2. Technical principles of MSCT

This report concerns multislice CT with 4-slice and 16-slice CT scanners. Advantages of multislice CT are increased speed and volume coverage, excellent opportunities for dedicated 2D and 3D visualisation, and post-processing. Disadvantages are occurrence of specific artefacts (multislice artefacts, cone-beam artefacts) and increased contribution to patient dose due to reduced geometric efficiency and more prominent impact of the additional tube rotations necessary before and after data acquisition over the planned scan range.

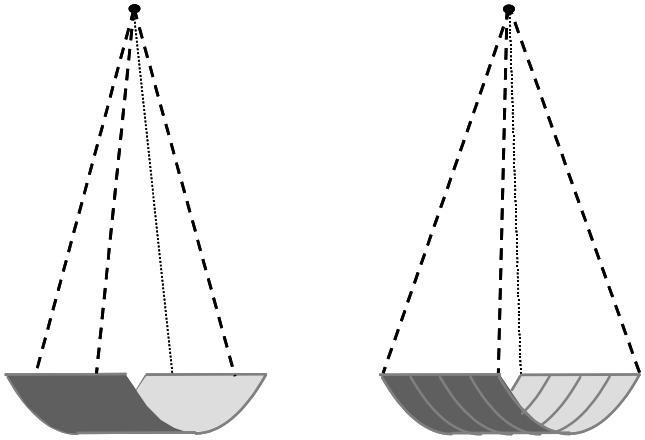

Figure 1. Computed tomography: acquisition with one single slice (left) and multislice CT with four active acquisition channels (right).

Computed tomography requires measurement, at different angles, of the dose profile of the fan shaped x-ray beam after attenuation by the patient. To achieve this a quickly rotating gantry containing an x-ray tube and detector array are used. In single slice CT, the curved detector array consists of about 800 – 1000 adjacent detector elements along the detector arc. In multislice CT, from 4 up to 16 dose profiles can be measured

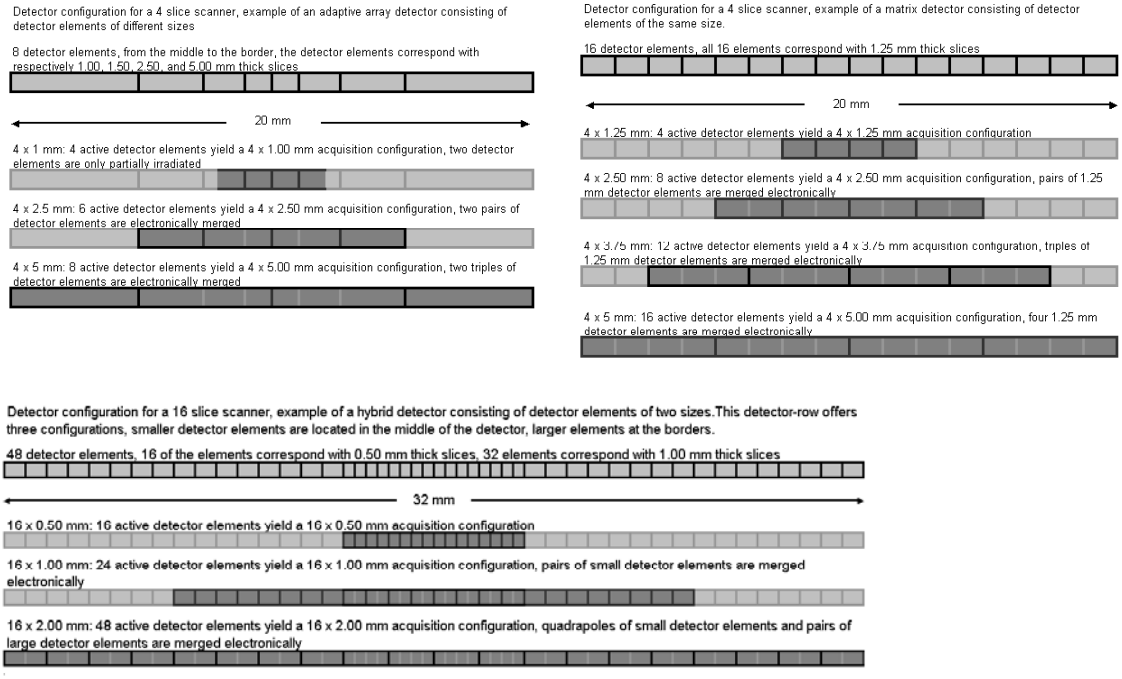

Figure 2. Examples of multislice detectors, an adaptive array 4-slice detector, a matrix array 4-slice detector and an hybrid array 16-slice detector. The figure shows for each detector array the possible acquisition configurations.

simultaneously (figure 1). Figure 2 shows examples of multislice detectors, i.e. an adaptive array 4-slice detector, a matrix array 4-slice detector and an hybrid array 16-slice detector. The figure shows for each detector array the possible acquisition configurations. Manufacturers have already announced the availability of 32-40 slice scanners by the end of 2004. Multislice CT is achieved with detector systems that provide not only hundreds of detector elements along the detector arc, but also multiple rows of detector elements perpendicular to the detector arc, e.g. resulting in arrays with 4 to 16 active detector rows (sections) along the z-axis. In MSCT, solid-state detector elements (scintillators) are used exclusively; such detectors are 20-30% more dose efficient than the gas-filled detectors sometimes used in single slice CT.

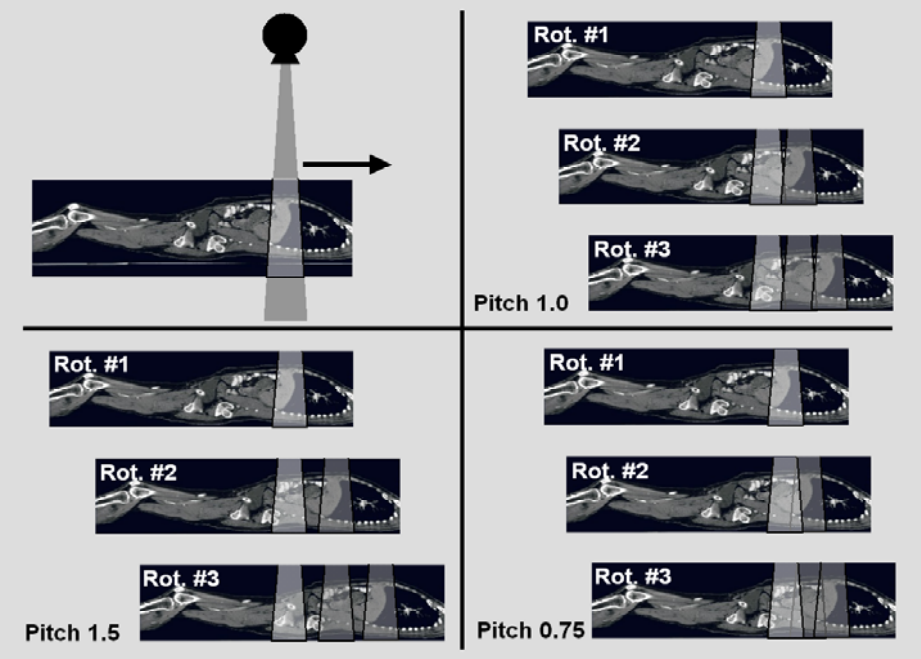

Figure 3. The figure shows the irradiated volume for three contiguous rotations at the moments with the x-ray tube in the frontal position. The table speed and the beam collimation determine the pitch pactor. A high table speed introduces interspaces (e.g. pitch factor 1.5), a low table speed yields an overlapping acquisition (e.g. pitch factor 0.75). A pitch factor of 1.0 yields a contiguous acquisition. The arrow indicates the direction of the table speed.

During a helical MSCT acquisition the patient moves at constant speed through the gantry. The table speed, rotation time and the total width of all simultaneously imaged sections determines whether the transverse slabs of the patient, which are exposed sequentially during data acquisition, are overlapping, contiguous or with interspaces. The acquisition parameters table speed, rotation time, section thickness and number of simultaneous acquired sections determine the CT pitch factor. For an overlapping acquisition, the CT pitch factor is smaller than one, for contiguous acquisition the CT pitch factor equals one and for an acquisition with interspaces the CT pitch factor is larger than one (Figure 3).

Dosimetry in MSCT

The product of tube current and rotation time is referred to as the radiographic exposure (mAs) and it is a key acquisition parameter since it strongly affects the noise in the reconstructed images as well as the local absorbed dose. However, it is not useful to compare values of radiographic exposure (mAs) for different types of scanner since this quantity does not take into account the considerable differences in scanner design, such as the composition and shape of the beam filter and the beam geometry (short versus long geometry or large field versus small field scanning). In other words, radiographic exposure is not well correlated on an absolute scale with either patient dose or image quality. Dedicated dosimetric quantities provide a better means for the evaluation of acquisition protocols for different MSCT scanners with regard to local absorbed dose.

Currently three dosimetric quantities are widely recognised in CT. They provide an indication of the average dose in the scanned region, the exposure from the complete CT examination and the radiation risk of the entire CT scan. These quantities are respectively the volume computed tomography dose index ($CTDI_{vol}$, mGy), the dose-length product (DLP, mGy.cm) and the effective dose (E, mSv), as discussed more fully and defined in the Appendix on CT dosimetry. Most MSCT scanners provide the operator with an indication of the $CTDI_{vol}$ and DLP for each sequence/series (e.g. different phases of contrast enhancement) and each CT scan.

The $CTDI_{vol}$ depends primarily on technical acquisition parameters such as tube current, rotation time, tube voltage, beam filtration and geometric efficiency. The $CTDI_{vol}$ is derived from measurements in two cylindrical PMMA CT dosimetry phantoms representing the attenuation of respectively the adult head (16 cm diameter PMMA) or the body (32 cm diameter PMMA). The $CTDI_{vol}$ is an excellent parameter for comparison between different protocols and different MSCT scanners.

However, quantitative values of $CTDI_{vol}$ for the two different phantoms, i.e. head and body, cannot easily be compared with each other. Measured under the same exposure conditions, the $CTDI_{vol}$ is higher when measured in the relatively small head phantom, compared with values for the larger body phantom. Note that in the previous quality criteria the quantity $CTDI_{w}$ was used, nowadays it is current practice to use the $CTDI_{vol}$ which is the $CTDI_{w}$ divided by the pitch factor.

In addition to considerations of local absorbed dose, a quantity that expresses patient exposure from the complete MSCT examination should also take into account the extent of the exposed range and the exposures during all sequences of the examination. A quantity that fulfils these conditions is the dose-length product (DLP, mGy.cm). This is the $CTDI_{vol}$ multiplied by the length of the exposed range for each sequence. The DLPs from component sequences can be summed to provide a measure of the exposure from a complete CT examination, although only DLP values associated with the same CT dose phantom, i.e. either the head or body phantom, can be added together. $CTDI_{vol}$ or DLP values measured within different sized CT dose phantoms should never be compared with each other.

Table 1. Nominal probability coefficients for stochastic effects [ICRP60].

Exposed population	Detriment (10^{-2} Sv^{-1}) *)			
	Fatal cancer	Non-fatal cancer	Severe hereditary effects	Total
Adult workers	4.0	0.8	0.8	5.6
Whole population	5.0	1.0	1.3	7.3

**) A dose and dose rate effectiveness factor (DDREF) of 2 has been included in the probability coefficients.*

Table 2, Typical effective doses for exposures to natural and medical sources of ionizing radiation..

Natural exposures	
Annual background	2.5 mSv/year
Intercontinental flight (London-Los Angeles)	0.080 mSv
One week skiing	0.015 mSv
Medical exposures	
Knee AP radiograph	< 0.001 mSv
Chest PA radiograph	0.03 mSv
Abdomen AP radiograph	0.3 mSv
CT cranium (acute stroke)	2 mSv
CT chest (metastases)	6 mSv
CT abdomen (abscess)	9 mSv

In radiation protection, effective dose (E, mSv) is often applied as a dose quantity that correlates well with the radiation risk, i.e. the carcinogenic as well as the hereditary risk. Table 1 provides the link between effective dose and radiation risk (risk coefficients derived by ICRP (1991)). These risk coefficients apply to a general population and consequently they may not be applied to patients since life expectancy of the general population is not representative of the life expectancy for specific subgroups of patients. A reduced life expectancy would reduce risk coefficients considerably. Effective doses from MSCT can easily be compared with those for other exposure conditions, such as exposure during radiography or fluoroscopy, exposure to radionuclides and exposure to natural sources of radiation (table 2). For the calculation of effective dose, the average absorbed doses to the 20 most sensitive organs must be assessed, which is not feasible in clinical practice. In general, assessment of effective dose is based on corresponding measurements of the $CTDI_{vol}$ or $CTDI_{air}$. Effective dose is then calculated from these CTDI values using established conversion factors. An overview of dosimetric quantities and their applicability is provided in table 3; for an overview of CT dosimetry see also Appendix 1.

Table 3. Application of the three dosimetric quantities used in multislice computed tomography as indicators of CT technique, patient exposure and radiation risk

Quantity	CT technique (e.g. mAs, kVp, pitch)	Patient exposure (e.g. scan range, phases)	Radiation risk (e.g. organ dose)
$CTDI_{vol}$ (mGy)	+ +	--	--
DLP (mGy.cm)	+	+	--
E (mSv)	--	+	+ +

++: good; + moderate, -- bad correspondence between the dose quantity and its application for the assessment of practice.

Exposure factors in relation to dose in MSCT

$CTDI_{vol}$ is influenced primarily by the radiographic exposure (C, mAs) and the CT pitch factor (or the volume radiographic exposure (C_{vol}, mAs)), rotation time, tube voltage, beam flat filter and beam shaping filter (additional filtration, small field versus large field), and the detector acquisition configuration, i.e. section thickness and number of simultaneous acquired sections. These parameters are the primary technical scan parameters in CT and should be optimised to yield the required image quality at a dose level that is as low as reasonably achievable (ALARA principle). Note that C_{vol} is also known as the 'effective mAs', i.e. the radiographic exposure divided by the pitch factor.

Radiographic exposure

Radiographic exposure (C, mAs), the product of tube current (I, mA) and rotation time (t, s), is one of the main determinants for patient dose and image quality. For helical scanning some manufacturers use the volume radiographic exposure (C_{vol}, mAs), which is the radiographic exposure corrected for the CT pitch factor (as defined in Appendix 1):

$$Cvol = \frac{C}{CT\ pitch\ factor}$$

If all other CT acquisition parameters remain the same, patient dose increases proportionally with (volume) radiographic exposure; this is true for the dose descriptors $CTDI_{vol}$, DLP and effective dose. Radiographic exposure is closely related to image quality, more specifically to the noise in CT images. Noise is generally expressed as the standard deviation of the CT numbers (in hounsfield units (HU)) in a region of interest within an image of an homogeneous part of a CT test object. In general a twofold increase in the radiographic exposure can be assumed to yield a 30% reduction in noise. On the other hand, a reduction in the radiographic exposure to 50% of its initial value would result in a 40% increase in the noise. These values are not valid at very low radiographic exposures since in this case other sources of image noise should also be taken into account, e.g. electronic noise. Noise also increases

with increasing pitch factor assuming a constant radiographic exposure (C, mAs). Either radiographic exposure or volume radiographic exposure is selected and displayed for clinical scans.
It is essential that CT scans are performed at an optimised level of radiographic exposure, i.e. optimised for the specific clinical problem and for the individual patient. However, there is not yet any generic scientific basis for the optimisation of radiographic exposure. Most (MS)CT scans are performed at radiographic exposures that have been established pragmatically (see for example the papers on optimal technique for paediatric CT by Siegel (2003), Suess (2002) and Westerman (2002)). Recent publications provide a preliminary scientific framework for the selection of radiographic exposure for particular CT studies, with most demonstrating potential for (sometimes substantial) reduction in radiographic exposure. These papers apply for example to screening for colon cancer (VanGelder2002, VanGelder2003), lung cancer, suspected chronic sinusitis (Tack2003) and suspected bronchiectasis (Yi2003).
Note that, depending on patient size, one and the same radiographic exposure may lead to different values of the effective dose, i.e. under the same scan conditions effective dose will be considerably higher for small patients and children and it will be relatively smaller for large patients. Note also that radiographic exposure for one type of CT scanner should not be compared with radiographic exposure on another type. Differences in scanner design (e.g. tube filtration, beam shaping filter and scan geometry) cause considerable differences in the beam characteristics.
In order to achieve consistent image quality for patients of different size, radiographic exposure should be adapted to the physique of the individual patient. The radiographer can achieve this manually by assessing patient physique and using tables that provide the optimal radiographic exposure for patients of different size, for example the recommended techniques for paediatric CT of the thorax (Siegel (2003)).

Automatic exposure control (AEC) in MSCT

Ideally, optimal radiographic exposure should be established by means of automatic exposure control, and this function has become available on recent models of CT scanners. Automatic exposure control provides the user with a tool for adapting the radiographic exposure to the size of the patient. It may also adapt radiographic exposure to the body area, e.g. by providing during chest CT scans higher exposures at the level of the shoulders and lower exposure at the level of the lungs. Advanced systems for automatic exposure control modulate radiographic exposure during each rotation, e.g. by increasing the radiographic exposure for lateral projections and decreasing it for frontal projections (Giacomuzzi1996, Kalender1999, Schmidt2003). Automatic exposure control aims to maintain similar image quality for patients of different size and to achieve optimal use of radiation. Systems for AEC can be based on scan projection radiography or alternatively on-line assessment of the attenuation during the helical scan and real-time adaptation of the tube output. An effective

provision for automatic exposure control should always be available and used on new MSCT scanners.

Tube voltage

Until recently most CT scans were performed with tube voltages between 120 and 140 kV. The advantage of a high tube voltage and the corresponding high (average) photon energy is good transmission of x-rays, which results in relatively high detector dose and thus contributes to relatively low noise level in images. Nowadays there is a trend to use lower tube voltages, i.e. between 80 and 100 kV. The advantage is better radiation contrast between lesions and surrounding tissue, particularly in the case of contrast enhanced lesions, either by using iodine, e.g. CT angiography, or air, e.g. CT colonography, as the contrast agent. A reduction in tube voltage from 120 kV to 80 kV, leaving all other scan parameters unchanged, would yield a reduction in effective dose for body CT scans by a factor of 3 – 4. However, a reduction of tube voltage alone would increase image noise considerably and image quality would probably deteriorate to a level that is unacceptable to the radiologist. In general, reduction of tube voltage should necessarily be accompanied by some increase in radiographic exposure (mAs). Such an increase is acceptable as long as the combination of reduced tube voltage and increased radiographic exposure yields an effective dose that is equal to or lower than the initial effective dose. Ideally, radiographic exposure should be increased enough to maintain the initial image quality, or initial contrast-to-noise ratio, whilst achieving a reduction in patient effective dose. Most papers published so far advocate low tube voltage scanning mainly for contrast enhanced studies of the chest, neck and brain (including brain perfusion), and for paediatric CT. Indeed, low tube voltages might be counterproductive when scanning large body parts (e.g. the abdomen of an adult) even in the case of contrast enhanced studies. This is because the transmission of an 80 - 100 kV x-ray beam becomes poor and, in order to maintain proper image quality, an excessive high radiographic exposure and effective dose might be required. Images acquired at low tube voltages might also be more susceptible to artefacts, e.g. beam hardening artefacts.

Rotation time

In general tube rotation time should be kept as low as possible in order to yield minimal movement artefacts, short scan time (and hence short breathhold), and the opportunity for scanning a range that is as large as possible. Longer rotation times might be necessary if the required radiographic exposure cannot be achieved

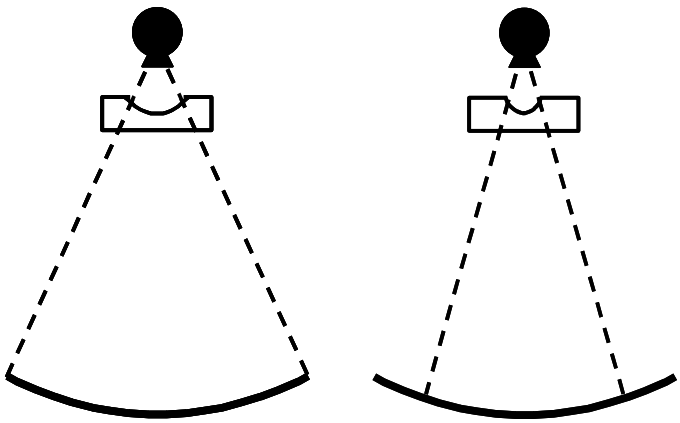

Figure 4. Acquisition with a standard beam shaping filter (left) and a small beam shaping filter (right).

for the shortest scan time. Longer rotation times might also be selected in cardiac CT to avoid synchronisation of the heartbeat and the tube rotation. Patient dose is in principle proportional to rotation time when all other CT scan parameters remain constant.

Beam filter and Scan Field Of View (SFOV)

Filtration of the x-ray beam with a flat filter and a beam-shaping filter is performed to reduce beam-hardening artefacts and to reduce the dynamic range of the signal from the detectors, respectively. Optional additional filtration might reduce patient dose even further, but it could result in a reduction in image quality by yielding a poorer contrast-to-noise ratio.

In addition to the standard scan field of view, some scanners offer the option of using a beam-shaping filter that creates a small scan field of view (figure 4). This is used for CT scanning of small body parts such as the neck, extremities and head. A small scan field of view might also be used if only a certain part of the body is of interest, e.g. when scanning the lumbar spine or the heart. In the latter case patient dose is certainly reduced, although image artefacts might also be induced.

CT Pitch factor

A large CT pitch factor, i.e. greater than one, implies the presence of interspaces between successive beam rotations and contributes to the ability to scan a long range and to achieve a short scan time and breathhold. However in some cases image artefacts and deterioration of z-axis resolution might compromise image quality. For a specific clinical problem, any change in the CT pitch factor should be compensated for by a corresponding change in radiographic exposure in order to yield constant volume radiographic exposure. This applies particularly to those CT scanners where the operator selects radiographic exposure instead of the pitch corrected radiographic exposure (volume radiographic exposure). However, any change in CT pitch factor is automatically compensated for in the case of those CT scanners where the operator directly selects volume radiographic exposure.

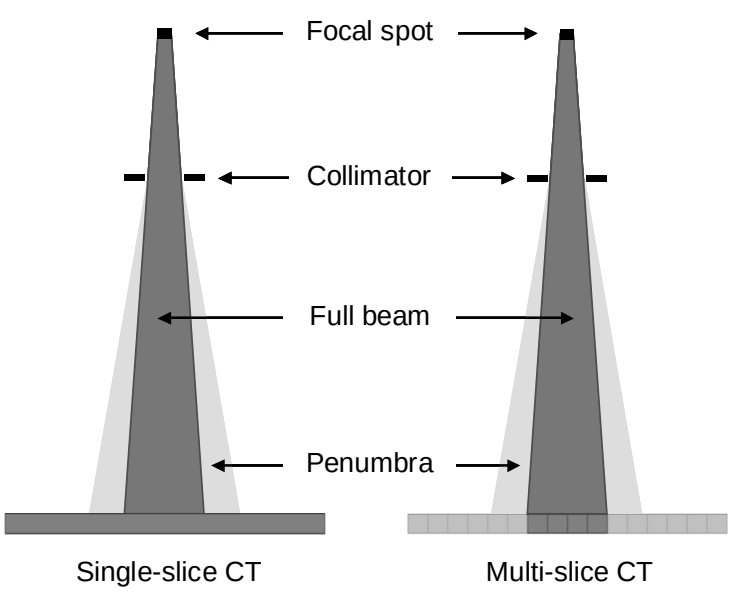

Figure 5. The penumbra goes undetected in MSCT resulting in a reduced geometric efficiency. The detector of the single-slice scannr (left) encompasses the entire x-ray beam. The figure shows a multi-slice acquisition with four active detector rows (right, the active detector arrays are dark grey), the penumbra irradiates detector arrays that do not contribute to the acquisition.

<u>Section thickness, geometric efficiency and x-ray beam penumbra</u>

Section thickness is the primary parameter affecting z-axis resolution. A small section thickness, e.g. 0.5 – 0.75 mm, yields an excellent 3D volume data set for image post processing with minimal artefacts. However, a small section thickness increases examination time and reduces the geometrical efficiency of the scan, with the latter effect being most notable for 4-slice scanners.

The x-ray focal spot, the beam collimator and their mutual location determine the geometry of the fan shaped x-ray beam in a CT scanner. The intensity of the unattenuated fan shaped x-ray beam in the scan plane depends on the distance from the focal spot and the beam-shaping filter (bow tie filter). Perpendicular to the scan plane, at a certain distance from the focal spot, the unattenuated fan shaped x-ray beam has an area of more or less constant intensity, referred to as the full beam, and an area with gradual decrease of the full beam, also referred to as the penumbra (figure 5). The geometric efficiency is defined as the percentage ratio of the total nominal width of all simultaneously acquired sections and the FWHM of the irradiated area along the z-axis. The total nominal width of all simultaneously acquired sections is the number of simultaneously scanned sections times their nominal section width. For a single-slice CT scan, the row of detector elements encompasses the entire x-ray beam, including the full beam and the penumbra. For data acquisition in multi-slice CT scanners, the area of the penumbra is usually neglected, i.e. it is not detected and so is not used for image reconstruction. A typical width of the penumbra is about 1 - 1.5 mm on either side of the full beam (measured as the full width at half maximum in the center of rotation). In single-slice CT scanning the geometric efficiency, i.e. percentage of radiation that is actually used for image reconstruction, is generally 100%, assuming that post patient collimation is not applied; however, this might be reduced down to 50% when operating at small nominal slice widths, i.e. smaller than 2 mm due to post-

patient collimation. In multi-slice CT scanning, the radiation within the penumbra goes undetected and this necessarily implies a decrease in the geometrical efficiency. Table 4 provides an indication of the geometrical efficiency for two fictitious CT scanners assuming a realistic penumbra width of 1.0 - 1.5 mm on either side of the full beam (full width at half maximum). Note that geometric efficiency for a 4 x 0.5 mm or 4 x 0.75 mm scan configuration could be as low as 40 – 60 %, implying that with these scan configurations only half of the radiation dose is effectively used for image reconstruction. For broader beams the effect becomes less pronounced and for a 16 x 2 mm beam the geometrical efficiency should be 90% or better. The table shows that as a consequence, the geometrical efficiency for 4-slice CT scanners is poor when compared with 16-slice scanners.

Overscan

In helical scanning images are reconstructed at any position along the z-axis by interpolation of adjacent transmission profiles. This implies that some additional pre- and post-scan rotations of the x-ray tube are required in order to reconstruct images at the borders of the scan range, i.e. at the start and the end of the selected volume. This is also referred as overscan. Therefore, in helical scanning, the exposed range is always larger than the reconstructed volume. Typically, the pre- and post scan is one rotation both at the start and at the end of the planned scan range. In some cases the pre- and post scan may consist of two rotations, both at the start and end of the scan range. The contribution to patient dose (DLP or E) from the pre- and post scan is relatively high for scans with a large beam thickness. For example, 4-slice scanning with a 2 or 3 mm section thickness could typically yield an additional exposed range of 16 – 24 mm. For 16-slice scanning, the effect of the pre- and post-scan on patient dose might be even more pronounced; with a 1 mm or 1.25 mm section thickness typically yielding an additional exposed range of respectively 32 mm or 40 mm. The proportional contribution of the pre- and post-scan is of course particularly high in the case of a relatively small scanned range for a 16-slice scanner or in case of a 16 slice scan with thick sections.

Table 4. Indication of the typical geometrical efficiency of fictitious multislice scanners.

4-slice scanners

Section thickness mm	Collimation Mm	Geometrical efficiency %	Configuration
0.5	2	40-50	4 x 0.5 mm
0.75	3	50-60	4 x 0.75 mm
1	4	55-65	4 x 1 mm
2	8	75-80	4 x 2 mm
2.5	10	80-85	4 x 2.5 mm
5	20	85-90	4 x 5 mm

16-slice scanners

Section thickness mm	Collimation Mm	Geometrical efficiency %	Configuration
0.5	8	75-80	16 x 0.5 mm
0.75	12	80-85	16 x 0.75 mm
1.25	20	85-90	16 x 1.25 mm
2	32	90-95	16 x 2 mm

**) The geometrical efficiency is the percentage of the radiation beam that is actually used for image reconstruction; the loss of geometrical efficiency is due to overbeaming.*

DLP and effective dose

Exposure factors that are primarily related to DLP are the length of the scanned range and the number of sequences. These aspects are discussed in the next paragraph on clinical principles of MSCT.

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