal NVs can be set. Conversely, if you cater to a skinny patient population, NVs can be toned down.

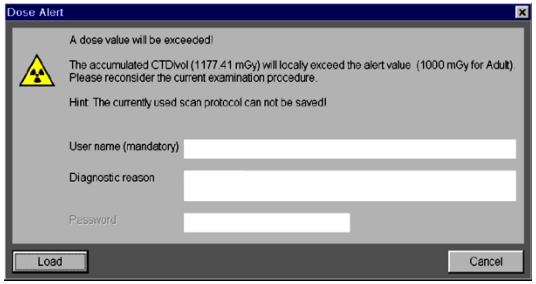
For scanners that do not have the software installed, NVs and AVs can still be implemented, by manually keeping an eye on post-scout $CTDI_{vol}$ values, prior to the actual scan, by the technologist. Note, however, that Medicare penalties will apply for scanners without the DoseCheck software, starting 1 January 2016.

NV and AV examples

NV

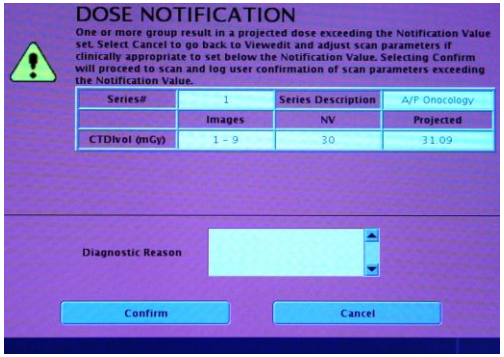


Siemens

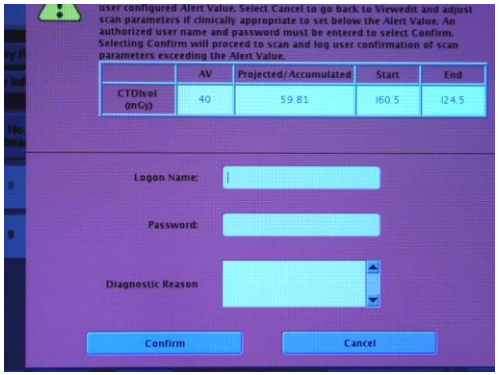


AV

NV

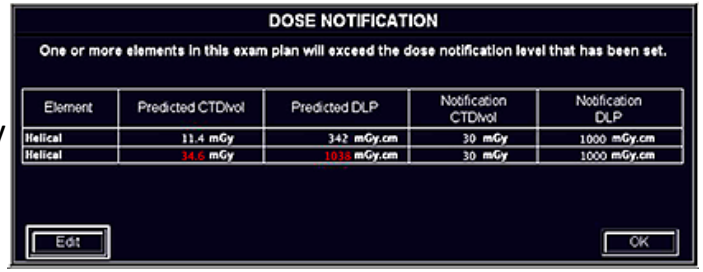


GE

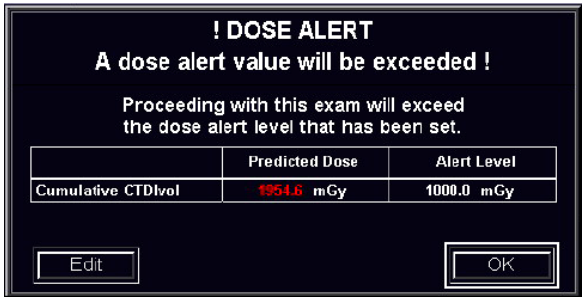


AV

NV



Toshiba



AV

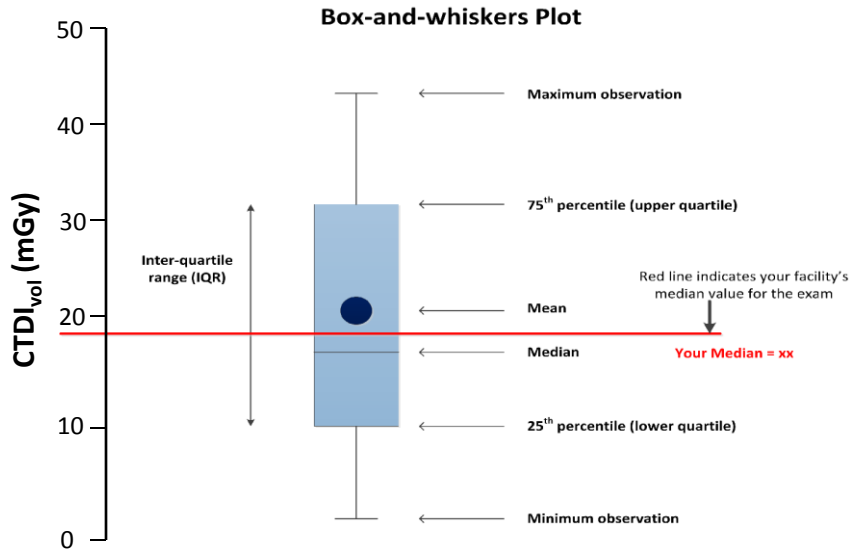
ACR: NRDR-DIR

The American College of Radiology (ACR) maintains the National Radiology Data Registry (NRDR), which has a number of components. One of these is the Dose index Registry (DIR), which is a repository of information on CT dose indices.

Participant facilities in the DIR are provided with software that collects CT Dose Index data for a variety of scans, both adult and pediatric (broken down by age), and forwards them to the DIR.

The DIR complies all this information, and provides a comprehensive report to the facility, containing comparisons of the facility to (a) similar sites in the community, (b) sites across the division (in our case, New England); (c) sites in the metropolitan area, as well as (d) all facilities registered with the DIR. Comparisons are made of $CTDI_{vol}$, DLP and SSDE, and reports are provided every 6 months, typically in January and July.

These reports can be helpful in checking how your facility fares in comparison with others with similar patient populations, and offers insights into how you can better manage your protocols.



Structure of DIR plotted data:

The blue box provides dose index from the 25th to the 75th percentile of the data in that category.
The whiskers provide the range of values (min and max).
Mean of all the data is shown with the blue dot, and median is indicated with the line across the box.
The red line shows where the median data of your facility lies. Each graph contains four such box and whiskers plots, one for each category (a-d) identified above. For each protocol, three such graphs are provided: $CTDI_{vol}$ (example shown here), DLP and SSDE.

CT Protocol Review

MA DPH Radiation Control Program (RCP) regulations require a **quarterly** review of protocols on each CT scanner. These reviews must be documented in writing, and a record of the reviews must be maintained for regulatory review (some facilities make it part of the quarterly RSC meeting minutes). The Joint Commission has also announced the requirement for such reviews, commencing on 1 July 2015.

The reviews should be conducted by the lead CT / QC technologist (or his /her designee). During the review, compare all programmed technique parameters against the master protocol manual:

- kV
- mA [for non modulated scans]
- settings of mA modulation
- tube rotation time
- Detector configuration
- Pitch
- image thickness – primary acquisition and all subsequent reconstructions
- reconstruction algorithms - primary acquisition and all subsequent reconstructions
- noise reduction settings (ASiR / SAFIRE / iDose / AIDR strength)
- Displayed CTDIvol for default protocol (as soon as the protocol is activated)

The purpose of the review is to ensure none of the settings have been corrupted: this is a more common occurrence than you realize. Switching of kVs and resetting the reconstruction algorithm are fairly frequent. If the facility has multiple scanners of the same make and model, it helps to ensure that all protocols are set identical (again, several differences are typically discovered).

The review process should convince you to get rid of protocols that are seldom / never / no longer used. This greatly reduces clutter in the protocol screen, and prevents the chances of inappropriate protocols being accidentally selected.

Some scanners may have password protection to ensure that nobody except the lead technologist (or other appropriate individual) can make permanent changes to protocols. Make sure that the password protection works. Such password protection of protocols is NOT mandated by any regulation *(do not confuse this password with the password set for Dose Check Alert Value)*. In the absence of lockout software, administrative controls on the accessibility of the protocols is acceptable.

Low Dose Lung Cancer Screening

The ACR has published guidelines for lung cancer screening CT scans, and offers accreditation for the protocols in conjunction with chest module accreditation.

The **pass/fail criteria** is a $CTDI_{vol}$ of **3.0 mGy,** for an average patient (5'7", 70 kg, 154 lbs). mA modulation is required; if not, the technique must be manually decreased for thinner patients, and increased for larger patients (mA modulation is necessary starting January 2016, or penalties apply. See Medicare requirements, next page).

The scanner should be multi-detector, running a helical protocol. Patient age should be 55-80, and patients should have a smoking history of 30 pack-years (e.g., 1 pack a day for 30 yrs, 2 packs a day for 15 years, 3 packs a day for 7.5 years, etc.). Former smokers should have stopped smoking in the past 15 years; screening should be discontinued if smoking has ceased for 15 years, of if the patient develops a health problem that limits life expectancy substantially, or limits the willingness to have curative lung surgery.

A mechanism must be in place to refer patients to smoking cessation counseling, or to provide smoking cessation materials.

The interpreting physician must have read at least 200 chest CT cases in the prior 36 months.

Massachusetts law requires any screening program to be registered and approved by the DPH Radiation Control Program prior to operation. In this context, screening is considered as a setup where a patient comes in for the scan without a physician's prescription (self-referral). If a physician prescribes the scan, it is not considered screening, and no prior DPH approval is necessary.

For self-referrals, a mechanism must be present to refer the patient to a health care provider if abnormal findings are present.

Upcoming Medicare Requirements

NEMA XR-29-2013 standard (also known as **MITA SmartDose Standard**) goes into effect on **1 January 2016.**

Equipment not in compliance by Jan. 1, 2016, will be subject to a five percent /per scan technical component reduction on diagnostic CT procedures billed in physician office and hospital outpatient settings. The reduction will increase to 15 percent Jan. 1, 2017.

If a facility bills for both inpatient and outpatient CT scans on the same scanner, the reduction only applies to those scans billed as outpatient procedures. The MITA SmartDose policy does not affect scans billed under the hospital inpatient setting or for interventional radiology procedures.

Source: ACR website

NEMA XR-29 requirements

- DICOM Radiation Dose Structured Reporting (RDSR) → Enables dose information ($CTDI_{vol}$ and DLP) to be captured in a standardized electronic format that can be included in the patient's medical record ***Most Scanners are already compliant***
- NEMA XR-25 Dose Check
  - Notification Values
  - Alert Values – with password requirement

  Many Scanners are not compliant. Check with manufacturers if upgrades are possible *
- Automatic Exposure Control (AEC) → mA modulation, or a method to set mA based on patient habitus Described in more detail in NEMA XR-28-2013 ***Most Scanners are already compliant***
- Reference adult and pediatric protocols → Pre-loaded protocols for adults and children ***Most Scanners are already compliant***

Existing **Neurologica CereTom scanners will be upgraded and brought into compliance with DoseCheck and AEC before the end of 2015*

All NEMA standards (XR-25, XR-28 and XR-29) can be downloaded for free from the NEMA website.

CT Fluoroscopy

CT-F involves 'live' imaging from tableside, for biopsies and other interventional procedures.

Two modes of imaging are possible (one, or both may be available on a suitably equipped scanner):

Single axial scans – each step on the pedal results in one discrete exposure, generally providing 3 images. Risk for increased patient dose is low, because the exposure terminates until the next pedal push. Dose indices displayed is $CTDI_{vol}$ (mGy/exposure) and DLP (mGycm, cumulative).

Continuous imaging – similar to a regular fluoroscopy system. The primary beam is on as long as the pedal is depressed. Potentially much more patient dose, if the operator is not careful to take the foot off the pedal. Dose index displayed is CTDI rate, as mGy/sec, as well as cumulative DLP.

Generally performed at 120 kV, but at much lower mA than a diagnostic scan (combined with tube rotation speed, to give typically 30-50 mAs).

GE: the single axial mode is called **Smart Step**, while the continuous mode is called **Smart View**. Only one or the other may be available on a scanner. Typical radiation beam thickness is 10 mm, providing 3 images of 5 mm *(how? Two consecutive 5 mm images, and one overlapping image spanning the two. The amount of offset of the overlapping image can be specified as 'bump distance').* In Smart Step mode, the total number of exposures may be capped at a certain level, such as 90. The Smart View mode on a GE OptiomaCT660 is <u>not</u> programmed to provide CTDI rate (this is a potential shortcoming).

Philips: In most cases, the single axial mode is used. Beam thickness is typically 12 mm, providing 3 images of 4 mm each. **Hand Care** is generally programmed in (and not selectable by the user) – this feature turns off the beam for a certain length when the x-ray tube is above the patient; designed to avoid direct irradiation to the user' s hands. The concept is similar to Siemens' X-Care.

CT Fluoroscopy cont'd

Toshiba: Only a continuous beam is available. Typically performed at 12 mm, providing 3 images of 4 mm each.

Siemens: A continuous mode is typically provided. Hand Care is selectable by the user, and the arc along which the beam is turned off can be modified – to the top, left, or right.

Elevated doses to internal organs are a potential risk from CT Fluoroscopy. Effective doses from a typical procedure are generally a fraction of the dose from a diagnostic scan, because of the lower mA used. However, the entire dose to the body is CTF is concentrated within a narrow band of the body's cross section.

Suppose a continuous beam CTF results in a CTDI rate of 6.5 mGy/min (a fairly typical value at 120 kV, 50 mAs, 10 mm beam thickness).

Stepping on the pedal for a mere 5 seconds will result in 32.5 mGy, exceeding the ACR pass/fail adult abdomen standard.

At 8 seconds of beam on time, the NEMA recommended adult torso NV of 50 mGy will be crossed.

Similarly, for single axial scans, around 10 exposures will cause the ACR or NEMA levels to be breached. None of the displayed dose indices will indicate a cause for concern. On the other hand, radiation beam-on times will be dictated by the complexities of individual cases, and the clinical benefit from the procedure should be weighed against any potential risks from radiation exposure.

Skin burns from CTF procedures are feasible only after tens of minutes of beam on time. Exposure rates at the skin are typically in the range of 4-8 R/min, which is higher than typical dose rates using a C-arm. However, since the x-ray tube is in continuous rotation, there is no 100% dwell time on one patch of skin, thereby decreasing the risks of skin reactions.

CT Fluoroscopy cont'd

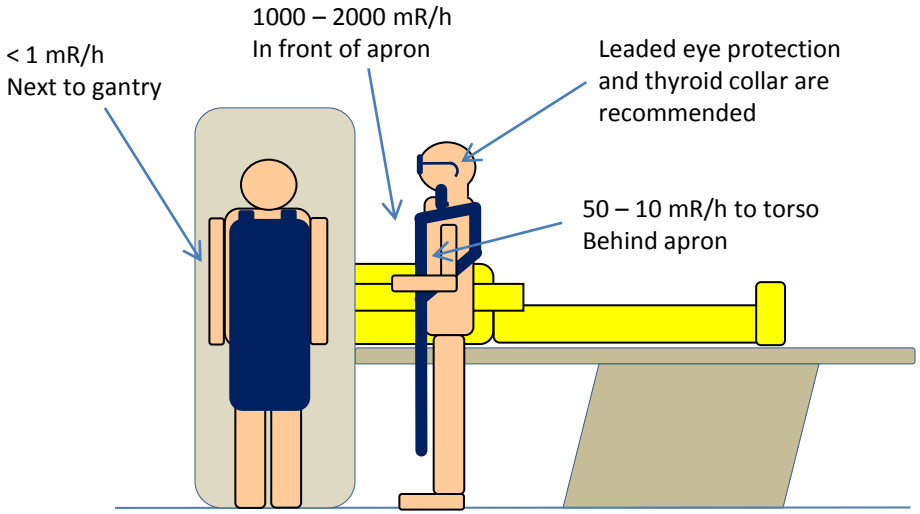
Scatter radiation rates experienced by tableside personnel during CTF are in the order of 1000 – 2000 mR/h. Unfortunately, this is the location in the CT room with the highest possible scatter radiation.

A typical 0.5 mm Pb-equivalent lead apron decreases scatter radiation by more than 95% . Therefore, aproned personnel would be expected to see no more than 50-100 mR/hr to their torsos.

Thyroid collars and leaded eye protection are strongly recommended for lengthy procedures.

The physician's hands are subject to excess extremity dose depending on his/her practice. Wearing a ring badge is recommended on a case by case basis.

Hands should be kept away from the primary beam to the extent possible. If the hands are in or near the primary beam, leaded gloves of no less than 0.5 mm lead equivalent are mandatory.



Only personnel whose presence is absolutely necessary in the room should be present when the beam is on. Scatter radiation immediately adjacent to the gantry are almost negligible, because of shielding from the gantry components. Essential personnel whose presence tableside is not required, should stand next to the gantry if possible, as long as it does not compromise their tasks. All personnel in the room must be lead-aproned, irrespective of where they stand. A portable leaded or leaded acrylic shield can provide additional head-to-toe protection.

Patient shielding

Aside from being exposed to the primary beam during a CT scan, the patient's body not included in the primary beam is exposed to scatter radiation. This takes two forms:

a. External scatter, impacting the skin and organs close to the surface ('remote scatter' for this tutorial) ; and
b. Internal scatter, from the irradiated organs to nearby non-irradiated organs and tissues

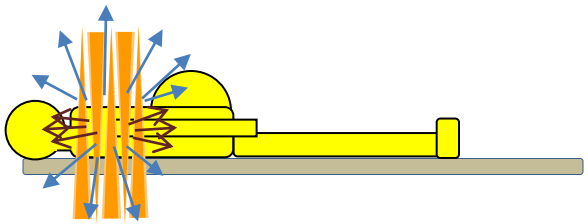
Shielding measures only protect from external scatter (out-of-plane shielding). In addition, some shielding products are meant to protect organs at the body surface from the primary beam itself (In-plane shielding).

Shielding to protect organs from 'remote scatter' is most effective when the irradiated region is close to the region to be protected. Examples:

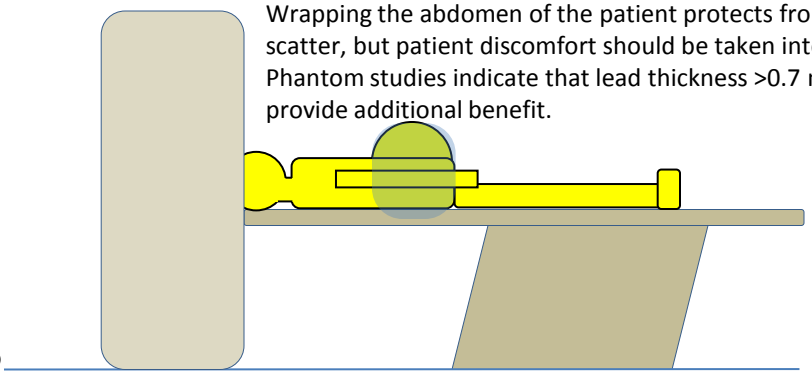
protecting the abdomen of a pregnant patient during a PE study

Protecting male reproductive organs during abdomen / pelvic scans extending close to the gonads, when the gonads do not need to be imaged

In the case of the pregnant patient, it is beneficial to wrap the abdomen in a lead apron, but carefully ensuring that the scan area does not extend into the region covered by the apron.



Schematic diagram showing external (⟶) and internal (⟶) scatter components potentially impacting the fetus. Out-of-plane shielding can only protect against the external scatter component. Setting the scan boundary to minimize overranging and overscanning are the only means to reduce internal scatter from reaching the fetus



Wrapping the abdomen of the patient protects from external scatter, but patient discomfort should be taken into account. Phantom studies indicate that lead thickness >0.7 mm do not provide additional benefit.

Patient shielding cont'd

Lightweight shields made of bismuth compounds are available for shielding of anterior radiosensitive organs (eye shields, breast shields and thyroid shields) in the primary beam. Their utility is very limited, and they can potentially cause more harm than good.

If they are used, they must be placed after the scout scan is done, otherwise the scout will consider the shield as part of the body, and increase mA (and therefore dose), thereby defeating the intention of shielding.

The above strategy is still defeated on systems and protocols that perform angular modulation using real-time data.

Options such as XCare on some Siemens scanners, which decrease the mA above the breast or eyes of the patient, essentially perform the same function of the bismuth shield, without any of the associated disadvantages.

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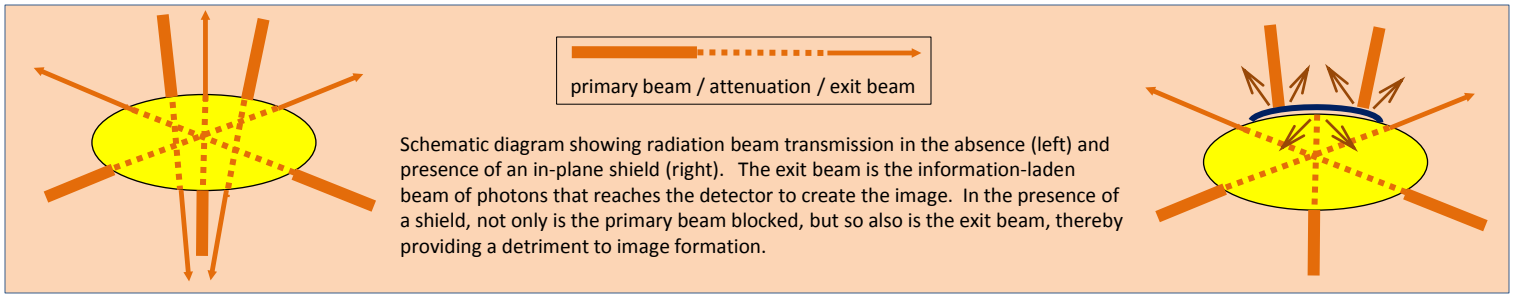
The shields always have the potential to introduce streak artifacts

While they offer some degree of protection from the primary beam when the tube is above the patient (by blocking the entrance beam), the shields act as filters when the tube is below the patient, blocking off the exit beam, thereby preventing the essential image forming photons from reaching the detector.



Schematic diagram showing radiation beam transmission in the absence (left) and presence of an in-plane shield (right). The exit beam is the information-laden beam of photons that reaches the detector to create the image. In the presence of a shield, not only is the primary beam blocked, but so also is the exit beam, thereby providing a detriment to image formation.

Carefully positioning the patient at isocenter, thereby enabling mA modulation to function as intended, and carefully setting scan boundaries to reduce overscannning and overbeaming are more than sufficient to minimize dose to all organs (not just the peripheral radiosensitive ones), without any of the pitfalls associated with bismuth shields.

Therefore, the overall recommendation is to stay away from primary beam (in-plane) shielding.

Image Gently ® and Image Wisely ®

These two campaigns are meant to create awareness of, and provide guidance to technologists, physicians and patients on best practices for CT scans on pediatric patients (www.imagegenty.org) and adult patients (www.imagewisely.org).

The Joint Commission (TJC) has put forth a requirement that starting **1 July 2015**, CT technologists must have annual training that includes "radiation dose optimization techniques and tools" addressed in these two campaigns.

This CT Tutorial covers the essence of the message contained in the Image Gently and Image Gently campaigns. In addition, the reader should consult the two websites for additional, in-depth coverage of any individual topic. The websites contain numerous downloadable articles, as well as links to associated websites and CT manufacturer's information. **Free CEUs** are available. Both sites also have sections dealing with fluoroscopy, interventional radiology and nuclear medicine.

For help with setting pediatric protocols, the Image Gently website has excel based calculators that can be downloaded, along with instructions. GE, Toshiba, Philips and Siemens provide manufacturer-specific instructions under the 'vendor specific modules' section.

Previous guidance (2008) on pediatric scan dose reduction by Image Gently focused on decreasing the mAs for different age and / weight groups, using the average adult mAs as the baseline. With the advent of SSDE, the current emphasis (2014) is on using pediatric patient AP and lateral diameter to consitute SSDE-based dose-reduction. A detailed description, together with spreadsheets for head and torso SSDE based- protocol setting can be found on the website: Under the 'procedures' tab, go to 'Computed Tomography' and click on the 'Pediatric CT Protocols & Instructions' link, on the box towards the right side

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Summary: Best Practice Guidelines

1. Check with your CT applications specialist about lower kV protocols approved for your scanner

2. Consult with your Radiologists about potentially using 100 or 80 kV for later phases of multi-phase iodine contrast studies. 80 kV will offer better iodine visualization, and can be used in bolus tracking

3. Using 100 kV for CT Angiography decreases patient doses without significantly compromising image quality, as long as the patients are not obese

4. For pediatric exams, consider cutting down the number of phases in multiphase exam (e.g., three phase instead of a four phase liver study)

5. Check for tube rotation times in excess of 1.5 seconds, especially for axial head protocols that are not mA modulated. These can easily be shortened to 1 sec or less, with the corresponding increase in mA (to maintain the same mAs). Short times will minimize the possibility of patient motion artifacts.

6. Ensure that widest radiation beams are used on protocols, except where clinically indicated. This increases geometric efficiency, minimizing overbeaming

7. Narrower radiation beams are preferred for short helical scan lengths, pediatric scans, and protocols with high pitch values. This represents a balance between overbeaming and overranging.

8. Narrow beam widths are preferred when narrow image thicknesses are required with considerable gap between successive images (e.g., high resolution chest axial scans).

9. Aggressive algorithms (such as Bone for GE scanners) are required for high resolution chest scans.

Summary: Best Practice Guidelines cont'd

10. Take time to evaluate the extent of overranging for your most commonly used protocols (already complete scans can be used to be most realistic), especially those extending close to radiosensitive organs (eyes, thyroid, gonads), shorter scan lengths, and especially for pediatric protocols. Evaluate the use of lower pitch values and narrower beam widths, while balancing against $CTDI_{vol}$.

11. The ACR recommends using pitch values higher than 1 for pediatric scans. However, this should be judiciously weighed against overranging, and corresponding decreases in mA (and kV, if applicable).

12. A single long helical scan delivers lower Effective Dose than fractionating scan lengths.

13. When excessive image noise is an issue, try fine tuning the algorithm if possible. Second option: consider increased image thickness. Neither come at the cost of increased dose.

14. Ensure that the left and right lasers are in alignment: if the lasers are turned on with no obstructions, they should appear as one overlapping line. Visualizing two lines indicates misalignment.

15. Take time to properly position the patient it isocenter, making use of the coronal lasers. If the scout indicates that the patient height is off-center, do not merely adjust the image on the scanner display; go back into the room and adjust table height. Performing a repeat scout costs negligible dose in comparison to scanning a patient off-center.

16. Delete old and unused protocols to reduce clutter on the protocol screen, and minimize the chances that they will be used.

17. It helps to list all master protocol parameters in a tabular form on paper; this makes protocol review easier, and can act as a checklist.

Summary: Best Practice Guidelines cont'd

18. While evaluating pediatric protocols, ensure that the reference phantom diameter is 16 cm. If a 32 cm diameter is used, double the displayed $CTDI_{vol}$.

19. If protocol-locking password system is not available, establish a chain of command administratively to determine who has the authorization to add, delete or alter protocols.

20. If the scanner is equipped with Dose Check software, ensure from applications personnel that it is activated, and reverence values entered.

21. Establish a chain of command for who has the password to override DoseCheck Alert Value. The individual (or his / her designee) who possesses the password must be available or reachable 24/7. All technologists (including part time, *per diem*, overnight and weekend techs) must be aware of this command structure.

22. For CT fluoroscopy, ensure that the mAs is in the 30-50 range. Update action times (provided in the annual physics report) on the fluoro log book, and promptly notify Masse Associates if the action time is exceeded during a case, so a skin dose estimate can be made.

23. Ensure that all personnel in the room during CT fluoroscopy wear lead aprons, and that the fluoroscopists are badged. If the wearer uses a single whole body badge or single collar badge, these must be worn at the neck level outside the apron. If the wearer is on a 2-bdage system, the waist badge must be worn under the apron at the waist level, and the collar badge outside the apron at neck level.

24. If the physician's hands are in the primary beam, heavy lead gloves (0.5 mm Pb equivalent) must be worn.

25. If off-the-plane patient shielding is used, ensure that it is kept away from the scan boundary, and is not part of the scout scan or main scan. If in-plane shielding is used (although it is recommend that you not use it), ensure that it is placed after the scout is done.

THE END