Master of Science Thesis

2003

LUJI-RADFYS-EXE-2/2003

Reduction of Patient Dose in CT: Adjustment of Tube Current to Patient Size

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Abstract

In this study we have investigated the possibilities of dose reduction in CT at a constant level of image quality by adjustment of the tube current to the phantom size. The motivation with the study is the high radiation doses to the patients and the increasing frequency of CT examinations and the variety of examinations in CT, and that the same tube current is used for different patient sizes. According to UNSCEAR (2000) the effective dose from CT examinations are in general relatively high, typically 1- 30 mSv. These doses can often approach or exceed levels known to increase the probability of cancer. The conclusion based on the results from a survey of the frequency of radiological examination in Sweden during the period 1998-99 (Leitz, 1999), is that a relatively small fraction of all examinations (11 %) gives rise to a large contribution to the collective effective dose (68 %).

The aim of the study was to reduce the patient dose without reducing the image quality. This is possible with adjustment of the tube current to the patient size.

To establish the relation between phantom size, image quality and dose we scanned tree different phantoms, two of these were the standard CT dosimetry phantoms (16 and 32 diameter) and the third was a similar phantom with 24 cm diameter. The scans were made on a Siemens Somatom Plus 4, with standard scan technique for abdomen. To estimate the image quality at different levels of dose, we made measurements with reduced mAs and determined the signal-to-noise ratio, SNR. Finally for the different phantom sizes, we determined the necessary dose to keep the SNR at the same level as when the body phantom was imaged with 185 mAs.

The results of the study showed that the mAs settings can be reduced to about 41 mAs in the phantom with diameter 24 cm (the medium size phantom) and to 20 mAs in the 16 diameter phantom (the small phantom) and still get the same level of SNR as in the body phantom at 185 mAs. This means that by using a lower mAs setting for smaller patients a considerable dose saving can be achieved.

Determinations of SNR in patient images have also been performed. Images of twenty-four patients who had a CT examination of the liver were evaluated. The result was that the dose to the patient can be reduced to half for patients weighing about 60 kg and still get the same SNR as for an 80 kg patient with the original mAs setting.

With the knowledge from this study, a measuring-tape graded with mAs setting can be produced. Before examining a patient the technician would measure the circumference of the body part of interest for the patient with the mAs-tape to find what mAs setting should be used. More than a 50 percent reduction in patient dose is possible by appropriate choice of scan parameters.

Abstrakt

I denna studie har vi undersökt möjligheten för dos- reduktion i CT med samma bildkvalité genom att anpassa rörströmmen efter patientens storlek. Motivationen med studien var den alldeles för höga stråldosen som används vid CT-undersökningar, samt att antalet och variationen av datortomografiundersökningar ökar ständigt inom röntgendiagnostiken. I de allra flesta fallen anpassas man inte rörströmmen efter patientens storlek och därmed får smalare patienter en onödigt hög stråldos.
Enligt UNSCEAR är den effektiva dosen från CT- undersökningar i allmänhet relativt hög, 1-30 mSv. Stråldoser av denna storlek orsakar en högre sannolikhet för att patienten ska utveckla en cancersjukdom som en direkt följd av bestrålningen. Av Statens strålskyddsinstituts rapport, 2001:01, kan man dra slutsatsen att en relativt liten fraktion av alla röntgenundersökningar är CT-undersökningar (11 %), men att de står för en hög andel av den kollektiva effektiv dosen (68 %).

Studiens mål var att reducera stråldosen till patienter som undersöks med datortomografi utan att minska på bildkvalité, genom att anpassa rörströmmen efter patientens storlek.

För att undersöka relationen mellan patientstorlek, bildkvalité och stråldos använde vi tre olika storlekar av fantom. Två av dessa var standard dosimetrifantom för CT (16 och 32 cm diameter) och det tredje fantomet var ett liknande fantom, 24 cm diameter. Fantomen är cylindriska och tillverkade av PMMA. Mätningarna gjordes på en Siemens Somatom Plus 4 CT-scanner med standardinställningar som används för buk-regionen. För att bestämma bildkvalitén vid olika dosnivåer gjordes mätningar med reducerade mAs-tal och signal-till-brus förhållandet, SNR, bestämdes. Till slut bestämde vi det mAs-tal som för olika fantom standardinställningar som används för buk-regionen-storlekar gav samma SNR-nivå som för det stora fantomet ("body" fantom) vid det ursprungliga mAs-talet 185 mAs.

Studien visar att mAs-talet kan reduceras till 41 mAs för det mellanstora fantomet (24 cm diameter) och för det lilla fantomet (16 cm diameter) till 20 mAs och ändå få samma nivå av SNR som i body fantomen vid 185 mAs. Därmed kan en stor dos reducering göras för smalare patienter genom att använda lägre mAs.

Bestämning av SNR gjordes också i patientbilder. Tjugofyra bilder från patienter som gjorde CT-undersökningar av lever utvärderades. Vi fann att mAs-talet som används för en smal patient (60 kg) kan reduceras till hälften och ändå få samma SNR som för en 80 kg patient med den ursprungliga mAs -inställningen.

Med denna kunskap kan vi producera ett mAs -måttband. Innan en undersökning mäter man patientens omkrets i kroppsregionen som ska undersökas. mAs-måttbandet är graderat med mAs-tal och det mAs-tal som ska användas visas på måttbandet. En stor minskning av stråldosen till patienter kan göras genom ett lämpligt val av scan parametrar.

Introduction

When computed tomography (CT) was introduced in clinical practice in the seventies it revolutionised diagnostic radiology. CT first became possible with the development of modern computer technology in the sixties, but some of the ideas on which it is based can be traced back to the 1917, when the Bohemian mathematician J.H. Radon proved in a research paper of fundamental importance that the distribution of a material or material property in an object layer can be calculated if the integral values along any number of lines passing through the same layer are known. The physicist A.M. Cormack, in 1963, carried out the first experiments on medical applications of this type of reconstructive tomography. He worked on improving radiotherapy planning at Groote Schuur Hospital, Cape Town, South Africa. Without knowledge of previous studies he developed a method of calculating radiation measurement (Cormack, 1963). Cormack discovered that Radon had been unaware of even earlier work on the subject by the Dutch physicist H.A. Lorentz, who had already proposed a solution of the mathematical problem for the 3D case in 1905 (Cormack, 1992).
The English engineer G.N. Hounsfield, who is generally recognized as the inventor of CT, first achieved a successful practical implementation of this theory in 1972. He, like his predecessors, worked without knowledge of the abovementioned earlier findings (Kalender, 2000). In 1979, G.N. Hounsfield and A.M. Cormack received the Nobel Prize in medicine for the invention of computed tomography.

In the early days of computed tomography, the technique was only used for head examinations. The first CT scanner allowed distinguishing infarcts and tumours from normal brain tissue without the administration of contrast agents. These new abilities can explain the rapid development of CT. The primary concern in CT was to define strategies for examinations and to develop standards for scanning protocols, thereby attributing a lower priority to dose aspects (Nagel, 2000). It was not until the end of the 1980s that a larger number of CT users became aware of the relatively high radiation doses to the patient in CT examinations and the increasing frequency and variety of examinations (EC, 2000).

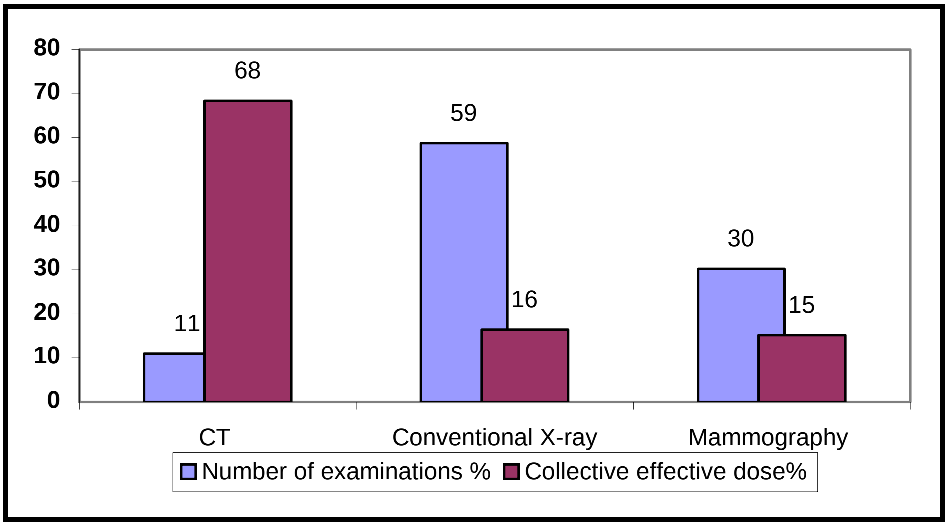


Fig.1
Percentage contribution from various X- rays examinations to the collective effective dose.

In fig. 1 an estimation of the contribution from CT to the radiation exposure of patients is presented. The data is based on a survey of the frequency of radiological examination in Sweden, performed by the Swedish Radiation Protection Authority (SSI) during the period 1998-99 (Leitz, 1999). The result for CT is a bit remarkable: a relatively small fraction of all examinations gives rise to a large contribution to the collective effective dose. CT is by far the single largest contributor to the radiation exposure to the population in Sweden from diagnostic medical source. This is also true for many other industrialised countries, e.g. Germany (Nagel, 2000), according to UNSCEAR (2000) effective dose in CT are typically 1-30 mSv.

The past few years' savings in collective dose in conventional radiography have been achieved by e.g. introducing film-screen combinations of higher speed index. However, this dose saving have been compensated by the increasing dose contribution from CT examinations. In spite of a growing awareness of radiation dose to patients among radiologists, the collective dose is increasing, due to an increasing number of CT examinations and more advance CT examinations.

The dose distribution in the patient is completely different between CT and conventional projection radiography. In conventional projection radiography the absorbed dose decreases continuously with the depth from the entrance of the X-ray beam to the exit. In the case of CT, with its rotational geometry, the dose is almost equally distributed in the scanning plane. The patient is equally irradiated from all directions during a complete rotation of the X-ray tube. The dose at the points of entry is higher than in the centre and the greatest variations occurring are only by a factor of 2 to 5 compared to conventional where the variations are as great as a factor of 100 between entrance dose and the exit dose (Kalender, 2000).

Image quality in CT depends primarily on two types of scan parameter: dose related parameters and parameters, which are related to processing and viewing of the image.
The slice thickness, inter-slice distance, pitch factor, volume of investigation, exposure factors and gantry tilt are all dose-related parameters.
Slice thickness for clinical requirements lies in the range of between 0.5 mm and 10 mm. The larger the slice thickness, the greater the low contrast resolution in the image due to higher signal-to-noise ratios. The smaller the slice thickness, the greater the spatial resolution in the z-direction. Artefacts, due to partial volume effects, can occur if the slice thickness is large. On the other hand, if the slice thickness is 1-2 mm the image may be significantly affected by noise.
Pitch factor in clinical practice lies in the range between 1 and 2. If we have a constant volume of investigation and use a low pitch factor, both the local dose and the integral dose to the patient is increased.
Volume of investigation depends on the clinical needs; the greater the size of this volume, the higher the integral dose to the patient.
Exposure factors are defined as the settings of x-ray tube voltage (kV), tube current (mA) and exposure time (s). At given values of tube voltage and slice thickness, the image quality depends on the product of x-ray tube current (mA) and exposure time (s). An increase in mAs gives a proportional increase in the dose to the patient.
Gantry tilt can be used to reduce the radiation dose to sensitive organs or tissues (e.g. eye lens, uterus) and/or to reduce or eliminate artefacts (EC, 2000).

Processing parameters are field of view, number of measurements, reconstruction matrix size, reconstruction algorithm and window settings for viewing the image.

Field of view, the choice of a small FOV allows increased spatial resolution in the image, because the whole reconstruction matrix is used for a smaller region than in the case of larger FOV. The selection of the FOV must take into account not only the opportunity for increasing the spatial resolution but also the need for fitting in the whole area of possible disease.
Reconstruction matrix is the array of rows and columns of pixels in the reconstructed image, typically 512 x 512.
Reconstruction algorithm: In most CT scanners, several reconstruction algorithms are available depending on clinical requirements, e.g. algorithms for head or body examinations.
Window width is defined as the range of CT numbers converted into grey levels and displayed on a monitor; a large window width represents a good choice for acceptable representation of a wide range of tissues.
Window level is defined as the central value of the window used for the display of the reconstructed CT image (EC, 2000).

Image quality, which is to be seen as the "minimum necessary" image quality for a particular application, must be ensured for all patients and without exception for all slices of the volume to be examined, with minimum dose. In the current report we use signal-to-noise ratio (SNR) to compare the quality of the images. SNR is the ratio of the strength of the signal for information content in the image and the noise level (the standard deviation of the signal). In those cases where a high SNR is necessary, high values of mAs should be selected.

Local dose quantities are indicators of the intensity of the irradiation inside the limits of the irradiated body region. The most specific dose quantity for CT that has been used is the Computed Tomography Dose Index (CTDI). Ideally, the radiation profile would be box-shaped (fig. 2), but due to scattering the radiation profile looks like a peak with tails in the z-direction.

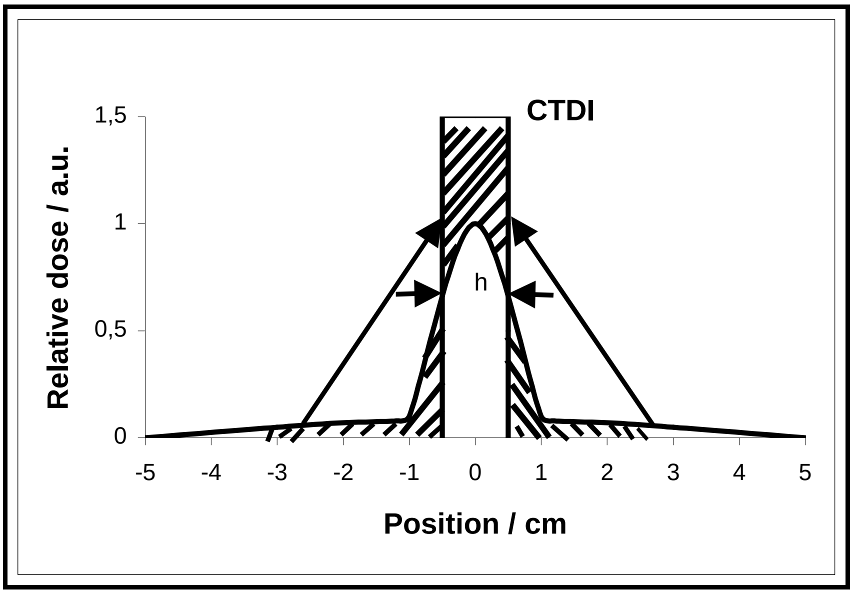


Fig.2
Illustration of the term CTDI

CTDI is equivalent to the dose value that would result if the absorbed radiation dose profile were entirely concentrated within a width equal to the nominal slice thickness (fig.2). All contributions from the dose profile are summed and divided by the nominal slice thickness. The CTDI is defined by the following equation:

$$\mathrm{CTDI} = \frac{1}{h} \cdot \int_{-\infty}^{+\infty} D(z) \cdot dz \ ,$$

where D(z) is the dose at a given location z, and h is the nominal slice thickness (Nagel, 2000).

In practice, CTDI is measured with a pencil-like ionisation chamber with an active length of 100 mm. A measurement performed with such a chamber is denoted by the subscript 100: $CTDI_{100}$. CTDI can be measured free in air or in a standard CT dosimetry phantom. When measurements are performed free in air, the pencil ionisation chamber can be placed on or parallel to the centre of rotation of the scanner ($CTDI_{100,\ air}$). The standard CT dosimetry phantom is made of a cylindrical block of PMMA of two different diameters: 16 cm representing "head", and 32 cm representing "body". At five different locations in the phantom, measurements with a pencil ionisation chamber can be performed (fig 3). A measurement in the centre is denoted $CTDI_{100,\ c}$ position A (c=central) and at positions B – E, which are at 10 mm depth from the surface the measurements are denoted $CTDI_{100,\ p}$ (p=peripheral).

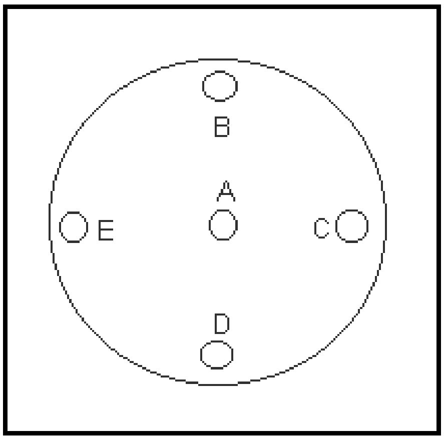


Fig.3
Arrangement of the locations A to E for the determination of CTDI in a standard CT dosimetry phantom
($CTDI_{100,\ c}$ -locations A; $CTDI_{100,\ p}$ -locations B - E)

Generally measurements of CTDI are normalised to unit radiographic exposure, Q, (i.e. normalised to the mAs setting) and this is represented by the subscript 'n', (${}_{n}CTDI$), (EC, 2000).

Measurements of CTDI in the standard head or body CT dosimetry phantom may be used to provide an indication of the average dose over a single slice. If we assume that the dose in a particular phantom decreases linearly with radial position from the surface to the centre then we can approximate the dose to a slice by the weighted CTDI (EC, 2000) ($CTDI_w$), defined by the following equation:

$$CTDI_w = 1/3 \cdot CTDI_{100,\,c} + 2/3 \cdot CTDI_{100,\,p}\ (mGy)$$

where $CTDI_{100,\,p}$ represents an average of measurements at the four different locations at the periphery of the phantom (locations B – E).$CTDI_w$ is determined separately for the two phantom sizes. It is important to differentiate between absolute and normalized values of $CTDI_w$. The normalised weighted CTDI, (${}_nCTDI_w$) is defined as

$${}_nCTDI_w = (1/3 \cdot CTDI_{100,\,c} + 2/3 \cdot CTDI_{100,\,p})/Q\ \ (mGy/mAs),$$

where Q is the radiographic exposure (mAs setting). This quantity is the most unambiguous definition of the dose output from a CT scanner and is therefore used in this report.

To allow comparisons with other types of radiological examinations, the effective dose (E) is determined. The effective dose may be derived from values of DLP for an examination using appropriate conversion factors and the following equation:

$$E = E_{DLP} \cdot DLP\ (mSv),$$

where E_{DLP} is the conversion factor (mSv/mGy · cm), and DLP is the dose-length product defined as

$$DLP = CTDI_w \cdot h \cdot n\ (mGy \cdot cm),$$

where h is the slice thickness and n is the number of slices.

General values of the conversion factor, E_{DLP}, appropriate to different anatomical regions of the patient (head, neck, chest, abdomen or pelvis) are given in Table 1, (EC, 2000).

Table 1
Conversion factors (effective dose per DLP) for various body regions

Region of body	Conversion factor, E_{DLP} (mSv/mGy · cm)
Head	0.0023
Neck	0.0054
Chest	0.017
Abdomen	0.015
Pelvis	0.019

The absorbed dose to the patient depends strongly on the selected examination parameters, and also on the clinical and physical characteristics of the patient. There is a simple relationship between the absorbed dose to the patient and the product of the tube current (mA) and the exposure time (s): the absorbed dose to the patient is a linear function of the current-time product (mAs). If we reduce the mAs value by a factor of two, then the dose is reduced by the same amount. The noise level, however, increases by a factor of $\sqrt{2}$, when the dose is reduced to 50%, i.e. the relationship between the noise in an image and the dose is not linear. The total dose to the patient depends linearly on the size of the examined volume, so the examination volume should always be kept as small as possible.

The collective dose from CT examinations is unnecessary high in industries countries (Kalender, 2000). One reason for this is that CT scanners in general are not equipped with automatic exposure control. The same exposure parameters are used no matter the patient size, e.g. the mAs settings are often the same for large patients and for small patients. The images for the smaller patients therefore have a higher signal-to-noise ratio than what is needed, and the patient dose is also unnecessary high. The higher signal-to-noise ratios for the smaller patients do not necessarily result in a higher diagnostic quality.

Aim

The aim of this project is to estimate the possible dose savings by adjustments of the mAs setting to the patient size.

Materials and methods

The CT scans were made on a Siemens Somatom Plus 4, with a standard scan technique for abdomen (120 kV, 10 mm slice thickness). 185 mAs was used as a standard exposure setting. Measurements on phantoms were also made with reduced mAs (170, 150, 130, 110, 90, 70, 50, 34 mAs) while the other parameters were kept constant. By comparing the SNR from the different images, a conclusion can be made on how much dose is needed for imaging a particular phantom size (or patient size). It was decided to keep the SNR at the same level as when the body phantom was imaged with 185 mAs. By adapting the mAs value to the phantom size (keeping all other scan parameters constant) the same level of SNR could be achieved and thus the dose to the smaller phantoms (see below) could be reduced. At the same time we measured CTDI for the different phantom sizes for calculation of the effective dose.

Measurement of CTDI

CTDI is usually determined in two sizes of the standard CT dosimetry phantom: a body phantom (32 cm diameter) and a head phantom (16 cm diameter). In this study we have also included a third size of the phantom, which is 24 cm in diameter, for achieving more solid results (fig. 4) (i.e. more data points). The CTDI is determined by measurements of the dose at the five different positions in the phantom. The CTDI phantoms include a central bore, and a peripheral bore at a depth of 10 mm from the surface. The dose at the peripheral position is measured at 0°, 90°, 180° and 270°, and a mean value of these four measurements is calculated. The central bore allows estimates of organ dose for organs on or near the axis of the body, whereas peripheral bore is representative for the doses to organs which are located at or near the surface of the body. The smaller phantom ("head phantom") is representative for

Fig.4
Cylindrical standard CT dosimetry phantoms. The smaller phantom can be fitted the in larger phantoms to achieve all the phantom sizes used in this study.

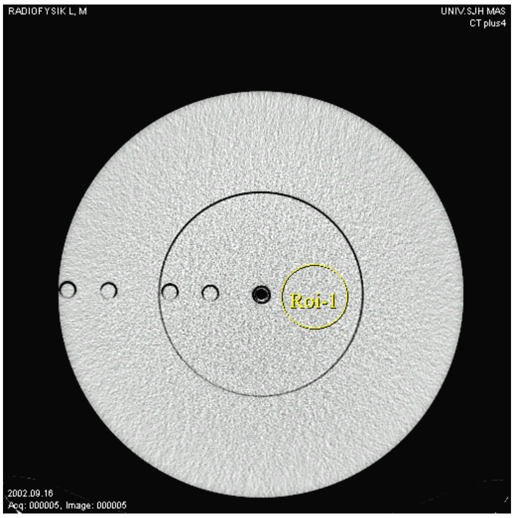
CT examinations of the head and neck, and for children. The large phantom (“body phantom”) is representative for CT examinations in various regions of the trunk for a normal sized adult. The longitudinal dimensions of the phantom must be larger than the active length of the detector used for the dose measurements. Measurements were made with a pencil ionisation chamber (Radcal Corporation, model 90X6, calibrated by the producer), fig.5, and the measured absorbed doses were recorded. Three measurements were performed at each mAs-value to estimate the standard deviation.

Fig.5
Pencil ionisation chamber for measurements of CTDI

The measurements needed for calculating the CTDI were performed parallel to the axis of rotation during one complete tube rotation, while the table rotation remained stationary. When the ionisation chamber was inserted into one of the bores, the other ones were filled with inserts of PMMA.

Determination of SNR

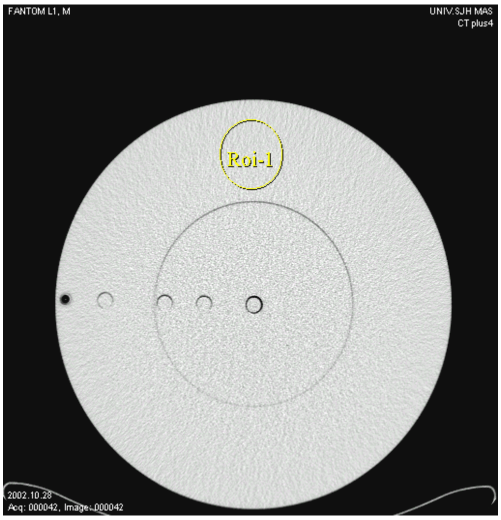
All images produced during the determination of CTDI were transferred to a workstation for further evaluation. To assess the signal-to-noise-ratio, a region of interest (ROI) was placed in a homogeneous region in the central part of the phantom (fig. 6) and in the periphery of the phantom (fig. 7). The software package Osiris (http://osiris.glorifyjesus.com) was used for the estimation of the signal (mean Hounsfield unit, HU) and the standard deviation of the signal inside the ROI.



ROI Data:

Min:	Mean:	Max:	Standard deviation:
46	**125**	203	**22**

Fig.6
Illustration of phantom and ROI's emplacement



ROI Data

Min:	Mean:	Max:	Standard deviation:
20	**131**	262	**32**

Fig.7
Illustration of phantom and ROI's emplacement

Figures 6 and 7 shows where we placed the ROI's. Adjacent to these two figures are examples of data obtained from the ROI's. The SNR is calculated by dividing the mean HU value by the standard deviation. The maximum HU and the minimum HU are also shown for illustration.

Determination of SNR in patient images

Forty-eight images of patients who had a CT scanning of the abdomen were collected and investigated. Both the images with and without contrast agent were examined. The scans were made on a Philips Tomoscan SR 7000, with a standard scan technique (250 mAs, 120 kV and slice adjustment 7-10-7 (slice thickness-pitch-reconstructed slice thickness)). Images with contrast medium are taken with standard contrast concentration 300 mg/ml, volume 95 ml, injection speed 3,0 ml/s and delay 65 s.

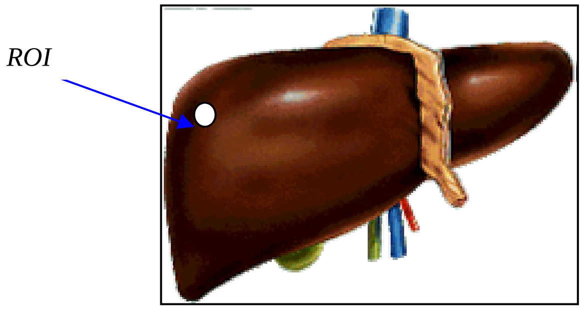


Fig.8 Liver

For the determination of SNR, we placed an ROI in the right liver lobe in the same height for each patient just above vena portea in an area without big vessels. The mean HU and the standard deviation of the HU in this ROI were recorded, and the ROI was calculated for each image. Weight and length of the patient were also noted for comparisons of SNR with the patient size.

Results

Determination of SNR

Fig.9 shows the relationships between SNR and mAs in the centre of three CT phantoms.

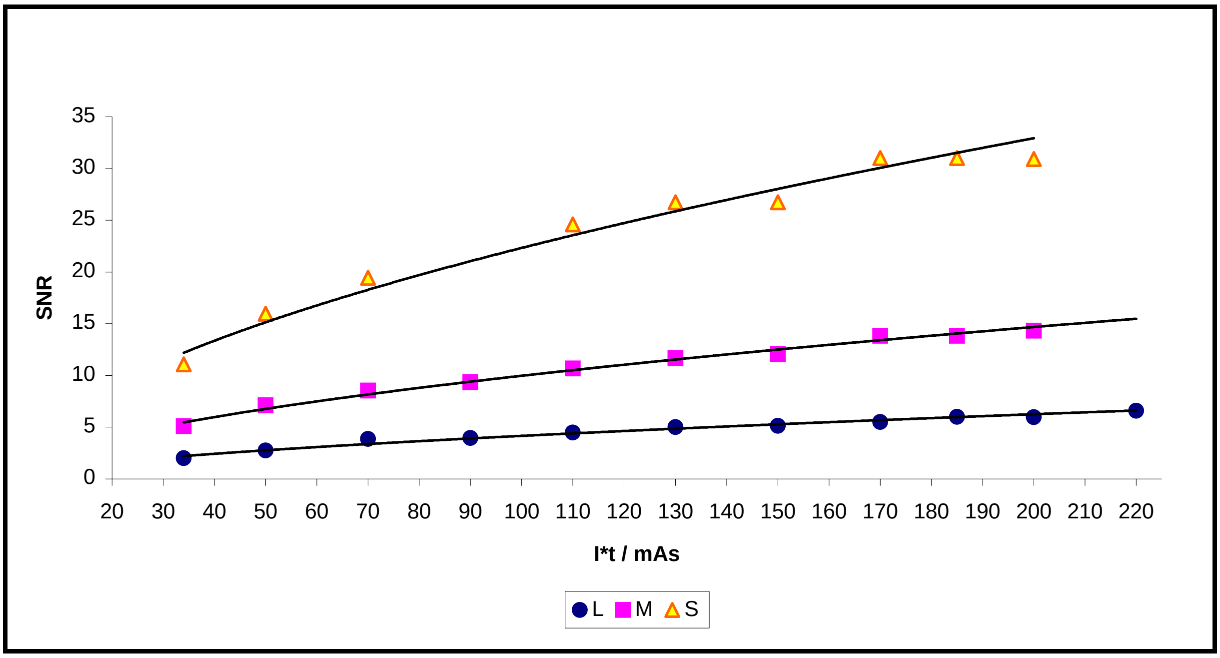


Fig.9
SNR as a function of mAs.–Measurements are performed centrally in the three phantoms (L- body phantom with diameter 32cm (large); M- phantom with diameter 24 cm (medium); S-head phantom with diameter 16 cm (small)).

An important conclusion to be drawn from fig.9 is that we can reduce the mAs setting to about 41 in the phantom with diameter 24 cm (the medium sized phantom) and 20 mAs in the head phantom (the small phantom) and still get the same value of SNR like in the body phantom (the large phantom) at 185 mAs.

Fig.10 shows the relationships between SNR and mAs in the periphery of three CT phantoms. The slopes of the three curves are the same as for the curves in fig.9.

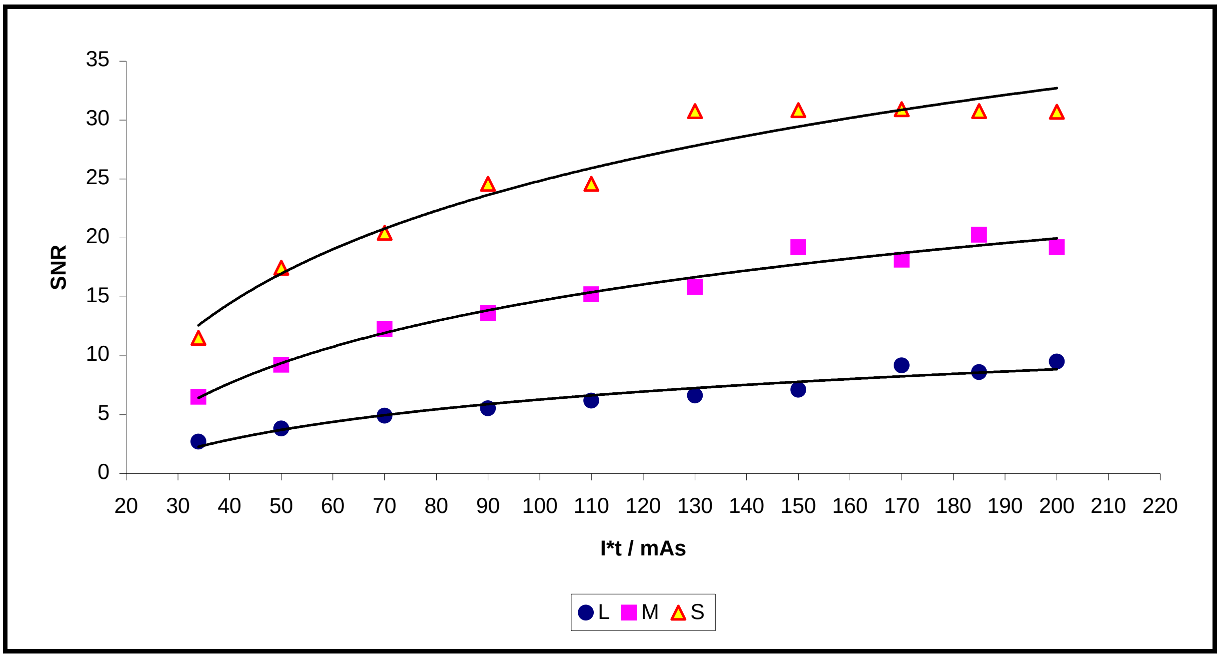


Fig.10
SNR change with mAs - periphery (L- body phantom with diameter 32cm; M- phantom with diameter 24 cm; S-head phantom with diameter 16 cm)

In order to assess the dose reduction that can be achieved by using a proper mAs setting with respect to the size of the patient a calculation example is presented below (example 1). The effective doses to the medium sized phantom exposed with the original mAs setting and the reduced mAs setting are compared.

Example 1:
Number of slices = 10; slice thickness 10 mm and E_{DLP} = 0.015 mSv/mGycm (table 1)

At 185 mAs:

$CTDI_{100,c}$= 14.69 ± 0.02 mGy
$CTDI_{100,p}$= 21.26 ± 0.08 mGy
$CTDI_w$ = 19.07 ± 0.08 mGy

E = 19.07 mGy*10*1 cm*0.015 mSv/mGycm = **2.86 ± 0.01** mSv

At 50 mAs: (50 mAs since 41 mAs cannot be selected on this particular CT scanner)

$CTDI_{100,c}$= 3.94 ± 0.13 mGy
$CTDI_{100,p}$= 5.54 ± 0.08 mGy
$CTDI_w$ = 5.01 ± 0.15 mGy

E = 5.01 mGy*10*1 cm*0.015 mSv/mGycm = **0.75 ± 0.01** mSv

At the same level of SNR as for the large phantom, the dose to the medium sized phantom can be reduced to a quarter of the original dose. Obviously, the dose reduction is the same as the percentage decrease in mAs.

Fig.11 shows the necessary mAs setting to achieve a constant level of SNR as a function of the circumference of the phantom (patient). The values in the figure are from experimental data. The figure shows the pronounced dose savings that can be achieved by reducing the mAs setting for smaller patients.

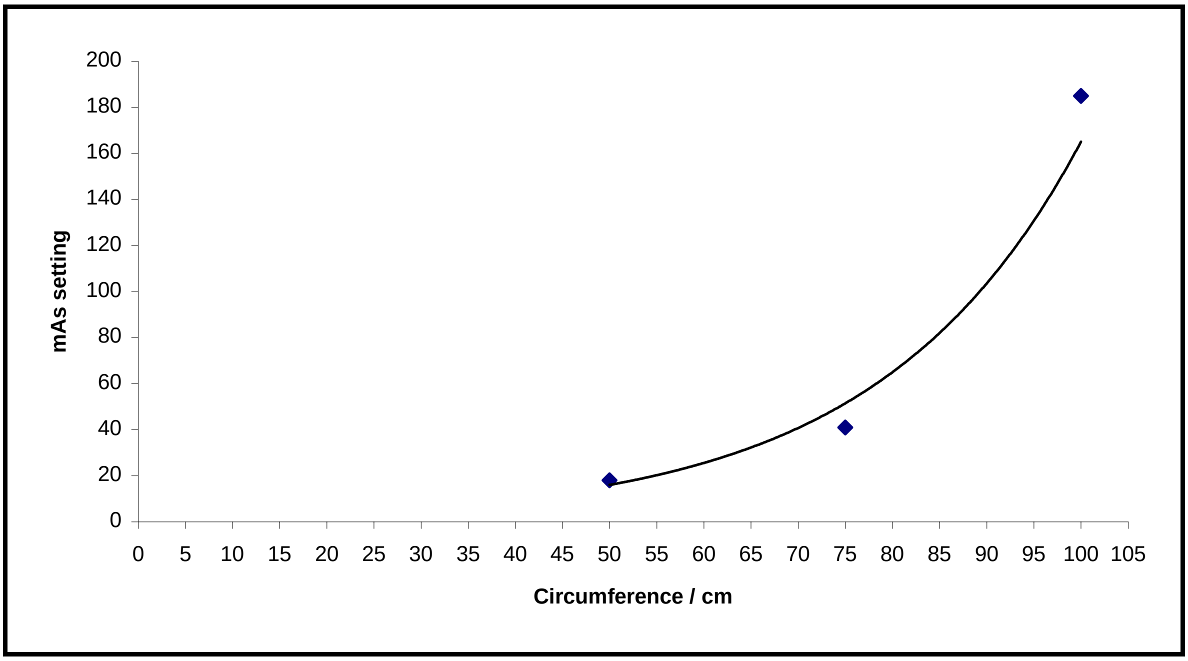


Fig.11
mAs setting as a function of phantom circumference (data points indicate experimental values) at a constant SNR level

Determination of SNR in patient images

Fig.12 shows the results of the determination of SNR in the patient images of the liver. The SNR of the images from smaller patients is higher than for larger patients. This means that it would be possible to reduce the mAs settings and thus the patient dose for smaller patients in accordance with the results above, and still get the same SNR as for larger patient at the original mAs setting. Example 2 illustrates this finding.

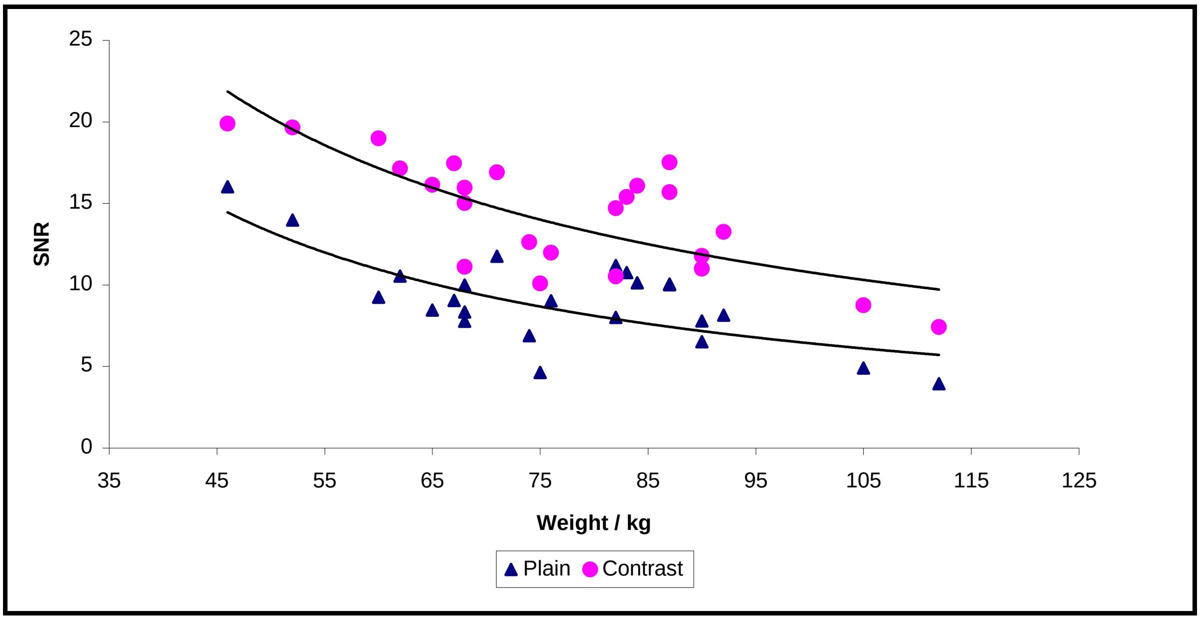


Fig.12
SNR measurements in the liver of patient images (Plain = without contrast)

Example 2

(measurement in patient images)

Effective dose to a 60 kg patient, with original mAs and with mAs reduction.

Scanner: Philips Tomoscan SR 7000

${}_{n}CTDI_{w}$ (body) = 0.08 mGy/mAs (Nagel, 2000)
Number of slices = 10
Slice thickness = 10 mm
E_{DLP} = 0.015 mSv/mGycm (table 1)

At 250 mAs

E = 0.08 mGy/mAs * 250 mAs * 10 * 1 cm * 0.015 mSv/mGy = **3** mSv

At 110 mAs

E = 0.08 mGy/mAs * 110 mAs * 10 * 1 cm * 0.015 mSv/mGy = **1.32** mSv

Under the conditions given above for this particular scanner, the effective dose for standard CT examination of the abdomen can be reduced to less than half if the mAs settings is adjusted properly to the patient size. Again the SNR is the same as for a large sized patient at the original mAs setting, and thus the necessary diagnostic quality of the images is not affected.

Discussion

The radiation dose from CT is relatively high and the frequency of CT examinations and the different types CT examinations are growing. The absorbed dose in tissue can vary between 10-100 mGy and these doses can often approach or exceed levels known to increase the probability of cancer (EC, 2000). Different CT scanners have different standard protocols for similar examinations, and therefore the patient dose can vary significantly between different types of CT scanners (EC, 2000). There is a continuing need to balance the benefits and risk to patients. In principle, this means to eliminate unnecessary exposures but still, to keep clinical justification. More than a 50 percent reduction in patient dose is possible by appropriate choice of scan parameters. A strategy for reducing the patient doses from CT is needed.

Too often, all patients, regardless of their size, are examined using the same protocol, including the same mAs setting. For smaller patients this results in images where the image quality (signal-to-noise ratio) is unnecessary high. The increased SNR do not improve the diagnostic quality of the images. In this study we have investigated how to choose an appropriate tube current and exposure time for different sizes of patients. By adjustment of the mAs setting to the patient size, the image quality can be kept good enough to satisfy the clinical needs. By choosing the mAs setting wisely there is a great potential for reduction of patient dose. Instead of examining a patient with a circumference of 75 cm (this corresponds roughly to a 70 kg patient) with the standard mAs setting (which gives satisfactory image quality for a patient with 100 cm circumference or roughly 80 kg) the mAs setting and also the effective dose to the patient can be significantly reduced (fig.11). Note that the important parameter to measure is the circumference of the body part where the medical examination will be done (e.g. the abdomen) instead of the weight of the patient. The mAs setting should then be adjusted to the circumference of the body part of interest. A reduction of the mAs significantly reduces the patient dose and also lengthens the x-ray tube life.

Determinations of SNR in patient images have also been performed. Images of twenty-four patients who had a CT examination of the liver were evaluated. The weight and height of the patients were recorded and SNRs were determined. Again the relationship between SNR and patient size could be verified (fig.12). The figure shows that a reduction in patient dose can be achieved with a modulation of the tube current. The dose to a patient can be reduced to half for patients weighing about 60 kg and still get the same SNR as for an 80 kg patient with the original mAs setting. It should be noted that the dose savings given in fig.12 are smaller than what was predicted in the phantom study. This is probably caused by the fact that with human subjects the uncertainties increase and therefore a slightly higher dose is needed for the patient images than what was predicted.

Efforts to reduce the patient dose in CT have been done by other research teams. Two of these studies are presented below. Witling et al. (2001) scanned 19 patients on a Philips Tomoscan AV EU, using a 5-mm abdominal scan protocol and a soft reconstruction kernel. As nearly all scans with a standard scan technique (140 kV, 225 mAs) are of sufficient quality in their practice, they altered the scan parameters to 120 kV, 175 mAs and thereby they reduced the dose. They confirmed that the measured noise in a patient image increase with the patient diameter. In the other part of their study they tried to achieve constant image quality in a group of 22 patients by adjusting the individual dose to the patient size. They found that the dose could be reduced (mean 28%, range 0-76%) in 19 of the 22 patients.

Greess et al. (2000) investigated the potential of using an attenuation-based on-line modulation of the tube current to reduce dose, i.e. the tube current was adjusted to the thickness of the patient in a particular projection The tube current was reduced when the tube was situated in “frontal projection” and increased when the tube was in “lateral projection”. They chose six different anatomical regions which are typically covered by routine CT examinations. Thirty tumour patients were examined on a Siemens Somatom Plus 4 CT scanner. The authors found that the mean dose reduction by using the on-line tube current modulation compared to using standard constant tube current was 18 % in the skull base, 53 % in the shoulder region, 22 % in the thorax, 15 % in the abdomen, 25 % in the pelvis and in the knee region. Both of these studies show that it is possible to reduce the patient dose from CT examinations significantly. The high doses in CT stem mainly from two different facts: 1. CT scanners have not until recently been equipped with “automatic exposure control” (like e.g. on-line tube current modulation) and 2. imaging in CT have so far always aimed at achieving the best possible image quality, independent of the radiation dose to the patient and of the clinical requirements. With the advent of new CT scanners, equipped with automatic exposure control, the first of these two problems will disappear. The second problem, however, will need further work by the scientific community for coming up with a proper solution.

In the future it would be interesting to investigate the necessary dose for a particular examination and a particular patient size. A group of patients should be examined with a special CT protocol: With a stationary patient couch three slices are produced with different settings. The first slice is produced with the standard protocol. This slice will be used as a reference. The next slice will be produced with a tube current which is adapted to the circumference of the patient. This image will be compared to the first one, both with respect to SNR but also with respect to the diagnostic quality. A third slice will finally be produced with an even lower dose. This slice will be used for finding the lower limit of dose that is required for producing an image which necessary diagnostic quality. With the knowledge from this new study, a measuring-tape graded with mAs settings can be produced. Before examining a patient the technician would measure the circumference of the body part of interest for the patient with the mAs-tape to find what mAs setting should be used.

Conclusion

A 60 kg patient can be examined with only half the dose of an 80 kg patient and still the same level of image quality can be achieved.

Acknowledgement

I would like to acknowledge all the people at the Department of Radiation Physics and the Department of Radiology, Malmö University Hospital for all your help, support and patience.

There are some people I would like to mention explicitly:

My supervisor, Anders Tingberg, for great support and that you always find time for discussion and encouragement.

Professor Sören Mattsson for your time and enthusiasm.

Mats Nilsson for great knowledge in the subject.

Peter Leander and Raffaele la Mantia for discussions and collection of clinical data.

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ISBN3-89578-081-2

Leitz Wolfram Leitz and Helene Jönsson
Patient Doses of X-rays examinations in Sweden
SSI rapport: 2001:01 ISSN 0282-4434

Nagel Radiation Exposure in Computed Tomography
Fundamentals, Influencing Parameters, Dose Assessment,
Optimisation, Scanner Data, Terminology
Edited by H.D. Nagel Hamburg with contributions from
M. Galanski, N. Hidajat, W. Maier, H.D. Nagel. Th. Schmidt, 2000

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