Contents

1. Introduction

At the X-ray department at Sahlgrenska University Hospital/SU Östra in Göteborg, Sweden, a MultiDiagnost Eleva FD manufactured by Philips Medical Systems was taken into clinical use at lab 8 in January 2007. It is a direct radiography (DR) system, using a digital flat panel detector (FPD). The most frequently performed procedure at the lab is the double contrast barium enema (DCBE) examination. This optimization study is applied to that examination.

Radiological departments at modern hospitals are today mainly digital and the former standard technic with screen-film systems has continuously been phased out. Today mainly two digital devices for radiography are in use, the computed radiography (CR) and as in this case, the direct radiography (DR) systems.

The DCBE is a colon examination assessing for example polyps and colon cancers. A patient undergoing a DCBE examination is fasting since the beginning of the day before the examination and has taken two doses of a laxative medicine. An empty colon is of great importance. During the DCBE examination a barium solution as well as air is used to achieve contrast, thereby the name of the examination. First the entire colon is filled up with barium solution by pouring the solution through a tube inserted in anus. Thereafter as much solution as possible is drained out again, leaving a thin layer of barium at the inner surface of the colon. The colon is then filled up with air. Both fluoroscopy and exposures are used during the examination, but fluoroscopy is mainly used to see if the contrast material has filled the colon as well as drained out. Almost all images are taken after the air has filled up the colon. The images give a good view of the inner wall of the colon. At SU Östra about 18-25 images are taken at each examination.

Optional examination types to the DCBE examination are coloscopy and CT colography. Coloscopy is more sensitive than the DCBE examination. It gives direct viewing of the colon as well as the possibility to do a biopsy during the examination time. Coloscopy does not include any radiation unlike both CT colography and DCBE examinations. The coloscopy is cumbersome for the patient and risks for puncture of the colon and infections exists. The CT colography is more sensitive than the DCBE examination and possibly even better to detect pathologies than coloscopy is. Another advantage with the CT colography is the possibility to detect other pathologies in the abdomen. However, maybe surprisingly the patients don't consider the DCBE more cumbersome than they do the CT colography [1]. The drawbacks both for DCBE and CT colography are that no biopsies can be taken during the examination and that a radiation dose is given to the patient. An effective dose of 5-10 mSv, are reported for both DCBE examination and CT colography [2], but the variation between sites are considerable.

The DCBE examination is one of the examination types that the Swedish Radiation Protection Authority (SSI) has decided to be subject to diagnostic standard dose (DSD) measurements [3]. The DSD is a standardized way of surveying the radiation dose given to the patients at a given lab by measuring the average dose area product (DAP) given to the patient during the examination. The DSD is compared to a diagnostic reference level (DRL) that aims to be an upper limit of acceptable dose. About 70 % of the radiation dose during the examination is due to exposures and the rest is due to fluoroscopy. The skin dose during the examination is not at risk to cause acute effects and is not given any particular attention during this study. To risk acute effects skin doses of about 2000 mGy is needed [4], during the DCBE examination a skin dose over 100 mGy is rare.

The effective dose to a patient during an examination depends on many different factors. The technical parameters such as tube voltage, tube load, magnification, as well as the number of images taken and the fluoroscopy time. The image quality of an image depends on the same technical parameters but also on the post processing of the image data.

Previous work [5-7] has shown that added copper filtration reduces the effective dose to patients undergoing a DCBE examination. A study done in Scotland [5] from 2004 reports that the patient doses in the UK are steadily decreasing compared to previous surveys. The new X-ray equipments which include low dose options, such as pulsed fluoroscopy, digital imaging facilities and copper filtration, are stated as the most important factor for the decrease.

As new systems or methods are implemented in the daily routine at a radiological department an optimization process is a formal demand. The process has the aim to fulfil the **ALARA**-principle, that all doses due to radiological exposure shall be **A**s **L**ow **A**s **R**easonably **A**chievable. How low that is, is the matter of investigation in the optimization process, where the diagnostic accuracy as well as practical and economical factors should be taken into account [8].

The aims of this work were to

- Use a CDRAD-phantom to survey the parameter settings impact on effective dose and image quality for exposures.
- Use visual grading characteristics to investigate the possibilities to lower the effective dose from exposures to the patient by finding the best of a few different parameter settings while the image quality is high enough to make correct diagnosis.
- Survey the work procedure at the lab and if possible suggest improvements from a radiation hygiene point of view.

2. Material and Method

2.1 The X-ray equipment

The MultiDiagnost Eleva FD[1] is a DR equipment using a FPD, manufactured by Philips Medical Systems. The X-ray equipment is seen in Figure 1. It is a multi-purpose system, intended for several examination types. At SU Östra the system is mostly used for colon examinations as well as hypopharynx and oesophagus examinations. However the system may just as well be used for skeleton or interventional procedures therefore the system can be operated in both under- and over table mode.

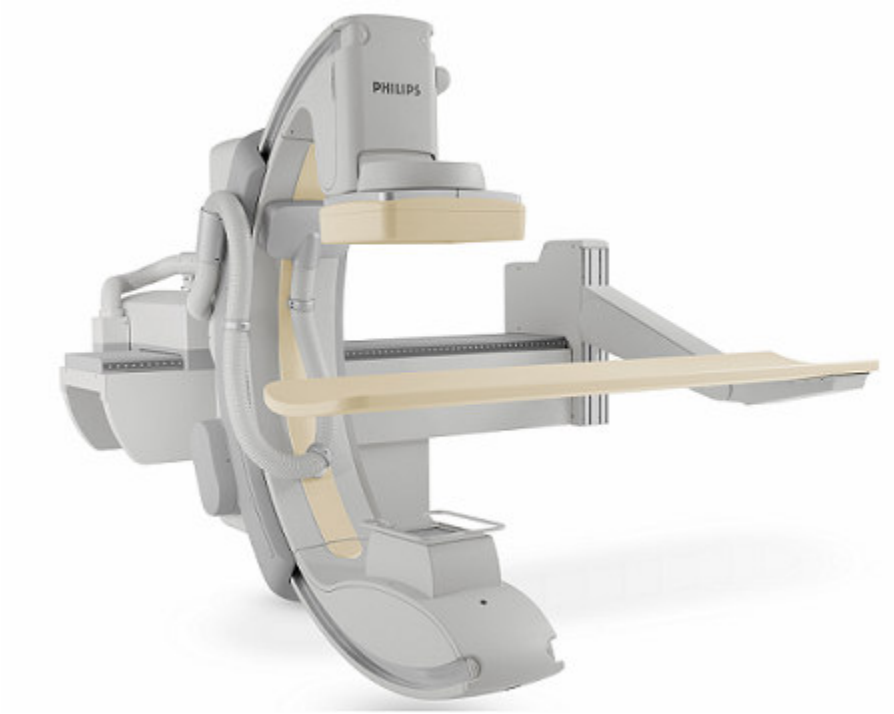

Figure 1: The MultiDiagnost Eleva FD system. Courtesy of Philips Medical Systems.

The FPD uses indirect conversion with a scintillating screen of cesium-iodine (CsI). The scintillating material absorbs X-ray photons and converts them to light. The light is thereafter converted to an electrical signal by means of a photodiode array [9]. The layer of CsI can be kept rather thick due to the possibility to construct it in a needle shaped way. A thicker layer increases the X-ray absorption and the needle shape decreases the lateral light diffusion, both important factors to obtain a good image quality in the end.

[1] X-ray tube housing assembly: SMR 0608 ROT-GS 505, SN: 11121 C 206292, Philips Medical Systems, Hamburg, Germany.
Generator: Philips Velara GFD 2T PEI9890 0000 62091, SN: 064309, Philips Medical Systems, Hamburg, Germany.
X-ray tube: type 989000080201, SN: 206292, Philips Medical Systems, Hamburg, Germany.
Detector: Pixium 4700, SN: 063447, Trixell, Moirans, France.
Grid: N 80/cm, ratio 15, focused on 100 cm, SN: K12816, Smit Röntgen, Eindhoven, Netherlands.
Total inherent filtration, without table and matress: 3.5±10% mm Al at 81 kV, measured using Barracuda, RTI Electronics AB, Mölndal, Sweden.

In the protocol used during the DCBE examination the fluoroscopy mode uses the maximal filtration, i.e. the inherent filtration (3.5 mm Al) and added 1 mm Al + 0.3 mm Cu. For exposures there is no extra filtration used, except for the inherent. During the clinical work the IQX (**I**ntelligent **Q**uality E**X**posure) is activated. The IQX is Philips' dose regulation system for exposures that alter the kilovoltage needed to achieve the detector dose. The IQX has a default value, for each patient size, and corrects the kilovoltage within the first milliseconds of the exposure pulse. The available tube potentials for each exposure are within the interval +25 kV and -15 kV from the default value. The default value for the normal patient size is 90 kV. A time is set (50 ms for the adult DCBE) as the maximum exposure time, if the in-pulse controlling system calculates that the kilovoltage used as a default, will not be enough to reach the detector dose in time, the kilovoltage is increased. If the detector dose will be reached with the default kilovoltage, before the minimum exposure time (10 ms for adult DCBE), the kilovoltage will be lowered.

The detector dose is defined as the air kerma (see 2.4) at the detector plane without the grid and when calibrated it should be measured using 20 mm Al as prefiltering. The prefiltering shapes the X-ray spectrum roughly as a patient, and is used when calibrating since the detector response is spectrum dependent. Throughout this thesis the 100 % detector dose for exposures has a value of 660 nGy, when measured as explained above.

2.2 CDRAD-phantom

For the phantom study the CDRAD[2]-phantom was used [10]. The phantom is a contrast-detail phantom and consists of a square PMMA (poly methyl metacrylate) tablet containing drilled holes of specific diameter and depth. The dimension of the tablet is 256x256 mm and 10 mm thick. The phantom consists of 15 rows and 15 columns that makes up a total of 225 small squares, separated with led-containing paint. A schematic representation of the phantom is shown in Figure 2. The small squares in the top three rows contain one hole in the centre, the other contain one in the centre and one, equal in size and depth, in one of the four corners, randomly chosen. Each row has holes of the same diameter and each column has holes with a constant depth. The widest and deepest hole is at the top right corner and the narrowest and shallowest is at the bottom left corner. The CDRAD phantom as well as the CDRAD Analyser (see below) was supplied by Philips Medical Systems. As the X-ray energy changes, by prefiltering for example, the visibility of low-contrast objects are at greatest risk to be affected. A contrast-detail phantom with low contrast objects has the potential to quantify these changes.

[2] CDRAD 2.0 07030, Artinis Medical Systems B.V.

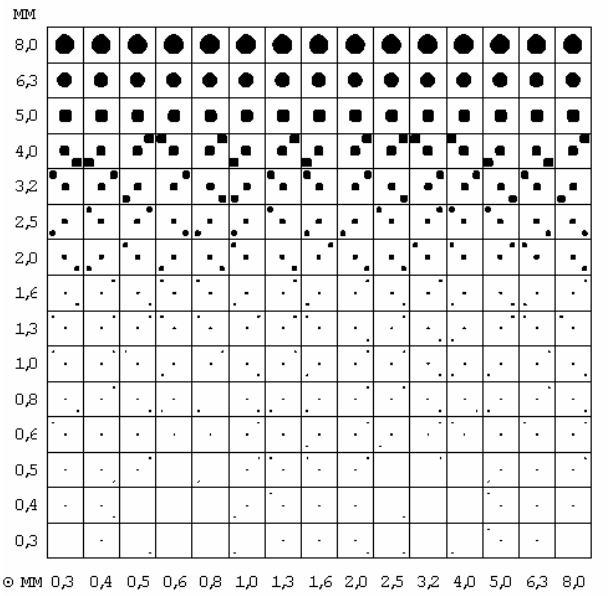

Figure 2: The CDRAD-phantom, presented in a schematic way [11].

CDRAD Analyser

To evaluate the images of the CDRAD phantom the software, CDRAD Analyser[3], was used [12]. The software automatically locates the corners of the phantom or you may manually indicate the four corners in the image. The correct location is of great importance because the software uses that information, to decide where the centres of all the small squares are located. In these small squares the algorithm calculates the average greyscale value and the standard deviation for the signal as well as for the background. The program decides whether or not the spot is "visual" by application of a Welch Satterthwaite test (student t-test with Welch correction) [13]. Every square where the algorithm finds a "visual" spot is marked with a red dot in the contrast-detail diagram, shown in Figure 3.

The contrast-detail curve is fitted according to the dots in the diagram and the hole-depth is plotted against the diameter of the hole, as seen in Figure 3. There are two statistical values that can be set by the user, the Alpha level of significance (Alpha) sets the threshold for the significance level and the a priori difference of mean (APD) was used to compare images stored with different bit-depth. The default value ($1 \cdot 10^{-8}$) for the Alpha was used [14] and (0) for the APD, since all images were taken with the same bit depth.

[3] CDRAD Analyser (version 1.1), Artinis Medical Systems B.V.

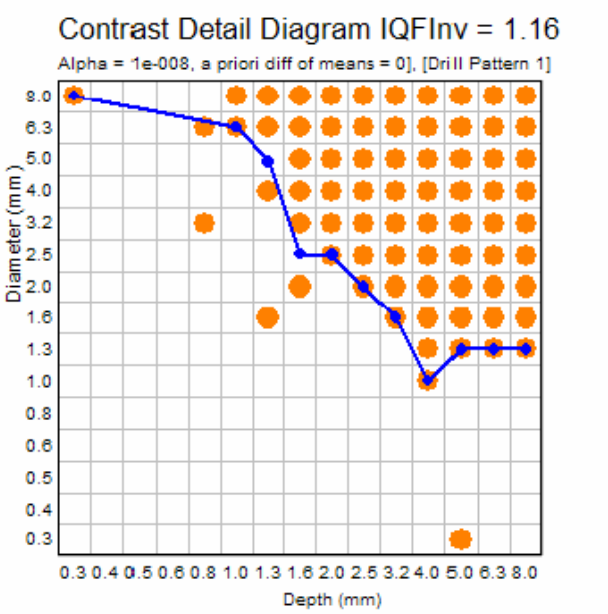

Figure 3: The contrast-detail diagram and curve as shown by the CDRAD Analyser.

The software calculates a score for each image. The score is named the inverse Image Quality Figure (IQF_{inv}), and is calculated according to

$$IQF_{inv} = \frac{100}{\sum_{i=1}^{15} C_i \cdot D_{i,th}}, \qquad (1)$$

where $D_{i,th}$ is the threshold diameter in contrast column C_i. If no spot is visible in a column, $D_{i,th}$ =10.00 mm and if all is visible $D_{i,th}$ =0.3 mm, for a hole depth (contrast) between 0.3 mm to 8.0 mm [12]. The IQF_{inv} ranges from 0.27 to 8.89, and a higher score indicates a better image quality.

The CDRAD Analyser software gives the opportunity to construct a group of several images and a group total IQF_{inv} is given by the software. More than one image, successively taken, of the CDRAD phantom is recommended by the manual, these images are placed in the same group to reduce noise in the result. The group score and the group contrast-detail curve is a fit to the different contrast-detail curves.

2.3 Effective dose calculations

PCXMC 1.5 is a PC based Monte Carlo program, used to calculate absorbed and effective dose to patients in diagnostic X-ray examinations [15]. The program calculates the effective dose according to the formalism given by ICRP 60 [16], as well as the absorbed doses to a number of organs for patients differing in age and size. The phantom in PCXMC is based on the mathematical hermaphrodite phantom of Cristy [17] but a few modifications have been done. Six different ages are available and for these the height and weight are possible to change. Input data for the calculations are:

patient and setup data:	age, height, weight, focus-skin distance (FSD), field size and projection
spectrum data:	kV, anode angle, filter materials and thicknesses
calculation data:	dose area product (DAP), entrance air kerma or mAs

To estimate the effective dose in the phantom study in this thesis, PCXMC was used. The "effective dose" for the phantom study is a comparative value and is not an absolute value. A patient 174 cm high and 110 kg was used in PCXMC simulating a patient 24 cm thick. The thickness was chosen as a mean between the lateral and the PA projection of a patient of normal size. During an examination both the lateral and the PA projection as well as oblique projections are used. To simplify the calculation in PCXMC, the only projection used was the PA projection. A square X-ray field (22.53x22.53 cm at the entrance plane in PA projection) was used, as seen in Figure 4. The same patient setup was used for all combinations of parameter settings.

The simulations throughout this thesis included 50 000 photons per energy level, keeping the statistical error in the effective dose calculations as given by the software down to around 2 %. The anode angle for the X-ray system was 12 degrees. The use of a phantom, not having the true proportions of a real man, insert a systematic error in the method as well as the uncertainty in the organ dose conversion factors, the definition of effective dose and the probability for the different interaction processes for example.

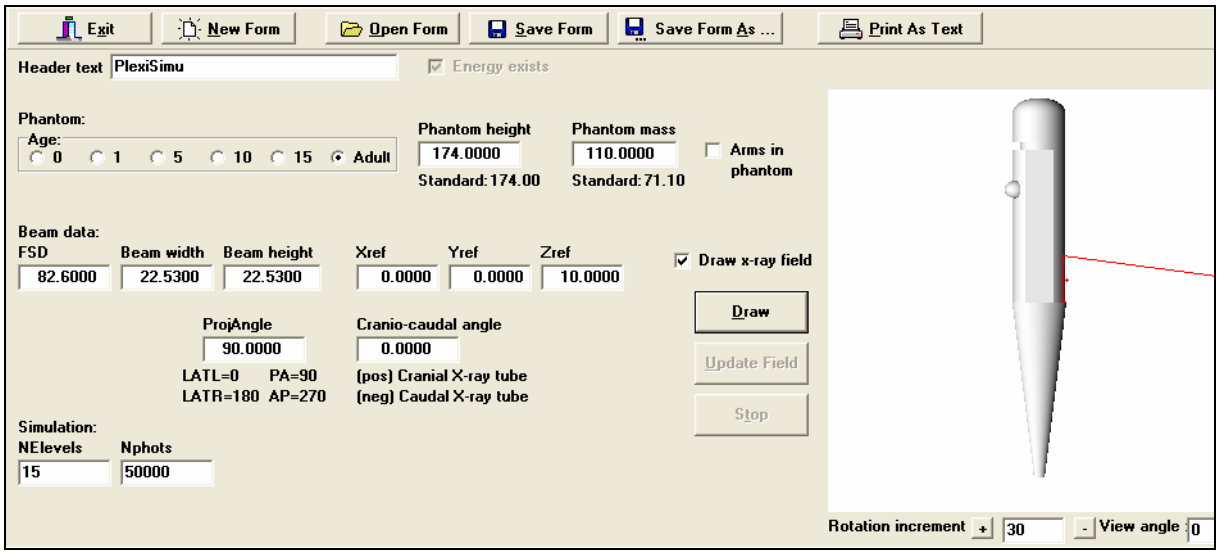

Figure 4: The setup in PCXMC to simulate the effective dose during the phantom study

PCXMC was also used to calculate the effective doses delivered to the patients due to exposures, in the patient study in this thesis. During an examination several different projections are used as well as several different kilovoltages. A reconstruction of this in PCXMC is not possible with the data available. The mean DAP value given by the system for the exposures during the examinations was used and only one projection (PA), the same as for the phantom study, was used for the calculations. The patient used was 174 cm high and weighing 71.10 kg, which was the standard size patient in PCXMC. The image format used was 30x30 cm and the source to image distance (SID) was 110 cm. The tube voltage used for calculations was 98 kV. Even though the calculated values for effective dose is an approximation it gives an indication of the effective doses delivered due to the exposures during the DCBE examination.

2.4 KAP (Kerma Area Product) or DAP (Dose Area Product)

The Kerma Area Product is the integral of the air kerma over the field area at a plane perpendicular to the beam axis. Kerma is short for kinetic energy released per unit mass [18]. In the case of X-ray photons (0-150 keV), kerma is the initially released kinetic energy by photons, transferred to electrons. For field sizes where the heel effect can be neglected the air kerma is approximately constant over the field area and can then be expressed as the product of the air kerma and area as seen in Eq.(2). [19].

$$KAP = \int_A K_a(A) \cdot dA \approx K_a \cdot A \qquad (2)$$

The unit for KAP is Gy·m^2 or more often Gy·cm^2 [20]. For photon energies used in diagnostic X-ray the air kerma is approximately equal to the absorbed dose. This approximation holds well when the X-ray photons mean free path is much longer than the range of the secondary electrons, i.e. charged-particle equilibrium exists [21]. Therefore the KAP is often interchangeably called DAP (Dose Area Product).

The KAP meter

The KAP measurements[4] during the phantom study were performed using a plane-parallel ionisation chamber that fully covered the X-ray field, together with an electrometer. The KAP meter was mounted on the X-ray tube and the KAP value was recorded during all measurements in the phantom study.

To correct the effective dose for attenuation (and scattering) in the table and the mattress the absorbed dose per KAP was measured for different tube potentials. At 10 cm depth of PMMA with and without table and mattress, absorbed dose measurements were made using a R100[5], which is calibrated to show kerma free in air. Thus a factor 1.80 [22] was multiplied by the measured value to get absorbed dose, in this case this resembles the absorbed dose to a tissue equivalent organ at depth 10 cm**.** The correction for the table and mattress ranges from 0.82-0.85 for tube potentials between 73-125 kV. The derived correction term is a linear fit to the experimental data of the quotient of $\left(\frac{D}{KAP}\right)_{with\ table} \Big/ \left(\frac{D}{KAP}\right)_{without\ table}$ taking the tube potential into account. The linear fit is shown in Eq.(3).

$$E_{corr} = E_{PCXMC} \cdot (0.0006 \cdot kV + 0.78) \qquad (3)$$

It should be noted that the assumption made is that the effective dose is expected to vary in the same way as the absorbed dose to an organ at depth 10 cm.

The mathematical DAP meter

Included in the X-ray device, MultiDiagnost Eleva FD, there is a mathematical DAP meter. The mathematical DAP meter uses data from the generator and the jaws positions to calculate the DAP value [20]. The calibration is done without consideration to the table and mattress. The parameters that influence the calculation of DAP are kV, mA, ms, prefiltering and collimated area. During all examinations, the DAP meter records the total DAP for the

[4] Ionization chamber: type 70 157, Fabr.nr. 01 096 , VacuTec Meßtechnik GmbH, Dresden, Germany.
Electrometer: Doseguard 100, s/n1195, RTI Electronics AB, Mölndal, Sweden.

[5] Dosemeter: R100 s/n 1777; RTI Electronics AB, Mölndal, Sweden.
Electrometer: Solidose 400, s/n 471, RTI Electronics AB, Mölndal, Sweden

examination. The recorded DAP values from the integrated DAP meter are used during the DSD measurements (see 2.5) as well as in the patient study.

To correct for the differences in measured values between the mathematical DAP meter and the KAP meter, the DAP meter values were corrected according to the quotients shown in Table 1.

Table 1: The output from the mathematical DAP meter was corrected according to the quotients in the table..

kV	$\left(\frac{KAP}{DAP}\right)_{without\ .0.2mmCu}$	$\left(\frac{KAP}{DAP}\right)_{with\ .0.2mmCu}$
90	0.86	0.93
96	0.85	0.90
102	0.85	0.88

2.5 DSD (Diagnostic Standard Dose)

The measurement of diagnostic standard doses is a formal demand from SSI since 2002 [3] and is a consequence of the European Unions *Directive 97/43/Euratom* [8].

In conventional X-ray, for six different, commonly performed examinations, DSD measurements shall be performed every third year or when a change of equipment or method takes place. One of these six is the DCBE examination. The DAP is used to measure the DSD. The DSD's are compared to a Diagnostic Reference Level (DRL). The DRL is an efficient tool to control and survey the doses delivered in X-ray examinations. If the DSD is higher than the reference level, an investigation shall be performed to find out why and actions shall be taken to lower the doses, if there are no obvious clinical reasons for having a higher DSD. The DRL is not applicable to an individual patient and the DSD measurements shall include at least 20 patients [3]. The patients' weight, are recommended to be in the interval of 50-100 kg. From a plot with the DAP value at the y-axis and the patients weight at the x-axis a linear fit to the data points is made and the interpolated DAP value for a person of 70 kg is the DSD. The present DRL for the DCBE examination is 50 $Gycm^2$.

From the protocol of a DSD measurement it is possible to extract more information than the standard dose for a 70 kg patient. The protocol used during the study included the patient data; age, height and weight as well as the examination data; fluoroscopy time, number of images taken and the operating resident.

2.6 Visual Grading

Visual Grading Analysis

Visual grading analysis (VGA) is a commonly used method to evaluate image quality of clinical images in radiology, in which the reproduction of important anatomical structures are judged by an observer. Two major ways of doing this exist, either the observer uses one or several images as a reference and then others are compared to the reference (relative VGA) or where a reference image is not used and each image is evaluated in terms of an absolute scale (absolute VGA) [23].

Visual Grading Characteristics (VGC)

In a VGC study each image is evaluated by stating the certainty of fulfilment of a few criteria by a multi-step rating scale. The VGC analysis is using statistical methods from receiver operating characteristic (ROC) analysis to evaluate the outcome. VGC can be used to evaluate data from an absolute VGA study. The multi-step rating results in a distribution corresponding to each imaging method tested. The cumulative distributions of the scores from two different imaging methods can be plotted against each other. Using ROC software, the area under the VGC curve can be calculated. An area value of 0.5 means that the two imaging methods are equal [24], if a higher value is obtained the compared method is better then the reference and if smaller the opposite is true.

2.7 ViewDEX

To evaluate image quality in the patient study, a software, ViewDEX (**View**er for **D**igital **E**valuation of **X-ray** images), was used [25,26]. The software shows the images to an observer in a random order and the answers to the criteria are collected in a log file. The log file can then be used for analysis. When an image has been evaluated the observer can not go back and change the answers. Nor is there any possibility to continue to the next image before the current image's criteria is evaluated. During evaluation of an image the observer has the possibility to adjust the window/level as well as use the pan and zoom function.

2.8 The phantom study setup

The aim of the phantom study was to survey the parameter settings impact on effective dose and image quality for exposures of the CDRAD phantom. The study setup intended to resemble the clinical situation as much as possible, therefore the phantom was placed on the table and mattress, with the X-ray device in under table tube mode. The experimental setup is shown in Figure 5. The source to image distance (SID) was 110 cm and the distance between the bucky and phantom surface was 6 cm. The detector format, 30x30 cm, was used, making sure that the relevant part of the phantom was visible in the images. To simulate scattering conditions in a patient the CDRAD phantom was used together with 19 cm of PMMA, making up a total phantom thickness of 20 cm. The CDRAD phantom was placed in the middle of the PMMA tablets, as recommended by the CDRAD manual [10], simulating objects in the body.

Dose calculations for the phantom study were performed using PCXMC, using a patient 24 cm thick. The patient thickness was chosen as a compromise between the lateral and the PA projection of a patient of normal size.

For the effective dose calculations the correct correspondence between PMMA and water is not critical, since the aim of the effective dose calculations is to get a comparative estimate of the effective dose to a patient. The attenuation in 20 cm PMMA and 24 cm of water is almost the same for monoenergetic X-ray photons with the energy 100 keV. For a spectrum with peak energy 100 kVp the situation is a little bit different. Less water is then needed to attenuate the same amount of photons as 20 cm of PMMA. However, if scaling with the

density ($\rho_{PMMA} = 1.190 \frac{g}{cm^3}$ [27]) 20 cm of PMMA corresponds to about 24 cm of water. Such an approach is motivated if the only interaction process is Compton scattering. For energies of diagnostic X-ray the photoelectric effect is slightly more dominant in water than in PMMA.

Figure 5: The experimental setup for the phantom study.

Image collection and analysis

During all exposures the tube voltage was set manually and the automatic exposure control (AEC) was used. Exposures with four different prefiltering settings, included in the system, were performed. Five different detector dose settings were used and for each setting of prefiltering and detector dose, six tube potentials were examined. In Table 2 the different detector doses, prefilters and tube potentials used are shown. All combinations were tested. During the phantom study the 100% detector dose corresponds to the air kerma 660 nGy.

Table 2: The different parameter settings used during the phantom study. All combinations were tested, making up 120 different combinations.

Detector dose	Prefilter	Tube potential
40 %, 60 %	No prefiltering	73, 85
80 %, 100 %	1 mm Al and 0.1 mm Cu	90, 102
120 %	1 mm Al and 0.2 mm Cu	109, 125
	1 mm Al and 0.3 mm Cu	

The phantom study included 120 combinations of parameter settings and for each setting, five consecutive exposures were taken. The images were analysed using the CDRAD Analyser where the five images were grouped together and the group score was used as the IQF_{inv} for that setting. The IQF_{inv} error estimation was calculated as the standard error of the individual scores. For each setting the mean KAP value for the five exposures was used as an input to PCXMC for calculation of the "effective dose" to the phantom. The "effective dose" was corrected for attenuation in the table and mattress (see Eq.(3)).

2.9 The patient study setup

Based on the results from the phantom study the patient study was decided to consist of three different parameter settings aiming to find the best of them in aspects of effective dose while keeping the images' diagnostic value. The following settings were used:

- Setting 1 100% detector dose and no prefiltering (the original)
- Setting 2 100% detector dose and prefiltering with 1 mm Al and 0.2 mm Cu
- Setting 3 80% detector dose and prefiltering with 1 mm Al and 0.2 mm Cu

For each setting, images from 20 patients were collected and from each patient eight projections were used to build up the study material. An experienced radiologist decided which projections to use. The following projections were used:

1. Lateral projection of rectum
2. Left posterior oblique (LPO)
3. Splenic flexure
4. Hepatic flexure
5. Overview of lower part in supine position
6. Overview of lower part in prone position
7. Crosstable (AP) view of lower abdomen, patient in right decubitus position
8. Crosstable (AP) view of upper abdomen, patient in right decubitus position

All eight projections were not present for all patients. The images from the 60 patients were viewed by a resident with help from a senior radiologist and the images considered "not

normal" were excluded. Exclusion criteria were images in which distracting foreign objects existed, images where there were unsatisfactory drain of barium contrast or when the patient had faeces left in the colon.

The effective dose to the patients was calculated using PCXMC. For each setting the mean DAP value due to exposures for one examination with 20 images was used as input to the calculation. The percentage of the total DAP value that were due to exposure varied for the different settings, since no changes were made in the fluoroscopy protocol. To estimate the percentage, examination reports were used. An examination report shows the total DAP separated in fluoroscopy and exposure DAP. From each setting the mean $\frac{DAP_{\exp}}{DAP_{tot}}[\%]$ given by examination reports, were used to calculate the exposure DAP for the patients in the study.

Image quality evaluation

The images were evaluated on a DICOM calibrated 8-bit greyscale monitor with the resolution 2048x1536 pixels. The software used for the evaluation, ViewDEX, shows the images in a random order, unique for each observer and the evaluation is saved in a log-file used for analysis. Each image was evaluated using three different criteria with four possible levels of fulfilment. The criteria were:

1. The image quality is good enough to discover pathological changes of the fine mucosal.
2. The reproduction of the fine mucosal line is good enough.
3. The noise level, in the relevant parts of the colon, is acceptable.

The possible options of fulfilment were:

A. Strongly agree
B. Agree
C. Disagree
D. Strongly disagree

The setup of the study in ViewDEX is shown in Figure 6. The criteria and optional answers were written in Swedish when the radiologists evaluated the images.

Six radiologists evaluated the images, two were experienced radiologists and four were residents.

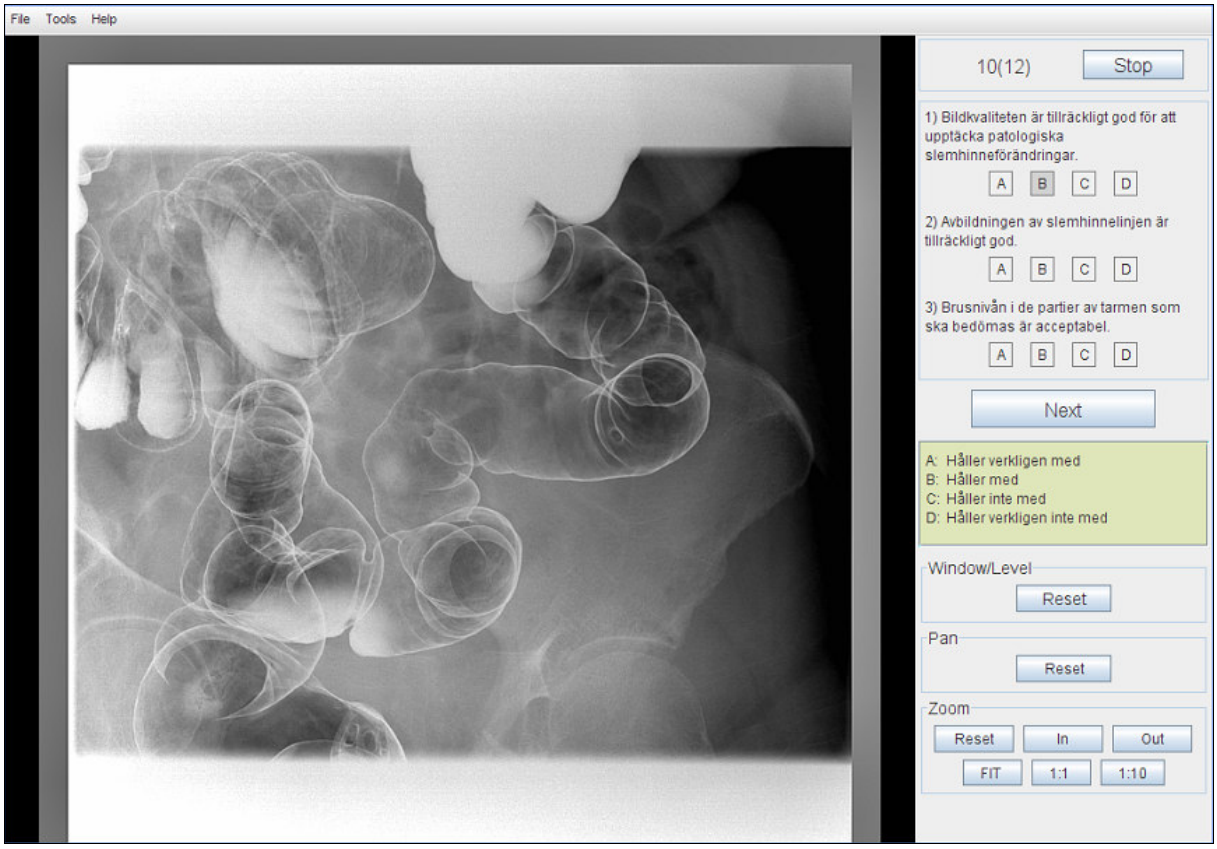

Figure 6: The setup for the evaluation study, as shown by the software ViewDEX. The criteria and possible answers are shown to the right in Swedish (for translation see the text above).

To analyse the outcome of the patient study, a 2AFC (two alternative forced choice) method was used since the ROC software was unable to calculate the area under the curve for the observers. The area under a ROC curve has the same meaning as the outcome in the corresponding 2AFC experiment [28]. A rating scale with four steps was used but the most part of the images for all three settings were rated with the two scores that agreed with the criterion, giving a small spread in the ratings, making it impossible to use the ROC software. The number of images from the compared methods had to be the same to be able to perform the analysis. To simulate the 2AFC, the ratings for the two modalities were randomly compared in pair. The number of times that the compared modality was rated as better, added to the half of the times they were rated as equal, divided to the total number of compared pairs gave the outcome. Setting 1 was set as the reference method that the other two settings were compared to. Each observer was analysed separately for each criterion and setting and the mean from the six observers were taken as the result for that criterion and setting. If two compared methods are considered of equal quality the outcome will be 0.5. The compared method is the better one if a result above 0.5 is obtained and if below the reference method is the better. The standard error (S_E) was calculated from the results from the six observers and the 95 % confidence interval was the result $\pm 1.96 \cdot S_E$.

2.10 The work procedure

The aim for the examination is to leave a thin layer of barium solution at the inner surface of the colon to create contrast and take images that reproduces the entire colon, making it possible to detect pathological changes for example in fine mucosal. The exposures are overviews as well as images of smaller areas of the colon.

At SU Östra it is the residents that perform the DCBE and they evaluate the examination images together with a senior radiologist. During a DCBE examination at SU Östra, 18-25 exposures in different projections are normally taken and 1-5 minutes of fluoroscopy is used. The Swedish Radiation Protection Authority, SSI, have published an example of good radiological practice for patient of normal size [3]. No more than 16 exposures and less than 5 minutes of fluoroscopy are recommended for normal cases. Hence, the examination method used at SU Östra includes a few more images than recommended by the SSI. The combination of relatively inexperienced operators and examinations that are frequently cumbersome to perform are possible reasons to the larger amount of images taken per patient.

Diagnostic standard dose measurements (see 2.5) were made in October and December, before and after the image collection for the patient study, to see how the working procedure at the lab had changed during the period. Both times the measurements were made with the original setting, setting 1. A DSD measurement does not only show the difference in the total DAP value but also how many exposures that are taken per examination as well as the fluoroscopy time used per examination. These data are of major importance when the work procedure is considered. Effective dose was calculated using PCXMC.

The fluoroscopy during an examination is mainly used to see whether the barium contrast has filled and drained out of the colon, for this good image quality is not a great demand. Included in the X-ray system there are a few different fluoroscopy programs possible to choose from. The fluoroscopy program used during the DCBE examination (program I) provide three different settings of frames per second (fps) and three different levels of detector dose per pulse. This gives the operator nine different detector dose settings (image qualities) ranging from 60 nGy/s to 1594 nGy/s, possible to alternate during the examination. The setting with the lowest detector dose is the default and it is normally never changed during the examination. The default setting uses 2 fps and 30 nGy/pulse. As an alternative to the program used in the clinic at the moment another program (program II) was tested clinically during the examination of one patient. Program II used a lower detector dose setting, 22 nGy/pulse as default, but the same frame speed.

3. Results

3.1 The phantom study

The results from the 120 different combinations of parameter settings from the phantom study are all shown in Figure 7. The further to the top left corner of the plot, the lower the dose and the better the image quality. The line connects the six points with the original setting, 100% detector dose and no prefiltering. In each series the six points correspond to different kilovoltages, normally the lowest kilovoltage is at the top right corner and the higher the kilovoltage the further to the lower left corner.

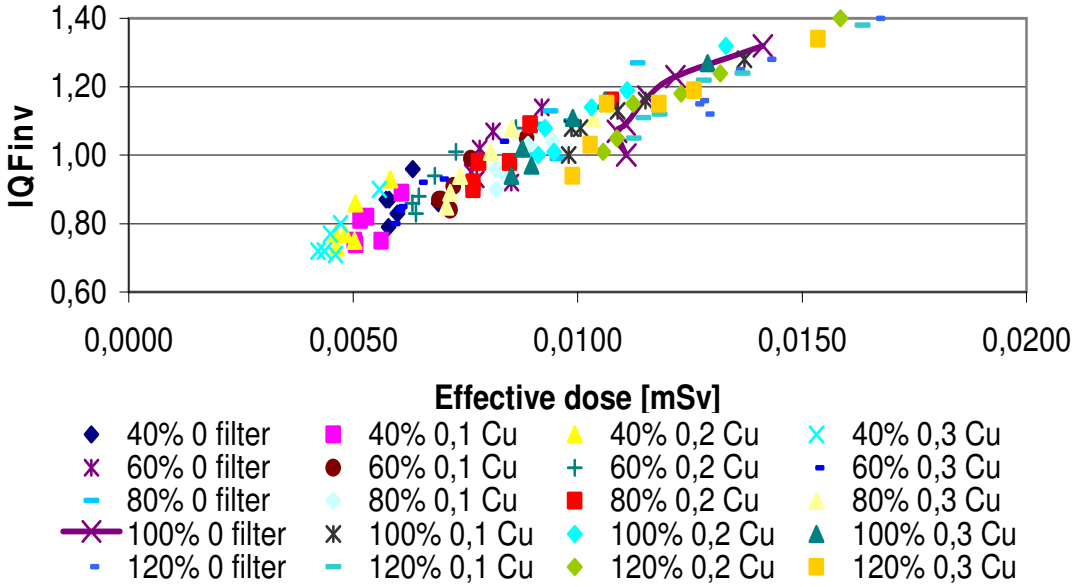

Figure 7: The results from the phantom study. All 120 data points divided in series where the prefiltering and detector dose is constant. The line connects the points for the original setting, no prefiltering and 100% detector dose. (When prefiltering (mm Cu) is added 1 mm Al is also included.)

When prefilter was added both the effective dose and the IQF_{inv} was affected. The effective dose as well as the $IQF_{inv,}$ decreased when prefilter was added, as shown in Figure 8 for the case of 100% detector dose.

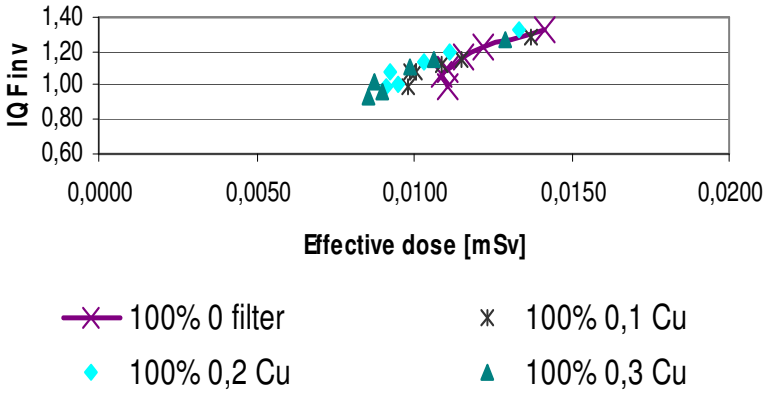

Figure 8: The results from the phantom study showing the effect of adding prefiltering at 100 % detector dose. Both the effective dose and the IQF_{inv} decreases when prefilter are added. The line connects the points for the original setting. (When prefiltering (mm Cu) is added 1 mm Al is also included.)

The detector dose setting influenced both the image quality and the effective dose. When the detector dose was increased the IQF_{inv} as well as the effective dose increased and the opposite when decreased, as shown in Figure 9.

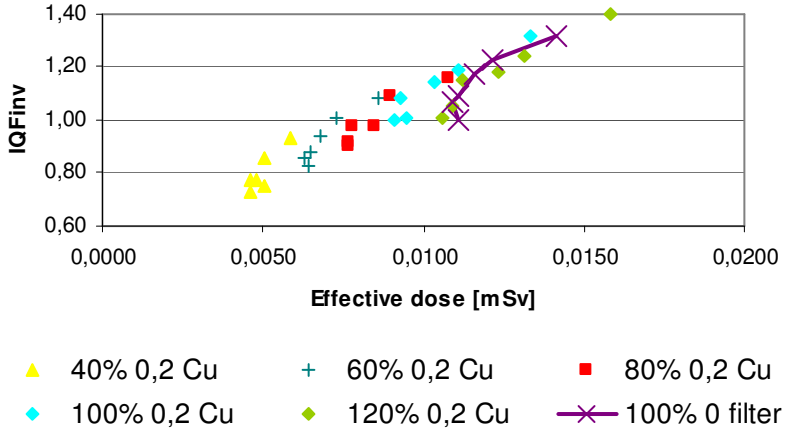

Figure 9: The results from the phantom study showing the effect of changing the detector dose setting. When the detector dose increases the IQF_{inv} as well as the effective dose increases. The opposite is also true. The line connects the points for the original setting. (When prefiltering (mm Cu) is added 1 mm Al is also included.)

Three settings from the phantom study were decided to be tested in a patient study. The three settings, 100% detector dose and no prefiltering (the original); 100% detector dose and prefiltering with 1 mm Al and 0.2 mm Cu; 80% detector dose and prefiltering with 1 mm Al and 0.2 mm Cu, are shown in Figure 10. By adding 1 mm Al and 0.2 mm Cu as prefiltering the effective dose was lowered between 6-18% with small impacts on the IQF_{inv}. When the detector dose was lowered to 80 % as well as prefilter added, a decrease in effective dose of 24-31 % was seen. In this plot ± 1 standard error in IQF_{inv} is included for the original setting.

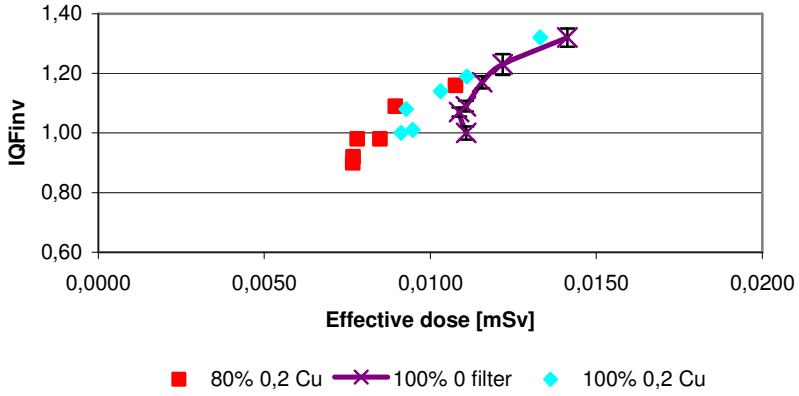

Figure 10: The parameter settings chosen from the phantom study to be included in the patient study. The line connects the data points for the original setting, the error bars are ± 1 standard error. (When prefiltering (mm Cu) is added 1 mm Al is also included.)

3.2 The patient study

Image quality analysis

A total of 415 images were included in the observer study, 137 images from setting 1, 134 images from setting 2 and 144 images from setting 3. To be able to perform the 2AFC analysis the number of images from each setting had to be equal. To get groups with the same number of images, three images from setting 1 and ten images from setting 3 were excluded prior to the analysis. The ones excluded were randomly chosen. A total of 402 images were included in the analysis, 134 images from each setting.

The results from the patient study are shown in Figure 11. Each pair of bars show the result for one criterion. Setting 1 was set as a reference and a value of 0.5 indicate equal fulfilment of the criterion for the reference and compared setting. If lower than 0.5 the reference setting was better and if higher, the compared setting was better. With a 95 % confidence interval the fulfilment of the criteria was not considered worse, when comparing setting 2 and 3 with the reference. For criterion 3 "*The noise level, in the relevant parts of the colon, is acceptable*" the reference setting seems to be considered slightly better compared to setting 2 and 3, but not with statistical significance.

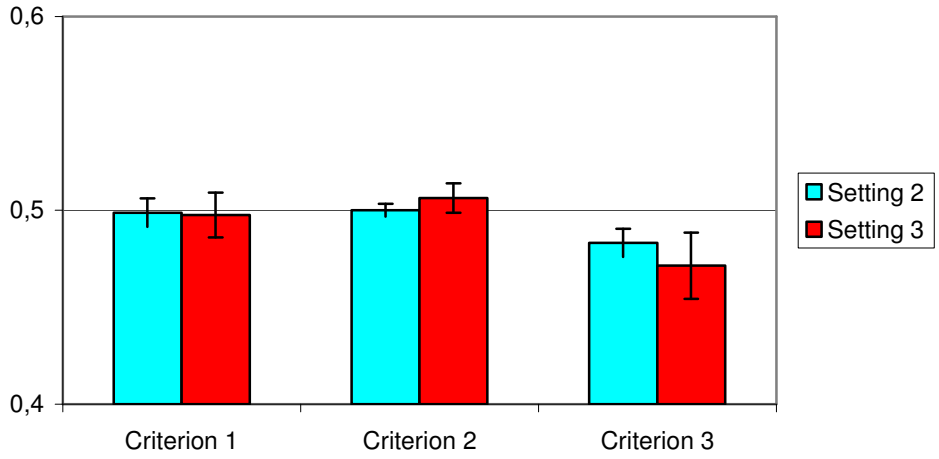

Figure 11: The results from the patient study for criterion 1-3. Both setting 2 and 3 are compared to the reference setting. Error bars are ± 1 standard error. ***Criterion 1****: The image quality is good enough to discover pathological changes of the fine mucosal;* ***Criterion 2****: The reproduction of the fine mucosal line is good enough;* ***Criterion 3****: The noise level, in the relevant part of the colon, is acceptable.*

Effective dose calculations

The average percentage of the total DAP value that were due to exposure during the patient study, was 80 % for setting 1, 70 % for setting 2 and 60 % for setting 3, rounded off to closest multiple of ten.

The effective dose to the patients was calculated using the DAP value for 20 exposures, DAP_{exp}, per patient, for each of the three settings included in the patient study. The DAP_{exp} and corresponding effective doses for the three settings are shown in Table 3 together with the reductions compared to setting 1.

Table 3: The DAP values for 20 exposures, DAP_{exp} and the corresponding effective doses, E_{exp}, for the tree different settings used in the patient study. The reductions compared to setting 1 are also shown.

	DAP_{exp} $(Gy \cdot cm^2)$	**Δ DAP**	**E_{exp}** (mGy)	**Δ E**
Reference **100 %; 0 filter**	10	0 %	1.6	0 %
Setting 2 **100 %; 0.2 mm Cu**	5.6	- 44 %	1.4	- 15 %
Setting 3 **80 %; 0.2 mm Cu**	4.3	- 57 %	1.1	- 34 %

3.3 The work procedure

Figure 12 shows the data points and their linear fit from the DSD measurement done in December. The equation in the upper right corner is the equation for the linear fit in the figure, y=DSD and x=weight in kg. The upper (red) thick line indicates the diagnostic reference level ($50\,Gy \cdot cm^2$) set by the Swedish Radiation Protection Authority (SSI) and the lower (green) thick line shows the diagnostic standard dose level for a 70 kg patient ($13.4\,Gy \cdot cm^2$) at SU Östra in December 2007.

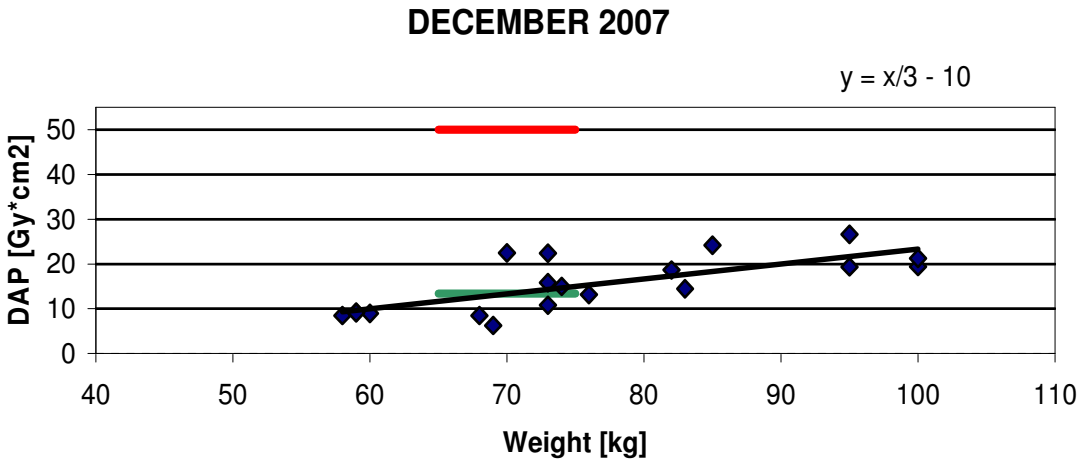

Figure 12: The DSD measurement in December 2007 at SU Östra. The upper (red) thick line marks the diagnostic reference level (50 Gy·cm^2) and the lower (green) line marks the standard dose, (13.4 Gy·cm^2) for a 70 kg patient.

Table 4 shows the average number of images taken per patient, the mean fluoroscopy time, the diagnostic standard dose and the effective dose from the two DSD measurements performed in October and December 2007. The fluoroscopy time decreased with almost 50 % and the number of images decreased as well, during the period. A total decrease in DSD and effective dose of 32 % without any change in settings was seen during the period.

Table 4: The differences between the two standard dose measurements for the DCBE examination done in October and December 2007 at SU Östra.

	December	**October**
Mean number of images/patient	20	26.5
Mean fluoroscopy time/patient [s]	116	227
DSD [$Gy \cdot cm^2$]	13.4	19.7
Effective dose [mSv]	2.1	3.1

For the fluoroscopy program II, tested during one examination, the operator performing the examination found the image quality during fluoroscopy satisfactory.

4 Discussion

4.1 The phantom study

In previous work [29, 30] the CDRAD phantom has been used with satisfactory results to assess image quality prior to a clinical study or to compare different detectors. The CDRAD phantom together with the CDRAD Analyser has been subject to investigation in homogenous backgrounds and the scoring has been compared to human observers [14]. The results indicate that relative variations in image quality can be identified, but the human observer and the software does not mimic each other. They also conclude that intra-software variability is insignificant, meaning that the software nearly always gives the same output for the same image. The intra-sample variability on the other hand is considerable, meaning that the software does not give the same result for successive images taken with the same parameter settings. The precision of the method is strongly dependent on the number of images used.

Experiences similar to those mentioned above have been apparent in this study as well. The intra-sample variability is a draw back of the method and decreases its precision, as well as it increases the time consumption when using a greater number of images.

The results from the phantom study showed degradation in image quality when prefilters were added, but the impacts were small (see Fig 9). The “effective dose” to the phantom decreased when prefiltering was added, as expected. Changes in detector dose substantially affected both the effective dose and the image quality (see Fig 10), making it possible to get rather large decreases in effective dose, but at the expense of worse image quality.

4.2 The patient study

From the results in the phantom study two new settings (prefiltering with 1 mm Al + 0.2 mm Cu (setting 2) and prefiltering with 1 mm Al + 0.2 mm Cu and 80% detector dose (setting 3)) along with the original was decided to be tested in a clinical study with patients (see Fig 11). By adding prefiltering (setting 2) the effective dose were lowered between 6-18% with small impacts on the IQF_{inv}. The results from the phantom study implied a slightly better combination of image quality versus effective dose for the 0.2 mm Cu compared to 0.3 mm Cu that have been used in previous work [6,7]. When the detector dose is lowered to 80 %

and prefilter is added (setting 3) the IQF_{inv} as well as the effective dose decreases. A decrease in effective dose to the phantom of 24-31 % was seen with the last setting.

The reason not to go even further down than 80 % in detector dose was simply not to jeopardize the diagnostic certainty. The procedure at the examination is not easy and if an examination has to be repeated the effective dose to the patient will increase. Another method to address the image quality is to take two images of the same patient with different settings in the same projection immediately after each other and then compare them, as was done in a paediatric study [7] in Sweden. That approach would require an approval from an ethical committee since it include extra radiation for the patient. Instead of lowering the detector dose further towards the diagnostic limit an optional approach is to simulate a lower detector dose by adding noise [31]. The method uses a noise image that is added to the image taken at normal settings and it has been used in previous studies [32,33].

The lack of quality criteria for the DCBE in the European guidelines [34] made it more difficult than expected to find suitable criteria to evaluate the images in the patient study. No normal, none-colon anatomy is visible in all images possible to have as criteria. With a 95 % confidence interval, the results for all criteria did not show any significant impact on image quality for the studied settings. However, as was expected the result for criterion 3, considering noise, indicates a trend towards noisier images when the radiation dose decrease. The lack of significant differences between the compared settings for criteria 1 and 2 may depend on several things. Firstly, one may expect that the study was performed at detector-dose levels too high to reveal differences and seccondly that the criteria were too easy to detect. This implies that if the used criteria are suitable, a further decrease in effective dose is possible.

Even though the phantom study showed degradation in image quality when the detector dose was lowered the clinical images did not seem to loose diagnostic value. The usage of a contrast-detail phantom can be strongly questionable during optimisation studies in X-ray imaging [35] since it is often the anatomy itself rather than the quantum noise that has the greatest impact on image quality. A probable reason why the previously visible degradation in image quality during the phantom study, was no longer visible in the patient study, is that the exposures during DCBE are “quantum saturated”.

In the phantom study a decrease of 6-18 % in effective dose between setting 1 and 2 could be seen and a decrease of 24-31 % between setting 1 and 3. The result from the patient study showed decreases of 15 % between setting 1 and 2 and 34 % between setting 1 and 3. The results from the phantom and patient study agree well to each other. The decrease in effective dose was evidently smaller than the ones in DAP. The decrease in DAP between setting 1 and 2 in the patient study was 44 % while the effective dose decrease was 15 %. Thus it is important not to mix them up. The results from the patient study agree well with a review of effective doses and DAP values for patients done in Scotland [5]. It states that increasing the prefiltering with 0.2 mm Cu, using a X-ray beam of 80-100 kVp decreases the DAP value with 44-50 % but the effective dose with “only” 12-22 %, corresponding to the decrease in this study between setting 1 and 2. Other studies with 0.3 mm Cu as prefiltering have also shown decreases in effective dose, without substantial loss of image quality. For example a DCBE study in England [6] estimated a decrease of 11% in effective dose when 0.3 mm Cu

was added in the examination protocol for adults. A paediatric study [7] in Sweden found a decrease of 44% in effective dose with added filtration of 0.3 mm Cu.

No information about patient sizes was gathered during the image collection and the total DAP value was not available for all 60 patients. The mean DAP value was calculated for 12, 16 and 19 patients for setting 1, 2 and 3 respectively. If DSD measurements were done for all 20 patients per setting, the effective dose calculations could have been done for the interpolated 70 kg patient. Instead an estimate of the effective dose to the patients was done in the patient study, with the mean DAP as input data to PCXMC. Corrections for attenuation and scattering in the table and mattress were made as well as correction for the differing output between the mathematical DAP meter and the plane-parallel KAP meter.

In conventional X-ray examinations a conversion factor, $E_{DAP,}$ may be multiplied with the dose area product, DAP, according to Eq.(4) to estimate the effective dose to patients'

$$E = E_{DAP} \cdot DAP, \qquad (4)$$

$E_{DAP} = 0.28 mSv \cdot Gy^{-1} \cdot cm^{-2}$ for colon examinations [36]. Since the kerma (or more precisely μ_{tr}) is very dependent on the photon energy at diagnostic X-ray energies, the E_{DAP} value above is not directly applicable because it does not take prefiltration into account for example. Therefore it is not a very good way to compare the effective doses to the patients in this study. Calculating the effective dose with this E_{DAP} value would only show the change in the DAP value for the examination.

A decrease of 20 % for the detector dose gives the same amount of change in effective dose, if the same tube potential is used. During the effective dose calculations throughout this study the same tube potential was assumed. However, in the clinic, when the IQX is used that is not necessarily true (more about the IQX see 2.1.). When the detector dose setting is lowered, keeping the same maximum exposure time (i.e. 50 ms), the kV used by the system may decrease.

4.3 The work procedure

At Sahlgrenska University Hospital, lab 8 at SU Östra is the only lab currently performing DCBE examination. The operators that perform the examinations are residents doing their residency within the radiological field. At various times, new residents are introduced to perform the examination as a part of their training to become radiologists. The DSD may be hugely influenced by this rather rapid change of operators.

In October a DSD measurement (see 2.5) was done for the DCBE examination and a second one was done in December. Both of them were made with the original setting of examination parameters. During this period of time a decrease of 32 % in DAP was seen. A change in attitude to and awareness of the dose-area product was noticed among the residents during this period, probably due to the study within this thesis. The residents performing the examinations started an informal competition, in which everyone tried to get the lowest DAP value for an examination, still producing examinations that were diagnostic. Competing may not be a recommendation, a low DAP value is not the most important achievement during an examination but for this specific case it gave the residents feedback about the radiation dose.

For a DCBE examination at SU Östra the contribution to the DAP value due to fluoroscopy is in average 20% of the total DAP. Since exposures contribute the most to the total DAP as well as the effective dose to the patient, the patient study as well as the phantom study dealt with exposures. As the DAP value is reduced for the exposure part the percentage of the total DAP from fluoroscopy becomes greater (up to 30-40% of the total DAP).

The reference level for the DSD measurement was implemented in Sweden in 2002 [3]. It is likely to believe that the radiation dose levels have decreased since then in Sweden, as reported for example from Scotland [5]. A decrease of the reference level (50 $Gy \cdot cm^2$) for the DCBE examination is probable.

During a clinical test, fluoroscopy program II was tested. The frame speed 2 fps was the same as before but the detector dose per pulse was lowered. When program II is chosen, the kV-mA-ms curve is also changed and moved to higher tube potentials. This leads to that for a given patient and detector dose the stabilized tube potential would be about 15 % higher than with the original setting. But since the detector doses are not equal for the two settings, changes in both kilovoltage and mAs is probable to achieve the lower detector dose. Whether it is the tube potential or the mAs that increases has influence on the effective dose. A 25 % decrease in detector dose per pulse will not necessarily correspond to a 25 % decrease in effective dose, if the tube potential is changed. There were no effective dose calculations done during this study. A more thorough study would be needed to calculate an accurate decrease in effective dose. During the clinical test, program II was only tested for one examination and one operator and the result was only based on the operator's momentary experience. If the fluoroscopy program is changed to program II, the possibility to obtain better image quality during the examination still exists. For example, the operator may change the detector dose per pulse, thereby obtaining a higher detector dose per pulse.

6. Conclusions

In this work, the result from two studies of the DCBE examination is presented. A phantom study, using the CDRAD phantom, was used to survey the impact of the system settings on image quality and radiation dose. For three different system settings a patient study was performed, when clinical images was graded using VGC. The studies showed that adding 1mm Al and 0.2 mm Cu as filtration and decreasing the detector dose setting to 80 % of its current value, will decrease the effective dose to the patient. The image quality was not found to be negatively affected. For the case of homogenous background, as in the phantom study, degradation could be seen. A state of "quantum saturation" [35] for the examination is a probable explanation to the divergent results. The clinic is recommended to change the settings for the DCBE examination protocol.

The results from this study indicate a possible reduction of 34 % in effective dose from exposures, using parameter setting 3 (80 % detector dose and prefiltering with 1 mm Al and 0.2 mm Cu). An awareness of the dose-area product and a change in attitude among the residents performing the DCBE examination were noticed during the time of this study. DSD measurements were made before and after the patient study with the same parameter settings, a decrease of 32 % in effective dose was seen during this period of three months. Furthermore, a low-dose fluoroscopy program, that might decrease the contribution to effective dose with approximately 20 %, was briefly tested. By combining the operator improvements, the proposed change in exposure settings and the tested low-dose fluoroscopy program, a total decrease of about 40 % in effective dose may be obtained.

Continuous work to see how much further down in detector dose it is possible to go and still obtain diagnostic images would be interesting. With a method using simulated detector dose reduction [31] it might be possible to find the lowest limit without any risk for the patient. Other possible future studies are to investigate the image processing as well as investigate the fluoroscopy settings further. Furthermore, extending the work to other procedures at the lab would be of interest.

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