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PAEDIATRIC COMPUTED TOMOGRAPHY EXPOSURE OPTIMIZATION

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Summary

The utilization of Computed Tomography in paediatric examinations constantly increases. During the procedure, a high amount of dose is delivered to children, which could be avoided. This study examined whether the selection of scanning parameters – tube voltage and tube current-time product - could be based on patient size instead of patient age or weight aiming, for dose reduction.

The SRS78 spectrum processor software was employed to generate spectra with tube voltage between 80kVp and 120kVp and with tube current-time product between 50mAs and 165mAs. These spectra were attenuated by different thicknesses of polymethylmethacrylate (PMMA) phantoms. The simulation technique was validated with experimental measurements acquired on CTDI phantoms on a Siemens Somatom plus 4 scanner. The image quality was assessed in terms of noise, contrast and contrast-to-noise ratio (CNR). Furthermore the contrast of iodine, adipose tissue and cortical-bone relative to muscle were calculated in order to examine how the contrast of different materials was influenced when tube voltage changed.

The data analysis shows that there is a definite relationship between image quality and the size of a patient. When exposure settings are kept constant, the level of noise, contrast and Contrast-to-Noise Ratio (CNR) depends on the size of the phantom. Noise is increased exponentially and contrast is reduced linearly as the size of the phantom is increased. CNR is markedly higher in small size phantoms. Moreover, when tube voltage was reduced the noise level was increased less in the small size phantoms and the contrast of high atomic number materials is reduced more when tube voltage is reduced. The CNR for high atomic number materials presents modest improvements when tube voltage is increased therefore examinations with contrast agents could be performed at lower tube voltages. Furthermore the high CNR in small size phantoms could be traded off with
lower mAs. In particularly the mAs could be reduced by up to 95% while maintaining the same CNR as for adults resulting in dramatic dose reductions for children.

Moreover, since Computed Tomography stands out from all the other X-ray techniques due to its ability to detect structures of similar densities the detect ability of low contrast details was investigated. The Catphan phantom and particularly the CTP515 module was employed. The phantom was scanned with the Siemens Somatom plus 4 scanner at 80kVp, 120kVp and 140kVp and with tube current-time product between 43mAs and 165mAs. The image quality was assessed subjectively and objectively.

It is observed that when 120kVp and 140kVp are applied there are not sufficient differences on image quality which justify the selection of 140kVp in paediatric protocols. When 80kVp is applied structures with contrast lower than 10HU are not detected. Concerning mAs does not contribute to the detection of low contrast details except if it is combined with high tube voltages. However, mAs contribute to the visualisation of smaller in size details but above a threshold value, higher mAs does not serve any purpose and the value of 300mAs employ in many protocols is not justified.

In conclusion, the reduction of dose during paediatric Computed Tomography examinations is more than probable since scanning parameters could be reduced without degradation of image quality. However in order to assure the reduction of dose without side effects, protocols must be constructed which will individualize the Computed Tomography examinations. That is, the optimum spectrum must be selected relative to the diagnostic task and the size of the patient.
Περίληψη

Η χρήση της αξονικής τοµογραφίας, λόγω της διαγνωστικής της αξιοπιστίας, ολοένα και αυξάνεται. Ένα ποσοστό της τάξης του 10% των εξετάσεων πραγματοποιείται στα παιδιά, ποσοστό το οποίο συνεισφέρει το 67% της συνολικής δόσης στον παιδικό πληθυσμό. Αυτό το υψηλό ποσοστό δόσης σε συνδυασμό με τη μεγάλη διάρκεια ζωής των παιδιών αυξάνει την πιθανότητα πρόκλησης καρκίνου. Για αυτό το λόγο, τα τελευταία χρόνια η διεθνής επιστηµονική κοινότητα πραγµατοποιεί έρευνες για την μείωση της δόσης.

Μέχρι στιγµής η επιλογή των παραµέτρων έκθεσης (π.χ διαφορά δυναµικού (kVp), ρεύµα-χρόνος περιστροφής (mAs) ) των ασθενών γίνεται από πρωτόκολλα τα οποία διαχωρίζουν τους ασθενείς είτε ανάλογα µε την ηλικία τους είτε ανάλογα µε το βάρος τους. Στόχος αυτής της εργασίας είναι η μελέτη, καταρχήν της σχέσης µεταξύ της ποιότητας εικόνας και του µεγέθους του ασθενή και κατά δεύτερον η δυνατότητα προσαρµογής των παραµέτρων έκθεσης στο µέγεθος του ασθενή µε απώτερο στόχο τη µείωση της δόση χωρίς την υποβάθµιση της ποιότητας της εικόνας.

Το υπολογιστικό πρόγραµµα SRS78 Spectrum processor χρησιµοποιήθηκε για την παραγωγή φασµάτων µε διαφορά δυναµικού µεταξύ 80kVp και 120kVp και µε ρεύµα το οποίο κυµάνθηκε από πρωτόκολλα τα οποία διαχωρίζουν τους ασθενείς είτε ανάλογα µε την ηλικία τους είτε ανάλογα µε το βάρος τους. Στόχος αυτής της εργασίας είναι η μελέτη, καταρχήν της σχέσης µεταξύ της ποιότητας εικόνας και του µεγέθους του ασθενή και κατά δεύτερον η δυνατότητα προσαρµογής µεταξύ των παραµέτρων έκθεσης στο µέγεθος του ασθενή µε απώτερο στόχο τη µείωση της δόση χωρίς την υποβάθµιση της ποιότητας της εικόνας.

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Τα αποτελέσματα δείχνουν μια ξεκάθαρη σχέση μεταξύ της ποιότητας εικόνας και του μεγέθους. Διατηρώντας σταθερές τις παραμέτρους έκθεσης το επίπεδο θορύβου, αντίθεσης και CNR καθορίζεται από το μέγεθος του ομοιώματος. Συγκεκριμένα, παρατηρήθηκε μία εκθετική αύξηση του θορύβου και μια γραμμική μείωση της αντίθεσης όταν το μέγεθος του ομοιώματος αυξάνεται καθώς επίσης το CNR είναι πολύ υψηλότερο στα μικρά μέγεθη σε αντίθεση με τα μεγάλα. Η αντίθεση των υλικών με μεγάλο ατομικό αριθμό μειώνεται γρηγορότερα όταν το μέγεθος αυξάνεται. Επιπρόσθετα όταν η διαφορά δυναμικού μειώνεται ο θόρυβος αυξάνεται σε μικρότερο ποσοστό στα μικρά μέγεθη συγκριτικά με την αύξηση που παρατηρείται στα μεγάλα μέγεθη. Επίσης όταν το δυναμικό μειώνεται η αντίθεση των υλικών με μεγάλο ατομικό αριθμό αυξάνεται περισσότερο. Αυτή η αλλαγή ημιούχως έχει ως αποτέλεσμα το CNR των υλικών με μεγάλο ατομικό αριθμό να παρουσιάζει μικρότερη βελτίωση όταν η τάση αυξάνεται. Ως εκ τούτου οι εξετάσεις που κάνουν χρήση υλικών αντίθεσης contrast agents μπορούν να πραγματοποιηθούν σε μικρότερο κυττάρο χωρίς σημαντικές επιπτώσεις στην ποιότητα εικόνας. Επίσης το υψηλότερο CNR στα μικρότερα μέγεθη μπορεί να αξιοποιηθεί με χρήση μικρότερου mAs. Συγκεκριμένα φαίνεται πως το mAs είναι δυνατόν να μειωθεί σε ποσοστό 95% διατηρώντας την ίδια ποιότητα εικόνας με ενήλικες ασθενείς. Αυτό θα έχει ως αποτέλεσμα την δραματική μείωση της δόσης στα παιδιά.

Η αξονική τοµογραφία εξερευνεί από άλλες διαγνωστικές μεθόδους λόγω της ικανότητας της να διακρίνει γιατικούς δομές με παρόμοιες πυκνότητες. Επομένως αντικείμενο μελέτης αυτής της εργασίας υπήρξε επίσης η ικανότητα ανίχνευσης δομών χαμηλής αντίθεσης. Συγκεκριμένα ερευνήσαμε πως οι παραγόντες έκθεσης επηρεάζουν την ανίχνευση τέτοιων δομών.

Για την πραγματοποίηση αυτού του μέρους χρησιμοποιήθηκε το Catphan phantom και συγκεκριμένα το CTP515 module. Το ομοίωμα (phantom) αυτό αποτελείται από 3 ομάδες με διαφορετικό επίπεδο αντίθεσης η κάθε μία (10HU, 5HU και 3HU). Κάθε ομάδα αποτελείται από 9 δομές διαφορετικών διαμέτρων από 2-15mm. Ο τομογράφος Siemens Somatom plus 4 χρησιμοποιήθηκε για την ακτινοβόληση του ομοιώματος στα
80kVp, 120kVp, 140kVp και στα 43-165mAs. Η ποιότητα εικόνας αξιολογήθηκε αντικειμενικά και υποκειμενικά.

Από τα αποτελέσματα διακρίνουμε ότι όταν χρησιμοποιούνται τα 120kVp και 140kVp δεν υπάρχουν σημαντικές διαφορές ως προς την ανίχνευση των δομών επομένως η χρήση των 140kVp σε πολλά πρωτόκολλα δεν δικαιολογείται. Επίσης όταν χρησιμοποιούνται τα 80kVp η ανίχνευση δομών με αντίθεση μικρότερη των 10HU δεν είναι εφικτή ως εκ τούτου η επιλογή της διαφοράς δυναμικού πρέπει να γίνει προσεκτικά.

Σχετικά με το mAs παρατηρήσαμε ότι δεν συνεισφέρει στην ανίχνευση δομών χαμηλής αντίθεσης εκτός και αν συνδυαστεί με το κατάλληλο kVp. Παράλληλα φαίνεται πως αυξάνοντας το mAs είναι δυνατό να ανιχνευθούν δομές μικρότερου μεγέθους. Παρόλα αυτά όμως παρατηρήσαμε ότι από μια τιμή κατωφλίου και άνω το mAs δεν συνεισφέρει ουσιαστικά σε ανίχνευση μικρότερων τιμών. Ως εκ τούτου η επιλογή των 300mAs στα πρωτόκολλα επίσης δεν δικαιολογείται.

Ως συμπέρασμα η μείωση της δόσης στην εξέταση αξονικής τομογραφίας σε παιδιά είναι κάτι παραπάνω από πιθανή. Για να διασφαλίσουμε όμως ότι παράλληλα με τη μείωση της δόσης διασφαλίζεται και η διάγνωση, είναι αναγκαία η κατασκευή πρωτοκόλλων. Για την κατασκευή όμως των κατάλληλων πρωτοκόλλων το φάσμα πρέπει να επιλεχθεί με βάση το μέγεθος το άσθενούς και με διαγνωστικό σκοπό. Επομένως πρέπει να υπάρξει εξατομίκευση της εξέτασης.
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Chapter 1 Introduction

Computed tomography is the first digital and slice imaging modality. It was introduced into clinical practice in Britain in the early 1970s and since then it has become an essential part of radiological diagnostics. Its high diagnostic reliability and the advances in the development of CT image quality through years, attributable to the technological evolution, led to the proliferation of CT examinations in paediatric population.

Despite all the benefits, Computed Tomography is responsible for delivering high amount of radiation dose, particularly to the paediatric population which is ten times more sensitive than adults. The radiation doses children are exposed to during a Computed Tomography examination are similar to that of the survivors of the atomic bomb. The high doses along with children’s longer life expectancy increase the probability of radiation damage. Therefore the optimization of Paediatric Computed Tomography is indispensable.

The patient dose depends on numerous parameters. One of these parameters is the size of the patient. This thesis focuses on the probabilities of dose reduction without degradation of image quality, by adapting scanning parameters namely, tube voltage and tube current-time product, to the size of the patient.

Computed Tomography stands out from all the other X-ray techniques due to its ability to detect structures of similar densities. Thus, this study scrutinizes the influence of scanning parameters on the detectability of low contrast details.

The ultimate aim is to optimize paediatric protocols by means of individualizing the examination. Individualization concerns the adjustment of the scanning parameters to the size of each child and the diagnostic task. Hopefully this will result in dose reduction without degradation of image quality.
Chapter 1. Introduction

1.1 Statement of the Problem

According to the current literature the theoretical risk for radiation-induced cancer from CT examination is very high. This is particularly important when the patient is a child, since children have more rapidly dividing cells than adults and have longer life expectancy, therefore the probability that children will develop cancers from X-ray radiation are significantly higher. Since the employment of computed tomography in paediatric examinations constantly increases and a continued increase is expected, a reduction in the radiation dose delivered during paediatric examinations is an important issue.

Scanning parameters, in most of the protocols they are selected relative to the age of the patient and the type of the examination. However, growth and development are variables of childhood and children of the same age can be at different growth and developmental stages. Consequently, when deciding the most appropriate health care approach it is important to allow for a child’s individuality and to avoid making assumptions about a child based upon preconceived ideas pertaining to specific chronological ages. The main criterion for the selection of these scanning parameters was the acquisition of the optimum image quality while the delivered dose was in second order of priority. As a result an unnecessary high level dose is delivered to children.

In view of the importance of radiation protection of children, the focus of this thesis is the optimization of paediatric Computed Tomography in order to reduce the delivered dose without degradation of image quality. The optimisation is under two considerations. The first investigate the relationship between image quality and patient size and the aim is threefold. Firstly, to demonstrate how image quality depends on patient’s size. Secondly, to investigate how scanning parameters influence the image quality relative to the size of the patient. Thirdly, to investigate whether is feasible to adapt the scanning parameters to the patient’s size in order to reduce the dose but at the same time without degradation of the image quality.
Chapter 1. Introduction

The second investigate the detectability of low contrast details and the aim is to understand how the scanning parameters influence the detectability of low contrast details and whether these parameters could be reduced.

1.2 Outline of Thesis

This thesis is organized in the following manner.

Chapter 2 gives a brief introduction to the basic principles of CT, involving the image reconstruction process and the technological evolution up to the most recent developments. This chapter also deals with radiation concepts such as the interaction with matter and radiation dose, which need to be considered in order to comprehend ensuing references.

Chapter 3 outlines the basic CT scanning parameters such as slice thickness, mAs, kVp, image matrix, reconstruction algorithms, exposure time and pitch. The influence of these parameters on image quality and dose is described and explained.

In Chapter 4 the basic characteristics such as contrast, noise and artefacts required for the description of image quality are defined and explain. The factors which influence these characteristics and their influence in turn, on the detection of structures are also described.

Chapter 5 is a review of the current status in paediatric CT examinations. The tactics followed in paediatric CT examinations and contribution to the dose excess are mentioned and discussed along with the protocols currently used. Moreover the clinical and experimental researches and simulation techniques undertaken the last years purposeful to examine potential dose reduction in relation to patient size are reviewed. The Outcomes and conclusions of these studies are present. The possibilities for dose reduction are emphasized, and new concepts such as automatic exposure control systems which intend to reduce dose are described.
Chapter 6 is concerned with the relationship between image quality, patient’s size and the scanning parameters. In order to fulfil the aims of this part, experiments and computer simulations were performed. The equipment used for the experiments, the software for the simulations and the methodologies followed are described. Computer simulations were validated with experimental measurements and the outcome is presented. Eventually the results with the discussion are laid out.

Chapter 7 involves the detectability of low contrast details. In particularly it is examined how scanning parameters influence the detection of low contrast details in relation to their contrast level and their size. The equipment used in order to accomplish the experiments is described. Image quality was subjected to objective and subjective assessment. The methodologies followed for both assessments are reported. Finally results are plotted in graphs and Low Contrast detectability histograms.

Finally, the last chapter gives a summary and conclusions of the current work together with proposals for future work and measures on how to reduce dose.

1.3 Publications

Chapter 2  Computed Tomography

Tomography (Greek tomos = section + graphien = to write) is defined as a method of producing an image of the internal structures of a solid object by recording the differences of the passage of X-rays through these internal structures. It was first introduced as a clinical tool in 1971 when Drs. Godfrey Hounsfield and James Ambrose successfully diagnosed a brain tumor in a 41 year old woman. The establishment of Computed Tomography (CT) in the commercial market for radiological imaging in the mid 1970s had profound effects on the subsequent development of other medical devices such as PET (Positron emission Tomography), MRI (Magnetic Resonance Imaging) and SPECT (Single Photon Emission Computed Tomography).

2.1 The Principles of Computed Tomography

Computed tomography is a diagnostic imaging procedure that uses a combination of X-rays and computer technology to produce detailed cross sectional images (slices) of the body. A CT scanner is an open “doughnut” shaped ring called a gantry through which the patient (who lies on a couch) passes. Inside the doughnut shaped gantry of a CT scanner, there is an X-ray tube which rotates in a continuous arc around the patient. Carefully aligned and placed directly opposite the X-ray tube are the electronic detectors which detect and measure the transmitted photons. The transmission data are related to the sum of all the attenuation coefficients in the X-ray beam path. The average transmission for each beam path is measured and stored in the computer. The patient’s cross section is divided into a matrix of voxels. By collecting data at many angles and paths, enough information is obtained to calculate the attenuation coefficient of each voxel. When data
are collected at one particular angle, Θ, the intensity of the transmitted beam $I_\Theta$ will be related to the incident intensity $I_o$ by

$$P_\Theta = \ln\left(\frac{I_\Theta}{I_o}\right)$$  \hspace{1cm} Eq. 2:1

Where $P_\Theta$ is the projection of all the attenuation coefficients along the line at angle Θ.

For example for $\Theta=90^0$, one of the values of $P_{90}$, say $P_{90}(2)$ to indicate that it is the projection through the second pixel in the x-direction, is given by

$$P_{90}(2) = \mu(2,1) + \mu(2,2) + ... + \mu(2,512)$$  \hspace{1cm} Eq. 2:2

The issue is to obtain sufficient values of P in order to solve the equation for the numerous values of $\mu(x,y)$. The importance of computer technology to this development now becomes apparent, since correlating and analysing all this information is beyond human brain. The attenuation coefficient for each voxel (which is the average of all the attenuation coefficients contained within the corresponding voxel) is calculated using computer instructions called algorithms and afterwards is converted to a CT-number (specified in Hounsfield Units) which is calculated by:
Thereafter the image is formed by assigning gray scale values to the CT-numbers. Each pixel has a gray scale value corresponding to its CT-number and the final image is formed by combining and displaying the individual gray scale values of the pixels. On this scale, water and consequently each water-equivalent tissue with \( \mu_{\text{tissue}} = \mu_{\text{water}} \), has the value 0HU by definition. Air corresponds to a CT-number of -1000HU and bone up to 2000HU. The Hounsfield scale has no upper limit. For medical scanners a range from -1024HU to +3071HU is typically provided with 4096 grey levels.

<table>
<thead>
<tr>
<th>Material</th>
<th>CT number in Hounsfield Units</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bone</td>
<td>+100 to 2000 HU</td>
</tr>
<tr>
<td>Liver</td>
<td>40 to 60</td>
</tr>
<tr>
<td>Muscle</td>
<td>10 to 40</td>
</tr>
<tr>
<td>Kidney</td>
<td>30</td>
</tr>
<tr>
<td>Water</td>
<td>0</td>
</tr>
<tr>
<td>Fat</td>
<td>-50</td>
</tr>
<tr>
<td>Air</td>
<td>-1000</td>
</tr>
</tbody>
</table>

Human observers can typically discern up to a maximum of 60 to 80 gray levels. Therefore, the complete gray scale is assigned to the CT-number interval of interest only, the so-called window; values above the chosen window will be displayed as white, and the values below the window as black. To chose the desired CT-number interval only the centre and the width of the window have to be adjusted. The centre is chosen corresponding approximately to the mean CT-number of the interesting structures, while the window width determines the contrast in the image. For the display of very small attenuation differences as given in the brain, for example, a narrow window is chosen. For large differences, as presented by the lung or the skeleton, for example, a wide window is chosen. (Kalender 2000)
2.2 Historical overview - CT scanner evolution

In 1967 Hounsfield put forward a research proposal to EMI called “An Improved Form of Radiography”. In this proposal he wrote:

The gamma rays which penetrate the object would be detected by an accurately aligned collimator and sensing device, which would always be pointing towards the source of the gamma rays. The readings from the detector taken ‘round the edge’ of the object would be digitised and fed to a computer for processing. If sufficient scans and angles of scan are made, there should be enough information from the ‘edges’ readings in the detector to produce sufficient equations to calculate by computer the value of transmissions of each cubic millimetre of material within the slice (i.e. there would be more equations than variables). A crude picture could therefore be built up in matrix form of the absorption of the material within the ‘slice’

The proposal was accepted and in 1972 the first CT scanner was constructed.
Since 1972 CT scanner has been developed relative to the scanning times, the combinations of motions, geometric design and the performance of the detectors.

2.2.1 First generation

In early 1972, a clinical prototype EMI head scanner was installed at Atkinson Morley’s Hospital in London. These first generation scanners consist of two detectors and a single X-ray source assembly to generate the narrow beam X-ray attenuation coefficient data used for each slice. The two detectors were employed to allow two slices to be obtained simultaneously. The x–ray beam which was collimated to a pencil beam and the detectors were first translated across the object. Afterwards tube and detectors were rotated 10°, and another translational scan took place. The pencil beam scanners were typically characterized by both rotate and translate motions with a total rotation angle ranging from 180° to 240°. Around the head of the patient water bags were placed to provide a constant tissue equivalent path length for the X-ray beam and to minimize beam hardening problems. The first generation scan suffered from long scan time, allowing the possibility of patient motion.

Figure 2:3. First generation CT scanner
2.2.2 Second generation

The second generation systems (so called hybrid machines) used multiple detectors and a 10° fan beam. The rotate and translate motions were preserved. Multiple detectors except providing more views also enable the system to rotate a larger angular increment (10° instead of 1°). After each rotation the translation movement was repeated. Image quality was markedly improved over the first generation scanner because of several factors: more views, finer ray sampling, a larger image matrix (i.e. 320 vs. 80), a smaller detector aperture and reduced scan time. In this and all subsequent CT scanners, the bulky water bag was omitted. The fastest scanning time was 18 second. In this period if the patient can suspend breathing, the images will not be degraded by motion blur.

![Second generation CT scanner](image)

2.2.3 Third generation

In third generation scanners X-ray tube and detector array pivot around the patient in a single rotational movement during which the views are acquired. The rays of a view are all acquired simultaneously and each active detector is associated with a ray. Depending on the manufacturer the X-ray tube is either pulsed or on continuously. The X-ray tube and detectors rotate through 360° continuously. Top of the line models typically have
approximately 750 detector elements and $360^0$ scan times range from 2 to 4 seconds significantly shorter than the 18 second scan time obtained at the second generation scanner. Siemens introduced a scanner with a 0.75 second scan time which is achieved by employing slip rings.

![Third generation CT scanner](image)

**Figure 2.5. Third generation CT scanner**

### 2.2.4 Fourth generation

The fourth generation geometry consist of detectors which are kept immovable on a stationary ring and an X-ray source that moves on an inscribed circle within the detector ring. In fourth generation scanners larger fan angles are employed which reduces X-ray tube loading. Also shorter scanning times are possible. For instance the Picker 1200SX scanner can achieve a scan time of 1.6 second for a $360^0$ scan and 1.9 sec for the more generally preferred $398^0$ over scan. Shorter scan times considerably reduce problems of patient motion and moreover heart scanning is developed.
Chapter 2. Computed Tomography

A significant development for third and fourth generation scanners was the development of helical or spiral scanners. This approach incorporated a moving table during the rotation of the X-ray tube and as a result, image data is acquired as a spiral or helix rather than in the form of a series of slices. The major advantage of spiral scanning is the volume of coverage for a given rotation of X-ray tube. The gain in coverage is described by the pitch which describes the number of slice thicknesses the table moves during one revolution. Typical pitches range up to 2 or more in single slice systems, and N times that in multislice scanners with N slices.

A single slice CT scanner, produce a distinct slice, where the slice sensitivity may be well defined and reasonably constant across the CT slice. In spiral scanning, the slice is not so simply defined by the X-ray collimation. The nature of the moving table requires interpolation schemes to provide estimates of information within a given slice. This information is taken in an acquisition which includes information from the slice above and below the slice of interest and then interpolates the data to establish an effective slice at a given position.

Multislice CT permits multiple (up to sixty-four) slices to be acquired in a single rotation. These scanners may significantly reduce the scanning time for acquiring volume data and for improving longitudinal spatial or z-axis resolution. However a broader area is scanned at one time, which brings about more scatter radiation per slice and the consequent image
quality reduction and dose increase. One of the clinical advantages of multislice CT is the possible CT angiography. Slice thicknesses may range from 0.5mm to 10mm. However slice thicknesses can be determined not only by the slice width and the pitch but also the shape and the width of the interpolating filter in the longitudinal (z) direction.

It should be noted that CT scanners traditionally produce axial planes. Other planes of interest could only be generated by first acquiring a series of axial slices, and then reformatting the acquired data into a set of other planes. This process was generally called multi planar reconstruction, or MPR.

Figure 2:7. (a) single slice-Spiral CT, (b) multi-slice CT: a multi-slice CT scanner can acquire up to 64 slices simultaneously.

2.3 Radiation

Radiation can be categorised as non-ionizing or ionizing radiation depending on the effect of interaction between radiation and matter. Ionising radiation is radiation with enough energy to remove tightly bound electrons from the orbits, causing the atom to become charged or ionized. There are two sources of ionizing radiation, natural and man-made radiation sources. Man-made radiation accounts for 15% of our average annual radiation. By far, the most significant source of man made radiation exposure is medical procedures, such as diagnostic X-rays
2.3.1 Interaction of X-ray Radiation with Matter

An X-ray beam proceeds through matter interact with matter via a number of mechanisms, which include: the Photoelectric effect, the Compton Scattering, Coherent (Rayleigh) scattering and Pair Production. Depending on the energy of the X-ray beam different mechanisms will dominate. In the energy range considered for computed tomography imaging, photoelectric and Compton effects are the dominating mechanisms.

The photoelectric effect is the complete absorption of an incoming photon upon interaction with an atomic electron. All the energy of the photon is transferred to the electron, which is emitted from the atom as photoelectron with energy \( E_e \) given by

\[
E_e = E_\gamma - E_b
\]

where \( E_\gamma \) is the energy of the photon and \( E_b \) is the atomic binding energy of the electron.

In a Compton event, the incoming photon with energy \( E_\gamma \) is scattered by a free electron through angle \( \Theta \) and part of its energy \( E_e \) is imparted to the electron which recoils, and the remaining \( E_\gamma \) stays with the scattered photon.

As a result of the interactions described above the photons are either Compton scattered with some loss of energy or completely absorbed in a photoelectric interaction. Absorbed energy is derived from the photoelectron and from the recoil electron in the Compton event.

Therefore when an X-ray beam passes through matter, it becomes attenuated as photons are progressively removed from it. Consider a narrow beam of monochromatic photons which travelling in a homogeneous medium. This beam is attenuated according to Beer’s law:

\[
\Phi(x) = \Phi_0 e^{-\mu x}
\]
Where $\Phi_0$ is the incident photon fluence (photons per square centimetre), $\Phi(x)$ is the incident photon fluence after travelling distance $x$ in the medium and $\mu$ is the linear attenuation coefficient. The linear attenuation coefficient provides an indication of how effective a given material is, per unit thickness, in promoting photon interactions. The larger the value of the attenuation coefficient the more likely it is that photons of a given energy will interact in a given thickness of material.

The value of the linear attenuation coefficient ($\mu$) varies with material and, with photon energy. While specific values of the attenuation coefficient will vary among materials for photons of a specified energy, the generalized shapes of plots (neglecting fine details) of attenuation coefficient versus photon energy are similar among different materials. In general, such shapes show high values of the attenuation coefficient at low-photon energies that decrease as photon energy increases, go through a rather broad minimum value, and then increase as energy continues to increase (Figure 2:2).

![Figure 2:8. Mass attenuation coefficients of tissues](image)

The mass attenuation coefficient ($\mu/\rho$) is defined by:

Mass attenuation coefficient = linear attenuation coefficient / density = ($\mu/\rho$).
Chapter 2. Computed Tomography

It is useful for calculating the mass of material required to attenuate a primary beam by a prescribed amount, i.e.

\[ \Phi(x) / \Phi_0 = e^{(\mu/\rho)x_m} \]

where \( x_m \) is the mass of attenuator per unit area of beam. The quantity \( x_m \) is simply \( \rho \cdot x \). It has typical dimensions of \( g/cm^2 \) while \( \mu/\rho \) has dimensions of \( cm^2/g \).

The biological organs consist of complicated molecules. Thus, the mass attenuation coefficient of a compound or mixture of elements is of main importance and is given by (Knoll 1989):

\[ \mu/\rho = \sum w_i (\mu/\rho)_i \]

Where \( w_i \) represents the relative weight fraction of the i-th element in the mixture.

Table 2:2. Linear attenuation coefficients \( (\mu) \) for various materials at different energies

<table>
<thead>
<tr>
<th>Energy Absorber</th>
<th>Linear attenuation coefficient (cm(^{-1}))</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>80 keV</td>
</tr>
<tr>
<td>Water</td>
<td>0.184</td>
</tr>
<tr>
<td>Soft tissue</td>
<td>0.181</td>
</tr>
<tr>
<td>Bone (cortical)</td>
<td>0.428</td>
</tr>
<tr>
<td>Liver</td>
<td>0.193</td>
</tr>
<tr>
<td>PMMA</td>
<td>0.208</td>
</tr>
<tr>
<td>Air</td>
<td>2.03E-4</td>
</tr>
</tbody>
</table>

2.3.2 Radiation Dose

In general, when radiation interacts with matter part of its energy is absorbed by the matter and the remainder is scattered and was eventually escape the body. In an attempt to determine the physical effects due to the absorption of energy, the absorbed dose (or simply the dose) \( D \) is defined. Absorbed dose is the amount of energy deposited into a region of the body divided by the irradiated mass that absorbed radiation’s energy.
Chapter 2. Computed Tomography

\[
\text{Absorbed Dose}(D) = \frac{\text{Energy Deposited}}{\text{mass of absorbing material}} \quad \text{Eq. 2:8}
\]

It is equal to the energy deposited per unit mass of medium, and so has the unit \(\frac{J}{Kg}\), which is given the name gray (Gy).

However absorbed dose is not a good indicator of the likely biological effect. The same \textit{absorbed dose} delivered by different types of radiation may result in different degrees of biological damage to body tissues. Therefore the biological effect of radiation vary for different types and energies of radiation. As a result the \textit{equivalent dose} was introduced to take into account the dependence of the harmful biological effects on energy and the type of radiation being absorbed. The \textit{equivalent dose} is therefore a measure of the risk associated with an exposure to ionizing radiation. \textit{Equivalent dose} \((H)\) is calculated by multiplying the \textit{absorbed dose} averaged over organ \((D)\) with the evaluation factor. Formerly, \textit{The International Radiological Protection Commission} recommended the use as the evaluation factor of the quality factor \(Q\) only. The value of \(Q\) is 1 for X-rays, gamma rays, and beta particles, but higher for protons, neutrons and alpha particles.

In 1990, the organization issued new recommendations. The \textit{radiation weighting factor} \((wR)\) should be used as the evaluation factor. It is calculated by multiplying the quality factor \(Q\) and the modified factor \(N\). The modified factor has value of 1 when used for external sources of irradiation, but is defined arbitrarily by a competent authority when used for internal ones. The \textit{equivalent dose} is thus calculated by the equation:

\[
H_{T,R} = wR.D_{T,R} \quad \text{Eq. 2:9}
\]

the unit of \textit{equivalent dose} is the \textit{sievert} (Sv). In a mixed radiation field the following equation applies for the overall organ \textit{equivalent dose} \(H_T\) in a tissue or organ \(T\):

\[
H_T = \sum wR.D_{T,R} \quad \text{Eq. 2:10}
\]
Nonetheless the organs of the body have different radiation sensitivity. The contribution of the individual organs to the total risk when the body is exposed to homogenous irradiation can be described by the so-called tissue weight factors. Hence, in estimating the effect on the human body as a whole, the effective dose ($ED$) was defined. The effective dose is equal to the sum of the equivalent doses to the individual organs multiplied by their respective tissue weight factors.

$$ED = \sum_i H_i g_i$$  \hspace{1cm} \text{Eq. 2:11}$$

The gray and the sievert are relatively large units and for radiation protection purposes submultiples are generally used, e.g., mGy, mSv, or µGy, µSv.

### 2.3.3 Risks of Ionizing Radiation Exposure

Despite the benefits from the utilization of radiation in medical matters, such as radio diagnosis and radiotherapy, its utilization also constitutes severe health risks. It is now known that the biological effects of radiation are caused mainly by ionization of water molecules within cells. This produces highly reactive free radicals, which in turn damage macromolecules such as DNA. The effects on DNA include single and double strand breaks, cross linkages and other changes such as base damage, which will result in cell death or the induction of mutations. The risk of harm from radiation is largely dependent upon factors such as the magnitude of the dose, the rate at which dose was delivered, the type of radiation, the part of the body exposed (the most rapidly dividing cells such as the gut and bone marrow being most sensitive) and the age and health of the exposed individual.

A radiation risk is a concern in both adults and children. However when radiation risks refer to children, two major considerations must be taken into account. The first one is that children are considerably more sensitive to radiation, as demonstrated in epidemiological studies of exposed populations than adults, with the highest risk at the youngest age. This sensitivity is due to having more rapidly dividing cells than adults. At age 1 year, the lifetime attributable risk of mortality from cancer is 14% per Gray. This
risk falls to 10% at age 10 years, 8% at 20 years and 4% at 35 years. The second one is that children have a longer life expectancy, resulting in a larger window of opportunity for expressing radiation damage (National Cancer Institute). A research which performed by Brenner et al estimated that lifetime cancer mortality risks attributable to the radiation exposure from a CT in a 1 year old child, are 0.18 % concerning abdominal CT and 0.07 % concerning head CT – an order of magnitude higher than for adults (Brenner et al. 2001).

Risks associated with radiation exposure can be considered with regard to two main categories: namely, deterministic effects or stochastic effects. Deterministic risk results from cell death and is quantified in terms of radiation doses to a particular region that has a threshold level beyond which these effects occur. Deterministic risks are rarely seen with diagnostic X-ray based examinations, including CT, because radiation doses typically do not reach the threshold level, even for people who have multiple examinations. The main risk to the subject undergoing CT is due to the stochastic effect, which may result in cancer, and genetic effects, which occur in the offspring of the irradiated subject. There is a linear no-threshold relationship between radiation exposure and biological effect and the probability of occurrence of stochastic effects depends on the amount of absorbed dose.

It is common belief that there is no amount of radiation exposure that is considered absent of risk. Recent data from the atomic bomb survivors and medically irradiated populations demonstrate small, but significant, increase in cancer risk even at the low levels of radiation that are relevant to paediatric CT scans (National Cancer Institute).

Radiation effects are classified into acute and chronic. If the dose is acute (large dose during short time) serious damage may occur. On the contrary, if the dose is delivered over an extended period (chronic), the body can repair most of the small damages and the damages are not severe.

Two kinds of radiation effects can occur: somatic effects (damage to any of the cells of your body) and genetic effects (Genetic or heritable effects appear in the future
generations of the exposed person as a result of radiation damage to the reproductive cells)

There are a lot of somatic effects such as Cataracts, Skin Erythema (Skin dose> 1 Sv), Hematopoietic Syndrome (>1 Sv), Gastrointestinal Syndrome (>10 Sv), Central Nervous System Syndrome (>50 Sv), Permanent Sterility (gonads dose >5 Sv), Thyroid Bening Tumors (Thyroid dose > 0.5 Sv) but the most important long –term effect is cancer induction.
Chapter 3  Basic Scanning Parameters.

3.1 Slice thickness

Choosing the appropriate slice thickness for each type of examination it is of vital importance since it influences the sensitivity of lesion detection. With narrow slices, smaller lesions can be detected more easily. Furthermore if slice thickness is not suitable the lesion could get lost in the slice due to volume averaging effects. In addition to that if thin slices are obtained multi-planar reformatted images (MPR) of high quality may be constructed. In such cases, the reformatted images may obviate the need to carry out additional imaging in other planes, resulting in an overall dose reduction.

There are, however, several trades off in using thinner collimation. As slice thickness is decreased, there is a concomitant decrease in signal and an increase in image noise (Quantum noise). Quantum noise is the noise in an image that is related to the number of photons which are used to generate that image. It is a statistical phenomenon in which the more photons are used the lower the image noise. Consequently since slice thickness defines one of the three volumetric dimensions of a voxel, decreasing slice thickness, the volume of the voxel decreases. Therefore the number of photons in a voxel decreases, so the quantum noise rises. To the radiologists this appears as reduced contrast resolution and takes it greatest toll on the ability to distinguish tissues with relatively subtle density differences.

There is also another trade off to thinner slice collimation. In conventional CT, using a thin collimation may lead to long scan times due to the greater number of slices required to cover the anatomy. Also more time is needed for the reconstruction of these slices.
Chapter 3. Basic Scanning Parameters

Take, for example, the case of a chest CT study, where one may want to cover an area from the thoracic inlet to the diaphragm measuring 300mm. To scan this area using a slice thickness of 10mm would lead to a series of 30 slices. If, for example, it requires five seconds to acquire an image and then move the patient into the new position for the next slice, then it will take 150 seconds to complete the scanning. If, however, one elects to scan the patient using thinner slice thickness of 2.5mm in search of very small lung modules, then the study will be comprised of 120 images and it will require 600 seconds to complete the scanning. The extra time required may not only be burdensome, but also gives plenty of opportunity for patient motion to degrade the image. However this disadvantage has been overcome in modern scanners.

There is one final point of much interest that should be made, and this relates to dose. Each single scan produces a dose profile; when a series of contiguous slices is obtained, adjacent dose profiles derive. The dose from a series of contiguous slices is the sum of the individual dose profiles; this dose is greater than the dose of a single slice because of profile’s overlap. How much greater is the total dose depends on the volume scanned. Also, if the slices are not contiguous but they overlapped each other then the dose is even more elevated.

![Figure 3.1. Individual and total skin dose profiles for three contiguous slices of 10mm. The total dose profile is simply the sum of the doses from the individual slices.](image-url)
Thin slices, of 1mm to 4 mm are needed when the examination concerns complex anatomic regions, such as the larynx or very small anatomic structures, like distal pulmonary airways in a search for small adrenal masses. Moreover these slice widths are kept for detailed examinations or repeating examinations which are performed in order to verify suspected small lesions. For most examinations, however, 5mm to 10mm slice thickness is used.

### 3.2 Tube Current-Time Product (mAs)

The quantity of X-rays is directly proportional to the product of the tube current and exposure time (mAs). The mA controls the x ray intensity (photons per unit time) and the mAs parameter represents the number of photons (quantity) in the defined exposure time. High mAs is beneficial when imaging regions where soft tissue contrast is crucial (e.g. distinguishing one soft tissue structure from another). In evaluating high contrast regions (e.g. facial bones, spine), mAs is a much less important factor. As mentioned mAs directly relates to the number of photons emitted in an x ray beam, and therefore inversely to quantum noise. Consequently increasing mAs, quantum noise is decreased with concomitant increase in contrast resolution. Finally, mAs has a linear relationship with the absorbed dose. Halving the mAs means halving the radiation dose and doubling the mAs doubles the patient dose. In CT the mAs is the single most important factor for managing patient dose.

### 3.3 Tube Voltage (kVp)

The kVp is the maximum voltage applied across an X-ray tube in order to accelerate the electrons and produce the X-ray beam. The kVp is a reflection of the quality of the x ray beam which stands for the penetrability of the beam. The higher the kVp applied the higher the energy of the X-ray beam and the greater penetration through the patient so that a greater number of photons reach the detectors. However, the fact that more photons
reach the detectors has advantages and disadvantages. The benefit is that the quantum noise is lessens and the exchange is the lower contrast. This is a standard rule in radiology since the more energetic the beam, the less effect different levels of tissue density will have in attenuating that beam. Moreover the value of the kVp influences the absorbed dose. The higher the kVp the greater the absorbed dose.

### 3.4 Rotation time

Rotation time is the time it takes for the X-ray beam to complete one $360^\circ$ rotation. For conventional CT this can vary up to four seconds. Until recently, this had been one second for all spiral scanners. However recently, sub-second scanning has become possible. The benefit of a longer rotation time is the higher mAs it can deliver, leading to increased contrast resolution. However, longer rotation times may also allow greater patient motion, hereby leading to net loss in image quality and also higher dose since a greater amount of photons are delivered.

### 3.5 Pitch

Pitch is a new component introduced with helical CT. It is defined as the ratio of the patient’s movement through the gantry during one $360^\circ$ beam rotation relative to the beam collimation.

\[
Pitch = \frac{\text{Patient (table) advancement per } 360^\circ \text{ gantry rotation}}{\text{beam collimation}} \quad \text{Eq. 3:1}
\]

For instance, if the beam collimation is 5mm and the patient is advanced into the gantry 5mm during every $360^\circ$ rotation, then the pitch is 1.0.
Increasing the pitch leads to an overall decrease in administered dose since the scan time will be reduced as a result of the faster patient advancement through the gantry and less patient irradiation. However, increasing pitch leads to decreased spatial resolution. Modern CT protocols use thinner collimation with a higher pitch for an examination that has a higher resolution in a shorter scan time as compared with conventional CT.

### 3.6 Image Matrix

The image matrix relates to the number of pixels that form the image grid. When the field of view is kept constant, increasing the matrix will lead to smaller individual pixels and therefore increased image detail. However, spatial resolution is actually 3D, thus although the matrix defines the resolution in the cross sectional plane, the image detail is a function of the total voxel volume and, therefore closely related to the slice thickness.

One trade off of increasing the image reconstruction matrix relates to an increase in the number of data pieces that the computer has to handle and hence an increase in reconstruction time.

### 3.7 Reconstruction Filter (Algorithm)

During the course of a $360^\circ$ rotation, a huge amount of data is generated. These data are then passed through a mathematical filter algorithm as part of the image reconstruction process. A CT unit offers a lot of algorithms each optimized for different body parts and tissue types. A soft tissue algorithm lead to smooth images, optimized to enhance soft tissue contrast, while a bone algorithm will sharpen bone margins at the expense of creating soft tissue windows limited by noise and poor soft tissue contrast.
Each filter needs the same amount of time to process the data. Moreover, images can be processed sequentially in multiple different filters as long as the raw scan data is saved, but this adds time to the overall study time and affects patient throughput.
Chapter 4  Image Quality

Image quality is a concept of central importance for the evaluation of every imaging system. In principle, the central issue is always how accurately a system reproduces the object imaged. The human body contains many structures that are imaged; noise and artefact components, which can prevent the recognition of individual structures, will be superimposed on this image. If the structures are distinguishable from their immediate background and there is no loss of valuable medical information then the image is indicated of high Image Quality. High image quality is with good reason always demanded, but frequently evaluated only subjectively or in a general sense. Image quality is not a single factor but is composite of at least four factors: contrast, noise, spatial resolution and artefacts.

4.1 Contrast

Contrast is fundamental characteristic of the image. It is the visible difference between neighbouring structures on the image. As mentioned a beam of monoenergetic x rays traversing matter is attenuated exponentially according to the equation 2:5. The attenuation is affected by the composition of the material as represented by the linear attenuation coefficient, as well as the distance x through which the beam traverses. In general the contrast is given by the difference between signal and background attenuation coefficients. In CT procedures the attenuation coefficients are calculated using computer instructions called algorithms and converted to a CT number measured in Hounsfield Units given by:
Therefore in CT, image contrast is the CT number differences between two materials and is taken to be proportional to the Hounsfield Unit (HU).

Addition to the type of the material, contrast value depends on the X-ray beam’s energy. Increasing the energy of the x ray beam generally results in a reduction of image contrast. However, high and low atomic number materials are affected differently by the changes in x ray beam energies. Changes in beam energy are more important for high atomic number materials (e.g., iodine, Z_{effective} = 47) than for lower atomic number elements (e.g., soft tissue, Z_{effective} = 7.6). High atomic number materials have higher $\mu$ and therefore they have higher contrast. This is the reason that contrast agents (i.e. iodine solutions) are used during computed tomography examinations since they enhance the difference in density of various structures. The gastrointestinal track (GI) can be illuminated by giving the patient diluted water-soluble oral contrast material or rectally instilled colonic contrast material, which will help to distinguish the stomach and bowel from other soft-tissue structures and masses. Intravenous administration of water-soluble contrast material will produce a temporary increase in the density of vascular structures and highly vascularised organs. This effect is referred to as enhancement and is extremely useful. For example, a blood vessel and the tumour mass encasing and constricting it will appear as one homogeneously dense mass unless the vessel is enhanced with contrast material, when its narrowing will be apparent.

It is worth noticing that the imaging advantage of CT is more directly influenced by contrast than by spatial resolution\(^1\). That is, the contrast of the object often is more important than size in determining whether or not a specific lesion will be distinguishable. “Contrast Sensitivity” has to do with the ability to define subtle alterations in attenuation values reflecting small differences in the transmission of x-rays through the imaged object.

\(^1\) Spatial Resolution : has to do with the ability to define structures in an image that are only slightly separated in the imaged object.
The locations of small objects relative to the positions of the pixels in which they are seen may affect visualization. If a tiny object falls on the border between two pixels, its density will be averaged between the adjacent pixels, and it may go undetected, whereas if it falls squarely within a pixel, it is more likely to be seen.

### 4.2 Noise

Although the imaging system may have high contrast the radiologist will fail to identify even a large object if the noise level in the image is very high. There are two major contributions to the noise: statistical fluctuations in the number of X-ray photons detected per unit area (quantum or statistical noise) and fluctuations due to the properties of the image receptor and display system (processing noise). Today’s CT scanning systems come very close to efface the noise originating in the system itself therefore statistical noise dominates. In a CT image statistical noise is the fluctuations of the CT number that is, the standard deviation of the CT number.

In modern CT scanners the noise originating in the system itself are due to:

- Reconstruction algorithm or filter and

- Helical Interpolation Algorithm

Statistical noise is influenced by a large number of parameters, including:

- Tube voltage (kVp)

- Tube current-time product (mAs)

- Collimation/Reconstructed Slice Thickness

- Helical Pitch/Table speed
The number of X-ray photons detected is subjected to Poisson statistical fluctuations therefore statistical noise should be governed by an equation such as

\[ \sigma = a(mAs)^{1/2} \]

Eq. 4:2

where \( a \) is a constant. From the equation it is obvious that statistical noise reduces by increasing the number of photons used to form the image that is, increasing mAs. As the mAs is increased, more X-rays are detected, noise is reduced and lower contrast structures can be seen. Figure 2:2 shows two water phantom images scanned at different mAs. The phantom on the right hand is exposed at lower mAs and its image is noisier.

In cases where low contrast structures (typically from 4-10 HU difference) are imaged, the signal is so small that noise is a significant factor. Figure 4:2 illustrates just one aspect of the low contrast resolution test and how noise – as created by different mAs levels – can influence the results of this test (Note that the reader’s printer or computer monitor may also significantly affect the appearance of these images). Under good viewing conditions, the 4 mm diameter rods can be seen on the image on the left; while even the 6 mm diameter rods are questionable on the right hand image.
Chapter 4. Image Quality

4.3 Artifacts

Image quality in addition to being limited by the statistical noise, it may be degraded even further by artifacts introduced through malfunction of the detectors or problems in the reconstruction artifacts. Some of these will be summarized briefly.

4.3.1 Streak Artifacts

Streak artifacts can be caused either by misalignment of the mechanical scanning equipment or by patient motion during the scan. The equipment misalignment is most easily determined by scanning a steel pin or rod in the water phantom. The appearance of such streaks in the calibration image is an indication of mis-registration.

Patient motion is a common problem which results in inconsistencies in the view data. These streak artifacts are produced from high contrast interfaces such as bone interfaces moved during the scanning. The obvious way to reduce motion artifacts is to employ a short scanning time.
Chapter 4. Image Quality

4.3.2 Partial Volume Artefact

Because the X-ray beam diverges in a direction perpendicular to the slice, a projection measured in one direction may be slightly different from the projection taken along the same path but in the opposite direction. This provides one reason for requiring a full 360° scan of the patient. The inconsistencies in the data can be compensated by combining data from opposite directions.

A different but related partial volume effect arises from the observation that the anatomical structures do not in general intersect the section at right angles. A long, thin voxel could well have one end in soft tissue and the other end in bone. As a result the reconstructed $\mu$ would have an intermediate value that did not correspond to any real tissue at all.

4.3.3 Ring Artifacts

Ring artifacts are caused by faulty detector calibration or a drift in gain in third generation units. As a worst case example, consider a faulty detector with no response. Any miscalibration of the third generation detectors will result in ring artifacts.

4.3.4 Beam Hardening Artefact

CT algorithms assume a monoenergetic photon beam when they calculate the attenuation coefficients and CT numbers. X-ray tubes, however, produce heterogeneous beams containing all photon energies up to the maximum kVp. When such a beam passes through a patient, the low energy photons are selectively absorbed and the average energy of the beam increases with penetration into the patient. Patients are circular or elliptical in shape so X-ray beams passing near the edge of the patient pass through much less material than those passing through the centre. The centre beams become harder and the CT numbers calculated near the centre of the patient are too low. Additional filtration added to the X-ray beam can remove many of the low energy photons before they reach the patient. Filtrations in CT scanners are at least 4mm Al.
4.3.5 Aliasing

Aliasing occurs when the signal contains higher frequencies than the CT sampling frequency. Sharp, high contrast boundary edges such as edges of bone, metal clips or bowel gas contain very high spatial frequencies. If these spatial frequencies are higher than the sampling frequency, the aliasing artefact will be displayed as streaks from the interface edges producing a star-like pattern. These can be reduced by proper choice of the filter function used in the reconstruction algorithm.
Chapter 5  Review of Current Status in Paediatric Computed Tomography.

The introduction of computed tomography in the 1970s led to a revolution in imaging since it was able to provide more detailed anatomical images. As the percentage of CT examinations is increasing in the general population, the percentage of CT examinations is rapidly increasing in children as well. In 1989, approximately 4% of CT examinations were performed in paediatric patients; in 1993 this percentage increased to 6% (Brenner 2002). Currently, about 10% of all CT examinations are performed in paediatric patients, and they deliver about 67% of the overall collective radiation dose to this population (Mettler et al. 2000).

5.1 Doses During Paediatric CT

Computed tomography is generally associated with high delivered dose compared to other diagnostic techniques. The United Nations Scientific Committee on the effects of Atomic Radiation 2000 report on medical radiation exposure stated that, world-wide, CT constitutes 5% of radiological examinations and contributes 34% of the collective dose (2000). In 2002 the NRPB estimated that approximately 41.5 million medical X-ray examinations are conducted in the United Kingdom every year and that number is expected to increase (NRPB 2002). All these facts make clearly evident the high amount of dose is delivered during CT examinations and that will concern a bigger part of the population every year. Figure 2:2 shows the frequency of different types of X-ray examinations and the contribution of each one to the United Kingdom collective dose (Chapple et al. 2005).
Chapter 5. Review of Current Status in Paediatric Computed Tomography

Figure 5:1. Distribution of X-ray examinations in the UK and contribution of examination to collective dose

Absorbed dose in tissues from CT are among the highest observed from diagnostic radiology (i.e. 10-100 mGy). Consequently effective doses in CT are relatively high, typically 1-30 mSv (Rehani et al. 2000). To have a notion about the magnitudes of effective dose during a CT examination the results of typical chest and abdominal examination as Huda claims are quoted (Huda 2002; Walter 2002). The typical chest examination has an effective dose of about 5mSv which increases substantially as the size of the patient reduces. For newborns there is an increase by a factor of two to three in comparison to adult patients being scanned. Concerning abdominal examination the effective dose of an adult is about 4mSv and as patient’s mass decreases the effective dose may be increased up to a factor of two.

5.2 Current Paediatric CT Practice

Especially for children the high amount of dose delivered during the examination is associated with severe risks. It is a question whether it is possible to exploit all the benefits of this imaging tool and at the same time reduce the risks? Up to now, the outcomes from the undertaken researches show that the possibility exists. But firstly, let’s examine the reasons of this high amount of dose.
Undoubtedly the exposure settings utilized during that examination are higher than those in other X-ray techniques. But fortunately this is not the only cause of that high level dose. Despite the fact that there is increasing awareness among paediatric radiologists of the potential risks associated with ionizing radiation in medical imaging, there is still widespread underestimation of relative doses and risks by paediatricians (Thomas et al. 2006).

On several occasions, radiologists employ tactics which are not the proper ones resulting in unnecessary dose. For instance in many hospitals radiographers are not provided with paediatric protocols and they scan children with the same scanning parameters which are used for adults. Moreover, some times the scanning length is much bigger than the length necessitated for the examination and very often unnecessary repetitive examinations are performed. Besides, doctors decide on computed tomography while alternative examinations like MRI or ultrasound could be performed.

Often, CT scans are done before, during, and after injection of intravenous (IV) contrast material. When medically appropriate, multiple exposures may be reduced by eliminating pre-contrast images (i.e., un-enhanced images). Moreover the use of contrast material is also a risk and before an examination with contrast agent, must be assured that would not cause other side effects.

In addition, collimation is often not adjusted for examinations in children; many children are imaged with a collimation of greater than 5mm a value which is recommended for CT examinations performed for adults. This affects the spatial resolution and contributes to partial volume effect. On the other hand choosing collimation that is unnecessarily narrow will increase the radiation dose. Therefore if relative adjustments are made the dose would be lower without any effect on image quality.

All these considerations mentioned above, contribute to the overall absorbed dose and unfortunately they could be avoided. These considerations can and must be limited. The most effective way is to develop appropriate protocols for both, the type of the examination and the size of the individual.
5.3 Current Protocols

Until very recently the protocols were based on patient’s age. For instance, Siemens’ protocol was separated in four age groups: <1 years old, 1–5 years old, 5–10 years old and 10+ years old. However the categorization of patients into age group is not believed to be the appropriate. Recently, in an endeavour to reduce dose the companies attempted to base the protocols on patients’ weight. Undoubtedly this effort reduces the dose however much more can be done. Most of these protocols are separated in 6 groups. These groups are: under 15kg, 15–24kg, 25–34kg, 35–44kg, 45–54kg and above 55kg. In these protocols the exposure settings are adjusted to each individual and for each type of examination. In appendix 2 part of the protocols used by Siemens scanners are shown. We have to mention that, concerning head examinations all the protocols are based on age.

5.4 Literature Review

Dose reduction in CT has been an important issue since the early days of CT. The last few years many studies, purposeful to adjust exposure settings to patient size, have undertaken. These studies are based either on Computer simulations, experiments or clinical procedures. The outcomes from the whole of the studies show perspective dose reduction during CT examinations to be feasible.

5.4.1 Experimental Procedures

Numerous experimental studies have undertaken the last years in order to examine the effect of the patients’ size on image quality. In order to accomplish the experimental researches a variety of phantoms were used; Anthropomorphic, oval shape phantoms, and circular phantoms with various diameters. Diameters of circular phantoms were ranging between 6cm and 32cm (Rogalla et al. 1999; Donnelly et al. 2001; Morgan 2002; Boone et al. 2003; Siegel et al. 2004; Verdun 2004; Chapple et al. 2005) in order to cover the
entire range of population. These phantoms were exposed to several combinations of exposure settings. The exposure settings of which there effect on image quality and dose was examined, where tube voltage (kVp), tube current-time product (mAs), slice thicknesses and pitch. In particular the effect of kVp and mAs was extensively examined.

Most of the experiments were performed with the four tube voltages 80, 100, 120, and 140kVp which are available in most of CT scanners, and a wide range of mAs ranging from 10 to 800 mAs.

Some of the studies intended to examine whether it was possible to have the same image quality (CNR) as with the standard technique, by using lower exposure settings. Other studies aimed at achieving a compromise between satisfactory image quality and reduced dose. Dose calculation was accomplished using several means, such as TLD dosimeters (Rogalla et al. 1999; Chapple et al. 2005), Ionisation chambers, Monte Carlo packages and the Computed Tomography Dose Index (CTDI) (Morgan 2002; Boone et al. 2003; Siegel et al. 2004; Verdun 2004). Image quality assessment performed with the measurements of contrast, noise and Contrast-to-Noise Ratio (CNR). Contrast was measured as the difference between the CT numbers of two materials in Hounsfield Units (HU). Noise was the standard deviation of the CT number of the background. Measurements were taken by the placement of multiple ROIs across the image and eventually a mean value was calculated.

Furthermore some experimental studies examined how pitch or slice thickness influence image quality and dose (Vade et al. 1996; Donnelly et al. 2001; Frush 2002; Boone et al. 2003).

**5.4.2 Computer Simulation Techniques**

Concerning computer simulations, software which generates X-ray spectra was necessitated (Huda et al. 2000; Huda et al. 2004; Huda et al. 2004). These spectra were attenuated with either water or Polymethyl-methacrylate (PMMA) phantoms of various thicknesses. The number of X-ray photons at each photon energy $E$, $N(E)$, transmitted
along a path length $X$ was determined from the number of incident photons using the following equation

$$N(E) = N_0(E)e^{-\mu X} \quad \text{Eq. 5:1}$$

Afterwards the number of transmitted photons $N(E)$ was used either directly or indirectly to calculate contrast or noise.

Image contrast depends on photon energy and the material being imaged. For each material the CT number was calculated from eq. 5:2 for the corresponding energies.

$$\text{CT number (HU)} = 1000 \frac{(\mu_{\text{material},E} - \mu_{\text{Water},E})}{\mu_{\text{Water},E}} \quad \text{Eq. 5:2}$$

Eventually the contrast between two materials was the difference between their CT numbers in Hounsfield Units.

Statistical noise is the dominant contributor to image noise in CT. Researchers assumed that the relative amount of quantum mottled in a CT image was inversely proportional to the square root of the radiation exposure level incident on the CT X-ray detector. The relative noise level ($N$) was obtained by combining data on the X-ray tube output ($O$) and the percentage energy fluence $F$ transmitted through the patient $N \propto (O \cdot F)^{-0.5}$ (Huda et al. 2004).

Usually computer simulations were validated with experimental results.

### 5.4.3 Clinical Studies

The number of clinical studies is restricted. The reasons are not scientific but moral. You can not put human’s health at risk and especially children’s health in the name of public good, especially when the possibility harming their health is increased.
One clinical study involved patients aged 65–83 years old with cancer (Prasad et al. 2002); whereas another study involved children aged 0-13 years old (Shah et al. 2005). Some of the studies scan the patients at conventional exposure settings and then they re-scan them with lower exposure settings; other studies were accomplished by scanning the patients at reduced exposure settings and then the images obtained were compared with similar investigations at conventional settings. Most of the clinical studies were accomplished by following the second method because that way, they avoid the overexposure of the patient. Specifically this method is performed by scanning children or adults at lower exposure settings and then compare the results (noise value, contrast value or visibility of structures) either with examinations performed on patients with standard techniques or with the results derived from experimental studies using phantoms (Chan et al. 1999; Rogalla et al. 1999; Donnelly et al. 2001; Greess et al. 2002; Suess et al. 2002).

Clinical studies evaluate the effect of mAs and tube voltage on image quality. Apart from tube current and voltage another scanning parameter that has been examined by researchers, and it is of great importance in clinical studies, is pitch. Vade et al evaluated image quality using 1:1 pitch and 1.5:1 pitch and they also examined the impact on dose.

Almost in all clinical studies image quality assessment was carried out by expert radiologists who were unaware of the CT techniques, and they scored for structural resolution and diagnostic confidence. Dose calculations were based upon the displayed weighted CTDI (CTDIₜₐ) or measuring TLD dosimeters.

The aim of clinical studies was to examine whether it was possible to achieve acceptable image quality by changing exposure settings and additionally whether it was possible to adjust these exposure settings to patient’s characteristics (age, weight). Therefore patients had been categorized relative to their age or weight. Each category had a range of values in order to cover the entire spectrum of patients. Afterwards, for each group they had chosen the appropriate exposure settings in order to avoid overexposure or underexposure conditions which would affect both patient and image quality. We have to underline that none of these clinical studies categorized the patients relative to their size.
5.4.4 Conclusions

It is well known that smaller patients in diameter or age attenuate X-ray beams less than larger in diameter or age patients and absorb approximately 40% less energy. However, since children are considerably more sensitive to radiation energy their corresponding value of effective dose is nearly four times higher. This is the reason that adjusting exposure settings to patient’s size became the ultimate goal for many studies.

Experimental studies and computer simulation studies, as we mentioned above, use PMMA phantoms or water phantoms in order to simulate patients. Water phantoms simulate adequate the human body (Boone et al. 2003; Huda et al. 2004) in contrast to PMMA phantoms which do not simulate adequate the human body. Since the intention is to find a relationship between image quality and patient’s size exact calculations must be performed as far as possible. Therefore Boone et al propose a formula which denotes the relationship between the thickness of a water phantom and the thickness of a PMMA phantom which is

\[ T_{\text{water}} = 1.1207 \times T_{\text{PMMA}} \]

\[ \text{Eq. 5:3} \]

Where \( T_{\text{water}} \) and \( T_{\text{PMMA}} \) are the thicknesses of water and PMMA.

Siegel et al in order to assess the effect of body shape on radiation dose and image quality acquired CT scans in an oval shaped phantom which simulate more realistically the human body, and a circular one with the same cross-sectional area. Results indicate that there is no significant difference in radiation dose or image noise between the two shapes. Nonetheless, contrast values were higher, by about 1%, with oval phantom than with circular. However, this difference would not be detectable by observers therefore it is not of clinical importance.

A number of researches evaluated size-dependent exposure factors, including varying the tube current, tube voltage, rotation time, slice thickness and pitch. They concluded that when we scan small diameter patients and large diameter patients with identical exposure
settings the images for small diameter patients have lower image noise, higher contrast and higher dose (Huda et al. 2000; Kalra et al. 2002; Prasad et al. 2002; Boone et al. 2003; Siegel et al. 2004). The last few years many researchers aim to take full advantage of this and they propose different ways to achieve acceptable image quality and low dose.

Image contrast increases when tube voltage decreases. This is particularly apparent in smaller phantoms. In addition noise increases as tube voltage decreases and this increase is less in small diameter phantoms than in large diameter phantoms. Siegel et al claim that is possible to use a tube voltage of 80 kVp and maintain image contrast in small phantoms. However Cody et al reported that use of a tube voltage of 80 kVp resulted in beam hardening artefacts in young patients, thus they recommended the use of 100-120 kVp in paediatric settings. Nevertheless Suess et al allege that many institutions have extended the 80 kVp settings to all their paediatric contrast studies.

Apropos of tube current, Shah et al estimate that it can be reduced significantly in paediatric CT examinations by nearly 60%. They reduced mA to 91-130 for cranial and to 76-90 for thoracic, abdominal and pelvic examinations without any effect on image quality or reader confidence in the level of detail available to reach a diagnosis. There are plenty of studies that suggest different values of mA. However the lower proposed value was that of 12.5 mA (Rogalla et al. 1999) and they claim that although there was a statistically increase in the amount of noise on the low dose images, in none of the low-dose examinations was diagnostic information lost.

Shah et al also claim that cranial CT is more sensitive to tube current reduction than is pelvic abdomen or thoracic CT.

Additionally reduction in both, tube current and tube voltage, decreases the radiation dose even further in a constant size phantom. For example when the tube current was adapted for phantom size the radiation dose with an 80 kVp instead of 140 kVp was reduced by 82% in the 8 cm phantom (Siegel et al. 2004)
Another exposure setting which contributes to radiation dose is pitch. When the pitch is doubled, the radiation dose is reduced by half (Vade et al. 1996; Frush et al. 1998). By increasing pitch from 1:1 to 1.5:1 the radiation dose is decreased by 33% and no significant difference found in subjectively graded image quality between the two pitch factors of 1 and 1.5 (Vade et al. 1996; Donnelly et al. 2001).

Boone et al remark that the smaller the anatomic features, higher spatial resolution CT scanning is required and this is accomplish by reducing slice thickness. But when slice thickness is halved the mAs value needs to be doubled to maintain the same CNR (equal amounts of photons reach the detectors). Consequently halving section thickness and doubling the mAs values requires the acquisition of twice as many contiguous images therefore the energy imparted and mean radiation dose are double. But since they claim that a reduction in mAs values is possible to achieve the same CNR they propose a compromise between slice thickness and mAs values which will result in reduced dose and adequate image quality.

Another finding has to do with how the contrast of different materials is affected by changing exposure settings. When tube voltage is increased, the improvement in image CNR is more important for low atomic number materials such as fat and muscle and least for high atomic number materials such as iodine. Additionally high Z lesions are most likely to have sufficiently high intrinsic CNR for their detection and characterization, especially at the lowest kV values (Huda et al. 2000; Huda et al. 2004). Therefore in contrast enhanced paediatric scans the desired image quality can be achieved with lower kV at significantly reduce dose levels (Suess and Chen 2002). Concerning examinations with indications for low Z materials such as soft tissue lesions, the use of lower kV values result in marginal dose savings and it is not clear whether reducing clinical kV would be worthwhile (Huda et al. 2004).

Suess et al maintained that attenuation in the head mainly depends on the thickness of the skull, which changes with age. After the age of 6 years they suggest the adult settings because the size of the head and the ossification of the skull are almost at adult levels.
the contrary for pelvic or abdomen examinations they propose as criterion for the selection of appropriate settings, to be the weight of the patient and not age.

5.5 Technical Innovations

In the last decade many technological developments came about aiming to contribute to dose reduction. Faster scanners are constructed, improved X-ray utilization and also all major components of a CT scanner were optimized such as the flatness and shape of X-ray filters, axial beam collimators, detectors etc.

However the intention of some companies was to reduce radiation dose all the more, so they attempt not only to improve the scanners from technological aspect but they also tried to change the scanning procedure. Hence an important concept was suggested and implemented in 1995: the Automatic Tube Current Modulation (ATCM) during the scan according to the patient’s geometry. ATCM is based on the fact that a single tube current is not appropriate given the variation in cross sectional geometry (wider side-to-side than front-to-back), overall thickness, and regional attenuation. To maintain image quality based on noise, for example, more tube current will be required through the upper abdomen than the upper chest. The tube current can thus be modified across the x-axis (cross sectional differences) or the z-axis (regional differences). Minimum and maximum tube current values can be set to ensure good image quality and lower dose. To sum up this concept has the following features:

Is based on the fact that image noise is determined by X-ray quantum noise in the transmitted beam projection

Adjust tube current in an effort to maintain constant image quality at the lowest dose.

Automatic tube current modulation has the potential to enable radiation dose optimisation
A dedicated study on automatic exposure systems (Greess et al. 2002) found the mAs product to be reduced typically by 10 ~ 60% depending on patient geometry and anatomical regions, and no deterioration of image quality was observed. Also recent investigations report a dose savings as much as 20% to 60% with ATCM (Greess et al. 2000; Greess et al. 2002; Suess and Chen 2002; Tack et al. 2003). Moreover tests performed with and without tube current modulation were compared with respect to absorbed dose and image quality. In the anthropomorphic phantom measurements, the dose savings were 15% using care dose and artifacts were negligible.

It is important to understand, however, that ATCM modulation does not necessarily optimize the examination relative to dose. The amount of the minimum and maximum tube current may need to be chosen or adjusted to the patient’s size before the examination. For example using ATCM with a maximum level of 300mA in a five year old patient undergoing an abdomen CT will deliver a high dose as well. In addition, ATCM may not be as effective when the patient is very small or if is off-center in the gantry (Frush 2003).

Other technical advantages include more efficient use of the X-ray beam. MDCT beam geometry is such that some photons are not used for formation. Work is under way to capture and convert some of this unused beam. Another potential is the use of noise reduction technology. Karla et al recently reported on this technique, which segments the projection (or raw) data and processes the separated COM recombining the processed data (Kalra et al. 2003). One effect was the reduction in image noise compared with the unfiltered image data.
Chapter 6 Patient Size and Image Quality: Experimental and Computer Simulations.

In this study the optimization of computed tomography is under two considerations. The first concerns the adjustment of scanning parameters to patient size in order to achieve acceptable image quality at lower dose and the second concerns the detectability of low contrast structures. This chapter refers to the first case. The impact of patient size and the possibility of adjusting scanning parameters to patient’s size necessitated the investigation via simulation techniques. The simulation was validated with experiments. The equipment for the experiments, the softwares and the methodology followed to assess image quality are described. The results from the investigation are presented as well.

6.1 Materials

The investigation could not be performed only with experiments due to the limited access to the scanner at the Royal London Hospital and because phantoms of different sizes required. Therefore, to understand and further investigate the relationship between, image quality and patient’s size and how the scanning parameters affect the image quality of different in size patients a simulation of the experimental procedure has been developed. Simulation techniques were validated with experimental measurements.

6.1.1 Materials for Simulation Techniques

The Simulation techniques were required in order to generate spectra and to simulate different diameters of patients, aiming to represent the entire size-range of paediatric
population. The spectra were generated from the SRS78-Spectrum Processor (Reilly et al. 1997) and for dose calculations the ImPACT patient Dosimetry calculator was employed (Jones et al. 1993).

### 6.1.1.1 Spectrum Processor

The electronic version of IPEM Report 78-Spectrum Processor (Reilly and Sutton 1997) was used to generate the X-ray photon spectra. For this study the simulation was run using a range of tube voltages between 80kVp and 120kVp and tube current-time product in between 50mAs and 165mAs. X-ray spectra are generated from tungsten target, at an anode angle of 9 degrees and a 10mm thick aluminium filter. Every one of these features is consistent with the Siemens Somatom Plus 4 scanner.

Polymethyl-methacrylate (PMMA) phantoms-substituted for human body- were used to attenuate the spectra. The diameters of the PMMA phantoms are 8cm, 12cm, 16cm, 20cm, 24cm and 32cm. The particular material was selected, in order to provide a match for the experimental procedure which performed to validate the simulation. In order to simulate the attenuation along the central axis of a clinical scanner each spectrum was additionally attenuated by air. The thickness of air was equal to the distance between the X-ray tube and phantom and between the phantom and the detectors.

Figure 6:1 illustrates the output screen with the spectrum details and the materials used to attenuate the spectrum. The attenuated photon spectra are presented as the variation of the number of photons per mAs per mm² with energy. The spectrum processor also provides the mean photon energy of the spectrum.
6.1.1.2 ImPACT CT Patient Dosimetry Calculator

The ImPACT CT patient Dosimetry calculator (version 0.99v) was used to calculate the paediatric relative effective doses. The software is a tool which calculates organ and effective doses from CT scanner examinations using the data from SR250. The SR250 contains data which are generated from Monte Carlo simulations (Jones and Shrimpton 1993). To obtain the doses for a particular examination it is necessary, to input information on the spreadsheet; information such as scanner model, tube voltage, mAs, slice thickness, scans length etc. Figure 6:2 shows the spreadsheet of the ImPACT dose calculator.

For the calculation of normalised organ doses the program employs a hermaphrodite phantom. The phantom is based upon a mathematical representation of an 'average' adult. Impact provides the ability to calculate the relative effective dose to children of different ages and for different parts of the body.
6.1.2 Experimental Equipment

In order to validate the simulation techniques we performed experiments with CTDI phantoms. The phantoms were scanned with the Siemens Somatom Plus 4 scanner.

6.1.2.1 CT Scanner

The Siemens Somatom Plus 4 CT scanner is a third generation (rotate/rotate) high voltage transfer slip-ring scanner. The X-ray tube can be set at 80kVp, 120kVp and 140kVp tube voltages and a wide range of tube currents in between 43mA and 420mA. Table 6:1 contains the main features of the scanner.
<table>
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<th>Summary of manufactures specification</th>
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<tr>
<td>Rotation times (sec)</td>
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<td>Nominal slice widths (mm)</td>
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<td>Collimation</td>
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<td>Field of view</td>
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<td>Gantry aperture</td>
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<td>Focus-to-detector distance</td>
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<td>X-ray tube filtration</td>
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<td>Anode angle</td>
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<td>CT number scale</td>
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Window width and window centre are freely selectable but also anatomic specific window settings for soft tissue and bone windows are stored. The slip-ring contains 48 detector elements with 16 channels on each which constitutes 768 detectors. The active
size of each detector element is 22mm x 18.9mm. Therefore the total active area is 19958.4mm$^2$.

Reconstruction algorithms (kernels) are designed by a system of numbers and letters where “AB” stands for Adult Body and “AH” Adult Head, and “CH” and “CB” are the reconstruction algorithms for Child Head and Child Body respectively. These abbreviations are followed by a number which denotes the amount of smoothing or sharpness (e.g. CB50, CB90) by which the image will be reconstructed. As the number increases so the sharpness increases. Also real-time multi-planar reformatting of secondary views is feasible.

The software read the images, and displays them. Image analysis is carried out by the placement of circle regions of interest (ROIs) which registers the CT number and its Standard Deviation.

**6.1.2.2 CTDI Phantoms**

A body phantom simulates a human body or part of it with respect to size, shape, attenuation properties and radiation interactions. CTDI phantoms are made from polymethyl-methacrylate (PMMA) material. Compared to water, CTDI phantoms are not tissue-equivalent. However they are widely used in experiments either for dose measurements or image quality assessment. The density of PMMA is 1.19g/cm$^3$ and with mass composition of 8.05% H, 59.99% C and 31.96% O (C$_5$H$_8$O$_2$).

Two annular CTDI phantoms of 16cm and 32cm diameters were available at St Bartholomew’s hospital. The 16cm diameter phantom corresponds to the average size of a five-year-old child while the 32cm is approximately the average size of an adult. The axial length of the phantoms is 14cm. The 16cm phantom is used unaccompanied and also it may be slotted inside the larger annulus to make the 32 cm diameter phantom (Figure 6:3). Both phantoms have drill holes that are placed, one in the center and 8 holes around the periphery. Each hole is of 12mm diameter. PMMA rods are available in order
to fill the holes when needed. Figure 6:3 shows the CTDI phantom. The arrangement of the holes is distinguishable.

![CTDI phantoms](image)

**Figure 6:3 CTDI phantoms**

## 6.2 Methodology

### 6.2.1 Experimental Procedure

CTDI phantoms were scanned at various combinations of tube voltage and tube current-time products. All other parameters were kept constant. Table 6:2 shows the scanning parameters which the CTDI phantoms were exposed to. Window width and window centre were selected (at each combination of exposure settings) manually so that the optimum image quality, relative to the observers, was achieved.

<table>
<thead>
<tr>
<th>Scanning Parameters</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Tube Voltage</td>
<td>80kVp, 120kVp and 140kVp</td>
</tr>
<tr>
<td>Tube current-time product</td>
<td>50-135mAs</td>
</tr>
</tbody>
</table>

Table 6:2. The Scanning parameters the two CTDI phantoms were exposed to.
Chapter 6. Patient Size and Image Quality

---

<table>
<thead>
<tr>
<th>Slice thickness</th>
<th>5mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rotation time</td>
<td>1sec</td>
</tr>
<tr>
<td>Pitch</td>
<td>1.5</td>
</tr>
<tr>
<td>Reconstruction algorithm</td>
<td>CB50</td>
</tr>
</tbody>
</table>

Initially a topogram\(^2\) was performed and the same area selected to ensure that each phantom was scanned in the same region. Throughout the phantom scanning all the holes of the phantoms were plugged except for one hole around the periphery and the central hole. These holes were filled with air. For each set of exposure settings both phantoms were scanned successively five times. The 16cm phantom was scanned at 80kVp with 75mAs, 105mAs and 135mAs and also at 120kVp with 50mAs, 70mAs, 90mAs and 110mAs. The phantom was not scanned at 140kVp since the plan was to find out whether an acceptable image quality was achievable using lower exposure settings. The 32cm phantom was scanned only at 120 kVp with 50mAs, 70mAs, 90mAs and finally with 110mAs.

For consistency, at each scan the same slice was subjected to image quality analysis. Image quality was assessed by calculating a CNR for PMMA material and air. Firstly, the ROI is placed in the central hole of the phantom filled with air, at a position that lacks of edge artifacts. Then the same ROI is placed on the PMMA material (phantom background) close to the central hole (Figure 6:4). For each ROI the CT number and its standard deviation (SD) were recorded. The reason for the assessment of image quality at the centre is because the centre of the phantom has the highest level of image noise and the lowest levels of image contrast hence the inferior image quality.

\(^2\) Topogram: Survey radiograph with diagnostic image quality for patient position and planning the complete examination
Chapter 6. Patient Size and Image Quality

Figure 6:4 CTDI phantom. Placement of ROIs.

The procedure repeated for the five acquired images. The contrast calculated by subtracting the CT numbers of air from the CT number of the phantom background, while the noise is the standard deviation of the CT number in the background. For each image the Contrast-to-Noise Ratio (CNR) calculated (Eq. 6:1). Eventually its mean and its standard deviation from the five scans were calculated.

\[
\text{CNR} = \frac{HU_{\text{Phantom background}} - HU_{\text{Air hole}}}{SD_{\text{Phantom background}}} \quad \text{Eq. 6:1}
\]

In this part of the study our intention was also to examine how different contrast levels are influenced by patient’s size and the scanning parameters. Therefore solutions with different concentrations of Iodine in order to achieve various contrast levels with the PMMA material were prepared. However, the limited access to the scanner did not allow us to perform these experiments and the study was restricted to calculate the high contrast between PMMA material and air.

6.2.2 Simulation Technique

PMMA phantoms are “exposed” to the generated spectra. Likewise with the experimental procedure the contrast and noise were calculated in the centre of the phantom. Contrast was calculated between PMMA material and air by subtracting the CT number of air from the CT number of PMMA material. The CT numbers (Hounsfield Unit) were computed using the equation:
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\[ \text{CT number (HU)} = 1000 \rho \frac{\mu_{\text{object}} - \mu_{\text{water}}}{\mu_{\text{water}}} \]

The linear attenuation coefficients (\(\mu\)) are obtained from the National Institute of Science and Technology (NIST’s) XCOM photon database (Berger et al. 1998). The particular energies at which \(\mu\) are computed are the mean energy of the spectrum at the centre of each phantom. In the 8cm phantom the contrast of iodine, adipose tissue, and skeleton cortical bone relative to muscle were also calculated. The compositions of the materials are obtained from the International Commission on Radiation Units and Measurements (ICRP Report 44).

Image noise in CT is dominated by quantum noise which is inversely proportional to the square root of the radiation exposure level absorbed by the X-ray detector (Judy 1995; Huda et al. 2000; Huda et al. 2004). The relative noise is assumed to be inversely proportional to the square root of the energy fluence incident on the detector. The Contrast-to-Noise Ratio (CNR) for all PMMA phantoms is defined as

\[ \text{CNR} = k \cdot \frac{\text{Contrast}}{\text{Noise}} \]

\[ \text{Eq. 6:3} \]

Where \(K\) is a scaling factor and contrast is calculated in HU. The scaling factor was applied on the computer-modelled results in order to compare them with the experimental measurements. The scaling factor was calculated using the following equation:

\[ \frac{\partial}{\partial k} \sum_{i=1}^{N} (\psi_i - k \chi_i)^2 = 0 \]

\[ \text{Eq. 6:4} \]

Where \(\psi_i\) are the computed model CNR values and \(\chi_i\) the measured CNR values.
6.3 Other Considerations

6.3.1 Equivalent Thickness

Corrections for the differences in composition between PMMA phantoms and human patients were made. Patients in a model are considered as unit density water (H$_2$O). The relationship between the equivalent water thicknesses ($T_{\text{water}}$) and the PMMA phantom thickness ($T_{\text{PMMA}}$) was calculated. For the 5 X-ray spectra, the diameter of a water equivalent phantom was computed such that the photon fluence attenuation was identical to that of the PMMA phantom. The linear relationship found was used to convert the PMMA cylinder diameters to water cylinder diameters. The equation is:

$$T_{H,O} = 1.1125 T_{\text{PMMA}}$$  \hspace{1cm} \text{Eq. 6:5}

With this relationship, the 8 cm diameter PMMA cylinder was found to be equivalent in terms of X-ray attenuation to an 8.9 cm diameter water cylinder and a 24 cm diameter PMMA cylinder was found to be equivalent in attenuation to a 26.7 cm diameter water cylinder. This correction was necessary for the presentation of the results in terms of patient-relevant dimensions.

6.4 Statistical Analysis

The CNR measurements, acquired from the experiments performed on the two CTDI phantoms, were used to validate the Computer simulations. Differences in experimental mean CNR and computed CNR, for every scanning parameters and each phantom diameter, were tested with paired t tests. The two-tailed t distribution was used as a test for significant differences. $P \leq 0.05$ was considered to indicate a statistically significant difference.
Linear regression was performed for kVp and mAs settings versus equivalent diameter by using Excel (Microsoft Office XP). The regression lines resulted from the experiments were compared with the regression lines resulted from computer simulations by using the two sided t test with the t distribution data provided by the statistical functions in Excel.

For testing the equality of two slopes the following equation was used to calculate the t values:

$$t = \frac{b_1 - b_2}{\sqrt{s_{b_1}^2 + s_{b_2}^2}}$$  \hspace{1cm} \text{Eq. 6.6}

Where $b_i$ is the slope and $s_{bi}$ is the standard error of the slope\(^3\).

The differences in image noise, contrast and CNR were reported in simple percentiles.

\(^3\) For supplementary information see appendix
6.5 Validation of the Simulation Techniques

A good agreement between experiments and simulation is observed. Computed CNR and measured CNR are plotted for mAs settings at 120kVp for the 16cm and the 32cm phantom (Figure 6:5). Comparison of these values with a two sided t test indicated that there was no significant difference between any of the 8 pairs of points - measured and computer simulation result; Two-tailed P values (paired t test) ranged from 0.545 to 0.887. Apart from the comparison of the individual values, the experimental and computer simulation regression lines are compared showing that there is no statistical difference between their slopes.

Validation was also performed at 80kVp for the 16cm phantom. There is no statistical difference between experimental measurements and computer simulation as well (P value = 0.8212). However, there is a statistical difference between the slopes resulted from measured and computer simulation regression lines. This is probably due to the fact that the CT scanner at Royal London Hospital is not used at 80kVp and probably it is out of calibration for the specific value. Consequently, the values acquired at 80kVp were not
included in the calculation of the scaling factor ($K$). Figure 6:6 shows CNR versus mAs at 80 kVp for the 16 cm phantom.

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**Figure 6:6.** Graph shows CNR as a function of mAs at 80 kVp for the 16 cm phantom. Black marks denote the experimental measurements and grey marks the computer-modeled values.
6.6 Results and Discussion

Maintaining constant technique factors in CT, image quality is being determined by the size of the patient. Image quality assessment is performed in terms of noise, contrast and Contrast-to-Noise Ratio (CNR).

The size of the patient is an important factor that affects image quality in CT examinations. As infants grow into adults, the size increase markedly and the X-ray beam traversing their body is attenuated substantially. The increased X-ray attenuation results in a reduce detector exposure thereby image noise is expected to increased.

In Figure 6:7 the relative noise against phantom diameter is plotted at 135mAs for different tube voltages. Noise was normalised to the noise obtained at the 16cm phantom for 120kVp and 90mAs. Since noise was not in terms of Hounsfield Units, normalisation was necessary in order to have a sense about the magnitude of noise. Noise increases exponentially as the size of the phantom increases and it is inversely proportional to tube voltage. The exponential increase of noise is due to the exponential decrease of photons reaching the detector, when phantom thickness increases \( N = N_0 e^{-\mu x} \). Comparing the slopes of the regression lines (0.1169 at 80kVp and 0.1069 at 120kVp) is observed that at low tube voltages the exponential increase of noise with phantom diameter is steeper. Therefore, assuming reduction of tube voltage, there is less increase of noise in small size phantoms than in the larger ones. In particular, the reduction of the tube voltage from 120 kVp to 80 kVp results in 115.8% increase of noise level at the 8 cm phantom (infant size), 133.9% increase at 16 cm phantom and 174.9% increase at the 32cm phantom (adult size patient). Nevertheless is noted that the increase of noise when tube voltage is reduced is significant in all sizes.
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Figure 6:7. Graph shows the noise against phantom diameter for a variety of tube voltages. Data were normalised to unity at the 16cm phantom for 120kVp and 90mAs.

At 120 kVp the Noise level against mAs for different phantom diameters is plotted in Figure 6:8. Regression analysis was used to determine the associations between noise and mAs setting. Scatter plots of the data indicated that the data were curvilinear and that a straight line model would not provide the best fit. The simplest extension of the straight line model is a second order polynomial (Siegel et al. 2004). The average regression fit results $r^2$ value was 0.9957. Noise level, as expected, increases for low mAs values and for larger phantoms.

Contrast is the difference between signal and background attenuation coefficients. The value of the attenuation coefficient of a material depends on the energy of the X-ray
beam traversing the patient. In the CT diagnostic energy range, as energy increases the linear attenuation coefficient decreases as well.

Whilst the beam advances through matter its mean energy constantly increases due to beam hardening effects. The amount of beam hardening depends on the initial X-ray spectrum (i.e. tube voltage) and filtration, on the material and as well as the thickness of material traversed (i.e. phantom size). The Figure 6:9 shows the attenuation coefficients ($\mu$) for different materials at 80kVp and 120kVp when the X-ray beam undergoes beam hardening while transverse a PMMA material. The particular energies at which $\mu$ were computed were the mean energies of the transmitted spectra after the particular thickness of PMMA material was traversed. When the X-ray beam encounters the surface of an internal organ it has already undergone beam hardening from previous organs and tissues. This fact explains the reason we consider the beam hardening in the PMMA material instead the beam hardening in a single organ (i.e. beam hardening through the bone).

It is observed that as the thickness of the material traversed increases (bigger phantom size), the attenuation coefficient of the organs become smaller. In addition it is noticeable that the reduction rate it is not solely a function of the thickness traversed but is a function of the type of the material as well. The linear attenuation coefficient of high effective atomic number materials (i.e. iodine, $Z_{eff}$=47) presents a steeper reduction with
the traversed thickness, while low effective atomic number materials (i.e. water, $Z_{\text{eff}}=7.42$) present a smoother reduction. Moreover high effective atomic number materials are more sensitive to energy changes and it is evident in Figure 6:9 where the difference of water attenuation coefficients at 80kVp and 120kVp is very small compared to the difference presented in iodine. The reduction of $\mu$ with energy is shown better in Figure 6:10 where the linear attenuation coefficient against energy is plotted. It is observed that $\mu$ for high effective atomic number materials is reduced more quickly as energy is increased.

![Figure 6:10. Linear attenuation coefficient against energy.](image)

The behaviour of $\mu$ of different materials with energy is expected to influence their contrast. Figure 6:11 shows the contrast of adipose tissue, skeleton-cortical bone and iodine relative to muscle against tube voltage. When tube voltage was reduced from 120kVp to 80kVp image contrast was increased by approximately 64% for iodine, ~40% for skeleton cortical bone, and ~ 23% for adipose tissue. It is confirm that materials with high effective atomic number are affected to a greater extent than low effective atomic number materials when tube voltage changes. Therefore when tube voltage is increased the contrast of high Z materials is reduced more quickly. This results in lower improvement of CNR for the high Z materials when tube voltage increases. Therefore examinations with contrast agents should be preferred since they can be performed at lower kVp without significant impact on IQ and with dose savings.
Figure 6:11 Contrast against tube voltage for iodine, cortical bone and adipose tissue.

Data in Figure 6:12 show iodine contrast relative to PMMA as a function of phantom diameter for five tube voltages. All the regression lines result to $r^2$ values greater than 0.99. The data show that increasing both tube voltage and phantom diameter image contrast decreases and this is due to the aforementioned beam hardening effect. At 80kVp for example the contrast in an infant (8cm phantom) was approximately 13% higher than the contrast in an adult (32cm phantom). In addition increasing the tube voltage from 80kVp to 120kVp the iodine contrast reduces for the 8cm phantom by approximately 39%, for the 16cm phantom approximately 40% and 42% for the 32cm phantom. The corresponding values for the contrast between cortical bone and PMMA are 30.97% for the 8cm, 31.24% for the 16cm and 31.82% for the 32cm phantom; a modest increase as expected.
Figure 6:12 Iodine contrast Vs. phantom diameter acquired with five tube voltages. ♦ = 80kVp, ■ = 90kVp, ▲ = 100kVp, × = 110kVp and ✻ = 120kVp

Therefore contrast is a function of the material and the size of the phantom. The contrast of high atomic number materials presents a higher increase when the size of the phantom decreases. In particularly the Iodine contrast for the 8cm phantom increases by approximately 18% compared to the contrast for the 32cm phantom. Bone contrast increases approximately 12% at the 8cm phantom (Figure 6:13). Concerning examinations with high atomic number imaging (contrast agent) the big increase could be trade off by using lower tube voltage since in small size phantoms the contrast is less influenced by changing tube voltage.

Figure 6:13. The contrast of iodine and cortical bone relative to PMMA are plotted against phantom diameter.
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The dependence of contrast and noise on phantom diameter is such that enables us to acquire high image quality but with decreased dose. One of the most suitable characteristics to pass judgment on image quality is Contrast-to-Noise Ratio (CNR). In small diameter phantoms the contrast is higher and noise level is smaller. Therefore in small phantoms the contrast-to-noise ratio is expected as well to be superior. Figure 6:14 shows CNR against phantom diameter when phantoms are exposed to constant scanning parameters (120kVp and 135mAs). A second order polynomial provides the best fit between CNR and phantom size ($r^2=0.9943$). It is observed that when scanning parameters are kept constant, CNR depends on phantom size and as expected, in small diameter phantoms it is superior. Considering an 8cm and a 32cm phantom the CNR of the 8cm phantom is 90% higher.

![Figure 6:14 CNR against phantom diameter. Data were normalized to unity at the 32cm phantom.](image)

In this study the relationships between CNR, tube voltage and tube current-time product were investigated as well. Concerning tube voltage, it is observed that affects CNR but in different way for each phantom size (Figure 6:15). Particularly, in small diameter phantoms (8-24cm) CNR increases linearly with tube voltage. The linear regression fit results average $r^2 = 0.999$. However, the data shows that for larger phantoms the best fit is not a straight line. The best regression fit for the 32cm diameter phantom is a third order polynomial which results to $r^2 = 0.9732$. The explanation for this is probably the scattering effect which increases with phantom diameter. Thus, when tube voltage
changes the CNR does not behave with the same manner, for each phantom diameter. Therefore, if it is desired to find an equation that predicts CNR for every tube voltage, this equation should be extracted for each phantom diameter. This is an important outcome since it will enable the manufacturers to find the appropriate tube voltage for each patient size and consequently, to make the optimal protocols based on the size.

Moreover, a small change of CNR, in the order of 15% average in all phantom sizes, results from the reduction of tube voltage to 110kVp from 120kVp. It is worth to perform clinical experiments to examine whether that small or even smaller change of tube voltage does not lead to loss of medical information and at the same time to examine the contribution to dose reduction.

Concerning the tube current-time product, a linear regression line is applied for all of the six phantom diameters. The regression fit results in average $r^2=0.9923$. As tube current-time product increases a higher CNR is obtained. However as smaller the size of the phantom is, a steeper reduction of Contrast-to Noise Ratio is observed by reducing mAs (Figure 6:16).

Figure 6:15 CNR against tube voltage. Data were normalized to unity at the 32cm phantom and 120kVp.
Figure 6:16 CNR against mAs. Data were normalized to unity at the 32cm phantom for 120kVp and 165mAs.

Nevertheless, CNR in small diameter phantoms is much higher therefore this could be trade off for lower tube current-time product.

Figure 6:17 shows the experimental CNR acquired on the 16cm and 32cm CTDI phantoms. A linear relationship found between the CNR and mAs as well. The value of 300mAs is employed in most of the hospitals for adult’s abdominal CT examinations and in many cases this value is used for paediatric patients. If we extrapolate the linear relationship for both the 16cm and 32cm phantoms then the mAs for the 16cm phantom could be reduced by up to 95% while the same image quality as with the 32cm phantom is acquired. In that case as calculated from the ImPACT dose calculator the dose would decreased by 95% as well.
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Figure 6:17 experimental CNR against mAs for the 16cm and 32cm CTDI phantom.
Chapter 7  Detectability of Low Contrast Details

The detection of low contrast details represents one of the most important tasks of all sectional imaging methods. In practice, this is largely a question of the soft tissue contrasts which result from differences in density and depends only weakly on the energy of the X-ray quanta (Kalender 2000). The objective of this chapter is to investigate the impact of scanning parameters and in particular the tube voltage and tube current-time product, on the detection of low contrast details. The detection of the details has been examined in relation to the contrast level and in particularly, for three nominal contrast levels, 1%, 0.5%, and 0.3%. The detectability also has been examined in relation to the size of the details. The equipment employed for the experimental procedure is described. The detectability of the details hence image quality was assessed objectively and subjectively. The results are presented via graphs and Low Contrast Detectability (LCD) histograms.

7.1 Materials

The Siemens Somatom Plus 4 scanner which has been described in chapter 6, was used for the performance of experiments.

7.1.1 Catphan phantom

Catphan® 500 phantom (Figure 7:1) was used in order to examine how the contrast level and the size of a lesion are affect their detectability in relation to the scanning parameters.
Chapter 7. Detectability of Low Contrast Details

Particularly in this study we used the CTP515 module which is used to perform measurements and tests related to low contrast sensitivity.

CTP515 is 20cm in diameter and 4cm thick. It contains three contrast levels of 0.3% (3 HU), 0.5% (5 HU), and 1% (10 HU) to measure low contrast performance. Each contrast level consists of a group of 9 rods which are 15, 9, 8, 7, 6, 5, 4, 3, 2 mm in diameter (Figure 7:2). These insets are positioned on the periphery of the cylinder. They are long cylindrical objects (Supra-slice) and provide consistent contrast values at all z-axis positions avoiding any volume-averaging errors as you spiral through the section. This module also consists of internal insets which are short cylindrical objects (Sub-slice) and produce partial volume effects. These objects are cast from the same mix as the 1% supra-slice objects (nominal contrast level 10 HU). In this study we considered only the objects located on the periphery of the cylinder (Supra-slice targets) aiming for low contrast evaluation without partial volume effects.

![Figure 7:1 Catphan Phantom](image)

**Figure 7:1 Catphan Phantom**

### 7.2 Methodology

A topogram was performed at the Catphan® phantom and the CTP515 module area was selected to be scanned. Table 7:1 shows the scanning parameters at which the phantom was scanned.
Table 7:1. The scanning parameters which the CTP515 module was exposed to.

<table>
<thead>
<tr>
<th>Tube voltage</th>
<th>Tube current-time product (mAs)</th>
</tr>
</thead>
<tbody>
<tr>
<td>80kVp</td>
<td>75 105 135 165</td>
</tr>
<tr>
<td>120kVp</td>
<td>50 70 90 110</td>
</tr>
<tr>
<td>140kVp</td>
<td>43 77 94 111</td>
</tr>
</tbody>
</table>

All other parameters were kept constant. Slice thickness was 5mm, rotation time was 1sec, pitch 1.5 and all images were reconstructed with the algorithm CB50. For every set of scanning parameters the CTP515 modulus was scanned successively three times. Thereafter each acquisition, the section was reconstructed and the central slice of the low contrast modulus was selected and subjected to objective and subjective assessment of image quality. Window settings were selected to allow the maximum number of the circular objects to be seen.

7.2.1 Objective Assessment

The objective method is based on measurements of the CT-numbers and their standard deviations. As with the CTDI phantom, the image analysis was performed by the placement of ROIs inside and nearby the circular objects as can be seen in Figure 7:2. ROIs were placed in a way that avoids measuring edge artefacts. In order to examine how
Chapter 7. Detectability of Low Contrast Details

The detection of structures is influenced by the contrast two different low contrast levels were studied. Measurements were taken for the 1% (10HU) and 0.5% (5HU) contrast levels inside the 15mm diameter object. Our intention was also to examine how the size of a structure influences its detection. Therefore at the 1% contrast level apart from the measurements inside the 15mm object measurements were taken inside the 9mm and 8mm diameter objects as well.

The contrast is calculated by subtracting the CT number measured inside the circular object from the CT number measured nearby (phantom background). Noise is the standard deviation of the CT number in the phantom background. For each image CNR was calculated for each of the four objects. Figure 7:2 illustrates where measurements were taken.

![Figure 7:2 CTP515, low contrast module. The placement of ROIs in the objects is displayed.](image)

As already has been mentioned for each set of scanning parameters three images were available. Eventually, for each object a mean CNR and its standard deviation were calculated from the three images. That is, the CNR was computed for the 15mm, 9mm and 8mm diameter objects at 1% contrast level and for the 15mm diameter object at 0.5% contrast level. The CNR of each object and tube voltage is presented in plots as function of the tube current-time product.
Chapter 7. Detectability of Low Contrast Details

7.2.2 Subjective Assessment

The subjective method is based on the visibility of structures. In each image the number of circular objects that were visible at every contrast level was scored by two observers. The observers had to decide which and how many objects were recognized as separated or as not separated. The object was considered as visible when it was clearly distinguishable from its background or when the object was shaped in such a way that the observers were persuaded that the object was sufficiently distinct from its background and it was possible to place an ROI inside the object and avoid edge artefacts. The decision for the number of visible objects was taken by the two observers in common. At each contrast level the number of detectable objects was the average number from the three images obtained.

Outcomes are presented via Low Contrast Detectability (LCD) histograms. LCD is reported as the number of detectable objects that were visually detectable at each contrast level. For example five objects imply that the five largest targets were visually detectable.

7.3 Results and Discussion.

The detectability of low contrast details is one of the most critical and important diagnostic factors in computed tomography. The low contrast detectability is determined primarily by the noise level in the image. The objective assessment of image quality was carried out with the calculation of the Contrast-to-Noise Ratio.

Figures 7:3-6 shows the Contrast-to-Noise Ratio as function of tube current-time product for the 15mm, 9mm and 8mm diameter objects at 1% contrast level and for the 15mm diameter object at 0.5% contrast level.
Figure 7:3. CNR as a function of mAs for the 15mm object at 0.5% contrast level. CNR is plotted for three voltages.

Figure 7:4. CNR as a function of mAs for the 15mm object at 1% contrast level. CNR is plotted for three voltages.
Chapter 7. Detectability of Low Contrast Details

Figure 7:5. CNR as a function of mAs for the 9mm object at 1% contrast level. CNR is plotted for three voltages.

Figure 7:6. CNR as a function of mAs for the 8mm object at 1% contrast level. CNR is plotted for three voltages.

For all of the four objects the Contrast-to-Noise Ratio obtained at 120kVp and 140kVp are nearly equal. Some variations exist; however, they belong in the range of error. This justifies the selection of 120kVp instead of 140kVp for most of the protocols. As
expected the selection of 80kVp gives lower CNR especially when it is combined with low tube current-time product. When the 80kVp is combined with high tube current-time product the CNR approaches the CNR acquired when the 120kVp is combined with low tube current-time products. However the delivered dose is much higher and hence the selection of lower tube voltage and higher mAs in order to acquire the same image quality is not suggested. For example when a newborn is scanned for an abdominal examination, at 120kVp and 50mAs the effective dose, as calculated from the ImPACT dose calculator, is approximately 4.56mSv whereas if the newborn is scanned at 80kVp and 165mAs the effective dose is 10.26mSv.

Concerning the detection of details with different low contrast levels it is observed that the lower the contrast level is the bigger the uncertainty in the calculation of CNR. Therefore the lower the contrast level of a detail is, a higher CNR is demanded in order to assure the detection of a structure. For example the 0.5% contrast level detail was not detectable at 80kVp.

Our results do not allow us to draw a definite conclusion about the tube current-time product should be used in low contrast details. In our experiment the observers had to calculate the Contrast-to-Noise Ratio of objects known in size and position. The detection of abnormalities on clinical images is more complex and may require higher CNR levels. However, our results demonstrate that small size lesions and objects of contrast level around 5HU necessitate higher CNR levels.

The Low Contrast Detectability (LCD) histograms below show the number of detectable objects at various tube voltages and tube current-time products. The histograms are the output from the subjective assessment of image quality.
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Figure 7:7. Histogram shows LCD at 80 kVp & 105 mAs, 120 kVp & 110 mAs and at 140 kVp and 111 mAs.

Figure 7:8. Histogram shows LCD at 80 kVp for 75 mAs, 105 mAs, 135 mAs and 165mAs.
Figure 7:9. Histogram shows LCD at 120 kVp for 50 mAs, 70 mAs, 90 mAs and 110 mAs.

It is observed, that the high tube voltages increase the detectability of low contrast level objects. At 120kVp and 140kVp, the objects which belong at contrast level of 0.5% and 0.3% are detectable. In contrast, at 80kVp the detection of 0.5% and 0.3% objects is not achievable (Figure 7:7). It is observed that when 120kVp and 140kVp are applied there are not significant differences. It is also observed, that as mAs increases the detection of...
smaller size objects increases. This is effective in all tube voltages however, is more severe when high mAs and high tube voltage are utilized. However it is noted that above a threshold value high mAs does not serve any purpose. Particularly at 80kVp and 75mAs an object of 8mm diameter and contrast level around 10HU could be detected. Increasing the mAs up to 135mAs or 165mAs a 6mm diameter object can be detected. At 120kVp and 50mAs, an object of 4mm diameter with 10HU contrast level is detectable, but objects of lower contrast level would be overlooked. On the other hand, if tube voltage remains 120kVp but tube current-time product increases at 110mAs, an object of 2mm diameter and 10HU contrast and also an object of just 3HU contrast and 9mm diameters are probable to be detected. This is of high importance in cases of early stage tumours or small lesions. Therefore the size of a lesion affects its detectability and especially when the small size is combined with low contrast level the detection becomes more difficult. However the major role for the detection of a structure is its contrast level and its size play a minor role for the detection.
Chapter 8  Conclusive Comments

The utilization of Computed Tomography in paediatric examinations constantly increases. During the procedure, a high amount of dose is delivered to children, which could be avoided. This study is dedicated to examine whether the selection of scanning parameters –tube voltage and tube current-time product- could be based on patient size instead of patient age or weight. Moreover the detect ability of low contrast objects was investigated. The ultimate aim is to optimize the computed tomography exposure by achieving acceptable image quality while reducing the delivered dose.

The data analysis of our study shows that there is a definite relationship between image quality and the size of a patient. When exposure settings are kept constant, the level of noise, contrast and Contrast-to-Noise Ratio (CNR) depends on the size of the phantom.

Noise in small size phantoms is low and increases exponentially when the size of the phantom increases. Noise level also depends on the tube voltage (kVp) and tube current-time product (mAs). In particular, as mAs and kVp are increased the noise level is reduced. In particularly it was studied how noise level is affected if the previously mentioned scanning parameters were reduced. Nothing important was observed by reducing mAs. On the contrary, when the tube voltage was reduced the noise level was increased less in the small size phantoms.

Contrast relates to the linear attenuation coefficient ($\mu$). In this study it was examined how the linear attenuation coefficient of different materials changes when phantom diameter increases. The results indicate that there is a steeper reduction of $\mu$, when phantom size increases (higher X-ray beam energy), for the high Z materials. This behavior impacts the contrast. Firstly the contrast in small size phantoms is superior.
Secondly, since high atomic number materials are more sensitive to the energy variations, the contrast of high atomic number materials is more sensitive to changes in tube voltage (since the photoelectric effect depends on $Z^3$). That is, as tube voltage increases contrast reduces largely for high Z materials compared to the reduction occurs in low Z materials and it was also observed that for both low and high Z materials the reduction was more noticeable in bigger size phantoms.

However, this marked reduction of high Z materials, leads to modest improvement of Contrast-to-Noise Ratio (CNR) when tube voltage increases. This is confirmed in a recent study by Huda et al for head CT examinations (Huda et al. 2004). Therefore, for that reason the examinations that are aimed to detect high contrast level structures should be performed in low tube voltages since the requisite medical information can be acquired at the lower tube voltages without any consequences on image quality and with dose savings. In consequence, examination that could be performed, utilizing contrast agents must be preferred in cases in which the possibilities of other side effect such as allergies to the contrast agent are negligible.

The dependence of contrast and noise on phantom size is such that enables us to acquire the same image quality but at lower dose. Like with contrast, the CNR is superior for small size phantoms as well. Our data shows that the CNR for an 8cm phantom is 90% higher than the CNR in a 32cm phantom. It is also observed that the relationship between tube voltage and CNR depends on the size of the phantom (Figure 6:15). This is an important outcome since it will enable the manufacturers to find the appropriate tube voltage for each patient size and consequently, to make the optimal protocols.

Moreover, despite the fact that in small size phantoms the CNR decreases faster as the mAs product reduces, the much higher CNR acquired in small size phantoms, compared to the CNR acquired in bigger phantoms make feasible the reduction of mAs without degradation of image quality. The experiments performed for the evaluation of the simulation technique were carried out with tube current-time product range between 43mAs and 110mAs. A linear relationship found between the CNR and tube current-time product for both the 16cm and 32cm phantom. If the corresponding relationship is valid
for a wider range of mAs then the tube current-time product for the 16cm phantom could be reduced by up to 95% (compare to the settings hospitals use for adult CT examinations, namely 120 kVp and 300mAs) with the same image quality as for adults. It would be interesting to perform experiments in order to justify that reduction since the outcome would be a dramatic reduction of dose up to 95%. A recent study found similar dose reduction however they claim that so large dose reduction could be traded off for better spatial resolution (Boone et al. 2003). For example the 5mm section thickness can be adjusted to 1.3mm providing a four fold increase in Z axis resolution and still providing a dose reduction of 80%. However we must note that the objective measurements of noise, contrast and CNR might not respond to an accurate diagnosis since a higher level of diagnostic confidence appears to be demanded for paediatric imaging. Therefore experimental studies should be considered as guides and must be combined with clinical studies in order to select the optimum scanning parameters.

This study examined also the detectability of low contrast details. The factor that influences the detection of a structure is its contrast level foremost, and then its size. The results justified the selection of 120kVp instead of 140kVp in most of the protocols since the CNR acquired at 140kVp it does not make big difference. The lowest kVp that examined in this study and it was the 80kVp, gives much lower values of CNR and in cases where the contrast level is lower than the 1% the detection of structures was almost improbable. A study reported that in enhanced hepatic, CT, tumor to liver contrast was 5-40 HU; however, in some cases, it was only a few Hounsfield units (Awai et al. 2002). In that case, if 80kVp was applied radiographers might overlook the tumor. Therefore, tube voltage has a major role in the detection of low contrast objects and we must be very cautious in the selection of tube voltage.

Concerning mAs our data shows that mAs does not contribute to the detection of low contrast lesions except if it is combined with high tube voltages. For instance, the 0.3 % contrast level objects were detectable only at 120 kVp and 140 kVp with 110 mAs and 111 mAs respectively. However, our data demonstrates that tube current – time product contributes to the visualisation of small diameter objects. In all tube voltages, it is

83
observed that, increasing the mAs smaller objects are probable to be detected. However above threshold value high mAs does not serve any purpose and the value of 300mAs employ in many protocols is not justified. In the cases that exposure settings were kept constant the small size objects were difficult to be detect. This is of vital importance in paediatric computed tomography, since in some cases, the paediatric tumours or lesions are of smaller magnitude of those observed in adults. Therefore, the utilization of high mAs settings increase the probability of detection of smaller size objects. Moreover, the visualization of small size structures will lead to the detection of tumours in early stage in both adults and children.

The selection of the appropriate exposure settings – depending on the diagnostic task and patients’ size – should consider the detectability of low contrast structures since CT is beneficial because is able to distinguish structures of similar densities. One limitation of the study was that the detectability of a variety of low contrast level and size objects was performed with a 20 cm diameter Catphan phantom only. The limitation consist in that we did not examine how the size of the phantom influence the detectability of low contrast objects. Another limitation is that the Siemens CT scanner did not allow us to set the tube voltage at values between 80kVp and 120kVp. It is worthwhile to overcome these important limitations and perform an extensive investigation how patient size and exposure settings influence the detection of low contrast structures.

Below we present some suggestions in view of the risks of CT.

Firstly a radiologist must decide whether the selection of computed tomography is the appropriate examination and whether there is not alternative examination. For example, for abdominal and pelvic imaging, ultrasound and MRI are safer alternatives to CT in many cases.

Secondly, if the selection of Computed Tomography is justified then the radiologist must select the appropriate scanning parameters, not only in relation to tube voltage and tube current-time product but including the collimation, pitch, reconstruction algorithm and the scanning length as well. For instance, in examinations such as follow-up CT studies at
short intervals or examinations that studies the response of the organism after a chemotherapy treatment lower tube voltage and tube current-time product should be used in order to avoid unnecessary overexposure of the patient.

Chapple et al. (Chapple et al. 2005) investigated whether the use of lead shielding around the trunk, during head CT examination reduce the dose. Their investigation show that effective dose to the phantoms are reduced by up to 30% when 0.35mm lead protection is used. This suggests that a significant proportion of the scattered dose during CT impinges from outside the body, and thus can be effectively shielded. For small size patients this could lead to substantial reduction in the radiation risks associated with CT of the heat.

Concerning the manufactures, the last years aim to construct CT scanners which will provide high image quality and at the same time, to reduce the dose. An important concept was suggested and implemented in 1995: the Automatic Tube Current Modulation (ATCM) during the scan according to the patient’s geometry. The radiographer selects manually a range of tube current-time product (mAs) values which are distributed during the examination according to body geometry. This technique reduces the dose delivered during the examination. However if the selection of mAs values from the beginning takes into account the size of the patient then the dose savings would be even higher. Moreover a limitation of modern scanners is that there is not a wide range of tube voltages to be selected. For example, the tube voltages provided by the Siemens Somatom Plus 4 scanner are 80 kVp, 120 kVp and 140 kVp only. If intermediate values were available, for instance 90kVp 100kVp 110kVp or even 95kVp 105kVp etc. then probably it would be feasible to reduce dose without any compromise on image quality.

A new concept for Computed tomography which is under investigation and it might contribute to dose reduction is the statistical methods of image reconstruction. Researches that have been performed so far show that iterative reconstruction may have a significant impact on Computed Tomography. Applying this method noise and beam hardening artifacts are reduced therefore if we take advantage of this method we might be able to utilize lower exposure settings since the image quality from the outset is better(Elbakri et
al. 2002; De-Man 2005). However this method copes with two major challenges: the computation time (100 X FBP) and to demonstrate its clinical benefit.

In conclusion, the reduction of dose during Computed Tomography examinations is more than probable. However in order to achieve that we must optimize all the concepts considering the excess of dose ought to human factors and the employment of the technological innovations. However, in order to assure the reduction of dose, protocols must be constructed which will individualize the Computed Tomography examinations. That is, the optimum spectrum must be selected relative to the diagnostic task and the size of the patient.
APPENDIX I

Comparison of regression lines

For testing the equality of two slopes the following equation was used to calculate the t values:

\[ t = \frac{b_1 - b_2}{\sqrt{s_{b_1}^2 + s_{b_2}^2}} \]

\( t \) is distributed with \( n_1 + n_2 - 4 \) degrees of freedom, where \( n_i \) is the number of values on each regression line, \( b_i \) is the slope and \( S_{b_i} \) is the standard error for the slope. Indicators 1 and 2 refer for samples 1 and 2. The standard error for the slope was calculated with the following equation:

\[ S_b = b \cdot \sqrt{\frac{1 - r^2}{n}} \]

Where \( r \) is the Pearson’s coefficient and \( n \) is the number of values in the sample.

Reference Dosimetry

Reference dosimetry for CT is based on measurements in a standard CT dosimetry phantom of the computed tomography dose index made for a single slice with a pencil ionisation chamber (CTDI_{100}). and expressed in terms of absorbed dose in air. The appropriate combination of such CTDI measurements made centrally (CTDI_{100,c}) and
Appendix

peripherally at a depth of 1 cm (CTDI$_{100,p}$) can provide an indication of the average dose (mGy) within the slice of the phantom, as given by the weighted CTDI:

\[
CTDI_w = \left(\frac{1}{3} \cdot CTDI_{100,c} + \frac{2}{3} \cdot CTDI_{100,p}\right)
\]

Where CTDI$_{100,p}$ represents an average of measurements at four different locations around the periphery of the phantom

**Limitation of Doses**

<table>
<thead>
<tr>
<th>Annual Dose Limits</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Exposed Workers</strong></td>
</tr>
<tr>
<td><strong>Apprentices and Students</strong></td>
</tr>
<tr>
<td><strong>Members of the public</strong></td>
</tr>
</tbody>
</table>

Appendix II

Siemens Protocols

Part of the Protocols used in the Siemens Somatom Plus 4 scanner. The first separates the patients into age groups while the most recent protocol categorized the patients with their weight.

Table 1. Siemens Protocol based on age

<table>
<thead>
<tr>
<th></th>
<th>Slice</th>
<th>Feed /Rot</th>
<th>Length</th>
<th>Rotation</th>
<th>Time</th>
<th>Time</th>
<th>KV</th>
<th>mA/CTDlw</th>
<th>Kernel</th>
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<tbody>
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<td>7,5</td>
<td>202,5</td>
<td>1,0</td>
<td>29,0</td>
<td>120</td>
<td>50</td>
<td>/ 2.6</td>
<td>CB50</td>
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<td>Chest 1-5</td>
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<td>7,5</td>
<td>202,5</td>
<td>1,0</td>
<td>29,0</td>
<td>120</td>
<td>70</td>
<td>/ 3.6</td>
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<tr>
<td>Chest 5-10</td>
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<td>7,5</td>
<td>202,5</td>
<td>1,0</td>
<td>29,0</td>
<td>120</td>
<td>90</td>
<td>/ 4.7</td>
<td>CB50</td>
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<tr>
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<td>202,5</td>
<td>1,0</td>
<td>29,0</td>
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<td>110</td>
<td>/ 5.7</td>
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<tr>
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<td>29,0</td>
<td>120</td>
<td>50</td>
<td>/ 2.6</td>
<td>CB50</td>
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<td>/ 3.6</td>
<td>CB50</td>
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<tr>
<td>Orbits</td>
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<td>50</td>
<td>/ 2.4</td>
<td>CB50</td>
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</table>
### Table 2. Siemens Protocol based on weight

<table>
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<tr>
<th></th>
<th>kV</th>
<th>mAs</th>
<th>Scan time</th>
<th>Rotation time</th>
<th>Slice collimation</th>
<th>Feed/Rot</th>
<th>Pitch</th>
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<td><strong>Abdo &lt;15Kg</strong></td>
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<td>20</td>
<td>21.3</td>
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<td>2.5</td>
<td>10</td>
<td>4</td>
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<tr>
<td><strong>Abdo 15-24</strong></td>
<td>120</td>
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<td>21.3</td>
<td>0.5</td>
<td>2.5</td>
<td>10</td>
<td>4</td>
</tr>
<tr>
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<td>2.5</td>
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<td>4</td>
</tr>
<tr>
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<td>0.5</td>
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<tr>
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<td>2.5</td>
<td>10</td>
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<td>2.5</td>
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<tr>
<td><strong>Neck &lt;15Kg</strong></td>
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<td>11.3</td>
<td>0.5</td>
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<td><strong>Neck 15-24</strong></td>
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<td>30</td>
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<td><strong>Neck 25-34</strong></td>
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<td><strong>Neck 35-44</strong></td>
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<td><strong>Neck 45-54</strong></td>
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<tr>
<td><strong>Neck above 55</strong></td>
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<td>11.3</td>
<td>0.5</td>
<td>2.5</td>
<td>10</td>
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### Abbreviations

**ATCM**: Automatic Tube Current Modulation  
**CNR**: Contrast to Noise Ratio  
**CTDI**: Computed Tomography Dose Index  
**D**: Absorbed Dose  
**ED**: Effective dose  
**H<sub>T</sub>**: Equivalent dose  
**HU**: Hounsfield Units  
**FOV**: Field Of View  
**MDCT**: Multi Detector Computed Tomography  
**NRPB**: National Radiation Protection Board  
**PIXEL**: Picture Element  
**PMMA**: Polymethyl Methacrylate  
**ROI**: Region Of Interest  
**SD**: Standard Deviation  
**TLD**: Thermoluminence Dosimetry  
**VOXEL**: Volume element
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