Clinical Radiology (2007) **62**, 507—517

REVIEW

Dose reduction in paediatric MDCT: general principles

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Received 27 October 2005; received in revised form 11 December 2006; accepted 18 December 2006

The number of multi-detector array computed tomography (MDCT) examinations performed per annum continues to increase in both the adult and paediatric populations. Estimates from 2003 suggested that CT contributed 17% of a radiology department's workload, yet was responsible for up to 75% of the collective population dose from medical radiation. The effective doses for some CT examinations today overlap with those argued to have an increased risk of cancer. This is especially pertinent for paediatric CT, as children are more radiosensitive than adults (and girls more radiosensitive than boys). In addition, children have a longer life ahead of them, in which radiation induced cancers may become manifest. Radiologists must be aware of these facts and practise the ALARA (as low as is reasonably achievable) principle, when it comes to deciding CT protocols and parameters.

Introduction

In 1989, computed tomography (CT) accounted for 4% of diagnostic radiology examinations performed in the UK, contributing 40% of the collective population dose from medical radiation.[1] By 1999, one North American institution quoted that 11.1% of the department workload was due to CT examinations; a contribution of 67% to the collective dose.[2] CT can now be responsible for up to 17% of the departmental workload accounting for 70—75% of the collective dose from medical radiation.[3—5] With reference to the paediatric population, the British survey of 1989 stated that 4% of CT examinations were performed in children less than 15 years of age.[1] Mettler showed this figure to have risen to 11.2% by 1999.[2] Coren and colleagues reported a 63% increase in requests for paediatric CT between 1991—1994,[6] and McAllister a 92% increase in paediatric abdomino-pelvic CT examinations between 1996—1999.[7]

These often-quoted figures mirror the introduction of the first single detector helical CT machines (SDCT) in the late 1980s and multi-detector array CT (MDCT) machines in the late 1990s. The technological advances in commercially available CT machines have allowed the radiologist to increase the range of studies they perform using CT: peripheral and cardiac angiography; virtual endoscopy, including bronchoscopy and colonoscopy; multiplanar and volume reformats from isotropic data sets for skeletal examinations; and the more mundane evaluation of appendicitis and renal calculi are all now being performed using CT. However, such technical advances, although producing increased diagnostic accuracy (and some would argue beautiful, aesthetically stunning images) do not come without cost. One potential cost is the amount of radiation that can result from MDCT examinations.

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0009-9260/$ - see front matter
doi:10.1016/j.crad.2006.12.004

Radiation dose measurements

To interpret the radiation risks from CT, it is necessary to be familiar with descriptors of dose and their units of measurement.[8–12] The absorbed dose (measured in Grays) is the amount of energy absorbed per unit mass of an organ or tissue. The absorbed dose cannot be practically measured in patients. The equivalent dose (measured in Sieverts) takes into account the type or quality of radiation an organ is exposed to. It is numerically equal to the absorbed dose, when x-rays are involved. The effective dose equivalent (ED) takes into account all of the organs irradiated during an examination and incorporates a tissue weighting factor[13] based upon the individual organ radiosensitivity. ED allows us to estimate the amount of radiation a patient receives during a radiological procedure, such as CT, and also, provides a measure of stochastic risk.[10,12] The unit for ED is also the Sievert. Measurement of ED relies upon the use of phantoms or mathematical equations or models, such as Monte Carlo.[14–17]

Another measure of CT dose can be approximated with the CT dose index (CTDI), which gives a value for the dose within one ''section''. It is measured using an acrylic phantom (usually of two different sizes) and a pencil ionization chamber, over a length of 100 mm.[8,9] The weighted CTDI ($CTDI_W$) takes into account variations in absorbed dose from the periphery and centre of the phantom.[8,9,11] If variations in pitch are added to the equation, then the term volume CTDI ($CTDI_{VOL}$) is introduced. Numerically it is equal to the $CTDI_W$/pitch. CTDI values are expressed in milliGray. The dose—length product (DLP) has units of milligray per centimetre, and is calculated as the product of CTDI (such as $CTDI_{VOL}$) and scan length. DLP can be converted into ED measurements, using mathematical equations (see below).[8,9] A range of paediatric anthropomorphic phantoms are now available to calculate ED from either direct organ dosimetry or estimates of organ dosimetry based upon examination parameters for paediatric CT examinations.[18–20] At least three different phantoms are required to simulate examinations performed on babies, infants and children of different ages, as paediatric MDCT protocols and dose measurements must take into account weights ranging from less than 1 kg in the premature infant, to values of greater than 60 kg in some adolescents.

It is important to have a fundamental understanding of what CTDI and DLP represent for two reasons: (1) all CT machine manufacturers are now required to display CTDI values on the user interface and some systems will also display the DLP; (2), these dose estimates have been used to discuss dose delivery during MDCT. It is necessary to understand that these measures do not tell us the actual radiation dose a patient receives (that is, the CTDI depends only upon the parameters selected). However, these measures will give information about relative changes in dose that result from alterations in CT examination parameters and draw to the attention of the radiologist the proportionate change in radiation dose that will result, when an examination protocol is modified for an individual patient. For example, if the displayed DLP value decreases by 25%, the dose the patient receives will decrease by approximately 25% as well. In addition, estimates of ED can be obtained from the product of DLP and a factor, which is in part age-related.[21] It is imperative to understand the fact that, for a given set of CT parameters, the displayed CTDI and DLP will be the same, regardless of the patient's age and size, as these figures do not reflect the absorbed dose or the body's shape. Effective doses in CT can be higher for infants and small children than adults given identical CT parameters.[12,14,16,19,20,22] One reason is because the lower absorbed energy in a child is distributed in an even smaller organ (absorbed energy/organ weight = dose).[12]

Despite the inherent problems with CTDI and DLP values, as mentioned in the above paragraphs, these measures are readily available during every CT study performed, are still used by practising radiologists as indicators of radiation dose and are often quoted in the literature. Given these facts, we emphasize that such figures are merely estimates and are gross measures of radiation dose (particularly in the smallest children) at best.

Radiation risks

Bio-effects associated with radiation exposure, can be divided into two main groups: deterministic risk relates to cell death and can be quantified in terms of the radiation dose an organ or body region has received. Above a certain threshold dose, the effects of radiation are seen, and the higher the dose, the more severe the effect. Deterministic effects are rarely seen in diagnostic radiology, but may become a problem with angiographic procedures, including CT fluoroscopy.[23] In addition, temporary hair loss has recently been reported in patients undergoing MDCT brain perfusion studies, in combination with digital subtraction angiography.[24]

Stochastic effects are dependent upon a complex series of effects, including cell transformation and consist of the development of cancer in the irradiated individual or genetic problems in their descendents. The greater the absorbed radiation dose a patient receives, the greater the risk that a stochastic effect is to be seen. However, the severity of the effect is independent of the dose of radiation received.

The stochastic risk of developing cancer from low-level radiation, such as that resulting from CT examinations is debated. There are experts who conclude that there is a potential risk from low-level radiation (≤100–150 mSv) and experts who support an opposing view, who conclude that the risks are non-existent, or at best, highly speculative.[25] The data we have relating radiation risk to low-level radiation such as with CT examinations, are mainly taken from the studies of the atomic bomb survivors. These individuals have been followed up for almost 60 years.[26–28] These figures support a linear dose–response relationship with no threshold,[26,29] with excess risks of solid tumour development even at low doses (less than 150 mSv).[28] The data find that the relative risks persist throughout life, are greater in girls than boys, and decline with age, being highest for those exposed as children.[28]

The risk of developing a radiation-induced cancer has been estimated to be 5% per Sv at all ages,[13] though this figure is closer to 15%, if an individual is exposed in the first decade of life. Recently, the BEIR VII report[29] concluded that ''... it is unlikely that a threshold exists for the induction of cancers but notes that the occurrence of radiation-induced cancers at low doses, will be small''. In the report, evidence was provided supporting a risk of 1 in 1000 for an exposure of 10 mSv. In addition, children have a longer life expectancy, meaning radiation-induced cancers have more time to become manifest. Brenner et al. (2001)[7] assessed the risks of developing a fatal malignancy following a paediatric CT examination. Their estimates suggested that the risk of dying from cancer was 1 in 550 following an abdominal CT (performed with one particular technique), and 1 in 1500 for a brain CT (performed with one particular technique) obtained in infancy. This equates to an approximately 0.35% increase in cancer deaths over the background rate. However, it is important to note that these figures were calculated based upon the assumption of children being imaged using adult CT parameters and the risk would be lower if specific paediatric CT protocols were uniformly followed. Although these risks are small to the individual, they potentially pose a larger public health risk given the large (and increasing) number of CT examinations performed.[30]

Image quality and radiation dose in CT: basic principles

There is a complex relationship between image quality and the radiation dose imparted to the patient. Image quality in CT is determined by spatial resolution and contrast. The tube current (measured in milliAmperes) milliAmperes primarily affects spatial resolution and peak kilovoltage (kVp) affects both spatial and contrast resolution. The principle determinants of the dose a patient receives during a CT examination are due to these same factors: x-ray beam energy (related to the peak kilovoltage) and the x-ray beam intensity or the number of x-ray photons generated (related to the product of the tube current and time). A potentially major problem with CT (and other digital imaging techniques) is that unlike conventional film-screen systems, an excessive radiation dose to the patient does not result in a reduction in image quality (such as an overexposed or dark film). Rather, the higher patient doses in CT, lead to aesthetically more pleasing images. There is a theoretical and variable ceiling above which improvements in image quality have no affect on diagnostic capability. Clearly, doses well above this threshold are unnecessary. It is more problematic to determine what these thresholds are. In short, images should be diagnostic and not necessarily of the best possible quality.

Spatial resolution is the ability to observe small image details. In the image plane, this is determined by the focal spot size and the thickness of the detectors. In the *z*-axis (the long axis of the patient), the section thickness and the choice of pitch also influence spatial resolution. In addition, in-plane spatial resolution can be improved by reducing the displayed field-of-view. Thinner sections or a reduced pitch (<1) will improve *z*-axis resolution.[10]

Image contrast is related to the energy of the x-ray photons and hence to the peak kilovoltage, equipment filtration and patient size. Increasing the peak kilovoltage reduces the image contrast. Structures with high intrinsic contrast, such as bone and vessels containing iodinated contrast media may be better appreciated at lower tube voltages, compared with those with intrinsically less contrast, such as the soft tissues. Detection of low contrast lesions can also be improved by post-processing functions such as altering the display window and level settings.

Image noise or mottle, is the random fluctuation in beam intensity for the same radiation exposure. In CT, image noise depends upon the number of photons used to generate the image, and therefore to the tube current. Increasing the tube current, will decrease the amount of noise. Likewise, increasing the peak kilovoltage or the section thickness will decrease image noise. Conversely, larger patients have noisier CT images, as they transmit fewer photons. To visualize a lesion, the contrast should be sufficient to overcome any loss of image quality due to image noise.

Dose reduction and the radiologist

General considerations

The biggest dose saving in MDCT is when the examination is simply not performed. In children, if the clinical question can be answered by ultrasound or MRI (with no additional patient risks, such as sedation or anaesthesia to consider), then these methods should be used if available. Vetting of CT request forms by a consultant radiologist (and preferably by a consultant with a designated interest and specialist training in paediatric radiology) is potentially important in this regard, given that recent publications from the UK[31] and the USA[32] showed that referring clinicians have little or no knowledge of the radiation dose or risk that patients are exposed to during CT examinations. At a radiology conference held recently, a poll of delegates revealed that they considered around a third of paediatric CT requests to be unnecessary.[33]

Second, multiphase examinations in children should be avoided if possible. Earlier work has shown that more than one third of paediatric body CT is performed using multiple contrast phases,[34] and in the authors' experience both here in the UK and the USA, this is still a frequent occurrence, outside of specialist children's hospitals. It is not uncommon for children to be imaged without intravenous contrast media, and for a repeat study with intravenous contrast to be deemed necessary, if a lesion is suspected. A scenario commonly cited to the authors is the problem of inserting intravenous cannulae in small children in the CT suite. This can be easily overcome if the radiology department institutes a policy whereby the cannulae are sited by the ward staff before the child's arrival in the radiology department. Such a policy has the added benefit that children who require repeat examinations, for example, those children with tumours, do not then associate the CT suite with the trauma of having a cannula inserted.

Again, citing the authors' experience, a second commonly quoted reason for performing multiphase CT examinations, is to search for calcification. Although it may alter a list of differential diagnoses, it rarely changes a child's management to demonstrate calcifications within a mass lesion, using a pre-contrast enhanced examination. A post-contrast (single-phase study) is all that is necessary; a tumour mass that requires intervention will be biopsied or removed, regardless of the presence of calcification. If all other parameters are held static, then a multiphase examination will serve to increase the patient's radiation dose by a factor equal to the number of phases and number of separate examinations. In our practice, only around 1—3% of body examinations need multiphase technique.

Repeat CT examinations should also be avoided if possible. In the USA, around 1 in 3 individuals will have at least three CT studies.[2] Follow-up evaluation may be feasible using ultrasound in children, but if a repeat CT examination is considered necessary, then it may be acceptable to perform a lower-dose scan (for example, if CT is being used to follow up an inflammatory or traumatic lesion) and to accept noisier images than for a primary diagnostic examination.

Finally, the radiologist should be actively involved in designing the CT protocol. In this manner, the study can be limited to the appropriate body region. For example, follow up of Wilm's tumour may not need to include the pelvis, and aortic arch CT angiography does not need to extend to the lung bases. Even for a full abdomino-pelvic examination, it is important to limit the scan range. Extra images obtained above the domes of the diaphragm and below the symphysis pubis, rarely contribute to patient diagnosis.[35]

Shielding of superficial organs

Superficial radiosensitive organs, such as the eye lens, thyroid gland, and breast can be shielded using bismuth material. Studies in both adults[36—38] and children have shown[39] dose reductions of 29—57% to the breast with this technique, without loss in diagnostic quality. For example, the dose to the eye lens in facial CT can be reduced by half.[37] Lead testicular shields are also used in adult patients, but their use in paediatric CT examinations has not, to date been reported.[40,41]

Examination parameters

The list of parameters that contribute to radiation dose include tube current, gantry rotation time, kilovoltage, table speed and detector configuration (size and number of detectors and detector rows). The radiologist must select examination parameters tailored to suit a child's size (e.g., weight), the body region to be imaged, and the clinical question to be answered. Previously, radiologists did not routinely alter their protocols for paediatric patients,[34,42,43] though more recently, 43% of imaging departments report specific paediatric protocols.[4] Colour-coded formats[44] and weight-based protocols[45] are available to assist in this complex area, and some sample protocols have been listed in the Table 1a–d. Broadly speaking, the dose used can be reduced further for orthopaedic, angiographic and chest (compared with abdomen) and airway studies, as these are body systems that have higher intrinsic contrast. Lowering the tube current for these types of examination, will result in more image noise, but the image contrast is sufficiently high to overcome the detrimental effects of this additional noise. In other words, the signal (or image contrast) is high enough to maintain the signal-(or contrast)-to-noise ratio.

Tube current

Tube current is linearly related to radiation dose. If the tube current is halved, so too is the radiation dose. In paediatric CT, it is easy to see why the tube current should be reduced; you do not need as many photons to image a child, as less of them will be absorbed as they pass through the body. However, any decrease in tube current, will increase the image noise and the radiologist must balance savings in radiation dose, with increased noise. In paediatrics, some visible noise is generally accepted by radiologists (a soft-tissue filter can help to decrease visible noise)[10] and does not detract from the diagnosis. This is amply backed up by studies in the clinical and experimental literature, supporting a reduction in tube current for chest CT,[46–54] abdomino-pelvic CT,[47,55–58] and head CT.[47,59–63] Depending upon the clinical indication, then dose reductions of 50–75% may be possible, without compromising diagnostic quality.[64] In this vein, another simple way to decrease the radiation dose, is to reduce the values of the tube current and peak kilovoltage used to obtain the scout image(s), below those recommended by the manufacturer.[65,66]

Diagnostic quality, then, will depend upon many factors, including radiologist preference and clinical indication. The balance between diagnostic quality and radiation dose (including tube current) is an extremely complex one and this is beginning to be more systematically addressed in the paediatric population. For example, investigations include an innovative method for tube current reduction with lesion simulation.[67,68]

Gantry rotation time

Current MDCT machines have sub-second gantry rotation times. As mentioned, there is a linear relationship between tube current (mAs) and dose. Therefore, reducing the rotation time from 1 to 0.5 s, will also halve the radiation dose. Both are important bonuses in paediatric CT. The reduction in examination time has been an important factor in reducing the need to use sedation or general anaesthesia for children undergoing this type of

Table 1(a) Guidelines for multidetector row computed tomography parameters in children: Chest[a]

Weight (kg)	Peak kilovoltage	Tube current[b]		Section thickness (mm)	Pitch			Detector thickness[c] (mm)				Increment (mm)
		SDCT	MDCT		4-	8-	16, 64-[d]	4-	8-	16-	64-	
5–9.5	100–120	40	30	3.75–5	0.75	0.875	0.9375	2.5	1.25	1.25	0.625	2.5
10–19.5	100–120	50	30–40	3.75–5	0.75	0.875	0.9375	2.5	1.25	1.25	0.625	2.5
20–29.5	120	60	40	5	0.75–1.5	1.35	1.375	2.5	1.25	1.25	0.625	2.5
30–39.5	120	70	50	5	1.5	1.35	1.375	2.5	1.25	1.25	0.625	2.5
40–49.5	120	80	60	5	1.5	1.35	1.375	3.75	2.5	1.25	0.625	2.5
50–75	120	100–120	70–90	5	1.5	1.35	1.75	3.75	2.5	1.25	0.625	2.5
>75	120	120–140	≥110	5	1.5	1.35	1.75	3.75	2.5	1.25	0.625	2.5

[a] Parameters are based on GE single and multi-detector row CT machines.
[b] Use 0.5 s gantry time when an option; tube current are for four- and eight-section MDCT; 16-section weight-based colour-coded tube current are loaded on the machine.
[c] For anticipated multiplanar reconstructions or three-dimensional rendering, use thinnest detector width (e.g. 0.625 mm) with 16-section at all ages.
[d] For 64-section, pitch of approximately 1 under 20 kg, otherwise 1.375.

Table 1(b) Guidelines for multidetector row computed tomography parameters in children: Abdomen/pelvis[a]

Weight (kg)	Peak kilovoltage	Tube current[b]		Section thickness (mm)	Pitch			Detector thickness[c] (mm)				Increment (mm)
		SDCT	MDCT		4-	8-	16, 64[d]	4-	8-	16-	64-	
5–9.5	100–120	60	50	3.75–5	0.75	0.875	0.9375	2.5	1.25	1.25	0.625	2.5
10–19.5	100–120	70	60	3.75–5	0.75	0.875	0.9375	2.5	1.25	1.25	0.625	2.5
20–29.5	120	80	70	5	0.75–1.5	1.35	1.375	2.5	1.25	1.25	0.625	2.5
30–39.5	120	100	80	5	1.5	1.35	1.375	2.5	1.25	1.25	0.625	2.5
40–49.5	120	120	100	5	1.5	1.35	1.375	3.75	2.5	1.25	0.625	2.5
50–75	120	140–150	110–120	5	1.5	1.35	1.75	3.75	2.5	1.25	0.625	2.5
>75	120	≥170	≥135	5	1.5	1.35	1.75	3.75	2.5	1.25	0.625	2.5

[a] Parameters are based on GE single and multi-detector row CT machines.
[b] Use 0.5 s gantry time when an option; tube current are for four- and eight-section MDCT; 16-section weight-based colour-coded tube current are loaded on the machine.
[c] For anticipated multiplanar reconstructions or three-dimensional rendering, use thinnest detector width (e.g. 0.625 mm) with 16-section at all ages.
[d] For 64-section, pitch of approximately 1 under 20 kg, otherwise 1.375.

examination. In addition, faster imaging times mean the images are less likely to show motion artefact, which in the past may have required an examination to be repeated, with an obvious increase in the radiation dose the patient received. Reducing the tube current will increase image noise, and this must be factored in, when selecting these parameters.

Tube voltage

As noted previously, the tube potential (kVp) determines the energy of the incident x-ray beam. This parameter has not routinely been adjusted in the past for body CT exams in infants and children, with the majority traditionally being performed at 120–140 kVp. Reducing the peak kilovoltage can result in a substantial drop (due to an exponential relationship between radiation dose and peak kilovoltage) in the radiation dose, though the exact dose saving is in part related to the geometry of an individual CT machine, and thus, varies between different manufacturers. The physical distance between the x-ray tube and the patient, and the inherent tube filtration are important factors here.[9] Machines with greater tube filtration emit fewer low energy photons. Those low-energy photons that actually reach the patient, are absorbed, contributing to the radiation dose received by the skin and superficial organs, but not to image formation. Huda reported a four-fold decrease in the radiation dose, when the voltage was dropped from 140 to 80 kVp, showing this to be true for both body and head CT protocols.[10] The effect on image quality is more complex, as both image noise and tissue contrast are affected. An increased peak kilovoltage will increase the contrast-to-noise ratio for all tissues. However, the biggest differences are seen with soft tissue and fat.[69] For body regions with high inherent contrast, the peak kilovoltage can be reduced to 80 or 100, depending upon the child's

Table 1(c) Guidelines for multidetector row computed tomography parameters in children: Extremity skeletal examination[a]

Weight (kg)	Peak kilovoltage[b]	Tube current		Section thickness (mm)	Pitch			Detector thickness (mm)		Increment (mm)
		SDCT	MDCT		4-	8-	16, 64	4-, 8-	16, 64	
5–9.5	80–100	40	30	1.25–2.5	1.5	1.35	1.375	1.25	0.625	0.5–1.25
10–19.5	80–100	50	30–40	1.25–2.5	1.5	1.35	1.375	1.25	0.625	0.5–1.25
20–29.5	100	60	40	1.25–2.5	1.5	1.35	1.375	1.25	0.625	0.5–1.25
30–39.5	100	70	50	1.25–2.5	1.5	1.35	1.375	1.25	0.625	0.5–1.25
40–49.5	120	80	60	1.25–2.5	1.5	1.35	1.375	1.25	0.625	0.5–1.25
≥50	120	100–120	70–90	1.25–2.5	1.5	1.35	1.375	1.25	0.625	0.5–1.25

[a] Parameters are based on GE single and multi-detector row machines. Reconstruct 0.625 mm data set at 0.5–1 mm interval to use for additional planes (e.g. sagittal and coronal). There is no need with sub-mm thick images for imaging in more than one plane. Protocols generally for finer detail exams such as wrists and ankles. Thicker sections and increase interval for larger regions.
[b] Consider 80–100 kVp at all ages.

Table 1(d) Guidelines for multidetector row computed tomography parameters in children: CT angiography

Weight (kg)	Peak kilovoltage	Tube current[a]		Section[b] thickness (mm)	Pitch[c]			Detector thickness (mm)		Increment (mm)
		SDCT	MDCT		4-	8-	16, 64	4-, 8-	16, 64	
5–9.5	80–100	70	60	1.25	1.5	1.35	1.375	1.25	0.625	1.0–2.5
10–19.5	80–100	80	70	1.25	1.5	1.35	1.375	1.25	0.625	1.0–2.5
20–29.5	100	90	80	1.25	1.5	1.35	1.375	1.25	0.625	1.0–2.5
30–39.5	100	120	100	1.25	1.5	1.35	1.375	1.25	0.625	1.0–2.5
40–49.5	120	140	120	1.25–2.5	1.5	1.35	1.375	1.25	0.625	1.0–2.5
50–75	120	160–180	140–160	1.25–2.5	1.5	1.35	1.375	1.25	0.625	1.0–2.5
>75	120	≥200	≥170	1.25–2.5	1.5	1.35	1.375	1.25	0.625	1.0–2.5

Table reproduced from: Frush DP. Evidence-based principles and protocols for pediatric multislice computed tomography. In: Knollman F, Coakley FV, eds. *Multislice CT—Principles and Protocols*. Philadelphia, PA: Elsevier, 2005, pp. 179–201. Reproduced with permission from Elsevier.

[a] Tube current slightly higher than body CT protocols. Use 0.5 s rotation time when an option.

[b] Displayed thickness. For coronal and sagittal reformats and three dimensional reconstructions, reconstruct an axial data set at thickness of the detector (e.g. 0.625 for 16-section machine) at 0.5–1 mm intervals. Multiplanar thickness and interval should be similar to axial. For evaluation of larger structures, especially in larger children (e.g. aorta) the larger detector configuration (2.5 for 8- and 1.25 for 16-section machine) and a larger reconstructed thickness and interval can be used.

[c] For larger children and larger vessels, the highest pitch can be used for MDCT.

size. This could include chest, airway and skeletal evaluation, and CT angiography.

Section width and pitch

Beam collimation and table speed (determinants of pitch) are parameters that are linked to affect image quality and radiation dose. With MDCT, pitch is defined as the table movement (mm)/number of detector channels $\times$ section width (mm) (the denominator here is also referred to as the effective beam collimation). As with SDCT, the greater the pitch, the lower the radiation dose. Increasing the pitch from 1 to 1.5 reduces the radiation dose by a third and doubling the pitch halves the radiation dose.[34] Too great an increase in the pitch can lead to a reduction in image quality, by decreasing the z-axis spatial resolution, degrading the section profile and developing artefacts.[9,11,64] Higher pitches (>1) are often used in paediatric radiology and are increasingly used in adult imaging, such as for virtual endoscopy.[11] Some CT machines will automatically attempt to increase the tube current, if the pitch is altered to be >1; this is to maintain the contrast-to-noise ratio. The radiologist must recognize this and subsequently readjust the tube current downwards, to maintain the dose savings.

Thinner section widths may improve spatial resolution, but come at the expense of increased noise. Increasing the tube current (and hence the dose) can help to offset this problem. Increasing the reconstructed section thickness decreases noise also, but has no impact on dose. The capability of modern MDCT machines to produce images at narrow section widths means it is tempting for the radiologist to always obtain thinner sections with higher spatial resolution.

There are a variety of collimator thicknesses with modern MDCT equipment, affording sub-millimetre collimator thicknesses (with sub-millimetre reconstructed section thicknesses) yielding isotropic image reformations. In our experience, this issue of collimation and effect on radiation dose is not entirely explained by manufacturers. In general, the thinnest collimation used probably results in a slight increase in dose. This may be acceptable, however, given the advantage of improved reformations in selected circumstances, such as skeletal evaluation and CT angiography. However, just because sub-mm collimation exists, does not mean it always has to be used. If an infant has a 7 cm renal mass seen at ultrasound and a staging CT is required, then 1.25–2.5 mm collimation (with 5 mm section thickness) could be perfectly adequate.

Available number of detector rows

The number of detector rows also can affect dose. One inherent dose problem with MDCT relates to the shape of the x-ray beam; some of the beam extends beyond the confines of the detector rows, a concept referred to as "overbeaming".[3,8,11,70] A narrower effective collimation compounds this problem, but the effect decreases with more detector rows for the same scan distance (i.e. is less with 16- and 64-section machines, than four- and eight-section machines).

It is important to realize too, that parameters, such as tube current and peak kilovoltage, may not translate to equivalent doses between manufacturers or even between machines from the same manufacturer. For example, improvements in detector efficiency or changes in focal spot to iso-centre distances may result in a different dose for the same tube current.

Dose reduction and the manufacturers

In recent years, all of the major MDCT machine manufacturers have made alterations to their equipment, and attempt to have age or size-adjusted protocols, aimed at controlling the radiation dose, whilst maintaining image quality.[71–74] Most manufacturers now program paediatric protocols into their machines, and these act as a useful guide for paediatric dose reduction. More significant dose savings can be achieved when these protocols are modified by the radiologist, with the help of the local medical physics department or the radiology department of the regional children's hospital.

Automatic tube current modulation (ATCM) is another new technique for radiation dose management. This innovation has been shown to substantially reduce patient doses.[75–82] ATCM works on the premise that the visualized noise on the CT image is caused by quantum noise in the projections.[11] The tube current is modified to more closely follow the patient's anatomy and maintain a constant noise level on the images. The major methods are modulation in the *x* and *y*-axes (angular modulation), *z*-axis modulation and most recently, a combination of the two.

Angular modulation adjusts the tube current as the x-ray tube passes around the patient's body. In regions where the body has a non-circular cross-section (especially the shoulder region), the majority of the image noise comes from the lateral projections, with beam attenuation being much less in the anteroposterior projections. The tube current can be reduced for projections with low beam attenuation, without a noticeable affect on the amount of image noise. This ''intra-section'' current modulation is performed in real-time by modern machines.[11,77–79,82] This technique has shown dose savings of approximately 30% in the referenced studies.

With *z*-axis current modulation, the radiologist may first select a desired (acceptable) noise level (or index) for the examination to be performed or the equipment will provide a modulation based on pre-programmed levels. In regions such as the upper abdomen, where contrast resolution is intrinsically low, a lower amount of noise (lower noise index) is suggested. The converse would be true for orthopaedic work. The technique aims to equalize the noise level on the images obtained (during imaging, or through interrogation of densities seen on the scout, for example), and has been said to mimic the auto-exposure control systems used with conventional x-ray systems.[11] With some technology, the radiologist may also select and program both minimum and maximum acceptable tube currents, before the examination. The machine then calculates the adjustments that can then be made to the tube current within this selected range, using data from the scout (topogram) images. The desired noise level is thus maintained. Reductions in tube current of >40% have been reported with this technique.[75,76] The two techniques for tube modulation (angular and *z*-axis) can be used simultaneously, to further optimize the patient dose.

A caveat here is the use of tube modulation in conjunction with bismuth shields. If the shields are applied to the patient before the scout views are obtained, then theoretically, the CT machine (which modulates the tube current based upon the scout views) will increase the tube current to compensate for the presence of the shields. However, in the experience of one of the authors, the smart tube current technique of tube modulation on the GE Light Speed CT machine (General Electric Medical Systems, Milwaukee, WI, USA), will allow for tube modulation techniques, with bismuth shields *in situ* (Coursey et al. Data presented at the International Pediatric Radiology meeting, Montreal, Canada, May 2006). The effect of bismuth shields on automatic tube current modulation with other manufacturers is unknown at the present time.

The x-ray beam in CT, as with other x-ray systems, is filtered to remove low peak kilovoltage photons, which would otherwise be absorbed by the patient, increasing their radiation dose. Newer filters are contoured to shape the x-ray beam, further reducing the surface skin dose.[71,74]

Filters of a different kind can be used as a post-processing technique. Noise reduction filters allow the examination to be performed at a markedly reduced currents, and then to act to improve the quality of the images obtained, so that diagnostic capability is maintained.[83]

To repeat a previously stated fact: reducing the tube current reduces the patient dose, but increases image noise. The question ''how low can we go'' with regard to tube current, is not an easy one to answer, as there are obviously ethical concerns in repeatedly examining a child with different settings, in an attempt to optimize the balance between image noise and dose. It should be noted, that there are no currently established levels of noise on CT images, which have been deemed acceptable for practical purposes. However, as noted before, one step towards this end involves ongoing investigations in computer-simulated dose-reduction software, which provides the research radiologist with an opportunity to post-process the original examination by adding a controlled and variable amount of image noise in order to mimic the appearance that would have been obtained had a lower current been used.[61,67,84] This type of technology with can only help to further fine-tune our paediatric CT protocols.

Conclusion

With the continued development of MDCT, the use of CT technology and its contribution to the collective population dose from medical radiation have increased. It has been shown that the effective dose delivered during some CT examinations overlaps with those doses reported to increase cancer rates.[7,85] There are unique considerations with dose in children, as well. It must be recognized that children are more radiosensitive than adults and have a longer life ahead of them, in which radiation induced cancers may become manifest. Therefore, it is imperative that radiologists continue to work within the constraints of IR(ME)R[86] and practise the ALARA (as low as is reasonably achievable) principle, when it comes to CT protocols and parameters. We must work to maximize the yield from our CT images, whilst minimizing the risk to our patients.

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