Consequences of the Patient's Mis-centering on the Radiation Dose and Image Quality in CT Imaging – Phantom and Clinical Study

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Abstract The position of the patient highly influences the functioning of the ATCM in CT imaging. The effect of different mis-centerings on the CTDIvol dose was determined during PMMA and water phantom simulations, which were also used for noise assessment. The results show that a 50 mm mis-centering (with the phantom placed above the isocenter) can cause an increase of the CTDIvol by 47 % associated with the lower standard deviation of the HU signal, whilst a -50 mm mis-centering (with the phantom placed below the isocenter) leads to a decrease of the CTDIvol by 35% and is associated with a noisier image. A mean value of the mis-centering determined for 473 patients was -43 mm. A total of 470 of 473 patients were mis-centered below the X-ray tube, which shows an inclination of radiographers to place patients below the isocenter. This inclination was more significant for smaller patients.

Keywords CT, CTDIvol, Mis-centering

1. Introduction

Since its introduction in 1971, the computed tomography (CT) has developed from an X-ray modality with limited use to a widely used modality in medical imaging, mainly owing to its ability to visualize low-contrast structures. As regards the dose of the ionizing radiation, CT imaging is a high dose modality with an important impact on the overall population dose. According to the report of UNSCEAR [1], the annual per caput effective dose to the global population due to all sources of ionizing radiation is 3.1 mSv on average. Diagnostic medical radiology accounts for 0.62 mSv, which equals 20 % of the annual effective dose. The CT examinations contribute by approximately 47 % to the 0.62 mSv, although CT exams account only for 7.9 % of the total

number of diagnostic medical examinations in developed countries [1]. Due to the constantly increasing total effective dose from CT examinations [1], every step towards of lowering radiation doses is desirable.

During the last decade, many advances in the area of CT imaging led to a reduction of CT doses. The most important strategies are the automatic tube current modulation (ATCM) – angular modulation and longitudinal modulation, adjusting voltage based on patient attenuation, dynamically adjustable Z-axis beam collimation, and iterative reconstruction [2,3]. All these techniques can lower the dose to patients.

The X-ray spectrum used in CT imaging is different from the X-ray spectrum used in conventional radiography. The spectrum is much more filtered in order to reduce beam hardening. The total filtration includes an inherent filtration and an added filtration.

Figure 1. CT geometry of the scanned object and the bow-tie filter when the scanned object is well-centered (left) and mis-centered (right)

The total filtration is equivalent to 5-6 mm of aluminum [4]. The added filtration consists of a flat filter and a bow-tie filter. The flat filter ensures that softer X-ray photons, contributing to a higher radiation burden of patients without any benefit for the CT image are filtered out. The bow-tie filter shapes the flux of the X-ray beam. The flux of the X-ray beam incident upon a detector, after it has passed through the patient, should be as homogenous as possible.

The bow-tie filter attenuates the X-ray beam more on the peripheral sides of the field of view than in the central part. Therefore, the attenuation of the patient body is complementary to the attenuation of the bow-tie filter. A typical CT geometry with the static bow-tie filter is illustrated in Figure 1, showing a well-centered phantom and a mis-centered phantom. The bow-tie filter is usually static, but a study using the dynamic bow-tie filter has also been published [5].

The X-ray beam intensity is controlled by the ATCM (Care Dose4D, Siemens) which allows the scanner to modify the X-ray tube current in real time, in accordance with the attenuation in each projection [6]. The tube current modulation is performed in two ways. Firstly, the current is modified on the basis of a low dose scan projection radiograph (topogram), by comparing the actual patient attenuation to the attenuation of a standard-sized patient [7,8]. When only the AP topogram is required (AP or lateral), the perpendicular view is estimated using a mathematical model [9]. Secondly, the angular attenuation is measured and the current is adjusted according to the regions and angulations. Data from the online attenuation profile (measured in the previous tube rotation) is processed and sent to the generator control for the tube current modulation with a delay of 180° [9,10].

The Care Dose4D adapts the tube current automatically for a selected level of image noise (defined by the user through the parameter of ref. mAs). When patients are not placed correctly (mainly in the case of small patients), the ATCM can result in excessive dose reduction, leading to the deterioration of the image quality by increasing noise, or in an increase of the dose whilst keeping the preferred level of noise [10]. Therefore, the effective use of the ATCM is highly dependent on the patient's position, i.e. on their correct placing by the radiographer.

The first part of the study, performed on phantoms, deals with the influence of the phantom's mis-centering on the radiation dose and on the image quality. The second part of the study shows the real values of the patient's mis-centering. An assessment of the mis-centering for 473 patients is included.

2. Materials and Methods

A CT scanner Somatom Definition Flash (Siemens) in the mode with a single source was used for phantom simulations and also for patients' scans. All the simulations and scans were performed with 120 kVp, the ATCM switched on (CARE Dose4D) and in the abdomen mode. The setup of the abdomen mode was: rotation time 0.5 s, total collimation 38.4 mm, single slice collimation 0.6 mm, pitch factor 1.15, reconstructed slice thickness 5 mm. No iterative reconstruction was used. Each scan was preceded by a topogram in the antero-posterior (AP) projection.

2.1. Phantom Simulations

Simulations were performed on polymethyl-methacrylate (PMMA) and water phantoms. The PMMA phantom consists of 36 PMMA plates with a dimension of 30 cm x 30 cm x 1 cm. Three patient sizes were simulated by the PMMA plates. Small patients were simulated by 15 PMMA plates (phantom dimensions (X, Y, Z): 15 cm x 30 cm x 30 cm), medium patients were simulated by 23 plates (23 cm x 30 cm x 30 cm) and large patients were simulated by 30 plates horizontally and 3 plates vertically on both sides (30 cm x 36 cm x 30 cm). Due to the cubical shape of all PMMA phantoms, reconstruction artifacts were observed. Although a PMMA phantom with a circular cross-section would be more suitable, it was not available. Therefore, the second group of simulations was carried out on two cylindrical water phantoms, the first one with a diameter of 25 cm, the second one with a diameter of 40 cm. The second water phantom was more of an elliptical rather than cylindrical shape.

For the radiation dose assessment, only the CTDIvol [11] values, representing the output of the scanner, from the exam protocols were collected, a real measurement of CTDI was not performed.

The simulations with the PMMA phantoms were performed starting from the highest to the lowest position of the table, with an increment of 2 cm. The distance between the highest and the lowest position was 19.5 cm (the highest and lowest vertical position of the table enabled by the scanner). When the central plate of the phantom was positioned in the isocenter, the CTDIvol value was considered as a reference value of CTDIvol. For the small PMMA phantom, where the placement of the central plate in the isocenter was impossible (the patient table could not go higher), CTDIvol for the position closest to the isocenter was considered as a reference value. Other CTDIvol values were taken relatively to this reference value. The simulations were similar for all three sizes of the PMMA phantoms.

Three simulations with the water phantoms were performed for only three setups of the table: with the table in the highest vertical position, in the middle vertical position and in the lowest vertical position.

The position of the central plate of the PMMA phantom was determined from the antero-posterior (AP) diameter of the phantom (mm) and the DICOM tag (0018,1130) called "table height". This tag represents the distance from the top of the patient table to the center of the rotation in mm [12]. This parameter is henceforward referred to as "table vertical position". The position in relation to the isocenter was determined from the following formula:

$$Position = \frac{AP\ diameter}{2} - table\ height \quad (1)$$

Figure 2. The CT gantry with scanned objects well-centered (first), mis-centered in a negative direction (second) and mis-centered in a positive direction (third) When the parameter “position” gained from the formula (1) had a negative value, the central plate of the phantom was placed below the CT isocenter in the AP direction, whereas when the resulting position had a positive value, the central plate of the phantom was placed above the isocenter, as illustrated in Figure 2. Ideally the position should be equal to zero.

Figure 3. Relative CTDIvol values for different PMMA and water phantom simulations for different vertical positions of the patient table (The negative value of the position means that the phantom is placed below the isocenter.)

2.2. Patients' Data

473 consecutive patients who underwent the CT abdomen or abdomen+pelvis scan in our institute were included in this retrospective study. CT scans were performed by 6 radiographers in periods 03-05/2014 and 09-11/2014. Three of the radiographers usually work in the CT department on a regular basis, the others work there only during night shifts. All but one have more than 10 years training and experience.

The mis-centering of patients was determined from the patients' images in the same way as for the phantoms, using the formula (1), for all the included patients. The AP diameter was measured in the axial slice in which the iliac crest was first evident, when going through the reconstructed images in the head-feet direction.

3. Results

3.1. Results from Phantom Simulations

The relative CTDIvol values for all the phantom simulations (for PMMA and water phantoms) are shown in Figure 3. The relative value of CTDIvol is equal to the ratio of the CTDIvol value for a given table position to the reference CTDIvol value (the table position with the central part of the phantom placed in the isocenter of the CT scanner).

CTDIvol values are increasing when the phantom is mis-centered in the positive direction (positive mis-centering), e.g. when the centre of the phantom is placed above the isocenter, closer to the X-ray tube (the table is in a higher vertical position), when the phantom 'seems' to be larger for the X-ray tube. When the phantom was placed below the isocenter, the CTDIvol was smaller than the reference CTDIvol value. Due to the positive mis-centering the CTDIvol increased by up to 47 % for the largest phantom. For the smaller phantom, the increase of CTDIvol was between 10 and 30 %.

Figure 4. Signal values with standard deviations for PMMA and water phantoms (The negative value of the position means that the phantom is placed below the isocenter.)

Figure 5. The dependence of the patients' mis-centering on their AP diameter

When the phantom was mis-centered in the negative direction (negative mis-centering), the CTDIvol decreased by up to 60 %. In such cases, the patient's dose would not be increased, but the expected resulting image would be noisier.

The image noise was evaluated in standard deviation (SD) of the signal. The signal in Hounsfield Units (HU) was determined for all the phantoms for each of the table vertical positions. The values are shown in Figure 4.

The SD of HU for the table position with the phantom placed below the isocenter (negative mis-centering) was much higher than the SD of the signal for the phantom placed above the isocenter (positive mis-centering). According to the relative CTDIvol values, the CTDIvol for the positive position is higher and the image is less noisy. This result is in accordance with the results of the study Kalra [13].

The average HU value increases with the increasing negative mis-centering, as it is evident from Figure 4. The signal for the PMMA phantoms increases from around 135 HU to 155 HU for the negative mis-centering of 140 mm and more. A change in HU was not proved by the study Kalra

[13]. It can be explained by the fact that in the study [13], the signal of HU was determined for a mis-centering of only up to 60 mm, but the increase of the HU values is manifested for a higher mis-centering.

3.2. Results from Patients' Data

Patient's mis-centering in relation to the patient AP diameter is illustrated in Figure 5, where the higher mis-centering was recognized for larger patients.

Afterwards, the patients were divided into groups according to their AP diameter and the mean mis-centering was determined. The summary is included in Table 1.

In total, almost all patients (470 out of 473) were positioned in the negative direction, e.g. they were placed below the isocenter. The mean value of the mis-centering was -43 mm. The minimum value of the mis-centering (patient placed below the isocenter) was -88mm, the maximum value (patient placed above the isocenter) was 16 mm. The minimum and maximum of the mis-centering from

the study of Toth [6] was -66 mm and 34 mm. The mean value of the mis-centering was -23 mm. The study of Habibzadeh [14] published values of the mis-centering for 7 different CT scanners. The mis-centering ranged from -69 mm to 44 mm.

Table 1. Summary of the patients' mis-centering for different diameters of patients

Patients' AP diameter (mm)	No. of pts.	Mean mis-centering (mm)
<180	7	-60
181-200	25	-61
201-220	47	-52
221-240	80	-51
241-260	81	-48
261-280	82	-39
281-300	71	-35
301-320	41	-36
321-340	24	-28
Total	473	Mean -43

The result that almost all patients were placed below the isocenter led to the second approach of the determination of the mis-centering. In this second approach, the patient's AP diameter was measured in the slice where the patient's AP diameter was highest, because some radiographers could place patients according to the body part with the highest diameter, e.g. a diaphragm in the inspiration. The formula (1) was used again. The mean values of the mis-centering determined by the second approach equals -37 mm. Thus a higher patient's AP diameter leads to the decreased mis-centering, from -43 mm to -37 mm, which represents only about a 6 mm improvement compared to the first approach. For this reason, the second approach of the AP diameter determination was not further considered.

It is obvious from Figure 5 and Table 1 that when a patient's AP diameter was smaller, the radiographers inclined to the higher mis-centering, so the mis-centering increased with the decreasing patient's AP diameter. This was also stated in the other studies [6, 14].

As the tendency to mis-centering in a negative way (patient further from the isocenter) was evident in this study, the radiographers were asked about a possible reason. The explanation was that they strived to keep the patient table in the same position, without any consideration for the patient's AP diameter. During the shifts, some radiographers instructed the others (those who were working on night shifts) to do it in the same way. This might be the reason why almost all of our patients were placed below the isocenter. After the radiographers' explanation, all of them were instructed about the correct way of centering patients. After a certain time, the mis-centering will be checked again.

The mis-centering in the negative direction (negative mis-centering) does not lead to a higher value of CTDIvol, and therefore a higher dose to patients, but images can suffer from more noise, as the SDs demonstrate in Figure 4. This

finding was published in the study [6] as well. The noise in CT images not only for the phantoms but also for the group of the patients was determined. The patients in this group underwent CT exams repeatedly (2-5 CT exams). Only quite the obese patients (BMI 32.2-38.8) with repeated CT exams were included (N=5), because the SDs were determined for the area of fat on the back in the region of the hips. The area was between 30.1 cm² and 65.7 cm². The SD was determined for each CT exam.

Unfortunately, this comparison of the noise from repeated CT exams did not show an increase of the SDs with higher values of mis-centering. This is not in agreement with the results from the phantom simulations performed in this study, nor with the results of the study [14]. It can be caused by the fact that 3 of 5 patients had lost weight (up to 14 kg), causing the scanned volumes to differ. Therefore we were not able to prove the higher noise in the patients' images, as it was proved by the phantoms simulations, mainly due to the small number of repeated exams and changed physical conditions of the patients.

4. Conclusions

The mis-centering of the phantoms or patients can cause both an increase and a decrease of the CTDIvol values; therefore the associated doses can be higher or lower. The scanned object (phantom or patient) placed above the isocenter, closer to the X-ray tube in the AP direction, receives a higher dose due to the functioning of the ATCM. The object placed below the isocenter receives a lower dose, but the image quality might be impacted by the noise. The image is noisier when the mis-centering is higher. It was proved by the phantom simulations in this study and also in other studies [6,13,14]. But unfortunately it was not proved on patients in this study. The mis-centering of 50 mm can lead to the increase of CTDIvol by up to 47 %. The mis-centering of -50 mm can lead to a decrease of CTDIvol by up to 35 %.

In this study, the error in the centering of patients was between -88 and 16 mm, the mean value -43 mm. Almost all of the patients, except for 3 from a total of 473 patients, were placed below the isocenter, therefore the CTDIvol values were lower, and consequently the images were noisier. The values of mis-centering were a slightly higher than in other published studies [6,14], which was caused mainly by incorrect instructions given by radiographers.

As a result of the study, all radiographers were educated, therefore a better centering of patients is expected in the future. The issue of mis-centering does not seem to be a rare occurrence, but many CT departments may suffer from this problem, therefore radiographers should be instructed properly about importance of patient centering.

The right centering of patients is quite an important issue, which is often neglected, as was proved in few published studies [6,13,14]. There is made a lot of effort to lower the patients' doses from CT or to improve image quality by

different setups of CT scanners, but such a basic thing as mis-centering is often neglected, which results in ineffective optimization of CT exams.

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