Managing Patient Dose in Multi-Detector Computed Tomography (MDCT)

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Summary Points

Modern generations of CT scanners employ multiple rows of detector arrays allowing rapid scanning and wider scan coverage.

All new CT systems are MDCT, and a number of new dose reduction tools have become available commercially.

There are a number of new influencing parameters specific to MDCT which systematically increase or decrease patient dose compared to single-detector row CT scanners (SDCT).

As in earlier developments in CT, there is potential for dose reduction, but the actual dose reduction depends upon how the system is used.

It is important that radiologist, medical physicists and CT system operators understand the relationship between patient dose and image quality and be aware that often image quality in CT is greater than that needed for diagnostic confidence.

It must be remembered that "pretty" pictures are not essential for all diagnostic tasks, but rather a level of quality will need to be chosen – whether low noise, standard, or low dose, dependent on the diagnostic task.

Objective measures such as image noise or contrast-to-noise ratio may not completely capture all of the features relevant to making a correct clinical diagnosis. Thus, determining "optimal" image quality can be a complex task, as both quantitative metrics (e.g., noise) and observer perceptions are involved.

Initial reports after the introduction of MDCT indicated increased patient doses relative to SDCT; more recent reports show comparable or decreased patient doses.

If the user selects settings identical to those used in SDCT, there can be an increase in patient dose.

The increase in MDCT use has been faster than the decrease in dose per examination.

Physicians need to understand that thinner slices may increase patient dose, particularly if acquired using MDCT systems with less than 16 active detector rows.

There are indications that awareness on adapting exposure factors to manage patient dose is increasing but the rate at which technology is changing overtakes adoption of effective dose management.

Automatic exposure control (AEC) systems do not reduce patient dose per se, but enable scan protocols to be prescribed using measures related to image quality. If the image

quality is appropriately specified by the user, and suited to the clinical task, then there is a reduction in patient dose for all but the obese patient. In obese patients, the dose is increased to improve the image quality.

AEC does not imply total freedom from operator selection of scan parameters. While CT systems without AEC require operator selection of mA, AEC systems require understanding of newer concepts such as noise index, reference mAs and reference images in order for AEC to be operated effectively. Understanding of some parameters like the standard deviation of image pixels or noise index, is not intuitive and entails chances of error.

The selection of image quality parameters in AEC systems is not a straightforward process. There is lack of consensus on how image quality is to be specified; with the result that there are significant differences in the ways different companies achieve exposure control. It is important that users are aware of the behaviour of their system.

"One-size-fits-all" type protocols must not be used for any CT scanner.

Justification is a shared responsibility between requesting clinicians and radiologists. It includes justification of the CT study for a given indication, and classification of clinical indications into those requiring standard or high dose CT and those for which information can be obtained with low dose CT examination.

There are indications that awareness on adapting exposure factors to manage patient dose is increasing.

Scanning parameters should be based on study indication, patient age and body region being scanned so that radiation dose can be adapted based on these parameters.

Guidelines must be set so that inappropriate studies can be avoided and triaged to non-radiation based imaging technique.

Training of requesting physicians and CT staff can help in the optimization of scan indications, protocols and radiation dose.

1. MDCT TECHNOLOGY

Modern generations of CT scanners employ multiple rows of detector arrays allowing rapid scanning and wider scan coverage. All new CT systems are MDCT, and a number of new dose reduction tools have become available commercially.

1.1. Background

(..) Computed Tomography (CT) technology and its clinical applications have shown enormous resilience against alternative diagnostic methods and at the moment is stronger than ever. Enabled by technology that provides high power x-ray tubes, magnificent computing power, multi channel detectors to give sub millimetre slices with wider scan coverage, faster rotation times to complete one rotation in one third of a second, all have moved CT to dynamic applications in cardiology and 3-dimensional imaging of vascular and musculoskeletal anatomy.

(..) A number of terminologies are in use for this technology, namely multi-detector row computed tomography (MDCT), multi-detector CT (MDCT), multi-detector array helical CT, multi-channel CT and multi-slice CT (MSCT). The number of simultaneous but independent measurements along the patient long axis is often referred to as the number of “slices”, and this value is commonly used to represent the technical capabilities of a system (e.g. 64-slice MDCT). In this report, the Commission has chosen to use the terminology MDCT when referring to the technology generically, and 64-MDCT when referring to a specific technical implementation of MDCT.

(..) In 2000, ICRP published a report on “Managing Patient Dose in Computed Tomography” (ICRP, 2000). At that time there was an urgent need to focus the attention of radiologists, physicians, medical physicists and other personnel involved in CT on the relatively higher effective doses to individual patients, increasing frequency of CT examinations, changes in clinical applications and the increasing contribution of CT to the collective dose. Further, the technology in use dominantly utilised a single row of detectors (SDCT), permitting scanning of only a single slice at a time in either a discrete (sequential acquisition) or continuous fashion (spiral acquisition). Multiple-detector rows along the z-axis (longitudinal axis of the patient, i.e. head to toe) permit simultaneous scanning of more than one slice. MDCT was in its infancy at the time of the 2000 report (ICRP 2000) and thus there was brief mention in the report of its impact on radiation dose. The concrete data and experience was insufficient to make any judgement. In the following years there has been a phenomenal increase in use of MDCT and technology has been advancing very rapidly to move from 4 slice to 8, 16, 32, 40 and 64-slice. Furthermore, dual source MDCT has been recently made available and 256-slice MDCT is expected to be released soon. The improved speed of MDCT scanning has also meant new applications (cardiac CT, whole body scanning) as well as improved patient throughput and workflow. In the last two decades, use of CT scanning has increased by more than 800% globally (Frush 2003). In the United States, over the period of 1991 to 2002, a 19% growth per year in CT procedures has been documented. Also in the United States during this period, CT scanning for vascular indications has shown a 235% growth, followed by a 145% growth in cardiac applications An increase has also been demonstrated in abdominal (25%), pelvic (27%), thoracic (26%) and head & neck (7%) applications (Fox 2003). With 64-slice MDCT a further substantial increase is expected in cardiac applications. A 10% annual growth in the global CT market was reported in the year 2002 and this trend seems to continue.

1.2. Introduction to MDCT Technology

(..) MDCT systems are CT scanners with a detector array consisting of more than a single row of detectors. The “multi-detector-row” nature of MDCT scanners refers to the use of multiple detector arrays (rows) in the longitudinal direction (that is, along the length of the patient lying on the patient table). MDCT scanners utilize third generation CT geometry in which the arc of detectors and the x-ray tube rotate together. All MDCT scanners use a slip-ring gantry, allowing helical acquisition at rotation speeds as fast as 0.33 second for a full rotation of 360 degrees of the X-ray tube around the patient. A scanner with two rows of detectors (Elscint CT Twin) had already been on the market since 1992 and MDCT scanners with four detector rows were introduced in 1998 by several manufacturers. The primary advantage of these scanners is the ability to scan more than one slice simultaneously and hence more efficiently use the radiation delivered from the X-ray tube (Fig.1.1). The time required to scan a certain volume could thus be reduced considerably.

The number of slices, or data channels, acquired per axial rotation continues to increase, with 64-detector systems now common (Flohr et al., 2005a; Flohr et al., 2005b). It is likely that in the coming years even larger arrays of detectors having longitudinal coverage per rotation > 4 cm will be commercially available. Preliminary results from a 256-detector scanner (12.8 cm longitudinal coverage at the center of rotation) have already been published (Mori et al., 2004). Further, an MDCT system with two x-ray sources is now commercially available, signaling continued evolution of CT technology and applications (Flohr et al., 2006).

(..) MDCT scanners can also be used to cover a specific anatomic volume with thinner slices. This considerably improves the spatial resolution in the longitudinal direction without the drawback of extended scan times. Improved resolution in the longitudinal direction is of great value in multiplanar reformatting (MPR, perpendicular or oblique to the transaxial plane) and in 3-dimensional (3D) representations. Spiral scanning is the most common scan acquisition mode in MDCT, since the total scan time can be reduced most efficiently by continuous data acquisition and overlapping data sets and this allows improved multi-planar reconstruction (MPR) and 3D image quality to be reconstructed without additional radiation dose to the patient.

1.3. Differences between SDCT and MDCT

(..) One essential difference between SDCT and MDCT is how the thickness represented by an image, or slice, is determined. For a SDCT, slice thickness is determined by a combination of pre-patient and post-patient collimation. Therefore, the dimension of the detector array along the longitudinal axis can extend beyond the anticipated width of the x-ray beam or image slice (Fig. 1.1) (i.e. the detector width is greater than the beam width). For MDCT, the converse is true and the x-ray beam width must be large enough to allow irradiation of all “active” detector rows (i.e. all those being used for a particular scan acquisition); slice thickness is instead determined by the width of the individual active detector rows.

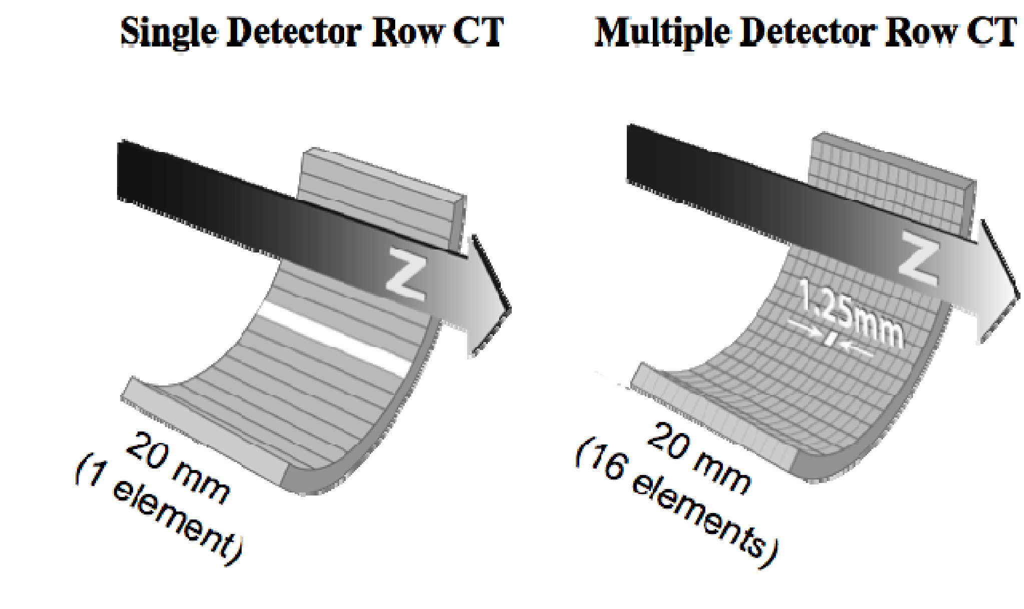

Fig 1.1 Schematics of detector rows and elements

(..) In Fig. 1.1 the single-detector row CT (SDCT) system on the left has one detector element along the longitudinal axis (indicated by z) and many (approx. 900) elements on the arc around the patient. The width of the detector (relative to the center of the gantry) is 20 mm, although the maximum beam width is only 10 mm. Thus the detector is wider than the x-ray beam. The multiple-detector row CT (MDCT) system on the right has 16 detector elements each of 1.25-mm along the longitudinal axis for EACH of the approximately 900 positions around the patient. The width of the detector is also 20 mm at isocentre. The four data channels allow the acquisition of 4 simultaneous slices, of 1.25, 2.5, 3.75 or 5-mm width.

(..) Larger slice thicknesses (2.5 mm, 5 mm, 10 mm) can be generated by electronically combining the signal from several of these rows. Therefore the slice thickness used for the purposes of image review often differs from the slice thickness used for data acquisition. It may be larger, but never smaller. In this document, the term ‘slice thickness’ always refers to that used for data acquisition (slice collimation).

(..) Due to the narrow width of the rows and the use of 4th generation geometry, gas ionization detectors are not used for MDCT scanners. In order to generate an image of a 1-mm slice of anatomy, detector rows of not much more than 1 mm in width must be used (detector dimensions are normalized relative to their coverage at the center of the CT gantry). The detector arrays are made from multiple rows, each approximately 1-mm wide (e.g. sixteen 1.25-mm wide detector rows).

(..) Another design for 4-MDCT detector arrays is illustrated in Fig. 1.2. When small slices are desired, only the central portion of the array is used. It is therefore not necessary to have narrow rows in the outer portions of the array. The wider detectors at the periphery allow simultaneous acquisition of four slices each of 5 mm thickness. This design is somewhat less expensive and more geometrically efficient.

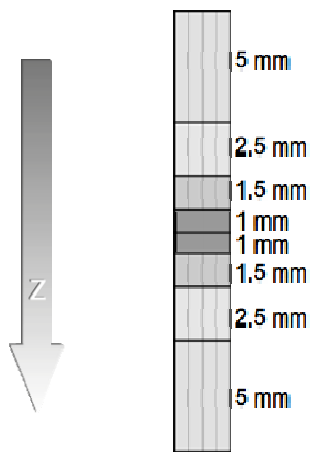

Fig. 1.2 Diagram of the detector geometry used in a 4-MDCT from two major manufacturers. The detector array is 20-mm wide along the longitudinal axis and uses eight rows of varying widths to allow simultaneous scanning of 4 slice up to 5-mm thick.

(..) Currently, MDCT systems are capable of acquiring up to 64 slices simultaneously in the z-direction (Fig.1.3). Three of the four manufacturers use 64 rows of either 0.625 mm or 0.5 mm detectors. The fourth manufacturer uses 32 rows of 0.6 mm detectors and oscillates the focal spot to acquire 64 overlapping slices (Flohr 2005). This results in the reduction of spiral artifacts and improved spatial resolution along the longitudinal axis (Flohr 2005).

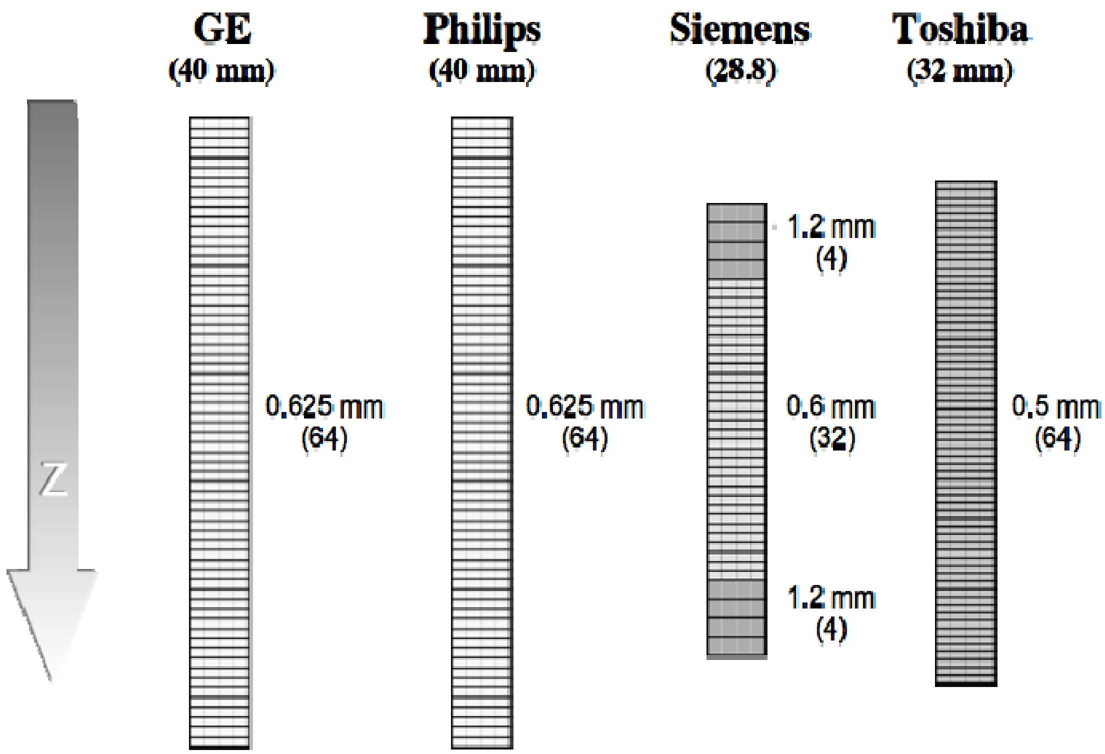

Fig. 1.3 Diagram of the detector geometries used in 64-MDCT from four major manufacturers. The Siemens 64-MDCT uses 32 sub-mm detectors and a moving focal spot to achieve 64 overlapping slice measurements.

(..) For sequential data acquisitions (e.g. the table is stationary during the rotation of the x - ray tube around the patient), each channel collects sufficient data to create one "slice" or image, so as many as 64 independent images along the z axis could theoretically be reconstructed. For narrow slice widths, geometrical "cone-beam" considerations may limit the number of allowed images per rotation to less than 64. For example, one manufacturer's 16-detector scanner allows only 12 data channels to be used in sequential scanning because of cone beam considerations (Flohr et al., 2005a; Flohr et al., 2005b).

(..) The primary attribute of MDCT systems is not the number of physical detectors rows, but the number of slices that are acquired simultaneously. The speed needed to cover a given volume is improved by a factor equivalent to number of slices included in the scan simultaneously. The reason why the number of simultaneous slices was initially limited to 4 was the amount of data to be acquired and transferred simultaneously. At that time, engineering and cost considerations limited the systems to 4 simultaneous data collection systems. Additionally, cone beam artifacts were not severe in 4-MDCT, but as the number of simultaneous slices increased, these artifacts become more problematic using conventional fan-beam reconstructions methods. Once 3-D cone-beam reconstruction algorithms (or advanced fan-beam algorithms with cone-beam corrections) and the increased computational power needed for these algorithms became available, 8- and 16-MDCT scanners were introduced.

(..) The advent of spiral CT introduced an additional acquisition parameter into the CT vocabulary, *pitch*. Pitch is defined as the ratio of the table travel per x-ray tube rotation to the x-ray beam width. With MDCT, a significant amount of confusion was introduced regarding the definition of pitch, as some manufacturers used an altered definition of pitch that related the table travel per x-ray tube rotation to the width of an individual data channel. The International Electrotechnical Commission CT Safety Standard specifically addressed the definition of pitch, reestablishing the original definition (table travel normalized to the total beam width) as the only acceptable definition of pitch (International Electrotechnical Commission, 2002; McCollough and Zink, 1999). This definition of pitch conveys the degree of overlap of the radiation beam: a pitch of 1 indicates contiguous radiation beams, a pitch less than 1 indicates overlap of the radiation beams, and a pitch greater than 1 indicates gaps between the radiation beams.

(..) Two manufacturers (Siemens and Philips) report the milliampere second (mAs) as the average mAs per unit length along the longitudinal axis, called either *effective* mAs or mAs/slice, and calculated as actual mAs/pitch. This distinction between mAs and mAs per unit length is important, because as the pitch is increased, scanner software may automatically increase the mA such that the image noise (and patient dose) remains constant with increasing pitch values (Flohr et al., 2003a; Flohr et al., 2003b; Mahesh et al., 2001). When the effective mAs or mAs/slice is displayed, the user may be unaware that the actual mA is increased. On General Electric MDCT systems, the mA value is automatically adjusted to the value that will keep image noise constant as pitch or slice width is changed, and the selection box is turned orange to alert the user of the change in the prescribed mA value.

1.4. What is the motivation for this report?

(..) After the publication of ICRP 87 in 2000, an editorial in British Medical Journal (Rehani and Berry 2000) and the February 2001 issue of AJR, considerable attention was focused on the topic of dose management in CT. Two papers addressed the lack of appropriate exposure factors selection in pediatric CT examinations (Paterson et al.2001, Donnelly et al. 2001). Further, Brenner et al. reported on the potential risk of cancer induction from the use of CT in the pediatric population (Brenner et al. 2001). These publications note that the use of CT has significantly increased in children (for good and clinically valid reasons), but they warned that this increased usage carries with it a potential for excessive exposure to radiation and an increased risk of cancer in the pediatric population. In the editorial by Lee F. Rogers in the same issue of AJR (Rogers 2001), he stated “sorry to say, but kids get overlooked”. These reports aroused media attention and the world became aware that radiation doses in CT should be more carefully scrutinized. The number of publications on radiation exposure in CT, and management thereof, has since seen a yearly increase. Manufacturers whose main focus had been on reducing scan time started to put radiation exposure reduction on their agenda. In recent years, improved management and optimization of radiation exposure in CT has been high on the agenda for all CT manufacturers.

(..) In 2005, the Commission realized that essentially all new CT systems are MDCT, and that a number of new dose reduction tools have become available commercially. Thus, to address these new tools, continued growth in CT applications, and the consequent growth in the contribution of CT to medical collective doses, it was decided to update ICRP publication number 87 (ICRP 2000b). In addition to reviewing these technology changes in CT, a number of issues will be addressed, such as:

- has MDCT caused an increase or decrease in patient doses?
- in cases where patient doses have increased, why is this so?
- how does new technology contribute to dose minimization?
- what actions are needed by scanner operator?

- are there dose management issues to be addressed?
- are there specific educational actions still required?

(..) As in its previous report (ICRP, 2000), the primary audience for this document is imaging professionals- radiologists, radiological technologists, medical physicists and researchers involved in patient dose management. However, this document provides reference material that may be useful for physicians such as cardiologists (as many own CT scanners), regulators and national authorities, manufacturers and hospital administrators

2. RADIATION DOSE IN MDCT

There are a number of new influencing parameters specific to MDCT which systematically increase or decrease patient dose compared to single-detector row CT scanners (SDCT).

Initial reports after the introduction of MDCT indicated increased patient doses relative to SDCT; more recent reports show comparable or decreased patient doses.
If the user selects settings identical to those used in SDCT, there can be an increase in patient dose.

The increase in MDCT use has been faster than the decrease in dose per examination and changes in technology have been faster than effective implementation of dose management strategies.

2.1. Introduction

(..) It is important to distinguish between the changes to collective dose attributable to CT examinations as a result of the increased usage of CT from the changes to the radiation dose imparted to an individual from a CT examination. In the practice of medicine, the individual patient dose is typically the focus, whereas for public health administration, management and planning, information on collective dose is more relevant. In this document greater emphasis is placed on individual patient doses, presuming that for medically appropriate CT examinations, the benefit to risk ratio will be maximized when individual patient doses are reduced to levels consistent with image quality appropriate to the diagnostic task. For CT examinations where the medical justification is questionable, the societal risk becomes a larger issue because the expected benefit to the individual is likely very small, if any.

(..) Biological effects depend among other things upon the absorbed dose to tissues and organs. Since absorbed dose within the patient cannot be measured directly, a number of indirect approaches are used to estimate these doses. These estimates are made using quantities that can be measured directly in an artificial patient or test object. Using these directly measured quantities, medical physicists can estimate mean organ dose and the quantity *effective dose*. The Commission is aware that the precise definitions of dose and exposure do not make them interchangeable quantities, and that new dose quantities such as CT air kerma index are being introduced (ICRU 2006). However, as the intended audience of this document is medical professionals and not necessarily medical physicists, it was deemed most appropriate to use the term dose in this document in a more generic manner similar to as in ICRP publication number 87. A detailed description of the dose quantities used in CT is provided in Appendix A.

Similarly the effective dose has been used but the readers are referred to Appendix A for applications of its use.

2.2. Are doses in MDCT different and why?

(..) Initial reports after the introduction of MDCT indicated increased patient doses relative to SDCT, whereas more recent ones have shown comparable or lower doses for the same examination. The principal reasons for higher doses in MDCT are dose inefficiencies in the early 4-MDCT systems, the use of higher doses to decrease image noise in the thinner slices used for 3D applications, and the increased ease of scanning larger patient volumes and multiple contrast phases. In 4-MDCT systems, a large percentage of the x-ray beam width is ”wasted” when thin (< 2 mm) slices are acquired. This inefficiency becomes small, of the order of few percent, in MDCT with 16 or more detector rows. MDCT systems acquiring 16 or more simultaneous slices should be used, whenever possible, for applications requiring narrow image widths (1 mm or less) to optimize dose efficiency.

(..) When acquiring data in the spiral mode, all CT scanners require an additional rotation or so of data collection at the beginning and end of the scan in order to obtain sufficient data to reconstruct images over the prescribed volume. As the total detector width of MDCT scanners increases or the total scan length decreases, the percentage inefficiency from this effect increases.

(..) After the introduction of 4-MDCT at the end of 1998, significant attention was given to new examination strategies and scan protocol parameters. Dose measurements made on the first commercial 4-MDCT system were reported within weeks of the installation of the system and called attention to the dose inefficiency at narrow slice widths (McCollough and Zink, 1999). Depending on the slice width, doses increases up to a factor of 2 were noted for comparable noise (McCollough and Zink, 1999). Depending on the scanner model and scan acquisition settings, higher doses were reported by others as well, attributable to a shorter x-ray source to patient distance, x-ray beam profiles that were greater than the detector width, and the use of overlapping radiation beams (e.g. a pitch of 0.75). Huda and Mergo reported an increase in patient effective dose of 30% for head examinations and 150% for body examinations (Huda and Mergo, 2001). A number of other studies also reported increases in patient doses (McCollough and Zink, 1999, Giacommuzzi et al., 2001, Brix et al. 2003, Dawson, 2004, Yates et al. 2004). The recently published results of the 2003 UK CT dose survey show that there has been a reduction in average patient doses from CT examinations since the last national UK CT dose survey published in 1991 (Shrimpton et al., 2005). In that survey doses from MDCT systems were generally slightly higher than dose levels from more modern SDCT scanners, demonstrating that from the 1980s to late 1990s doses fell in general for SDCT systems as the industry abandoned the use of more inefficient gas ionization detectors. The 4-MDCT systems temporarily reversed this downward trend in dose. The initial reports of higher doses in MDCT led to the perception that doses in MDCT are higher than in SDCT. An important aspect was that the early MDCT scanners had reduced dose efficiencies due to a large proportion of the x-ray beam width not being utilised for imaging (McCollough and Zink, 1999; Lewis and Edyvean, 2005). Modern MDCT systems are more efficient in this regard; the beam width not used for imaging has been reduced to at most 2-3 mm. This results in a dose increase of just a few percent for a beam width of 20 mm and above, but a doubling or more of dose for beams of less than about 4 mm.

(..) On a SDCT scanner, tube current and scan length are often limited by x-ray tube heat capacity. This increases noise when thinner slices are scanned, prompting many users to increase the tube current to offset the increase in noise from the narrow slice width. As x-ray tube technology has evolved, MDCT scanners have been able to operate at higher power levels, allowing both faster rotation times and longer total scan times. This reduction in the constraints on the x-ray tube in MDCT offers the potential to improve diagnostic image quality, but can also lead to increased doses if care is not taken to optimise scanning protocols.

2.3. What are considerations for users switching over from SDCT to MDCT?

(..) There are a number of parameters specific to MDCT that systematically increase or decrease patient dose compared to SDCT scanners (Nagel, 2002; Prokop, 2003), as described below.

2.3.1. Factors that can increase dose in MDCT

(..) If the identical mA settings are used for MDCT that were used in SDCT, even for a scanner from the same manufacturer, there can be an unnecessary increase in patient dose. This is primarily due to differences in the distance from the x-ray tube to the patient and detector array, although differences in tube and detector design between the scanner models also play a role. This underscores the fact that the "transfer" of scanning protocols from one scanner to another should always be performed with caution, so that image quality can be maintained with similar or lower radiation dose depending on scanner characteristics.

(..) The use of narrow collimation (e.g. 4 x 1 to 1.25 mm) decreased geometric efficiency with 4-MDCT scanners and lead to an increase in dose. The increase is approximately 30-60% for 4 x 1 mm or 4 x 1.25 mm collimation (30% is still acceptable compared to the typical dose variations between scanners), but may be as high as 145% with 2 x 0.5 mm or 2 x 0.625 mm collimation. This increase is no longer present for 16-MDCT scanners.

(..) The misleading use of the term "pitch" by a number of manufacturers for 4-MDCT systems (e.g. pitch values 3 and 6 were used) implied incorrectly that patient dose was reduced accordingly. These pitch values merely characterised the improved speed of the scanners. The International Electrotechnical Commission CT Safety Standard specifically addressed the definition of pitch, re-establishing the original definition (table travel normalized to the total beam width) as the only acceptable definition of pitch (International Electrotechnical Commission, 2002) (refer to section 2.1 for further details). This has eliminated many dose errors that were the result of user confusion concerning pitch definitions.

(..) In addition, two manufacturers made use of a modified, pitch-corrected definition of mAs (*mAs per slice* or *effective mAs*) and confusion with regard to these terms led to over or under specification of the correct technique factors. For example, the term effective mAs refers to the tube-current-time-product (mAs) divided by the pitch factor. Some users find that this term makes it easier to choose a given level of noise, as pitch is already taken into account. However, confusion of the two terms may lead to a substantial increase in dose. This can occur if a user chooses to employ the same mAs settings that he previously used on a SDCT scanner of the same manufacturer. For example a 200 mAs setting at a pitch of 2 (SDCT) will correspond to 100 mAs_{eff} (multi-detector). Choosing 200 "mAs" on the MDCT scanner actually means choosing 200 $mAs_{eff,}$ which will cause a twofold increase in patient dose compared to a SDCT system (all other determinants of the dose being unchanged).

(..) Operators need to be aware that reducing slice thickness can increase the dose exponentially. If the operator fails to realize that gain in longitudinal resolution decrease partial volume averaging and hence improves contrast for small objects. Consequently, images having higher noise levels do not necessarily undermine diagnostic accuracy; rather, the contrast to noise ratio may be similar or improved. For example, if slice thickness is reduced 5 mm to 1 mm, the fraction of the x-ray intensity falling on the CT detectors is reduced by a factor of five. The noise goes up by the square root of five, or from 100% to 224%. The only way to compensate for this is to give five times the dose.

2.3.2. Factors that can decrease dose in MDCT

(..) There are at least two situations where patient dose will obviously decrease with MDCT:

(..) By scanning thin slices, one single data set is acquired which can simultaneously be used for images with either high or standard longitudinal resolution, depending on the thickness of the slice that is reconstructed. In chest examinations, one scan series instead of two (standard plus a high-resolution) is sufficient. The same holds true for generating axial, coronal and oblique images of the facial bone and sinuses by secondary reformation from the same set of spiral MDCT scan data. In these cases, the ability to obtain the needed thin and thick images (for high longitudinal spatial resolution and high low contrast axial resolution, respectively) is met with one acquisition instead of two, reducing the total dose to the patient.

(..) With increased scanning speed, facilitated by both a shorter rotation time and a wider beam, the ability to cover the entire scan volume within a single breath-hold is much improved. Thus, the incidence of motion artefacts is reduced. This benefit likely has reduced the need for repeated examinations, although this has not been documented. However, the need to overlap by several centimetres the scans that can be acquired within each breath-hold time, in order to ensure that differences in long volume at the time of breath-hold do not cause gaps in the anatomy scanned, has been eliminated with MDCT.

2.4. Dose surveys and reference levels

(..) Several surveys have been performed in recent years to document the effect of MDCT on radiation dose compared to that of SDCT (Brix et al. 2003, Origgi et al. 2006, Papadimitriou et al. 2003, Shrimpton et al. 2006, Tasapaki et al.2001, Tsapaki et al. 2006). The results of one such survey are presented in Tables 2.3 and 2.4.

Table 2.3. Effective doses from various CT examination using SDCT and MDCT (Brix et al., 2003).

Examinations	Effective dose (mSv) SDCT	Effective dose (mSv) MDCT
Abdomen and pelvis	17.2	14.4
Liver/kidney	8.7	11.5
Aorta, abdominal	7.6	10.3
Coronary CTA	-	10.5
Brain	2.8	2.8
Face and sinuses	1.1	0.8
Face and neck	2.0	2.0
Chest	6.2	5.7
Pelvis	8.8	7.2
Calcium scoring	-	3.1
Virtual colonoscopy	-	10.2
Aorta, thoracic	5.8	6.7
Pulmonary vessels	3.6	5.4
Cervical spine	2.1	2.9
Lumbar spine	2.7	8.1

Table 2.4. Comparison of radiation dose of a recent multinational study, including both- SDCT and MDCT scanners (Tsapaki et al., 2006) with a wide scale national study in UK study (Shrimpton et al., 2005) and with dose reference dose levels recommended in the EUR 16262 report (European Commission, 1999). Data in DLP (mGy.cm)

Exam	IAEA study (Tsapaki et al. 2006)	UK study (Shrimpton et al. 2005)	Reference dose level EUR 16262
Head	544	787	1050
Chest	348	488	650
Abdomen	549	472	780

(..) With increasing contributions of MDCT scanners towards collective patient radiation doses, it is important for each centre to employ certain quality control policies. These quality control initiatives must be directed towards optimizing radiation dose while maintaining image quality necessary for confident diagnosis. There is evidence to support that low radiation dose CT can provide diagnostic information necessary in several clinical situations (Kalra et al., 2004). Recent studies have shown that reference dose levels currently recommended are on the higher side and separate reference levels may be required tailored to specific requirements of clinical indications for CT as well as patient size (Tsapaki et al., 2006). Aldrich et al. found that image noise is correlated with patient weight in abdominal CT (Aldrich et al., 2006). Using a 5 point image quality score (1 to 5 with 5 as excellent) they found that at an overall image quality score of 4.5, the noise at selected points in abdominal CT was 16 HU. Using this target noise value, they determined the required tube current for each patient weight and found that the use of this technique would have reduced radiation exposure for all patients weighing less than 70 kg. The dose reduction for the smallest patient (35.4 kg) was 72%. The International Atomic Energy Agency (IAEA), through a coordinated research project (CRP) that involved six countries and nine CT scanners across the world investigated the potential for patient dose reduction while maintaining diagnostic confidence in routine chest and abdomen CT examinations in adult populations (IAEA, in press). The main objective of the project was to develop a simple methodology whereby users could determine exposure factors that could be applied to patients of different body weight, rather than depending upon the current approach of using default values based upon standard sized patient. They developed a simple mAs prediction equation to optimize radiation dose for all patient weight categories. The results showed that patient weight can be a good predictor of required dose and that an agreement can be reached for a certain noise level to be acceptable and the value could be increased for larger patients. The project also developed recommendations on how to implement the methodology for dose estimation in a CT facility.

2.5. Perspective on radiation risks

(..) *Deterministic risk.* Although CT contributes a large part of the collective dose, in some countries it amounts to 70% of the dose from medical procedures, the individual patient skin dose in a single procedure is far below that which should cause concern for deterministic injury. This is unlike interventional procedures where peak skin doses in patients have been reported to cross threshold dose for skin injuries and a number of severe skin injuries have been reported (Rehani and Ortiz López, 2006). Still the deterministic effects cannot be ruled out as a patient may undergo more than one radiological procedure. In a recent paper, Imanishi et al. 2005 reported three cases of temporary bandage-shaped hair loss which occurred in patients who had combination of perfusion studies with MDCT and cerebral digital subtraction angiography (DSA) (Fig.2.1). In all these patients two cerebral angiographies had been performed in the same period as the serial CT examinations. The possibility of such deterministic effects cannot be excluded if multiple radiological procedures are performed on the same patient.

Fig.2.1 Bandage-shaped hair loss in a 53 year old woman with subarachnoid haemorrhage. Temporary hair loss lasted for 51 days was seen on day 37 after the first perfusion study of the head with MDCT. In this patient four perfusion studies of the head with MDCT and two angiographies of the head had been performed within the first 15 days of admission to the hospital. (Reproduced with permission from author, Imanishi et al.2005)

(..) *Stochastic risk*. It is not possible to prove that a particular cancer in a patient was caused by the few tens of mGy organ doses from a few CT examinations performed earlier in the life of an individual. However, on statistical bases, the exposures encountered in CT examination may increase the risk of certain cancers, especially in children (Brenner et al., 2001). The lifetime cancer mortality risks per unit dose vary with age. The BEIR VII report states that for the same radiation in the first year of life for boys, produces three to four times the cancer risk as exposure between the ages of 20 and 50 (BEIR, 2006). Further, female infants have almost double the risk as male infants. It is important that society protects those most at risk. CT examinations in children of up to 15 years of age, in many centres, account for nearly 15 to 20% of all CT examinations and the repeat rates of CT are increasing. Since the revelation in 2001 that exposure factors in CT of children are sometimes kept the same as for adults (Paterson et al., 2001; Rogers, 2001), there has been a definite increase in awareness about the need to tailor exposure factors for children, with new tools from manufacturers assisting users in this (McCollough, 2006) and accreditation and regulatory emphasis on the absolute necessity of adjusting CT doses to patient size.

2.6. Responsibilities for patient dose management

(..) The principles of radiation protection as stated by ICRP are justification, optimisation and dose limitation (ICRP, 1990). ICRP and the International Basic Safety Standards (BSS) require generic and individual justification (ICRP, 1990; IAEA, 1996). Professional bodies normally in consultation with regulatory bodies prepare guidelines for generic justification and also for individual justification. Justification for radiation based examination such as CT is perhaps the most crucial way of avoiding unnecessary exposure and thus a powerful radiation protection tool. Justification of an examination is the starting point and this issue is dealt with in Section 4.1. It is widely believed that many unjustified exposures are made both in developing and developed countries. There is a lack of published information on how much exposure from unjustified use of CT is occurring and how much of that can be saved through different actions. Professional societies of referring physicians and of radiology should work together with medical physics experts to survey the practice, estimate the magnitude of unjustified usage and evolve strategies for avoidance of unjustified exposures. In contrast to justification, optimization on the other hand has received great attention and there is substantial amount of information that is available in literature on the magnitude of the dose reduction that can be achieved through optimization actions. This publication itself contains a review of several of such reports. There is a need to achieve consensus among professional societies and provide recommendations. There are good reports from the Royal College of Radiologists (UK) and the American College of Radiology (USA) that provide justification for choosing particular examination over others and in what order depending upon the clinical situation (ACR 2000, RCR 2003).

(..) The ICRP and the International Basic Safety Standards (BSS) have always maintained that the system of dose limitation should apply to occupational exposures where they have specified appropriate dose limits (IAEA 1996, ICRP 1991). As far as patients are concerned, no dose limits are applicable and the exposure is to be kept as low as reasonably achievable through process of justification and optimization while achieving the desired clinical objective. There is no change on this policy.

(..) *What are the responsibilities of manufacturers*? Equipment design and compliance with applicable International Standards and National Regulations are the responsibility of the manufacturers. Unfortunately manufacturers did not consider radiation dose to patient an important issue until the media highlighted the issue (see Section 3.1). An editorial in AJR (Rogers, 2001) drew the attention of manufacturers stating "Equipment manufacturers should engage themselves in a campaign to see that CT in children is accomplished with the lowest possible radiation dose. This does not likely require any significant changes in hardware, if indeed it should necessitate any hardware changes at all. And, for that matter, no change or addition to the software should be necessary either. No purchase of a "paediatric package" should be required. The technician or radiologist should be able to accomplish the desired reduction in radiation dose simply by selecting the correct exposure factors. Manufacturers should see that this is available if they have not already done so". Manufacturers have certainly an important role to play and it is noted that following the ICRP publication 87 in 2000 and a number of publications in AJR in 2001, radiation dose in CT became an agenda for manufacturers because of the media attention these publications evoked. It is seen that every manufacturer is now showing consciousness to radiation dose to the patients and this emphasis is all the more important with increasing use of MDCT. Although manufacturers have accomplished commendable work in developing automatic exposure control (AEC, please see Section 3.3.2.1 in this publication) techniques, much work remains undone both by users and manufacturers in terms of defining the reference image quality for different diagnostic tasks.

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3. OPERATOR CHOICES THAT AFFECT PATIENT DOSE

As in earlier developments in CT, there is potential for dose reduction, but the actual dose reduction depends upon how the system is used.

It is important that radiologists, medical physicists and CT system operators understand the relationship between patient dose and image quality and be aware that often image quality in CT is more than what is needed for diagnostic confidence.

It must be remembered that "pretty" pictures are not essential for all diagnostic tasks, but rather a level of quality will need to be chosen – whether low noise, standard, or low dose, dependent on the diagnostic task.

Objective measures such as image noise or contrast-to-noise ratio may not completely capture all of the features relevant to making a correct clinical diagnosis. Thus, determining "optimal" image quality can be a complex task, as both quantitative metrics (e.g., noise) and observer perceptions are involved.

There are indications that awareness on adapting exposure factors to manage patient dose is increasing but the rate at which technology is changing overtakes adoption of effective dose management.

(..) MDCT represents state-of-the-art CT technology and offers a number of technical measures for dose reduction, the most important of which is Automated Exposure Control (AEC). AEC is analogous to photo-timing in general radiography, where the user determines the image quality (e.g., noise or contrast-to-noise ratio) requirements, and the imaging system determines the right mAs.

3.1. Tradeoffs between dose and image quality

(..) Excessive dose reduction can adversely affect image quality and decrease lesion detectability. Likewise, the visibility of lesions on "pretty pictures" acquired at higher doses is not necessarily greater than that on lower dose CT images (Kalra et al., 2004). Finally, an understanding of CT acquisition and reconstruction parameters can aid the radiologist, medical physicist and operator in maintaining image quality while imparting low doses to the patients.

3.1.1. General descriptors of image quality

(..) Image quality is a very broad term in the context of CT scanning. It may include several aspects that are related to radiation dose such as those which change the exposure. Some aspects of image quality such as motion artefacts are not related to patient dose. When motion artefacts are separated, image noise and image contrast are the most important descriptors of image quality. Image noise, or quantum mottle, is most directly related to the radiation dose used for CT scanning. An increase in radiation dose typically decreases noise and vice-versa. Image noise can be quantified as the standard deviation of the CT number (in Hounsfield units) and used for optimization of radiation dose and image quality. Image noise is specifically important for the detection of low contrast lesions, which may be obscured in by higher levels of image noise. On the other hand, studies such as chest CT, CT colonography and kidney stone protocol CT, have higher lesion-to-background contrast and therefore, higher noise can be accepted to reduce radiation dose. It is important to remember, however, that subjective acceptability of image quality in small patients (such as children) and large patients varies considerably at identical image noise level. These relationships between study indications, patient size, and image noise dictate that each CT imaging centre must have separate protocols based on patient size and study indications rather than a "one-size-fits-all" approach. CT manufacturers allow storing multiple labelled protocols on the scanner console that can be recalled within a few seconds. Thus optimized protocols for many different patient sizes and indications can be easily created and stored.

(..) Image contrast is determined by a more complex relationship to the scan and reconstruction parameters. It is dependent on the x-ray tube potential (kVp), but is independent of the photon fluence (mAs). A decrease in kVp can decrease radiation dose but increases image contrast, whereas an increase in kVp decreases image contrast. Image noise and image contrast can be used to adapt scanning parameters for managing radiation dose. In fact, different scanning options can be adopted for reducing radiation dose based on inherent contrast of structures in the scan region of interest.

3.1.2. Different imaging tasks require different level of quality

(..) In typical situations with high contrast, such as CT colonography and non-contrast-enhanced scan of the abdomen and pelvis for kidney stone evaluation, a lower dose CT can be performed because the increased noise levels do not affect lesion conspicuity due to their high inherent contrast (Iannaccone et al., 2005; Kalra et al., 2005a). Likewise, several studies have recently explored use of low kVp for CT angiography protocols, demonstrating that the high contrast between contrast-enhanced blood vessels and their surrounding structures allow evaluation even with high noise levels (Funama et al., 2005; Holmquist and Nyman, 2006). Further, routine chest CT studies should be performed with the use of lower radiation dose due to the high inherent contrast between air filled lungs and soft tissues (as well as less x-ray beam attenuation in the thorax compared to abdomen) (Kalra et al., 2005b). Conversely, in situations with low contrast between lesions and background structures, such as most liver metastases, increased image noise can affect lesion detection and/or characterization. In such circumstances, inadvertent dose reduction and higher noise levels may compromise the diagnostic acceptability of the study.

(..) While it may be prudent to reduce radiation dose particularly for young patients with benign diseases, a standard dose CT is most appropriate in life threatening situations or for patients with possible malignant diseases, where the risk of misdiagnosis from a low dose CT is much greater than the statistical risk of a radiation-induced cancer.

3.1.3. Differences on choice of CT parameters and perception of image quality?

(..) As a result of concerted action in Europe through number of projects of European Commission (EC) and of IAEA, there has been considerable attention to radiation dose optimization in radiology (Brix et al. 2003; IAEA in press; Tsapaki et al. 2001; Tsapaki et al., 2006). An IAEA study demonstrated different image quality requirements and preferences of radiologists in different countries (IAEA in press). In addition to the variation between radiologists' perception of image noise, patient related factors (U.S. patient distribution is typically of greater weight than in a European or Asian population) may also contribute to variation in setting up of scanning protocols. Surveys from the United States suggest that there are considerable variations between scan parameters and associated radiation doses between different scanning centres, although image quality and dose assessment programmes, such as that offered by the American College of Radiology (McCollough et al., 2004) have been effective in reducing these variations.

3.2. Equipment and protocol issues affecting patient dose

3.2.1. Overbeaming

(..) *Overbeaming* is the general term used to describe the situation when the x-ray beam incident to the patient extends beyond the active detector area and is hence not used for imaging purposes. In single-detector CT, this occurs when slice collimation is positioned between the patient and the detector in order to improve the slice sensitivity profile. This situation delivers a dose to the patient that does not contribute to image formation, and occurrs commonly for very narrow image widths (less than 2 mm) and occasionally for thicker image widths. With single-detector CT, however, there is no absolute need to exclude the penumbral (gradient) portion of the x-ray beam from the imaging detector. Consequently, most single detector scanners make full use of the entire x-ray beam or dose profile (Fig. 3.1a), at the expense of some degradation of the quality of the slice profile.

(..) With multi-detector CT, the radiation incident to the patient must be uniform across all active detector rows. Consequently, penumbra must be either totally or partially excluded from the useful beam (Fig. 3.1b). This requires that the width of the x-ray beam be increased to allow the penumbral region to fall beyond the active detector area. This is true for all multi-detector scanners with more than 2 simultaneously acquired slices. For dual-detector scanners there was no absolute need to 'overbeam' (Fig. 3.1c) provided that the total width of the detector array was sufficient to capture the penumbra. Nevertheless, overbeaming is found on the many dual-detector scanners.

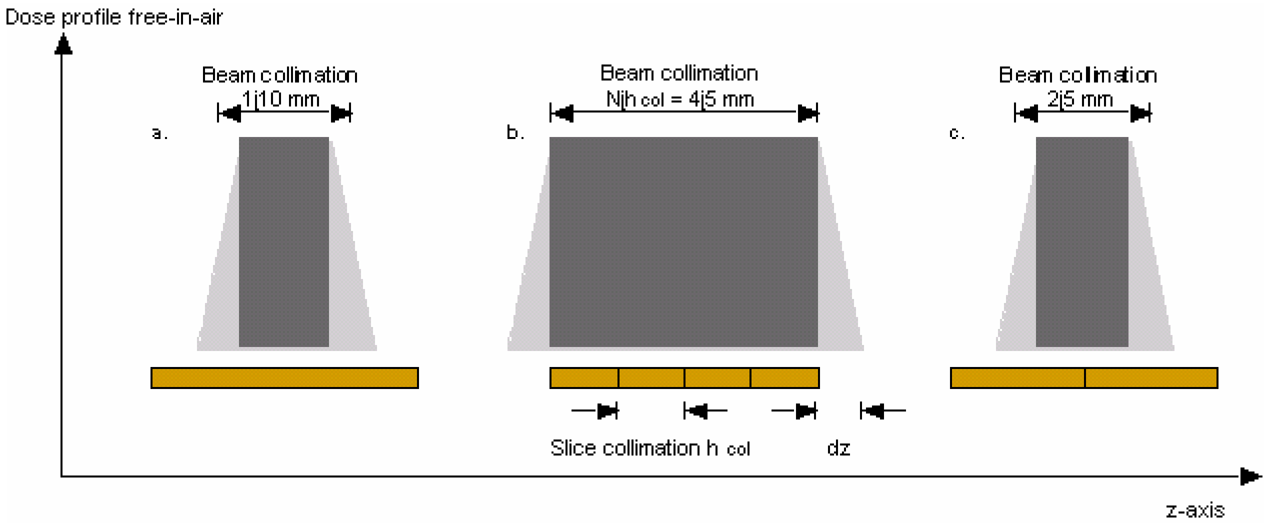

Fig. 3.1. Dose profiles free-in-air with umbra (dark grey) and penumbra (light grey) portions for a single-detector scanner (a.), a 4-detector scanner (b.), and a dual-detector scanner (c.). With single- and dual-detector scanners, the width of the active detector rows is sufficient to capture the entire dose profile, including the penumbra. For MDCT scanners with $N \geq 4$, penumbra is excluded from detection in order to irradiate all detector channels with equal values of incident irradiation. The combined width of the penumbra triangles at both sides is denoted by the overbeaming parameter dz (Actually dz is the total width which is obtained by half on both sides) (Nagel 2005)

(..) The dose consequence of overbeaming is largest when the total beam width is small. The worst case is found for single-detector scanners and 1-mm slice collimation when post-patient collimation is used. Though the overbeaming parameter dz (as depicted in Fig. 3.1) for single-detector scanners is relatively small (typically 1 mm), a 100% increase in dose results. Systems having a larger number of data channels (i.e., a greater number of slices that can be simultaneously acquired), can acquire narrow images while exposing a greater extent of the total detector width. The extent of overbeaming dz is larger about 3 mm for most multi-detector scanners (McCollough and Zink, 1999; Nagel, 2005). Generally, wider beam collimation in MDCT results in more dose efficient examinations, as overbeaming constitutes a relatively smaller proportion of the detected X-ray beam. However depending on the scanner model, wide beam collimation may limit the thinnest slices that can be reconstructed.

(..) Pre-patient control of x-ray tube focal spot motion and beam collimation improves scanner dose efficiency and thus reduces radiation exposure. This technique reduces overbeaming by measuring the position of the beam every few milliseconds and repositioning the collimating aperture as necessary. This allows a narrower x-ray exposure profile compared to systems with no focal spot tracking. All currently manufactured MDCT systems employ some mechanism for accomplishing this objective.

3.2.2. Overranging

(..) In spiral CT, data interpolation between two points must be performed for all projection angles (Fig. 3.2). Thus, the images at the very beginning and end of a spiral scan require data from z-axis projections beyond the defined "scan" boundaries (i.e. the beginning and end of the anatomic range over which images are desired). Commonly, an additional half rotation is needed at the beginning and at the end of the spiral scan, so the total number of additional rotations is 1.

(..) *Overranging* is the general term used to describe this increase in dose-length product due to the additional rotations required for the spiral interpolation algorithm. For MDCT scanners, the number of additional rotations is strongly pitch dependent, and the increase in irradiation length is typically 1.5 times the total beam width.

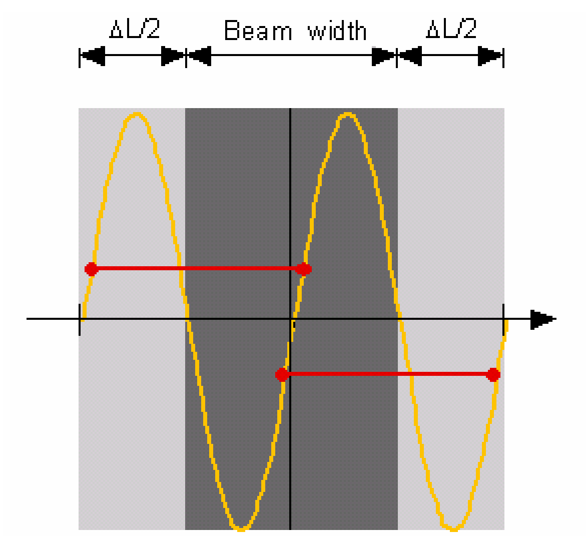

Fig. 3.2. Overranging for the special case of a single detector acquired in spiral scanning mode with a 360° interpolation algorithm and pitch 1. In general, half an extra rotation is required both at the beginning and at the end of the scan, thus causing an increase ΔL in scan length. ΔL itself varies depending on the selected pitch. (Nagel, 2005).

(..) The implications of overranging with regard to the dose-length product DLP (see Appendix A) depends on the length of the imaged body region. For spiral scans that are short relative to the total beam width, the dose efficiency (with regard to overranging) will decrease. Additionally, it is generally more dose efficient to use a single helical scan than multiple helical scans.

3.2.3. Slice thickness

(..) MDCT technology allows for the reconstruction of relatively narrow image widths in total scan times that are comparable or shorter than in single-detector CT. With 64-MDCT systems, for example, a typical adult male can be scanned from head to toe with sub-millimeter detector collimation in under 20 seconds. The detector collimation, however, must not necessarily be identical to the thickness of the reconstructed images. Thicker images, which are less noisy, can be generated from the thinner projection data. Nevertheless, the typical image thickness (typically 3 to 5 mm) is still smaller than those used with single-detector scanners (5 to 8 mm). Consequently, users may be tempted to compensate for the increased noise associated with thinner images by using elevated doses.

(..) With the reduction in image thickness, the magnitude of partial volume averaging also decreases. Thus, the CT number (image brightness) associated with objects that occupy less than one voxel increases as the voxel size decreases. For objects with z-axis dimensions less than one image width, the contrast of the object improves with reduced slice thickness, whereas quantum noise increases with reduced slice thickness. If a narrow image thickness is used, the contrast-to-noise ratio (CNR) and visibility of small details can improve despite increased noise (Wedegartner et al., 2004).

(..) In Fig. 3.3, images having 3, 5, 7, and 10-mm widths have been reconstructed at the same z-axis position from the same data (acquired with a 4-detector scanner and a detector collimation of 2.5 mm). The visibility of a liver lesion (approximately 3 mm in size) diminishes continually with increasing slice thickness – despite reduced image noise. Partial volume averaging is not restricted to objects which are smaller than the slice thickness, but is always involved due to the irregular shape and orientation of lesions, vessels, etc.

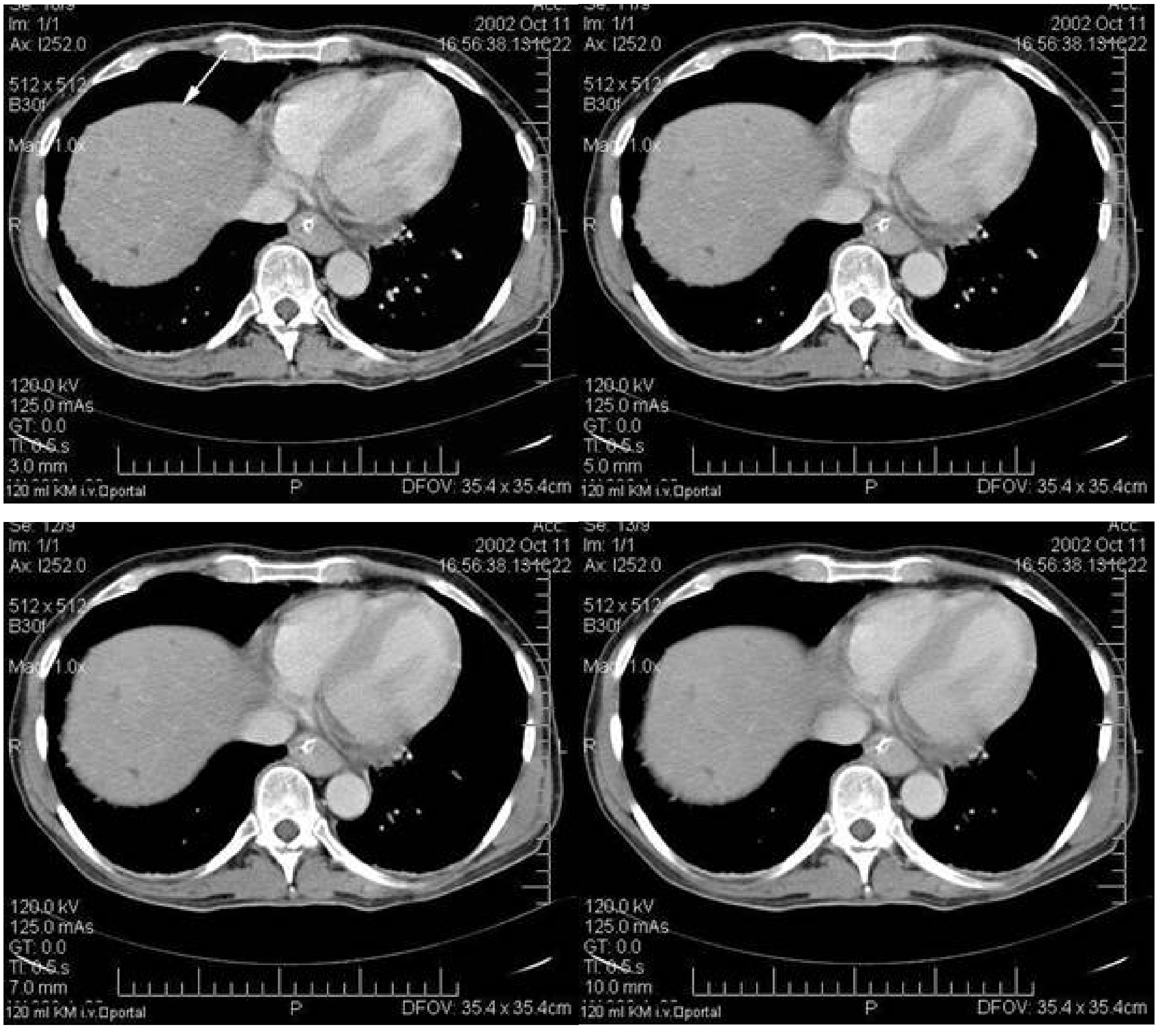

Fig. 3.3. MDCT examination of the liver performed at 120 kV, 4 x 2.5 mm slice collimation and 125 effective mAseff (CTDIvol = 11 mGy). From the same raw data set, slices of different thickness [3 mm (upper left), 5 mm (upper right), 7 mm (lower left), and 10 mm (lower right)] were reconstructed centered at the same position. Despite the increased noise pertaining for thinner slices, the visibility of small lesions improves remarkably owing to reduced partial volume effects. This is clearly demonstrated by a lesion approximately 3 mm in size (arrow) (courtesy of Dr. Wedegaertner, University Hospital Eppendorf, Hamburg, Germany).

(..) Consequently, if thinner image widths are required for multi-planar or 3-D reformations, or to reduce partial volume averaging, dose need not necessarily be increased to obtain the same image noise as achieved with thicker images widths. When reformations or partial volume averaging are not of concern, thinner images should be combined in order to reduce noise. With the advent of CT workstations that allow the user to manipulate the image thickness in real time (e.g., thick slab multi-planar reformations or maximum intensity projections from thin axial images), one can efficiently view thin images, in order to reduce partial volume averaging, and thicker images, to reduce image noise.

3.3. Operator choices that affect patient dose

(..) The technologist and/or radiologist monitoring the scan have control over several scan acquisition parameters that can be adjusted to obtain the desired image quality at optimum dose. Since increased exposure factors that result in better looking images go undetected there is considerable scope for the operator to perform the CT examination at a higher dose than necessary. As a result wide variations are observed in nationwide surveys even among those using the same CT system (Brix et al. 2003; Nagel et al. 2004; Shrimpton et al. 2005)

3.3.1. Scanner model and manufacturer

(..) There is a considerable difference between geometries of CT scanners that affect the distance between the focal spot of the x-ray tube and the center of rotation (isocenter) of the scanner. Differences also exist in filtration of the x-ray beams, efficiency of detection systems, noise levels in data acquisition electronics, and reconstruction algorithms. Thus, the image noise obtained at a given mAs, kVp and image width on one scanner model may differ considerably from that on another scanner model.

(..) While these attributes of a system are not “operator selectable”, it is not uncommon for large medical centers to have two or more scanners of different models, perhaps from different manufacturers. Thus, when scan protocols are prepared for a CT system, it is important to be cautious about that the “transfer” of parameters from one scanner model/manufacturer to another. Careful “migration” of protocols helps in maintaining image quality at similar or reduced radiation doses, depending on the scanner models being used.

3.3.2. Tube current (mA) and tube current-time product (mAs)

3.3.2.1 Manual (technique charts)

(..) Unlike traditional radiographic imaging, a CT image never looks “over-exposed” in the sense of being too dark or too light; the normalized nature of CT data (i.e., CT numbers represent a fixed amount of attenuation relative to water) ensures that the image always appears properly exposed. As a consequence, CT users are not technically compelled to decrease the tube-current-time product (mAs) for small patients, often resulting in excess radiation dose for these patients. It is, however, a fundamental responsibility of the CT operator is to take patient size into account when selecting the parameters that affect radiation dose, the most basic of which is the mAs (FDA, 2002; Linton and Mettler, 2003).

(..) As with radiographic and fluoroscopic imaging, the operator should be provided with appropriate guidelines for mAs selection as a function of patient size. These are often referred to as technique charts. While the tube current, exposure time and tube potential can all be altered to give the appropriate exposure to the patient, in CT, users most commonly (and appropriately) standardize the tube potential (kVp) and gantry rotation time (s) for a given clinical application. For example, the fastest rotation time is typically used to minimize motion burring and artifact, and the lowest kVp consistent with the patient size should be selected to maximize image contrast (Funama et al., 2005; McCollough et al., 2006; Nakayama et al., 2005).

(..) Although scan parameters can be adapted to patient size to reduce radiation dose, it is important to remember certain caveats when contemplating such adjustments. Firstly, body regions such as the head do not vary much in size in the normal population, so modification of scan parameters may not be applicable here based on head size. Secondly, recent studies have shown that there is poor correlation between patient size, image noise and mAs in chest CT studies (IAEA in press, Prasad et al. 2003). Several factors may be responsible for this aberration such as very little x-ray beam absorption by the lungs irrespective of their size, complex anatomic interfaces and motion patterns in chest compared to other body regions, and specific properties of reconstruction algorithms used for chest CT images. The poor correlation between patient size and mAs may lead to overestimation of radiation dose for chest CT, particularly for large patients, if size-based adjustment of scan parameters is performed.

(..) Numerous investigators and users have shown that the manner in which mA should be adjusted as a function of patient size should be related to the overall attenuation, or thickness, of the anatomy of interest as opposed to patient weight, which is correlated to patient girth, but not a perfect surrogate as a function of anatomic region (Boone et al., 2003; McCollough et al., 2002; Wilting et al., 2001). The exception is for imaging of the head, where attenuation is relatively well defined by age, since the primary attenuation comes from the skull, and the process of bone formation in the skull is age dependent.

(..) Clinical evaluations of mA-adjusted images have demonstrated that radiologists do not find the same noise level acceptable in small patients as in larger patients (Wilting et al., 2001). Because of the absence of adipose tissue between organs and tissue planes, and the smaller anatomic dimensions, radiologists tend to demand lower noise images in children and small adults relative to larger patients (Boone et al., 2003; Kalra et al., 2004; McCollough et al., 2002; Wilting et al., 2001). For body CT imaging, typically a reduction in mA (or mAs) of a factor of 4 to 5 from adult techniques is acceptable in infants (McCollough et al., 2002). For obese patients, an increase of a factor of 2 is appropriate (McCollough et al., 2002). For neurological CT imaging, the dose reduction from an adult to a new born of approximately a factor of 2 to 2.5 is appropriate. Sample technique charts are provided in Table 3.1 and 3.2. For body imaging (Table 3.2), the values are normalized to the mA values used in a standard adult (80 kg, 35 – 40 cm lateral width at the level of the liver). In neurological imaging, age is the preferred indicator of head attenuation. While typical ages are given for the sample body imaging technique chart, actual patient dimension is the preferred indicator of actual patient attenuation, as patient size and hence attenuation can vary markedly in the body for patients of the same age. To achieve increased exposure for obese patients, either the rotation time, or the tube potential, may also need to be increased. Importantly, if consistent compliance with the use of technique charts is not achieved in daily practice, the dose benefits are lost. Thus, methods of automating these adaptations to patient size have been investigated and implemented as discussed in the following section.

Table 3.1: Sample Technique Chart for Neurological CT (adapted from McCollugh 2002 and Boone 2003)

Age	Image width (mm) x Image Increment (mm)	Relative mAs
0 - 6m	3x3	0.46
6.1 - 18m	3x3	0.57
18.1m - 3yr	3x3	0.61
3.1 yr - 10 yr	3x3	0.73
over 10 yr	5x5	1.00
0 - 6m	7x7	0.40
6.1 - 18m	7x7	0.50
18.1m - 3yr	7x7	0.54
3.1 yr - 10 yr	7x7	0.65
over 10 yr	10x10	1.00

Table 3.2: Sample Technique Chart for Body CT (adapted from McCollugh 2002 and Boone 2003)

Approximate age (year)	Lateral patient width (cm) at level of the liver	Relative mAs
newborn	Up to 14	0.16
1	14.1 – 18	0.22
5	18.1 – 22	0.29
10	22.1 – 26	0.38
15	26.1 – 30	0.50
adult	30.1 – 35	0.71
adult	35.1 – 40	1.00
adult	40.1 – 45	1.42
adult	45.1 – 50	2.00

3.3.2.2 Automated exposure control (AEC)

(..) Extremely large variations in patient absorption occur both with projection angle and anatomic region. Since the projection with the most noise primarily determines the noise of the final image, it is possible to reduce dose (photons) for other projections without increasing the noise in the final image. This concept was introduced in 1981 by Haaga et al. (Haaga, 2001). In 1994 General Electric Medical Systems made available the first commercial mA modulation system, with dose reductions up to 20% (Kopka et al., 1995). Additionally, Kalender et al. reported on dose reductions up to 40% in elliptical body regions using anatomically-modulated mA (Gies et al., 1999; Kalender et al., 1999). Additional mA modulation products became available in late 2001, when, due in part to the public concerns over dose, dose reduction became a priority for purchasers of CT systems (Rehani and Berry, 2000; ICRP, 2000).

(..) Modulation of the x-ray tube current during scan acquisition is a very effective method of managing dose in CT. The modulation may occur angularly about the patient, along the long axis of the patient, or both. And, the system must use one of several algorithms to automatically adjust the current to achieve the desired image quality.

<u>Angular (x,y)</u>

(..) Angular (x,y) mA modulation addresses the variation in x-ray attenuation around the patient by varying the mA as the x-ray tube rotates about the patient (e.g., in the anterior-posterior. versus lateral direction) in order to equalize the photon flux to the detector. The operator chooses the initial mA(s) value, and the mA is modulated upward or downward from the initial value with a period of one gantry rotation.

(..) Kalender et al. demonstrated a decrease in shoulder streaks when the mA is increased through the shoulders such that the projection noise level is more uniform between anterior/posterior (AP) and lateral projections (Kalender et al., 1999). Some implementations, however, do not allow the mA to exceed the initial value prescribed by the operator. As the x-ray tube rotates between the AP and lateral positions, the mA can be varied sinusoidally, prospectively according to the attenuation information in the CT localizer radiograph, or in near real-time according to the measured attenuation from the 180° previous projection.

Longitudinal (z)

(..) Longitudinal (z) mA modulation addresses the varying attenuation of the patient among anatomic regions by varying the mA along the z axis of the patient (e.g., shoulders versus the abdomen versus the pelvis). Unlike angular mA modulation, where the mA is varied in a relatively cyclical fashion relative to the starting mA value, the task of z modulation is to produce relatively uniform noise levels across as the various regions of anatomy. Thus, the operator must provide as input to the algorithm the desired level of image quality. The methods used for this task differ considerably from manufacturers. For example, some manufacturers have the user indicate a reference noise value, or noise index, while others ask the user to indicate a reference effective mAs value or image data set (Kalra et al. 2005; McCollough et al., 2006). Details regarding these image quality selection paradigms are provided below.

(..) The mA is modulated to provide the desired level of image quality as the attenuation between anatomic regions varies. Because the tube current is adapted per gantry rotation, the exposure setting no longer needs to be fixed over the longitudinal scan range in a manner that satisfies even the most challenging portion of the scan range (e.g., the shoulder region in chest CT examinations). Dose reductions of up to 50%, depending on the type of examination and the default mAs settings, are thus achieved.

(..) Longitudinal dose modulation aims to ensure an appropriate noise level regardless of the local attenuation properties. By doing so, dose may be increased, for example, when proceeding from the upper abdomen to the pelvis in examinations of the entire trunk. Both an appropriate noise level and changes in inherent contrast should be taken into account in longitudinal modulation schemes, since in the pelvis, for example, the noise may be increased due to the improved inherent contrast (International Commission on Radiological Protection, 1991). The same holds true for the thorax region, where a dose reduction is possible not only as a result of the reduced attenuation, but also because of the high contrast characteristics of the chest.

Combined Angular and Longitudinal (x,y,z)

(..) Combined angular and longitudinal (x,y,z) mA modulation adds the previous two methods to vary the mA both during rotation and during longitudinal movement of the patient through the x-ray beam (i.e., anterior/posterior versus lateral *and* shoulders versus abdomen). The operator must still indicate the desired level of image quality. This is the most comprehensive approach to CT dose reduction because the x-ray dose is adjusted according to the patient attenuation in all three planes. An example of this approach is shown in Figure 3.4 for a 6-year-old child.

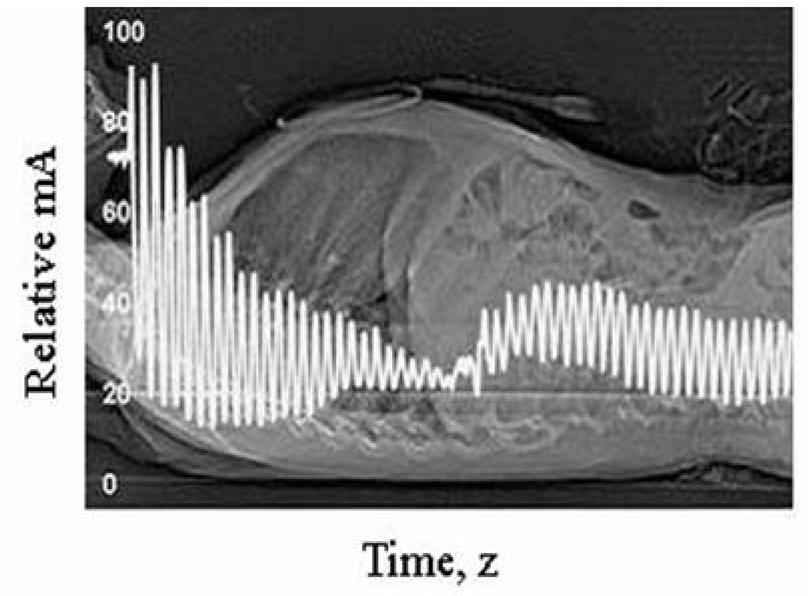

Figure 3.4. Tube current as a function of time (and hence position) for a spiral examination of a 6 year old child scanned with an adult protocol and an AEC system (CareDose 4D). Reference effective mAs* (mAs/pitch) = 165. Mean effective mAs of actual scan = 38. (courtesy of C. McCollough (McCollough et al., 2006).

(..) For longitudinal dose modulation component of the x,y,z dose modulation approach, the attenuation of the patient is measured in one direction (x or y) and estimated for the perpendicular direction with a mathematical algorithm from a single CT localizer radiograph. These attenuation profile measurements contain information regarding the patient's size, shape and attenuation at each z-axis position. Based on these profiles, tube current values are calculated for each gantry rotation. Tube current adjustment is based on a user defined image quality reference in order to maintain the desired image quality along the longitudinal direction. This is essentially the same process as for z modulation only. The technique then modulates these z-axis tube current values during each tube rotation according to the patient's angular attenuation profile (i.e. using the algorithm for x,y modulation). Dose reductions of up to 40-60%, depending on the type of examination and the default settings of image quality, are thus achieved (Kalra et al., 2004; McCollough et al., 2006; Mulkens et al., 2005; Rizzo et al., 2006).

Automatic exposure control (AEC) systems do not reduce patient dose per se, but enable scan protocols to be prescribed using measures related to image quality. If the image quality is appropriately specified by the user, and suited to the clinical task, then there is reduction in patient dose for all but obese patient. In obese patients, the dose is increased to improve the image quality.

AEC does not imply total freedom from operator selection of scan parameters. While CT systems without AEC require operator selection of mA, AEC systems require understanding of newer concepts such as noise index, reference mAs and reference images in order for AEC to be operated effectively. Understanding of some parameters e.g. the standard deviation of image pixels or noise index, is not intuitive and entails chances of error.

3.3.3. Image quality selection paradigms

(..) The use of an image quality selection paradigm allows the CT system to calculate the appropriate tube current values, as a function of angle and z-axis location, in order to deliver the desired image quality at the lower dose. This broad concept, implemented practically with some variation between manufacturers, is known as Automatic Exposure Control (AEC). In practice, it is relatively straightforward for the system to deliver the desired image quality (once defined). However, it can be quite difficult to determine the image quality requirement for various CT applications and patient sizes.

(..) In defining the required image quality, the user needs to remember that “pretty” pictures are not essential for all diagnostic tasks, but rather a level of quality will need to be chosen – whether low noise (higher dose), standard, or higher noise (low dose)-dependent on the diagnostic task. The CT system will then adjust the mA either during the rotation (x,y), along the z-direction, or in all three dimensions (x, y and z) according to the patient’s body habitus and the user’s image quality requirements. Thus one must differentiate between task of modulating the mA to achieve a defined image quality (tube current modulation) and the actual prescription by the user of the desired image quality.

The selection of image quality parameters in AEC systems is not a straightforward process. There is a lack of consensus on how image quality is to be specified; with the result that there are significant differences in the way different companies achieve exposure control. It is important that users are aware of the behaviour of their system.

(..) Each manufacturer of CT systems uses a different method of defining the image quality in the user interface. GE uses a concept known as the *Noise Index*. The noise index is referenced to the standard deviation of pixel values in a specific size water phantom. A "look-up-table" maps the patient attenuation measured from the CT localizer radiograph (Scout) image into mA values for each tube rotation according to a proprietary algorithm. This algorithm is designed to maintain the same image noise as the patient's attenuation changes from one rotation to the next. A different noise index may be required for different patient sizes and study indications (Kalra et al., 2003).

(..) Philips uses a *Reference Image* concept to help users select the desired image quality that should be matched. They refer to this as Automatic Current Setting (ACS). The user saves an acceptable patient exam, including the CT localizer radiograph (SurView), and the system saves the raw data. This information is saved as the Reference Case, on a protocol by protocol basis, to be matched on later exams using their proprietary algorithm.

(..) Siemens uses a *Quality Reference mAs* for users to establish the desired image quality level. The user selects, on a protocol by protocol basis, the effective mAs (mAs/pitch) typically used for an approximately 80 kg patient. (For pediatric protocols the *Quality Reference mAs* should be chosen for a 20 kg patient.) The noise target (standard deviation of pixel values) is varied on the basis of patient size using an empirical algorithm; hence image noise is not kept constant for all patient sizes but is adjusted according to the radiologist's impression of image quality. The CT localizer radiograph (Topogram) for individual patients is used to predict the mA curve (with variations in x, y, and z) that will yield the desired image quality for the specific patient size and anatomy. An on-line feedback system adjusts the actual mA value during the scan acquisition to match the precise patient attenuation at all angles, as opposed to the attenuation estimated by the one angle.

(..) Toshiba offers the user two methods of selecting the desired image quality in their *Sure Exposure* AEC product: Standard Deviation or Image Quality Level. Both are referenced to the standard deviation of pixel values in an attenuation-equivalent water phantom, which is created from the CT localizer radiograph (Scanogram) data. The reference value, index, or image can be stored with specific protocols on all manufacturers' implementations.

Assumptions regarding optimal noise levels

(..) Image quality is a non-specific measure of the subjective sense of "quality" of an image, which must be assessed by a trained observer. Objective measures such as image noise or contrast-to-noise ratio can be made relatively easily, but may not completely capture all of the features relevant to making a correct clinical diagnosis. Thus, determining "optimal" image quality can be a complex task, as both quantitative metrics (e.g., noise) and observer perceptions are involved. One straightforward approach is to require a specific noise level for a specific diagnostic task.

(..) Table 3.3 provides measured noise for a constant mAs (chosen to be 130) as the diameter of a water phantom was varied. Table 3.4 demonstrates the mAs required to yield a constant image noise (chosen to be 13.0 HU) as the diameter of a water phantom was varied. Tables 3.3 and 3.4 together demonstrate that it is not technically feasible to maintain a constant image noise over all patient sizes, even if this was clinically desired, because CT systems cannot reach these extremely low and high mAs values. The large range of mAs values required to maintain constant image noise as object size is varied is a consequence of the exponential nature of x-ray absorption.

Table 3.3: Measured noise for a constant mAs (130) as the diameter of a water phantom is varied (adapted from McCollugh 2002 and Boone 2003)

Diameter (cm)	10	14	20	25	30	40
Noise (HU)	1.9	3.5	5.1	8.2	13.0	33.6

Table 3.4. Tube current time product (mAs) required for a constant image noise (13.0 HU) as the diameter of a water phantom is varied (from McCollugh 2006)

Diameter (cm)	10	14	20	25	30	40
Tube current-time product (mAs)	2.9	6.2	19	50	130	869

(..) With empirically determined technique charts (i.e., appropriate mA values are determined for each patient size by a trained observer), both the extreme low and high mAs requirements are noticeably absent (see Tables 3.1 and 3.2). This is not only pragmatic with regard to the x-ray generator, it provides a more appropriate technique selection from both a patient dose and image quality perspective (compared to the criterion of having a fixed noise across all patient sizes). More aggressive dose reduction is not acceptable in children, and more aggressive dose increase is unnecessary. (Wilting et al., 2001). When Wilting et al. presented images with constant noise to radiologists for a variety of patient sizes, ranging from pediatric to obese patients, the pediatric images were found to be unacceptable, even though they contained the same level of image noise as normal and obese patients (Wilting et al., 2001). Kalra et al. observed a similar situation using the General Electric noise index paradigm, which for a given noise index attempts to deliver a constant noise across anatomic regions and patient sizes (Kalra et al., 2003). They found that readers required a lower noise index (less image noise) for smaller patients and a higher noise index (more image noise) for larger patients. Although lower image noise was found to be required for small patients, a dramatic level of mAs reduction is still appropriate to compensate for the decreased patient attenuation.

3.3.4. Temporal mA modulation

(..) Temporal mA modulation alters the tube current according to a time-based criterion. This is most-commonly used in CT examinations of the heart, reducing the dose for projections of limited interest, such as in early systole where the rapid cardiac motion compromises image quality. Based on the heart rate, such an ECG-based mA modulation scheme can reduce dose by up to 50% for a cardiac CT study for systems with one x-ray tube (Jakobs et al., 2002), and even more for dual-source systems (Flohr et al., 2006; McCollough et al., 2005).

(..) Usually, the tube current required for acceptable image quality is used for a time window that is somewhat wider than the desired temporal resolution (e.g., 330 to 350 ms time window for a 250 ms temporal resolution) in order to allow for some flexibility in the case of irregular heart rate. The window is centred over the cardiac phase desired for image reconstruction. Outside this time window, the tube current is not completely switched off, but is reduced to a much lower level (e.g., 20% of the required tube current). This ensures that data is available to perform dynamic studies over the entire heart cycle, although at increased noise outside of the time window selected for primary image reconstruction. However, in patients with higher heart rates (more than 60-65 beats per minute and irregular heart rates (premature ventricular contractions), where systolic or multiple reconstructions may be needed for primary interpretation, ECG based mA modulation will yield much noisier images in non-diastolic phases. Since the length of the data window is fixed, the dose reduction achieved by this feature depends on the heart rate.

3.4. Tube potential (kVp)

(..) Tube potential (kVp) determines the energy of the incident x-ray beam. Variation in the tube potential causes a substantial change in CT dose as well as image noise and contrast. In children and small adults, reducing the kVp leads to a dose reduction for a desired contrast to noise ratio, relative to higher kVp values (Funama et al., 2005; Huda et al., 2000; McCollough, 2005; Nakayama et al., 2005; Siegel et al., 2004). Most MDCT examinations are performed at either 120 or 140 kVp, with infrequent use of lower values. Recent reports suggest substantial dose reduction with use of low kVp (80-100 kVp) for CT angiography. In the abdomen, compared to 120 kVp, use of 100 kVp resulted in about 37% dose reduction for MDCT angiography of the abdominal aorta and iliac arteries (Wintersperger and Nikolaou, 2005). The use of lower kVp (80-100) for dose reduction has also been recommended for chest and abdominal MDCT in newborn and infants (Siegel et al., 2004). As a reduction in kVp can result in a substantial increase in the image noise, it can impair image quality if the patient is too large or if the tube current is not appropriately increased to compensate for the lower tube voltage. Thus, when implementing reduced kVp protocols, it is imperative that appropriate mAs values are determined as a function of patient size. For very large patients, relatively higher tube voltage is almost always needed to obtain diagnostically adequate studies.

3.5. Pitch, beam collimation and slice width

(..) These three factors are related to the detector configuration used for MDCT scanning. Generally, wider beam widths results in more dose efficient examinations, as overbeaming constitutes a smaller proportion of the detected x-ray beam. However, a wider beam width can limit the thinnest reconstructed sections for MDCT systems with < 16 data channels. On such systems, narrow beam widths decrease dose efficiency due to overbeaming, but are needed to allow reconstruction of thinner slice widths. Hence, beam width must be carefully selected to address the specific clinical requirements.

(..) In single-detector CT, increasing pitch decreased the dose without affecting the image noise (although spiral artifacts and image width increase at higher pitch values). In MDCT, an increase in pitch is associated with an increase in image noise. Hence, tube current must be adjusted upward to maintain adequate image noise. Thus there is no fundamental dose saving achieved in MDCT at increased pitch values unless lower tube currents are simultaneous employed. Most scanners allow the users to override the automatic adjustment of mA or mAs.

3.6. Scan mode

(..) Overranging of the x-ray beam with spiral MDCT leads to some amount of unused radiation extending beyond the beginning and ending of the region of interest. Due to this phenomenon the use of a single spiral acquisition (as opposed to multiple contiguous spiral scans) should be avoided in absence of overriding clinical considerations. However, this may be unavoidable in multi-region studies such as simultaneous neck and chest CT (position of arm) or simultaneous chest and abdomen CT (different delay times for optimal contrast enhancement).

3.7. Scan coverage and indication

(..) With the short scan acquisition times of MDCT, there is a tendency to increase the scan length to include multiple body regions either in part or completely (Kalra et al., 2004; Campbell et al., 2005). This increases radiation dose to patient. It is also essential to inform the patient's physicians of the dose consequences of repetitive studies or requesting exams of inappropriate anatomy or for non-medically-necessary indications (Katz et al., 2006).

3.8. System Software: Image reconstruction, noise reduction and metal artifact reduction algorithms

(..) Image-space (i.e., the reconstructed image) and sinogram-space (i.e., the raw projection data) smoothing filters can be used to reduce image noise and consequently allow the user to lower the dose to achieve the previously obtained noise level. Such methods, however, reduce spatial resolution. Special “adaptive” noise reduction filters allow for reduced settings while preserving spatial resolution (Raupach et al., 2005). Such filters analyze the image or projection data for high spatial frequency content (e.g., edges), and smooth regions where there is little edge information, while leaving intact the regions with higher spatial resolution information. Dose savings of 30% have been demonstrated with these techniques (Flohr et al., 2006; McCollough et al., 2005; Raupach et al., 2005). Similarly, ongoing work in the area of image reconstruction algorithms, presents substantial opportunities to reduce noise, and hence dose. Reconstruction algorithms with noise properties superior to those in images reconstructed by the conventional fan-beam filtered back-projection algorithm have been reported, and 3D cone beam algorithms, interactive reconstruction algorithms, and time-averaged Fourier methods for CT perfusion are all topics of active and encouraging investigations.

(..) A substantial decrease in detected signal amplitude is common in high attenuation regions, such as shoulders, due to beam attenuation in a particular projection. This leads to increased image noise with impaired image quality. Projection space filters, available on most scanners, increase the filtration of signal dependent noise in the reconstruction data and thus minimize the loss of resolution. Although there is some loss of image resolution (less than 5%) with the use of these filters, these reconstruction filters avoid an otherwise diagnostically compromised image. These filters can allow a 30-60% reduction in image noise without an increase in radiation dose, typically along the direction of the highest attenuation in non-cylindrical body regions like the shoulder (Kachelriess et al., 2001).

(..) Image post-processing filters have been designed to decrease image noise in scans acquired with reduced radiation dose. Unlike image reconstruction algorithms, these techniques do not require raw scan data for post-processing (Schaller et al., 2000). Different approaches have been adopted for noise reduction in scan volume datasets, which include linear low pass filter, non-linear smoothing and non-linear, three-dimensional filters.

(..) Image post-processing filters were designed on the basis of the principle that in any image, a group of structural pixels representative of structures of interest and a group of non-structural pixels representative of non-structural regions in the image are both present. The filter technique involves isotropic filtering of non-structured regions with a low pass filter and directional filtering of the structured regions with a smoothing filter, operating parallel to the edges and with an enhancing filter operating perpendicular to the edges. Two dimensional, non-linear filters decrease image noise in low-dose CT images but adversely affect the image contrast, sharpness and lesion conspicuity (Kalra et al., 2004). In addition, a three-dimensional filtration method, which generalizes the two-dimensional non-linear smoothing technique in all three directions (in x, y and z axes) in order to avoid loss of contrast and sharpness of small structures, has also been recently reported. Initial studies suggest that these filters may improve image noise without affecting image contrast and lesion conspicuity in low-dose CT (Rizzo et al., 2006).

(..) Streak artifacts from high-attenuation metallic implants are a common problem in CT scanning and can occur from metallic implants such as joint replacement prosthesis, dental implants, or surgical clips. To reduce loss of information from streak artifacts caused by dental implants, particularly in facial CT, a second series of images may be acquired with gantry angulation. This results in additional radiation exposure to the patients. In order to reduce streak artifacts from high attenuation objects, linear interpolation of reprojected metal traces and multi-dimensional adaptive filtering of the raw data have been developed (Mahnken et al., 2003; Watzke and Kalender, 2004). These algorithms reduce streak artifacts form metallic implants and may help in reducing radiation dose (Raupach et al., 2002).

3.9. Modification of scan acquisition and reconstruction parameters

(..) Where possible, CT should be obtained with the lowest achievable radiation dose to the patient. Multiphase examinations should be limited to the fewest phases necessary to make the diagnosis, as should the extent of anatomy imaged. The image width should be no thinner than necessary, in order to decrease image noise and hence avoid increasing the radiation dose to compensate for the increased noise levels. For children and small patients, the kVp should be as low as practical for the given patient, and automated exposure control should be used almost universally. In the case where a CT system is not equipped with automated exposure control, technique charts should be developed with the support of a knowledgeable medical physicist, and consistently used for all patients. This is absolutely essential for pediatric CT, in particular. Finally, providers of CT imaging services should be required to compare their dose levels and image quality measures, by patient size and exam type, against diagnostic reference levels or peer standards, in order to ensure that they are offering high quality examinations at appropriately low dose levels.

4. DOSE MANAGEMENT IN CLINICAL SITUATIONS

(..) “One-size-fits-all” type protocols must not be used for any CT scanner.

Justification is a shared responsibility between requesting clinicians and radiologists. It includes justification of CT study for a given indication, and classification of clinical indications into those requiring standard or high dose CT and those for which information can be obtained with low dose CT examination.

There are indications that awareness on adapting exposure factors to manage patient dose is increasing.

Scanning parameters should be based on study indication, patient age and body region being scanned so that radiation dose can be adapted based on these parameters.

Guidelines must be set so that inappropriate studies can be avoided and triaged to non-radiation based imaging technique.

Training of requesting physician and CT staff can help in optimization of scan indications, protocols and radiation dose.

4.1. Justification of examination

(..) Justification is a shared responsibility between requesting clinicians and radiologists (ICRP 2000b). With the continuous increasing data on suitability and efficacy of MDCT, it is important to ensure that only a qualified medical practitioner generates requests for CT examinations. The radiologist should be appropriately trained and skilled in optimization in CT to achieve an appropriate level of radiation protection, and with adequate knowledge concerning alternative imaging or laboratory techniques. Thus, each CT exam must be performed when the radiation dose is deemed to be justified by the potential clinical benefit to the patient as well as the availability of resources and cost. Clinical guidelines must be prepared, ideally at national level, to advise requesting clinicians and radiologists about appropriateness and acceptability of CT examinations. In the absence of national level agreement on these issues, local institutional guidelines must be developed. These guidelines must help radiologists and clinicians to triage patients to ultrasound or magnetic resonance imaging (MRI) and even conventional radiography, without unduly hindering clinical management. Such guidelines can also help in eliminating unnecessary CT examinations and must include a list of clinical indications for CT pertaining to diagnosis, treatment (surgical guidance and biopsy, drainage or other interventional radiology procedure), and follow-up of known or suspected pathologic processes. In this context, the American College of Radiology Appropriateness Criteria provide evidence based medicine based guidelines to help physicians in recommending an appropriate imaging test (ACR 2000). The European Commission and United Kingdom's Royal College of Radiologists (RCR) document titled "Referral guidelines for imaging" also provides a detailed overview of clinical indications for imaging examinations including CT and other radiation and non-radiation based imaging (RCR 2003).

(..) Justification of a CT examination may include justification of a CT study for a given indication, and classification of clinical indications into those requiring standard or high dose CT and those for which information can be obtained with low dose CT examination. In this respect, the introduction of informed consent for patients undergoing CT scanning as regards potential radiation risks may help in creating greater awareness amongst patients and greater responsibility for requesting physicians and radiologists. Unfortunately, most institutions do not take informed consent for radiation risks from the patients undergoing CT scanning. Introduction of informed consent for radiation risks, although challenging, may help to increase awareness about CT radiation dose and perhaps decrease some "unnecessary" CT from being performed. Such informed consent may include discussion of potential benefits and needs for CT scanning versus possible radiation associated harmful effects such as cancer.

(..) According to the charter on *Consumer Bills of Right and Responsibilities* developed by the *Advisory Commission on Consumer Protection and Quality in the Health Care Industry* appointed by the former United States President explicitly stated that the health care professional must "discuss all risks, benefits, and consequences to treatment or non-treatment" with the consumer or patient. In this context, despite low probability or risk of cancer with diagnostic radiation based procedures, there may be a need for informing patients about the benefits of the radiation based exam as well as the risk of radiation induced carcinogenesis from associated radiation exposure. In a survey of 82 cancer patients undergoing radiation therapy, Barnett et al (2004) reported that about half the patients (36/82) felt that information about severe side-effects (defined as critical organ damage, which are permanent, life threatening, require surgery or negatively affect quality of life) must be provided to them even if the risk is 0.1%. Interestingly, based on linear no threshold theory, Brenner et al (2001) have also estimated a 0.18% risk of lifetime cancer mortality in children receiving low dose radiation from CT scanning of abdomen. Another recent survey of the radiology chairpersons in the United States suggests that less than 15% (14/91) of the radiology departments currently inform patients about possible radiation risks and only 9% (8/88) of radiology departments inform patients about alternatives to CT (Lee et al., 2006).

4.2. Training issues

(..) Recent surveys suggest that there is a substantial lack of comprehension of CT radiation dose amongst requesting physicians (Lee et al., 2004; Thomas et al., 2006). Furthermore, there are considerable variations in the scanning protocols and radiation doses between different CT centres (Hollingsworth et al., 2002; Moss and McLean, 2006). Requesting physicians must be informed about appropriate indications for CT scanning, alternative imaging techniques for triage and radiation risks associated with CT scanning, so that they can justify benefits of CT examinations over potential harmful effects. The radiologists and CT technologists must be trained to adapt CT scanning techniques based on clinical indications (standard dose CT indications such as CT for liver metastases or low dose CT indications for screening CT studies, pediatric CT, kidney stone CT) and to assess associated radiation doses with different scanning parameters. With the constant upgrade of MDCT technology it is important to become acquainted with extrapolation or adaptation of scanning parameters from one scanner to another system. Interestingly, a Japanese survey recently reported that more CT centres are adapting parameters according to patient age and are more frequently using automatic exposure control techniques in order to manage radiation dose (Miyazaki et al., 2005).

4.3. CT dose and risk for individual situations

(..) Most studies on low radiation dose CT have investigated usefulness of reduced tube current, either with fixed tube current or with automatic exposure control techniques (Kalra et al., 2004). These studies have adapted tube current based on patient size (such as weight with fixed tube current scanning and attenuation profile with automatic exposure control techniques), or study indications (lower tube current for screening CT studies, kidney stone CT, and chest CT).

However, dose reduction has also been assessed with use of higher pitch values, lower kVp and use of special techniques such as two- and three-dimensional non-linear noise reduction filters.

Although this section provides some tabulated protocols for dose reduction with examples mostly from studies assessing 4 to 16 slice MDCT scanners, the same principles of dose reductions apply to other MDCT scanners including 32, 40 and 64-slice MDCT scanners. The purpose of these protocols is not to provide actual radiation doses which are likely to be variable for different vendors but to help the users to use these approaches for development of low dose scan protocols for their scanners. At the time of writing of this document, there was less data on similar dose reduction studies for higher end scanners such as 32 to 64 MDCT scanners. Further, the inclusion of certain types of examination in demonstrating dose management does not imply that these are common clinical applications of MDCT. It is based on availability of data on dose management studies in these applications.

4.3.1. Chest CT

(..) As described in preceding sections, image noise, a principle component of image quality, depends on attenuation of x-ray beam as it traverses through the body region being scanned. Less beam attenuation results in lower image noise for chest when compared to abdomen or pelvis, which causes greater beam attenuation. Therefore, compared to abdomen or pelvis CT, a lower radiation dose can be used to obtain a similar image quality for chest CT. Most studies have employed low tube current to reduce radiation dose with chest CT (Wormanns et al., 2005) (Table 4.1A). Prasad et al. 2002 have shown acceptable image quality for evaluating normal anatomic structures with 50% reduction in tube current (110-140 mAs compared to 220-280 mAs for 4-detector MDCT), irrespective of patient size. Studies have also employed different strategies to reduce radiation dose for chest CT based on patient size and clinical indications. Clinical indications for low dose chest CT include scanning young patients with benign diseases (Jung et al., 2000; Yi et al., 2003; Honnef et al., 2004), screening of lung cancer (Diederich et al., 2000; Picozzi et al., 2005), pulmonary nodules (Diederich et al., 1999; Leader et al., 2005), benign asbestos-related pleural based plaques and thickening (Michel et al., 2001; Remy-Jardin et al., 2004), emphysema (Zaporozhan et al., 2006), high resolution chest CT (Ikura et al., 2004), CT guided lung biopsy (Ravenel et al., 2001), and evaluation of patients with neutropenia

(Wendel et al., 2005) and cystic fibrosis (Jimenez et al., 2006). Most investigators have used reduced tube current in order to reduce associated radiation dose (Prasad et al., 2002, Ravenel et al., 2001). Recently, lower kVp (at 80 kVp compared to commonly used kVp of 120) has been described for CT angiography for pulmonary embolism to reduce radiation dose, and increase image contrast (Sigal-Cinqualbre et al., 2004) (Table 4.1B). Use of automatic exposure control techniques for chest CT, combined modulation and angular modulation, has been reported to reduce radiation dose by 20 and 14% compared to fixed tube current (Mulkens et al., 2005).

Table 4.1A. Tube current adjustment is the most frequently documented method to optimize radiation dose. Low dose chest CT with reduced tube current is generally sufficient for evaluation of pulmonary abnormalities. This table summarizes the use of low tube current CT (20 mAs versus 100 mAs for 80% dose reduction) for evaluation of pulmonary nodules (Wormanns et al., 2005). Due to high air-soft tissue contrast, lungs can be evaluated at considerably lower radiation dose. The data in all columns in the Table is from the CT units from a particular manufacturer.

Scanning parameters	Low tube current chest CT	Standard dose chest CT
Scanner	4-detector row MDCT	4-detector row MDCT
mAs	20 mAs (effective)	100 mAs (effective)
kVp	120	120
Rotation time	0.5 second	0.5 second
Pitch	1.75	1.75
Detector configuration	4 x 1 mm	4 x 1 mm
Scan coverage/area scanned	Chest	Chest
Slice thickness	5 mm	5 mm
CTDI vol	2.0 mGy	10.1 mGy
Effective dose	1.4 mSv	6.8 mSv

Table 4.1B. Compared to the use of low tube current to reduce radiation dose, kilovoltage adjustment has been assessed in fewer studies (REF). This table summarizes the use of 80 kVp in patients undergoing contrast enhanced CT of the chest (Sigal-Cinqualbre et al., 2004). Compared to conventionally used kVp of 120-140, use of 80 kVp can allow 2- 4 folds dose reduction if remaining parameters are held constant. The data in all columns in the Table is from the CT units from a particular manufacturer.

Scanning parameters	Low kVp chest CT	Low kVp chest CT	Standard dose chest CT
Scanner	4-detector row MDCT	4-detector row MDCT	4-detector row MDCT
mAs	135 effective mAs	180 effective mAs	90 effective mAs
kVp	80	80	120
Rotation time	0.5 second	0.5 second	0.5 second
Table speed	10 mm/rotation	10 mm/rotation	10 mm/rotation
Pitch	1:1	1:1	1:1
Detector configuration	4 x2.5 mm	4 x 2.5 mm	4 x 2.5 mm
Scan coverage/area scanned	Chest	Chest	Chest
Slice thickness	-	-	-
Effective dose (mSv)	1.54 (males), 1.88 (females)	2.05 (males), 2.51 (females)	2.05 (males), 2.51 (females)

4.3.2. CT for coronary calcium quantification and non-invasive coronary angiography

(..) For coronary CT examinations, it is important to reconstruct images during the phase of the cardiac cycle that will be associated with least motion of the coronary arteries. Current multidetector CT technology allows ECG gating of scan acquisition and reconstruction of images at any desired phase of the cardiac cycle. This needs scan acquisition at small, overlapping pitch, which leads to a higher radiation dose despite smaller scan length used for coronary CT. Effective doses or $CTDI_{vol}$ for some low dose coronary CT angiography and calcium scoring protocols are summarized in Table 4.2A, 4.2B & 4.2C.

(..) CT for coronary calcium quantification can be performed with low-dose CT due to high inherent contrast between coronary calcium and adjoining soft tissue, which allow interpretation even with high image noise. Several strategies can be adopted for reducing dose with coronary CT angiography and coronary calcium scoring CT, which include use of lower tube current

(Shemesh et al., 2005) (Table 4.2A) and tube potential (kVp) (Abada et al., 2006) (Table 4.2B), and ECG triggered tube current modulation (Jakobs et al., 2002) (Table 4.2C). Use of ECG triggered tube current modulation or ECG pulsing has been reported to save 20-50% radiation dose depending on the heart rate (Jakobs et al., 2002). Recent studies have also used patient size based adjustment of tube current for reducing radiation dose with coronary CT angiography (Jung et al., 2003). The authors added artificial noise to coronary CT angiography images of 30 patients. They noted that acceptable image quality and 17.9 (males)-26.3% (females) dose reduction could be achieved with weight based adjustment of tube current.

Table 4.2A. Radiation dose reduction for coronary calcium quantification can be accomplished with use of low fixed tube current or with ECG pulsing. In this study, there was excellent correlation between coronary calcium scores at 165 mAs and 55 mAs ($r = 0.9$, $p < 0.01$) (Shemesh et al., 2005). The data in both columns in the Table is from the CT units from a particular manufacturer.

Scanning parameters	Coronary calcium quantification	Low dose CT for coronary calcium quantification
Scanner	4-detector row MDCT	4-detector row MDCT
mAs	165 mAs	55 effective mAs
kVp	120	120
Rotation time	0.5 second	0.5 second
Detector configuration	4 x 2.5 mm	4 x 2.5 mm
Scan coverage/area scanned	Heart (120 mm)	Heart (120 mm)
Slice thickness	2.5 mm	2.5 mm
CTDI vol	12 mGy	4 mGy

Table 4.2B. Radiation dose reduction for coronary CT angiography with use of lower kVp (80 kVp versus 120 kVp used in most centers) as well as ECG modulated tube current (ECG pulsing) in slim patients. Use of lower kVp may result in inadequate signal and disproportionate image noise if used in patients with greater size (Abada et al., 2006).

Scanning parameters	Low dose coronary CT angiography
Scanner	64-detector row MDCT
mAs	520 effective mAs (with ECG pulsing)
kVp	80
Rotation time	0.33 second
Detector configuration	64 x 0.6 mm
Scan coverage/area scanned	Heart
Slice thickness	0.75 mm
Effective dose (mSv)	~ 2 mSv

Table 4.2C. ECG modulated tube current helps to reduce radiation dose. ECG modulated tube current is more efficient at lower heart rates therefore, administration of beta blockers helps to reduce dose. This table summarizes dose savings (45% for females, 48% for males) with ECG modulated tube current compared to CT performed without modulation in size-matched patients (Jakobs et al., 2002). The data in both columns in the Table is from the CT units from a particular manufacturer.

Scanning parameters	Coronary calcium quantification	Low dose CT for coronary calcium quantification
Scanner	4-detector row MDCT	4-detector row MDCT
Mean body mass index (kg/m^2)	25.59	25.65
ECG modulated mA	No	Yes
mAs	100 effective mAs	55 effective mAs
kVp	120	120
Helical pitch	1.5:1	1.5:1
Table speed	7.5 mm/second	7.5 mm/second
Rotation time	0.5 second	0.5 second
Detector configuration	4 x 2.5 mm	4 x 2.5 mm
Scan coverage/area scanned	Heart (120 mm)	Heart (120 mm)
Slice thickness	1.5 mm	1.5 mm
CTDI vol	12 mGy	4 mGy
Effective dose (mSv)	1.95 (male), 2.48 (female)	1.03 (male), 1.37 (female)

4.3.3. CT colonography

(..) CT colonography is being increasingly used as a screening technique for colorectal cancer. In order to reduce the number of false positive lesions and differentiate between a lesion and polyp, generally two acquisitions are obtained for CT colonography, which increases radiation dose. There is a need for reducing risk with screening techniques and the presence of high inherent contrast between air-distended or contrast tagged fecal matter (stool tagging with oral contrast) and colonic wall offer a unique opportunity to reduce radiation dose for CT colonography. Effective doses for some low dose CT colonography protocols are summarized in Table 4.3A, 4.3B & 4.3C.

(..) Compared to routine abdominal CT studies, CT colonography can be performed at a much lower dose. In fact, several strategies have been adopted for reducing dose associated with CT colonography including the use of higher beam pitch (Cohnen et al., 2004) (Table 4.3A), and lower tube current (Iannaccone et al., 2003) (Table 4.3B) and kilovoltage (Capunay et al., 2005) (Table 4.3C). Recently, automatic exposure control technique has been reported to reduce radiation dose with CT colonography (Graser A. et al. In press).

Table 4.3A. High inherent contrast between air or contrast filled colon and colonic lesions or mucosa allow use of lower tube current as well as higher beam pitch values (compared to beam pitch of less than 1 used in example illustrated in Table 4.3B) (Cohnen et al., 2004).

Scanning parameters	Low dose CT colonography
Scanner	4-detector row MDCT
mAs	10 effective mAs
kVp	120
Rotation time	0.5 second
Pitch	2:1
Detector configuration	4 x1 mm
Scan coverage/area scanned	Abdomen and pelvis
Slice thickness	1.25 mm
Number of acquisitions	2 (prone and supine)
Total effective dose (mSv)	0.7 (males), 1 (females)

Table 4.3B. Tube current reduction can lead to substantial dose reduction for CT colonography despite two CT passes. This table illustrates use of a very low tube current (10 effective mAs) to reduce radiation dose for CT colonography (Iannaccone et al., 2003). The effective dose includes total combined dose for localizer radiographs and CT acquisition in both supine and prone positions.

Scanning parameters	Low dose CT colonography
Scanner	4-detector row MDCT
mAs	10 effective mAs
kVp	140
Rotation time	0.5 second
Pitch	0.875:1
Table speed	17.5 mm/second
Detector configuration	4 x 2.5 mm
Scan coverage/area scanned	Abdomen and pelvis
Slice thickness	3 mm
Reconstruction kernel	B 20 (smooth)
Number of acquisitions	2 (prone and supine)
Total effective dose (mSv)	2.15 (males), 2.75 (females)

Table 4.3C. For pediatric applications of CT colonography, dose can be reduced further with use of lower kVp as well as lower mAs (Capunay et al., 2005).

Scanning parameters	Low dose CT colonography
Scanner	4-detector row MDCT
mAs	15-30 mAs
kVp	90
Rotation time	0.5 second
Pitch	1.5:1
Table speed	25 mm/second
Scan coverage/area scanned	Abdomen and pelvis
Slice thickness	3.2 mm
Number of acquisitions	2 (prone and supine)
CTDI vol	0.3-0.7 mGy
Total effective dose (mSv)	0.3- 0.6 mSv

4.3.4. CT for trauma

(..) Trauma is a major cause of morbidity and mortality in young people throughout the world. It is also a major indication for CT scanning in the young patients, accounting for over 8 million CT or MRI examinations each year in the United States alone (Kalra et al., 2005; McCaig et al., 2004). Indeed, CT has become the imaging technique of choice for patients with trauma to head, neck, chest, abdomen and pelvis. However, several studies have reported protocols for trauma CT and raised concerns about overuse of CT in emergency settings (Hadley et al., 2006, Kortesniemi et al., 2006). Hadley et al have reported that use of American College of Radiology (ACR) appropriateness criteria on CT for trauma can help in reducing radiation dose by about 44% and imaging costs by 39%. The study also reported an estimated effective dose of 16 mSv to a typical trauma patient undergoing CT scanning.

(..) The most important approach for reducing radiation dose associated with the use of CT in trauma is appropriate selection of patients for imaging and triage of patients, when possible to non-radiation based or low-radiation dose imaging techniques (Hadley et al., 2006). Radiation dose increases with number of acquisitions over the same area of interest. Therefore, efforts must be directed towards limiting the number of acquisitions and reducing radiation dose for the “less critical” phase of acquisition (Stuhlfaut et al., 2006).

(..) Often, patients with trauma undergo scanning of contiguous areas of interest such as neck, chest, abdomen and pelvis or chest and abdomen, in the same imaging session. It is important to remember that due to cone beam shaped x-ray beam, there is a small portion of the x-ray beam at the start and end of each helical run which is not incident on the detectors. These unused x-rays result in radiation exposure to the patients and increase with increasing number of helices acquired during a CT examination. Furthermore, radiation dose to patients also increase with overlapping of helices between two anatomic areas of interest (at the level of diaphragm for chest and abdomen CT). Therefore unless there are over-riding clinical indications (such as breath-holding), number of helices acquired during CT examinations must be limited. Indeed, Ptak et al have recently reported that a single-pass or –run, whole-body CT examination resulted in 17% dose reduction compared to the multi-helical, conventional segmented CT protocol for head, neck, chest, abdomen and pelvis (Ptak et al., 2003).

4.3.5. CT of the urinary tract

(..) CT has replaced conventional radiography and intravenous urography for evaluation of urinary tract calculi and urinary tract in many centres in the world, particularly in the United States (Akbar et al., 2004). Although CT does provide valuable information pertaining to the urinary tract, it comes at the price of higher radiation dose to patients with benign disease, who often undergo additional follow-up CT studies (Katz et al., 2006).

(..) Several studies in patients and phantoms have documented that urinary tract calculi can be imaged with low dose CT, as “radio-opaque” or dense calculi offer high contrast against soft tissue background structures (Table 4.4 A,B) (Kluner et al., 2006; Kalra et al., 2005). Since nephrolithiasis is a benign disease, all attempts must be made to reduce dose in young patients and limit the number of CT examination performed for its evaluation. Radiation dose for stone protocol CT can be reduced with the use of lower tube current time product (Kluner et al., 2006) and automatic exposure control (Kalra et al., 2005).

Table 4.4A. Radiation dose can be reduced for CT for evaluation of suspected urinary tract calculi with low tube current. High contrast between most urinary calculi and soft tissues allow evaluation in relatively noisy images at low doses (Kluner et al., 2006).

Scanning parameters	Low dose CT for urinary calculi
Scanner	16-row CT scanner
mA	20
kVp	120
Rotation time	0.5 second
Pitch	1.43:1
Detector configuration	16 x 1 mm
Scan coverage/area scanned	Abdomen and pelvis
Slice thickness	5 mm
Reconstruction kernel	soft tissue kernel
Total effective dose (mSv)	0.5 (males), 0.7 (females)

Table 4.4B. This table summarizes use of z-axis (longitudinal) automatic tube current modulation (Auto mA) for dose reduction in patients with suspected urinary calculi (Kalra et al., 2005). Noise index is the noise in the center of an image of a water phantom. Higher noise index requires less tube current and therefore less radiation dose. The data in all columns in the Table is from the CT units from a particular manufacturer.

Scanning parameters	Regular dose CT	Low dose CT	Low dose CT
Scanner	16-row MDCT,	16-row MDCT	16-row MDCT
Noise index	-	14	20
Average mAs	182.4 Range, 160–240	104 range, 50–160	62.6 range, 37.5–186.9
kVp	140	140	140
Rotation time	0.5 second	0.5 second	0.5 second
Pitch	0.938:1	0.938:1	0.938:1
Detector configuration	16 x 1.25 mm	16 x 1.25 mm	16 x 1.25 mm
Scan coverage/area scanned	Abdomen and pelvis	Abdomen and pelvis	Abdomen and pelvis
Slice thickness	2.5 mm	2.5 mm	2.5 mm
Reconstruction kernel	Standard	Standard	Standard
Effective dose (mSv)	25	15	8.8

4.3.6. CT guided interventions

(..) CT guided interventions pose special issues pertaining to radiation dose to the patient and to the radiology staff performing the procedure. Generally, two or more “passes” or scan acquisitions are obtained in the area of interest. With CT fluoroscopy, radiation exposure to the patient as well as the radiologist in the scanner gantry area is of concern (Table 4.5A,B). There is evidence to support that radiation dose can be reduced during CT guided intervention procedures by limiting the scan length, reducing mAs and fluoroscopic time, and use of alternative non-radiation based imaging guidance (such as ultrasonography) (Buls et al., 2003; Heyer et al., 2005).

Table 4.5A. Efforts must be directed towards reducing radiation exposure from CT fluoroscopy to both patients and physicians. This table summarizes patient and physician doses from CT fluoroscopy (Buls et al., 2003). Physician doses are average doses from CT fluoroscopy guided biopsy, aspiration and drainage, and radiofrequency.

Scanning parameters	CT fluoroscopy
Scanner	4-detector row MDCT
mA	90
kVp	120
Rotation time	0.75 second
Scan coverage/area scanned	Area of interest at the level of needle/catheter tip
Slice thickness	8 mm
CTDI vol	12 mGy
Average effective dose (50% range)	
Patients:	
Biopsy	18.3 mSv (9.8-23.0)
Aspiration and drainage	15.8 mSv (12.6-26.9)
Radiofrequency ablation	36.3 mSv (26.3-51.5)
Overall	19.7 mSv (10.8-27.1)
Physicians: Overall median doses	
Eyes	0.210 mSv (0.143-0.313)
Thyroid	0.240 mSv (0.155-0.406)
Left hand	0.176 mSv (0.118-0.260)
Right hand	0.759 mSv (0.445-1.41)

Table 4.5B. Radiation dose from CT guided biopsy can be reduced by reducing tube current or limiting the scan volume. This table summarizes application of low tube current for dose reduction in children undergoing CT guided biopsy (Heyer et al., 2005).

Scanning parameters	Low dose CT guided biopsy
Clinical indications	Chronic infectious interstitial pulmonary disease in children
Scanner	4-detector row MDCT
mAs	20 effective mAs
kVp	120
Detector configuration	5 x 2 mm
Scan coverage/area scanned	Region of interest (10 mm)
Maximum number of images	4
Slice thickness	10 mm
Effective dose	0.83 mSv (range, 0.38- 1.40 mSv)

4.3.7. CT in children

(..) Children are more susceptible to risk of radiation induced carcinogenesis compared to adults. Therefore, radiologists, medical physicists, and technologists, must pay special attention to CT scan protocols and radiation dose when imaging children. Radiation dose in children and small adults can be reduced without affecting diagnostic information obtained from the study. Image noise is proportional to the x-ray beam attenuation, which in turn is affected by the distance that x-rays traverse through the patient body region being scanner. Scanning parameters (mAs, kVp) can be adjusted to adapt dose to patient weight or age (Frush et al., 2002). Alternatively, automatic exposure control techniques can be also used to reduce radiation dose to children (Greess et al., 2002; Greess et al., 2004).

(..) In a recent review on radiation dose reduction in children, Vock recommends several strategies to accomplish this objective including rigorous justification of CT examinations, acceptance of images with greater noise if diagnostic information can be obtained, optimization of scan protocols, scanning of minimum length as needed, and reduction of repeated scanning of identical area (Vock, 2005). A recent study of CT evaluation of pediatric trauma suggests that more than one-half of the examinations were normal (Fenton et al., 2004). For follow-up CT studies, the scan volume can also be restricted depending on the clinical indication in order to reduce radiation dose. Jimmenz et al have reported substantial dose reduction (55%) by limiting the scan coverage to just 6 images per examination for follow-up CT of patients with cystic fibrosis (Jimmenz et al., 2006).

4.3.8. CT of the pregnant patients

(..) Common indications for CT scanning in a pregnant patient include suspected appendicitis, pulmonary embolism, and urinary tract calculi. To minimize radiation exposure to the fetus, it is important to triage the patient appropriately if diagnostic information can be obtained from an alternative non-radiation based imaging. Radiologists and physicians must also decide if immediate scanning is required or if scanning can be postponed until after the delivery. Strict x-ray beam collimation in modern CT scanners allows very little scattered radiation dose. For scanning body regions outside abdomen and pelvis, such as chest CT for suspected pulmonary embolism, shielding is not indicated as most scattered radiation comes from internal scattering and external scattering is minimal due to tight beam collimation. For abdominal-pelvic CT, scanning parameters must be selected to reduce the fetal dose (such as wider beam collimation and pitch, and lower mAs, kV and scan volume) (Table 4.6). For CT in a pregnant patient with suspected appendicitis, the scan volume must be restricted to the necessary anatomy, and dual-pass (with and without contrast) studies should be avoided (Wagner and Huda, 2004; Ames Castro, 2001). A “step-and-scan protocol” may help in terminating the study when the appendix or area of interest has been scanned (Wagner and Huda, 2004). Likewise, in CT for renal calculi in a pregnant patient, fetal dose must be reduced with use of low mAs, high pitch and a limited scan volume, without substantially compromising the study quality (Forsted and Kalbhen, 2001).

Table 4.6. Summary of typical scanning protocols and radiation doses for scanning used in a CT center for imaging pregnant patients with suspected pulmonary embolism, appendicitis, and renal stones, which represent the commonest indications for CT in pregnancy (Hurwitz et al. 2006). The radiation dose values were estimated using anthropomorphic phantoms simulating a pregnant woman.

Scanning parameters	Pulmonary emobolism	Appendicitis	Renal stone
Scanner	16-MDCT	16-MDCT	16-MDCT
mA	380	340	160
Gantry rotation time	0.8 second	0.5 second	0.5 second
kVp	140	140	140
Pitch	1.375:1	1.75:1	1.75:1
Detector configuration	16 x 1.25 mm	16 x 0.625 mm	16 x 0.625 mm
Scan coverage/area scanned	Chest	Abdomen- pelvis	Abdomen- pelvis
Slice thickness	2.5 mm		2.5 mm
Fetal dose at 3 months	0.07 cGy	1.5 to 1.7 cGy	0.4 to 0.72 cGy
Maternal effective dose (mean ± SD)	14.4 ± 2.1 mSv	13.3 ± 1.0 mSv	4.51 ± 0.45 mSv

4.4. Future directions

(..) CT vendors have invested efforts towards the development of dose efficient technologies (Kalra et al., 2004). Despite the efforts of CT vendors and users (radiology and referring physicians), contributions of MDCT scanning to radiation dose has been increasing. Further efforts at dose management should include the development of guidelines for indications for CT for the purpose of diagnosis, staging, or follow-up of patients, further optimization of automatic exposure control techniques and other dose management strategies, continued efforts of the international, national or regional organizations to educate physicians and medical physicists to realms of radiation dose associated with MDCT, as well as research and development of non-radiation based imaging techniques which will be able to replace CT by providing equal information in a timely and appropriate manner.

APPENDIX A

HOW TO DESCRIBE DOSE IN CT

A1. CT Dose Index (CTDI)

(..) The CT Dose Index (CTDI) is the primary dose measurement concept in CT. It represents the average absorbed dose, along the z axis, from a series of contiguous exposures. It is measured from one axial CT scan (one rotation of the x-ray tube), and is calculated by dividing the integrated absorbed dose by the total beam width. CTDI theoretically estimates the average dose within the central region of a scan volume, which is referred to as the Multiple Scan Average Dose (MSAD) (Shope et al., 1981), the direct measurement of which requires multiple exposures. The CTDI offered a more convenient, yet nominally equivalent method of estimating this value, and required only a single scan acquisition, which in the early days of CT, saved a considerable amount of time.

(..) The equivalence of the MSAD and the CTDI requires that all contributions from the tails of the radiation dose profile be included in the CTDI dose measurement. The exact integration limits required to meet this criterion depend upon the width of the total beam width and the length of the scattering medium. To standardize CTDI measurements, the FDA introduced the integration limits of ± 7T, where T represented the nominal slice width (Shope et al., 1981). Interestingly, the original CT scanner, the EMI Mark I, was a dual-detector-row system. Hence, the nominal radiation beam width was equal to twice the nominal slice width (i.e., N x T mm). To account for this, the CTDI value, while integrated over the limits ± 7T, was normalized to 1/NT:

$$\mathrm{CTDI}_{FDA} = 1/NT \bullet \int_{-7T}^{+7T} D(z)\,dz \qquad \text{(Eqn. 1)}$$

where D(z) represents the radiation dose profile along the z axis. However, the FDA definition neglected to account for the need to integrate over a longer limit (±7NT).

(..) The scattering media for CTDI measurements were also standardized by the FDA (United States FDA Code of Federal Regulations, 1984). These consist of two polymethylmethacrylate (PMMA, e.g., acrylic or lucite) cylinders of 14-cm length. To estimate dose values for head examinations, a diameter of 16 cm is used, and to estimate dose values for body examination, a diameter of 32 cm is used. These are typically referred to, respectively, as the head and body CTDI phantoms.

(..) The $CTDI_{100}$, like the $CTDI_{FDA}$, requires integration of the radiation dose profile from a single axial scan over specific integration limits. In the case of $CTDI_{100}$, the integration limits are ± 50 mm, which corresponds to the 100 mm length of the commercially available "pencil" ionization chamber (European Commission, 2000; Jucius and Kambic, 1977; Pavlicek et al., 1979).

$$CTDI_{100} = 1/NT \bullet \int_{-50mm}^{+50mm} D(z)\, dz \qquad \text{(Eqn. 2)}$$

$CTDI_{100}$ is acquired using a 100-mm long, 3 cm^3 active volume CT "pencil" ionization chamber and the two standard CTDI acrylic phantoms. The measurement must be performed with a *stationary* patient table.

(..) The CTDI can vary across the field-of-view. For body imaging, the CTDI is typically a factor or two higher at the surface than at the centre of rotation. The average CTDI across the field-of-view is given by the Weighted CTDI ($CTDI_w$) (European Commission, 2000; International Electrotechnical Commission, 2002; Leitz et al., 1995), where:

$$CTDI_W = 1/3\ CTDI_{100,center} + 2/3\ CTDI_{100,edge}. \qquad \text{(Eqn. 3)}$$

(..) The values of 1/3 and 2/3 approximate the relative areas represented by the centre and edge values (Leitz et al., 1995). $CTDI_w$ is a useful indicator of scanner radiation output for a specific kVp and mAs. One must use the f-factor (f) appropriate to the task at hand to convert air kerma (mGy) or exposure (R) to absorbed dose (mGy or rad). According to IEC 60601-2-44, $CTDI_W$ must use $CTDI_{100}$ as described above and an f-factor for air (0.87 rad/R or 1.0 mGy/mGy) (European Commission, 2000; International Electrotechnical Commission, 2002).

Volume CTDI (CTDIvol)

(..) To represent dose for a specific scan protocol, which almost always involves a series of scans, it is essential to take into account any gaps or overlaps between the radiation dose profiles from consecutive rotations of the x-ray source. This is accomplished with use of a dose descriptor known as the Volume $CTDI_w$ ($CTDI_{vol}$) (International Electrotechnical Commission, 2002), where

$$CTDI_{vol} = (N \bullet T/I) \bullet CTDI_w \qquad \text{(Eqn. 4)}$$

(..) In helical CT, the ratio of the table travel per rotation (I) to the total beam width (N•T) is referred to as pitch; hence,

$$CTDI_{vol} = CTDI_w / \text{pitch}. \qquad \text{(Eqn. 5)}$$

(..) So, whereas $CTDI_w$ represents the average absorbed radiation dose over the x and y directions, $CTDI_{vol}$ represents the average absorbed radiation dose over the x, y and z directions. It is conceptually similar to the MSAD, but is standardized with respect to the integration limits (±50 mm) and the f-factor used to convert the exposure or air kerma measurement into dose to air. $CTDI_{vol}$ is the parameter that best represents the average dose at a point with the scan volume for a particular scan protocol for a standardized phantom (International Electrotechnical Commission, 2002). The SI units are milli-Gray (mGy). It is a useful indicator of the dose for a specific exam protocol, because it takes into account protocol-specific information such as pitch. Its value is required to be displayed prospectively on the console of newer CT scanners. The problem when measuring $CTDI_{vol}$ in MDCT is that occasionally the length of irradiation goes beyond the 100mm that the pencil chamber is designed. There are new chambers that are designed to overcome this problem.

(..) While $CTDI_{vol}$ *estimates* the average radiation dose within the irradiated volume of a CT acquisition for an object of similar attenuation to the CTDI phantom, it does not well represent the average dose for objects of substantially different size, shape, or attenuation. Additionally, it does not indicate the total energy deposited into the scan volume because is independent of the length of the scan.

A2. Dose Length Product (DLP)

(..) To better represent the overall energy delivered by a given scan protocol, the $CTDI_{vol}$ can be integrated over the scan length to compute the Dose-Length Product (DLP), where:

DLP (mGy-cm) = $CTDI_{vol}$ (mGy) • scan length(cm) (European Commission, 2000) (Eqn. 6)

(..) The DLP reflects the total energy absorbed (and thus the potential biological effect) from a specific scan acquisition. Thus, while an abdomenal CT might have the same $CTDI_{vol}$ as an abdominal and pelvic CT, the latter exam would have a greater DLP, proportional to the greater anatomic coverage of the scan.

A3. Organ dose and effective dose

(..) The effective dose is a "dose" parameter that reflects the risk of a non-uniform exposure in terms of a whole body exposure. It is a concept used to normalize partial body irradiations relative to whole body irradiations to enable comparisons of risk (International Commission on Radiological Protection, 1991). The calculation of effective dose requires knowledge of the dose to specific sensitive organs within the body, which are typically obtained from Monte Carlo modeling of absorbed organ doses within mathematical anthropomorphic phantoms, and recently also voxel phantoms based on real humans. Effective dose is expressed in the units of milliSieverts (mSv), and can be compared to the effective dose from other sources of ionizing radiation, such as that from background radiation level (e.g., radon, cosmic radiation, etc.) which is typically in the range of 1 to 3 mSv depending upon the location. Typical values for common CT and non-CT exams are given in Table A.1

Table A.1 Typical effective dose values in diagnostic radiological and nuclear medicine examinations (adapted from McCollough and Schueler 2000).

Head CT	1 - 2 mSv
Chest CT	5 - 7 mSv
Abdomen CT	5 - 7 mSv
Pelvis CT	3 - 4 mSv
Abdomen and pelvis CT	8 - 11 mSv
Coronary artery calcium CT	1 - 3 mSv
Coronary CT angiography	5 - 12 mSv
Hand radiograph	< 0.1 mSv
Dental bitewing	< 0.1 mSv
Chest radiograph	0.1 - 0.2 mSv
Mammogram	0.3 - 0.6 mSv
Lumbar spine radiograph	0.5 - 1.5 mSv
Barium enema exam	3 - 6 mSv
DiagnosticCoronary angiogram	5 - 10 mSv
Sestamibi myocardial perfusion	13 - 16 mSv
Thallium myocardial perfusion	35 - 40 mSv

(..) Although effective dose calculations require specific knowledge about individual scanner characteristics, a reasonable estimate of effective dose, independent of scanner type, can be achieved using the relationship:

$$\text{Effective Dose} = k \bullet \text{DLP} \qquad \text{(Eqn 7)}$$

where k is a weighting factor ($mSv \times mGy^{-1} \times cm^{-1}$) which depends only upon body regions (Table A.2) (McCollough, 2003).

Table A.2: Head, neck, thorax, abdomen, or pelvis values of k (European Commission, 2000; Geleijns et al., 1994)

Region of body	k (mSv • mGy^{-1} • cm^{-1})
Head	0.0023
Neck	0.0054
Chest	0.017
Abdomen	0.015
Pelvis	0.019

The Commission wishes to emphasize that effective dose is intended for use as a protection quantity on the basis of reference values and therefore should not be used for epidemiological evaluations, nor should it be used for any specific investigations of human exposure. Rather, absorbed dose should be used with the most appropriate biokinetic biological effectiveness and the risk factor data. The use of effective dose for assessing the exposure of patients has severe limitations. Effective dose can be of some value for comparing doses from different diagnostic and therapeutic procedures and for comparing the use of similar technologies and procedures in different hospitals and countries as well as from use of different technologies for the same medical examinations. For planning the exposure of patients and risk-benefit assessments, however, the equivalent dose or the absorbed dose to irradiated tissues is the more relevant quantity.

(..) Effective dose, however, does not tell the complete story with regard to the potential effects of ionizing radiation. Specific organs and tissues are known to be more radiosensitive than others. While this is reflected in effective dose, the absolute doses to specific organs or tissues are also an important consideration.

A4. Dose estimation tools

(..) Modern CT systems display the $CTDI_{vol}$ and DLP information for every scan acquisition. From these values, an estimate of effective dose may be obtained, as discussed above. For more complete calculations of organs dose, data from Monte Carlo dose calculations must be used. These are available from different sources as: the NRPB of the United Kingdom (Hart et al., 1994; Shrimpton et al., 1991)); the GSF of Germany (Zankl et al., 1991; Zankl et al., 1993; Zankl and Wittmann, 2001); CT-EXPO (Stamm and Nagel 2002). Several software programmes have been developed to integrate the dose to target organs for each slice irradiated in the CT exam (Kalender et al., 1999) and those from ImPACT (www.impactscan.org) .

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