Radiation dose issues in multi-slice CT scanning

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Introduction

Multi-slice or multi-detector row CT scanners, capable of imaging four simultaneous, parallel slices in a single rotation, were first introduced in 1998. Since then, scanners with 6, 8, 10 and 16 slice capabilities have become available and have made a marked impact on the role of CT in the diagnostic radiology department. Most recently 32, 40 and 64 slice scanners have been announced and the trend for increased capabilities and hence extended applications of CT is certain to continue.

CT scanning has been recognised as a high radiation dose modality, when compared to other diagnostic X-ray techniques, since its launch into clinical practice more than 30 years ago. Over that time, as scanner technology has developed and its use has become more widespread, concerns over patient radiation doses from CT have grown [1, 2]. The introduction of multi-slice scanners has focussed further attention on this issue, and it is generally believed that it will lead to higher patient doses [3].

The amount of radiation dose a patient receives from a CT scan depends upon two key factors, the design of the scanner and also on the way that the scanner is used. The designs of single slice and multi-slice scanners are similar in most aspects that affect radiation dose, but multi-slice scanning can potentially result in higher radiation risk to the patient due to increased capabilities allowing long scan lengths at high tube currents.

The aim of this leaflet is to explain how the differences between single and multi-slice systems will impact on patient doses. The term 'dose' is used here in its broad context and, unless otherwise stated, generally refers to both absorbed dose and effective dose.

The leaflet also includes a description of the principles of automatic tube current control systems available on modern scanners. Although not exclusive to multi-slice, they are now an integral part of most of these systems, and their development has progressed significantly alongside them.

Factors affecting radiation dose

CT scanner design factors

The majority of scanner design features that affect dose and dose efficiency need not differ between single and multi-slice systems. Indeed, some manufacturers have a range of systems from single to 16-slice which are identical in terms of most of the features listed in Table 1. The exception is that the single bank of detectors of a single slice scanner is replaced by multiple detector banks along the z-axis (Figure 1). It is this factor which primarily causes differences in dose efficiency between single and multi-slice scanners and will be discussed in a further section.

Table 1: Principal scanner hardware design features affecting radiation dose and dose efficiency

Effects on dose	Effects on dose efficiency
X-ray tube filtration	Detector material
X-ray beam shaping filters	Number, width and spacing of detectors
Collimator design	(in scan plane and along z-axis)
Focus to axis distance	

Figure 1: Schematic diagram showing single slice and multi-slice detector layout (not to scale)
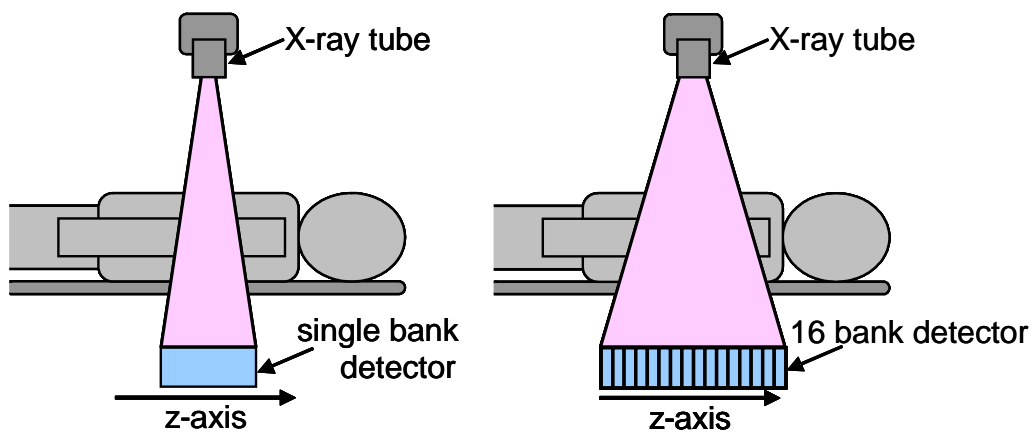
Scanner use

User selected scanning parameters, such as kV, mAs and scan length, will largely determine the radiation dose to the patient. Variation in the scan parameters used from site to site results in dose differences much greater than those due to scanner design factors. The increased capabilities of multi-slice scanners, which allow higher mAs values, longer scan lengths and multi-phase contrast studies, have the potential of directly increasing patient doses. Another indirect but significant effect on dose can result from the imaged slice width. Scanning is usually performed with narrower slices than on single slice scanners, so for the same noise, higher mAs values would need to be used. Furthermore, multi-slice scanners introduce new applications previously not possible in CT. The most notable of these is in the field of cardiology where high doses can result from the use of low pitch values in some applications.
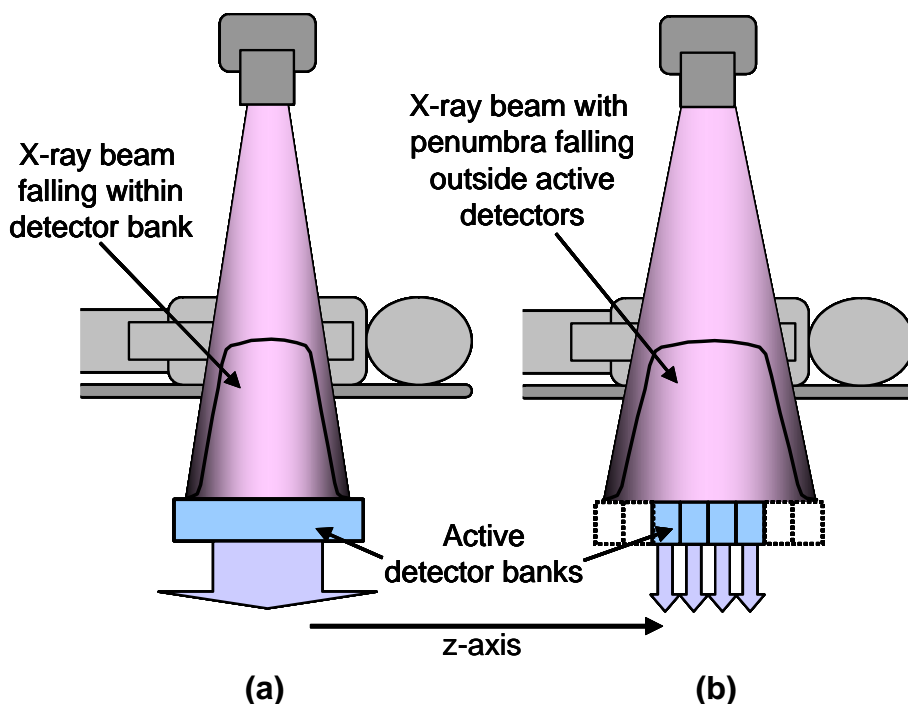
Geometric efficiency

One of the main dosimetric differences between single and multi-slice scanners is in geometric efficiency. The geometric efficiency of an X-ray beam is the proportion of the total beam that is utilised in the imaging process. If the geometric efficiency decreases from 100% to 50% whilst all other factors are kept equal, then, to maintain the same image quality, the dose would need to be doubled.

The overall geometric efficiency is sub-divided into two aspects. The first is the z-axis geometric efficiency, where the proportion of the overall X-ray beam width (dose profile) utilised along the long axis of the patient is considered. The second aspect, often overlooked, is the detector array geometric efficiency. This defines the proportion of the overall detector area that contains active detector material.

Z-axis geometric efficiency

Due to the finite size of the focal spot, an X-ray beam always has a reduced intensity at the periphery of the field and this region is referred to as the penumbra. On single slice systems, in most situations, the entire X-ray beam, including the penumbra, is utilised in image formation (Figure 2a). If the penumbra were utilised in image production on multi-slice scanners, the outer detectors would receive a less intense X-ray beam than the inner ones. This would lead to the images from these detectors being narrower and noisier. To avoid this, the collimation of the X-ray beam on multi-slice systems is increased such that the penumbra lies beyond the active detectors and they are all irradiated uniformly (Figure 2b).

Figure 2: Slice width collimation for (a) 10 mm single slice, and (b) 4 x 2.5 mm multi-slice CT scanner

X-ray beam falling within detector bank

X-ray beam with penumbra falling outside active detectors

Active detector banks

z-axis

(a)   (b)

In the latest IEC standard on CT safety [4], z-axis geometric efficiency is defined as the percentage of the X-ray beam width in the z-direction that is 'seen' by the detectors. On multi-slice scanners, the width seen by the detectors will be equal to the total active detector width. Where post-patient collimation is used to partially irradiate active detectors, the width 'seen' will be the nominal post-patient collimation width. The definition is given below as an equation and presented diagrammatically in Figure 3.

$$\text{Z-axis geometric efficiency} = \frac{\text{area under z-axis dose profile falling within active detectors}}{\text{area under total z-axis dose profile}}$$

On single slice systems the geometric efficiency is generally close to 100%, unless post-patient collimation is utilised. Current multi-slice scanners generally have dose profiles 2-3 mm wider than the total imaged width, which is generally defined by the nominal collimation setting. For example, with 4 x 2.5 mm acquired slices a nominal total collimation of 10 mm is used and the full width at half maximum (FWHM) of the dose profile is typically about 12-13 mm. As the extent of the penumbra is approximately constant, regardless of the total collimation, the proportion of 'wasted dose' is greater at narrow collimations and so the z-axis geometric efficiency is lower than for wider ones.

Table 2 shows an example of the extent to which z-axis geometric efficiency is affected by the collimation width on a 16 slice scanner [5]. Similar figures are obtained on other scanners that acquire between four and sixteen slices.

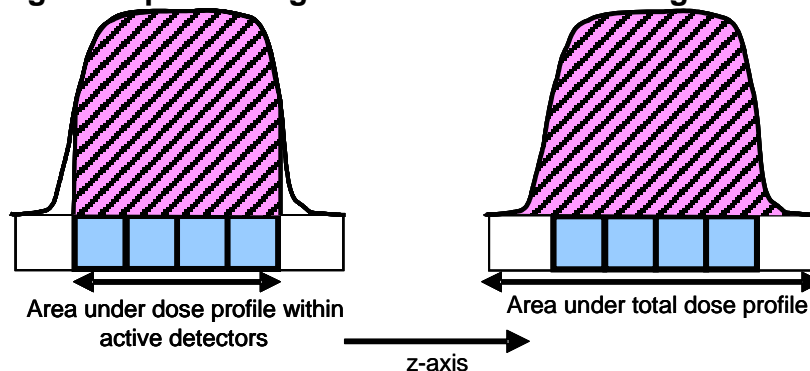**Figure 3: Diagram representing IEC definition of z-axis geometric efficiency**

Area under dose profile within active detectors

z-axis

Area under total dose profile

Table 2: Effect of collimation setting on z-axis geometric efficiency

Total nominal collimation width (mm)	Z-axis geometric efficiency (%)
1.25	54
5	66
10	83
15	89
20	97

Table 3: Number of acquired slices and z-axis geometric efficiency

| Scanner	Number and width of acquired slices (mm)	Total nominal collimation width (mm)	Z-axis geometric efficiency (%)
4-slice	4 x 1.25	5	66
8-slice	8 x 1.25	10	83
16-slice	16 x 1.25	20	97

Scanners that acquire a greater number of simultaneous slices have an advantage in terms of z-axis geometric efficiency. This is because for narrow slice widths a wider total collimation can be used. An example for a 1.25 mm acquired slice width is shown in Table 3.

On multi-slice scanners z-axis geometric efficiencies are generally in the range 80-98% for collimations of 10 mm and above, and about 55-75% for collimations of around 5 mm. For collimations around 1-2 mm z-axis geometric efficiencies are as low as 25% on some systems, although in dual slice mode they can be much higher.

Therefore, the reduced z-axis geometric efficiency of multi-slice scanners, for the wider collimations most commonly used, leads to dose increases of around 10% when compared to single slice systems. However, very narrow collimations can result in a tripling, or more, in dose.

The graphs in Figure 4 demonstrate the effect on the absorbed dose of a reduced z-axis geometric efficiency at narrow collimations. The benefit of acquiring a larger number of simultaneous, narrow slices is demonstrated in Figure 4b.

Figure 4: Data for typical multi-slice CT scanner showing dose relative to 20mm collimation plotted against (a) collimation, and (b) imaged slice width

(a)   (b)
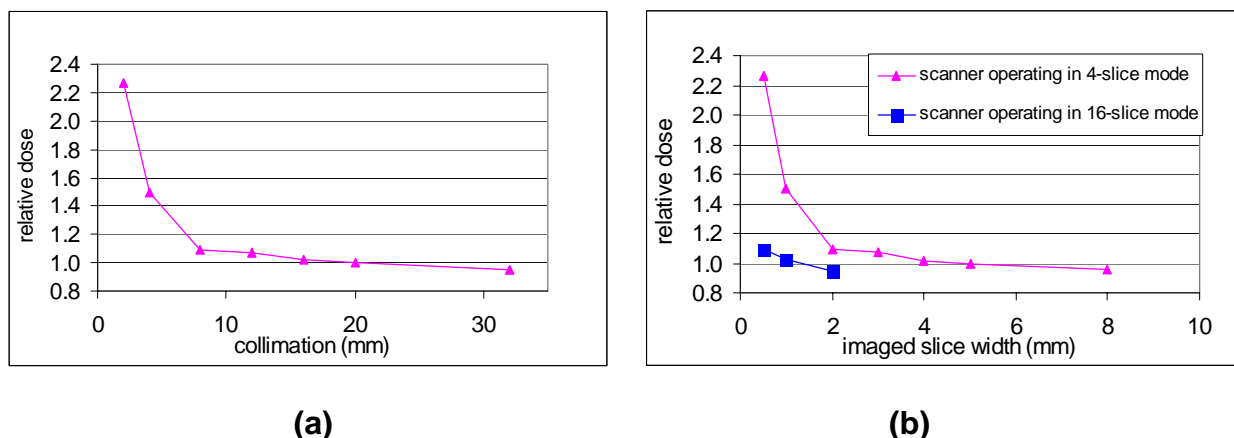
Detector array geometric efficiency (GE)

The total detector area consists of detector material and septa between the detector elements both in the scan (x-y) plane and, on multi-slice scanners, in the axial (z-axis) direction. Generally, as the number of simultaneously acquired slices increases, the size of the detector elements in the z-direction decreases. This leads to an increase in the number of septa and therefore a reduction in detector array geometric efficiency (Figure 5). The definition of detector array geometric efficiency is:

$$\text{Detector array GE} = \frac{\text{active detector area}}{\text{total detector area}}$$

An illustration of this effect is shown in Table 4 for the Philips Mx8000 range of scanners [6]. In this example the reduction in efficiency when moving from a 2-slice to a 16-slice system is only around 6%. This figure will vary with detector array design, with a trend towards lower efficiency values as the number of detectors per imaged length increases.

Figure 5: Diagram of section of detector array from (a) an 8-slice and (b) a 16-slice CT scanner. The 16-slice scanner has approximately twice the number of septa along the z-axis
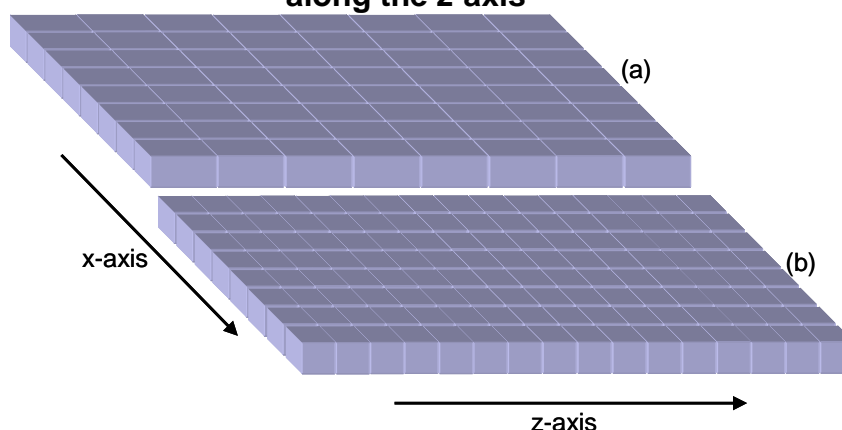
Table 4: Detector array efficiency for the Philips Mx8000 scanner range

Scanner model	No. of detector banks	z-axis coverage (mm)	Detector array efficiency (%)
2-slice	2	20	80
4-slice	8	20	78
16-slice	24	24	75

Considerations for helical scanning

Helical pitch

On multi-slice scanners, as on single slice systems, the absorbed radiation dose is inversely proportional to pitch if the tube current – time product (mAs) and tube potential (kV) are kept constant i.e. the dose will be halved if the pitch is doubled. Due to the nature of the reconstruction method used, on single slice scanners the imaged slice width increases as the pitch increases. As a result, noise on single slice systems remains constant with changing pitch. However, on multi-slice scanners different image reconstruction methods are used [7], and some of these allow for the imaged slice width to remain constant with pitch. In these circumstances the noise will increase as the pitch increases. Some multi-slice scanner manufacturers automatically adjust the tube current to compensate for changes in pitch. This maintains a constant noise and dose with changing pitch.

A parameter termed 'effective mAs' or 'mAs per slice' is sometimes quoted on multi-slice scanners. This quantity is designed to reflect the effect on the average absorbed dose in the scanned volume when the pitch is changed. The adjusted mAs value is obtained by dividing the true mAs per rotation by the pitch value.

When making dose calculations it is important to be aware whether true mAs or 'effective mAs' is being quoted. The two equations below show how the average absorbed dose within the scanned volume, $CTDI_{vol}$, is calculated from the $_nCTDI_w$, the average absorbed dose in the scan plane per mAs, from either the true mAs or the 'effective mAs'.

$$CTDI_{vol} = {_nCTDI_w} \times \frac{mAs}{pitch}$$

$$CTDI_{vol} = {_nCTDI_w} \times \text{'effective mAs'}$$

An illustration of the relationship between pitch, mAs, 'effective mAs' and $CTDI_{vol}$ is given in Table 5.

Table 5: Relationship between pitch, mAs, 'effective mAs' and CTDI$_{vol}$

a. for constant mAs

Pitch$_x$*	mAs per rotation	'effective mAs' or 'mAs per slice'	Relative CTDI$_{vol}$
Axial scan	100	100	1.0
0.5	100	200	2.0
1	100	100	1.0
1.5	100	67	0.67
2	100	50	0.5

b. for constant "effective mAs"

Pitch$_x$*	mAs per rotation	'effective mAs' or 'mAs per slice'	Relative CTDI$_{vol}$
Axial scan	100	100	1.0
0.5	50	100	1.0
1	100	100	1.0
1.5	150	100	1.0
2	200	100	1.0

*Pitch$_x$ = <u>table travel per rotation</u>, as defined in ImPACT Technology Update No.1 [8]
X-ray beam collimation

Additional rotations for helical interpolation

In helical scanning, additional information is required at each end of the planned image volume in order to provide interpolation data for the first and last images. On single slice scanners a half, or one, extra rotation is generally required at each end of the imaged volume. On multi-slice scanners the number of additional rotations required can potentially depend on a number of factors such as the interpolation method, the pitch and the reconstructed image width [9]. Each additional rotation on a multi-slice scanner generally contributes a greater percentage to the dose than that on a single slice scanner because the total collimated beam width is usually greater. This is shown diagrammatically in Figure 6 where the planned scanned length is equal for both the single and multi-slice scanner.

When planning scan lengths to avoid sensitive organs such as eye lenses or gonads, the additional rotations associated with helical scanning should be borne in mind. The additional dose will be particularly significant for short scan lengths and in these situations it may be preferable to perform the scan in conventional slice by slice mode which does not require the extra rotations.

As an example, for a collimated X-ray beam of 20 mm with one extra rotation at each end of a scan run, the increase in effective dose, relative to axial scanning, will be 10% for a scan length of 400 mm, and 40% for a scan length of 100 mm. This compares to increased dose figures of 2.5% and 10% respectively, for the same scan lengths on a single slice scanner, assuming half an extra rotation at each end of the scan run.

Figure 6: Additional irradiation outside the imaged volume (a) a single slice and (b) a multi-slice CT scanner operating in helical mode

Planned scan length

(a)

Planned scan length

(b)
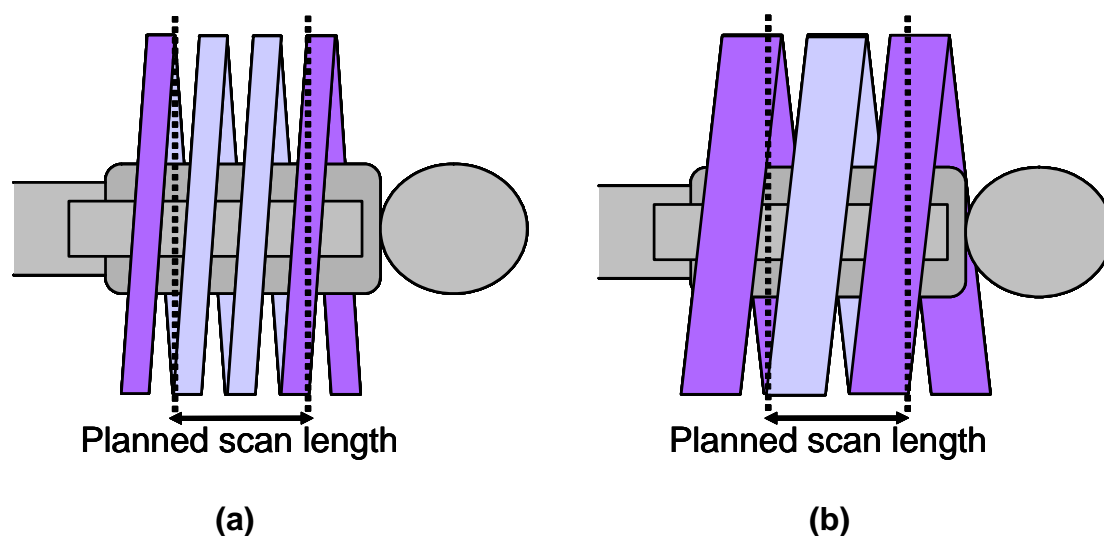
Dose optimisation

On any CT scanner the patient dose is highly dependent on the scan parameters used: kV, mA, rotation time, focal spot size, scan field of view, slice width and pitch. On multi-slice scanners there is the additional variable of X-ray beam collimation, as the same imaged slice width can be achieved from data acquired at a number of collimations. All these parameters must be carefully selected so that the given diagnostic requirements are met at the optimum level of radiation dose. In addition to the scan parameters, the reconstruction parameters such as the reconstruction matrix, reconstruction field of view and reconstruction algorithm must be considered. Although these do not affect dose directly, they may have an indirect effect by altering the image characteristics.

Automatic tube current control

A key parameter affecting dose to the patient is the selected tube current – time product (mAs). Automatic exposure control systems (AECs) have been an integral component of conventional X-ray units for many years. They operate by adjusting the length of the exposure, to produce X-ray films with a consistent level of optical density, regardless of the patient size. This avoids over- and under-exposures of the radiographic film and the necessity for repeat examinations. Digital systems, such as CT scanners, do not incur an image quality penalty at high exposures, and mAs values have, until recently, always been selected manually by the operator. Manufacturers generally recommend tube current and time settings for different examinations for a standard sized patient and the operator must decide whether, and if so, by how much, to adjust these settings to take into account variations in patient size. More recently, following concerns about dose levels in CT, particularly to paediatric patients [10], age or weight based tables of exposure settings are often provided as a guide to the user.

Over the last few years manufacturers have developed techniques for AEC in CT. On CT scanners the approach is to vary the tube current (mA). The overall aim of these systems is to achieve a more consistent level of image quality from patient to patient and also to optimise the use of X-rays, thereby reducing dose.

The attenuation of the X-ray beam increases with the thickness of material in its path, and for approximately every 4 cm of soft tissue, the X-ray beam intensity halves. In order

to achieve the same transmitted X-ray intensity, and thereby the same level of image noise, changing from a 16 cm to a 20 cm diameter phantom requires a doubling of the mA. Changing from a 32 cm to a 48 cm phantom, the mA should, in theory, be increased by a factor of 16. Systems which automatically adapt the overall tube current based on actual patient attenuation remove the guesswork from selecting the appropriate mA setting.

The adjustment of tube current can be considered on three levels. At the first level the mA is adjusted to take account of overall patient size (Figure 7a).The aim is to have a similar value of image noise for patients of different sizes. If used in isolation, this level of AEC maintains the same mA throughout the scan. However it is usually used in combination with the other levels of mA adjustment described below, which take into account attenuation variations within the patient.

Figure 7: Automatic adjustment of tube current to account for (a) overall patient size, (b) attenuation variations along patient's long axis and (c) varying attenuation throughout a rotation

(a)   (b)   (c)
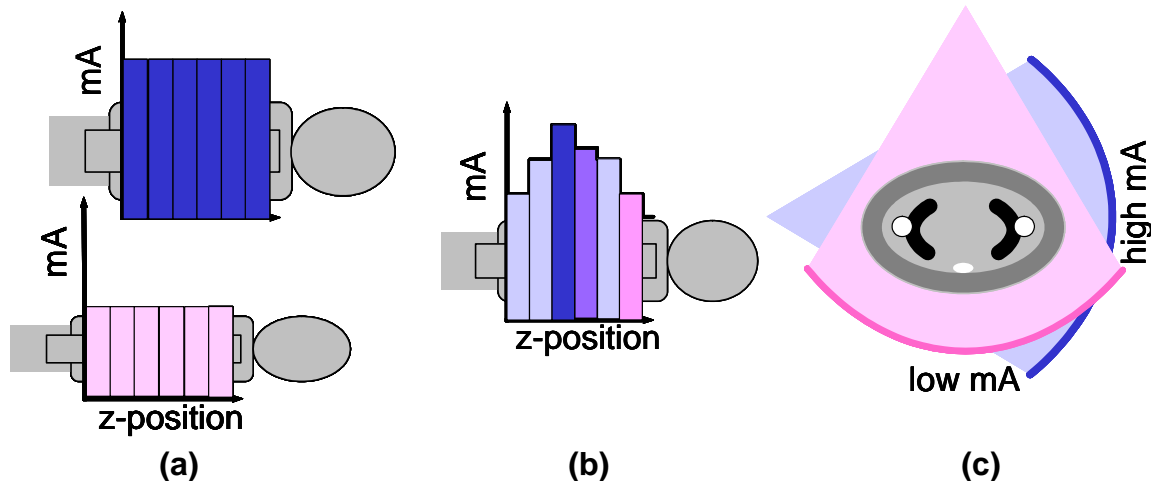
At the second level of automatic mA control, z-axis modulation, the mA is varied on a rotation by rotation basis to account for variations in attenuation along the patient's long axis (z-axis) (Figure 7b). With this type of tube current control, the mean level of image noise within a slice should remain approximately constant for different positions along the z-axis.

At the third level of automatic mA control, angular modulation, the mA is varied during the course of each tube rotation to compensate for the varying attenuation at different angles through the patient (Figure 7c). At some anatomical levels, e.g. the shoulders and pelvis, there is a considerable difference in attenuation between the lateral and anterior-posterior (AP) directions through the patient. The highest tube current is usually required for the lateral projections. Using this technique, a more uniform level of image noise is obtained across the imaged plane, and a given noise level can be achieved at a lower average mA.

Automatic mA adjustment requires prior knowledge of the attenuation characteristics of a patient. The attenuation information to adapt the mA for patient size is obtained from the planning scan projection radiograph (SPR), referred to by different manufacturers as ScoutView, Scanogram or Topogram. The SPR information is also used to adjust the mA for each rotation. There are two different methods currently used for obtaining

information to vary the mA during a rotation. The first method uses the SPR to calculate the relative AP and lateral patient dimensions in each rotation. The mA is then varied sinusoidally to best match this ratio. The second method uses 'online' data from the preceding 180° of rotation to modulate the mA.

When using these AEC systems, some way of prescribing the mA, and thereby the desired level of image quality, must be adopted. Manufacturers have approached this using a variety of methods. On some systems the mA that would be used without AEC on an standard-sized patient is input, and this value is used as the base-line for calculating the mA needed to obtain the same noise level for different patient sizes. Other systems require an 'image quality' index to be input. This index is usually related to the noise value (standard deviation of the CT numbers in a water phantom), and the mA is adjusted to maintain this prescribed noise level for all patient sizes. A third approach is to select a 'reference image' with the required level of image quality for a particular examination. This image is then used to adjust the mA such that the same level of image quality is achieved for each patient. In addition to the various methods for prescribing the tube current there is usually an option to set the upper and lower limits of the mA used by the AEC system.

Automatic adjustment of tube current is not only a useful dose reduction tool but can also lead to benefits in image quality. It should result in more consistent image quality from patient to patient, from slice to slice, and within a slice. Modulation of mA throughout a rotation can also reduce streaking artefacts caused by low photon flux in the lateral projections of anatomical areas such as the shoulders and pelvis. An added advantage of automatic mA control is that since it is designed to use lower overall mAs values, the heat capacity of the tube is preserved, allowing for longer scan lengths if these are deemed necessary, and for a longer tube life.

However, it is also important to note that although AEC systems are often marketed as dose-reduction tools, it is perfectly possible to operate them at a higher radiation dose then would be obtained without their use. Careful prescription of the mAs or image quality is needed to ensure that doses are optimised and within reference dose values [11].

Manufacturers are in the process of further developing their AEC systems. Different scanner models within a range may have different levels of AEC available. The most advanced systems are usually offered on the premium line scanners, and the current status of these is shown in Table 6.

Table 6: Capabilities of AEC packages available on top-of-the-range 16-slice CT scanners

Manufacturer	Name	Level of automatic mA control	Prescription of mA	Attenuation calculation
GE	SmartmA	• patient size • z-axis • angular modulation*	• noise index specified by user	• SPR • angular modulation: sinusoidal
Philips	DoseRight ACS and DOM	• patient size (ACS) • angular modulation (DOM)	• reference image with desired noise level selected by user	• SPR • angular modulation: online, based on previous 180°
Siemens	CAREDose 4D	• patient size • z-axis • angular modulation	• mA for standard patient specified by user	• SPR • angular modulation: online, based on previous 180°
Toshiba	SUREExposure	• patient size • z-axis	• noise index specified by user	• SPR

* On LightSpeed Pro[16] only

Appropriate image quality

Automatic exposure of tube current is an invaluable dose optimisation tool but relies on the operator selecting either the mA for a standard patient or the required level of noise for a given examination. Because CT does not carry an image quality penalty for over-exposure, there has been a tendency to err on the side of lower than necessary noise levels and hence higher doses. The current challenge in CT is to identify an appropriate image quality. This is the optimal value of noise for an examination i.e. the level at which a diagnosis can reliably be made at a minimum dose level.

A relatively new approach to determining these optimal noise levels is through the addition of simulated noise to images obtained at higher mAs values. Images from the same patient at a range of noise levels can then be viewed and scored for diagnostic quality, without subjecting the patient to multiple exposures. A number of studies using this approach have been undertaken and suggest that, in some cases, it is possible to significantly reduce mAs values without affecting the diagnostic quality of the scan [12, 13].

ECG-gated cardiac studies

Cardiac imaging is one of the most rapidly developing areas of CT. Due to the rapid motion of the heart, images would suffer from severe blurring if standard reconstruction methods were used. To overcome these difficulties, techniques have been developed to produce images using data from just a fraction of the cardiac cycle. This is achieved in conjunction with ECG monitoring, so that only data acquired in mid to late diastolic phase, where there is least cardiac motion, is used in image reconstruction. One way in which significant dose reduction can be achieved is by using the ECG information to control the tube current so that the selected mA is maintained during the heart phase for which data is required, but drops to a lower level, such as 20% of its nominal value, for

the remainder of the cardiac cycle (Figure 8). It is claimed that dose savings in the order of 30% and above can be achieved using this technique [14].

Figure 8: An example of ECG-controlled tube current modulation
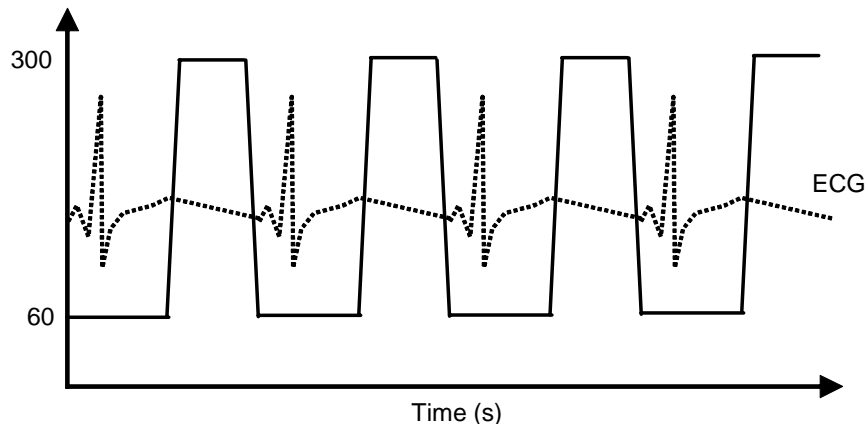
Other considerations

Room shielding

The shielding requirements for room design on multi-slice scanners do not in principle differ from single slice systems. Isodose curve patterns should be similar, but absolute scattered dose rate values may be higher due to the wider collimated widths available. If the overall examination length and mAs per rotation remain the same however, the total amount of scattered radiation will remain constant.

Changes in protocols in terms of kV, mAs and scan length, as well as a possible increase in patient throughput, need to be considered, and shielding requirements calculated on the basis of typical usage patterns.

Summary

Multi-slice technology has lead to a considerable advance in the capabilities of CT scanners. In terms of intrinsic dose characteristics they are very similar to single slice systems, although some differences exist in terms of z-axis geometric efficiency and detector array geometric efficiency.In helical scanning there may be extra dose from additional rotations at each end of the scan run.

In most situations the above factors would increase the dose on a multi-slice scanner by around 20% compared to a single slice system of equivalent design. In some applications they could lead to a doubling or more of dose and in these cases the justification for that particular technique should be carefully considered.

Automatic control of tube current is a feature available across the range of CT systems. It has the potential to significantly reduce doses, particularly when adjusting mAs values for paediatric and small patients.

Multi-slice CT scanners have the capability to scan long lengths with narrow slice widths and to perform multiple-phase contrast studies. This can lead to high patient doses. Careful consideration of the diagnostic requirements for a particular examination is essential, together with an appropriate selection of scanning parameters, to ensure that doses from multi-slice scanners are optimised, and their use in patient care justified.

References

1. Mettler FA, Wiest PW, Locken JA. CT scanning: patterns of use and dose. J Radiol Prot 2000; 20: 353-9.

2. Kalra MK, Maher MM, Saini S. CT radiation exposure: Rationale for concern and strategies for dose reduction. Applied Radiology 2003; August, online supplement; Proceedings from the SCBT/MR.

3. Golding SJ, Shrimpton PC. Radiation dose in CT: are we meeting the challenge? Br J Radiol 2002; 75: 1-4.

4. International Electrotechnical Commisssion. Medical electrical equipment – Part 2-44: Particular requirements for the safety of X-ray equipment for computed tomography 2001; IEC International standard 60601-2-44 Ed.2 Amended.

5. Keat N, Edyvean S, Lewis MA et al. GE LightSpeed[16] CT Scanner Technical Evaluation. MHRA Evaluation Report 2004; MHRA 04015.

6. Morgan HT. Dose reduction for CT pediatric imaging. Pediatr Radiol 2002; 32: 724-8.

7. Kalender WA. Considerations for Multi-slice Spiral CT. In: Kalender WA, Computed Tomography: fundamentals, system technology, image quality, applications. Publicis MCD Verlag; 2000. p. 76-8.

8. ImPACT. Technology Update No. 1, 2nd edition, Multi-Slice CT Scanners, 2002; MDA 02021.

9. Nicholson R, Fetherston S. Primary radiation outside the imaged volume of a multislice helical CT scan. Br J Radiol 2002; 75: 518-22.

10. Linton OW, Mettler FA. National conference on dose reduction in CT, with emphasis on pediatric patients. Am J Roentgenol 2003; 181: 321-9.

11. European Commission study group, European guidelines on quality criteria for computed tomography, EUR 16262 EN, Office for official publications of European communities, 2000.

12. Frush DP, Slack CC, Hollingsworth CL et al. Computer-simulated radiation dose reduction for abdominal multidetector CT of pediatric patients. Am J Roentgenol 2002; 179: 1107-13.

13. Britten AJ, Crotty M, Kiremidjian H et al. The addition of computer simulated noise to investigate radiation dose and image quality in images with spatial correlation of statistical noise: an example application to X-ray CT of the brain
Br J Radiol 2004; 77: 323-8.

14. Barrett JF. ImPACT Special Interest Report. Cardiac CT Scanning. MHRA Evaluation Report 2003; MHRA 03076.