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ORIGINAL PAPER

Impact of miscentering on patient dose and image noise in x-ray CT imaging: Phantom and clinical studies

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Abstract The operation of the bowtie filter in x-ray CT is correct if the object being scanned is properly centered in the scanner's field-of-view. Otherwise, the dose delivered to the patient and image noise will deviate from optimal setting. We investigate the effect of miscentering on image noise and surface dose on three commercial CT scanners. Six cylindrical phantoms with different size and material were scanned on each scanner. The phantoms were positioned at 0, 2, 4 and 6 cm below the isocenter of the scanner's field-of-view. Regression models of surface dose and noise were produced as a function of miscentering magnitude and phantom's size. 480 patients were assessed using the calculated regression models to estimate the influence of patient miscentering on image noise and patient surface dose in seven imaging centers. For the 64-slice CT scanner, the maximum increase of surface dose using the CTDI-32 phantom was 13.5%, 33.3% and 51.1% for miscenterings of 2, 4 and 6 cm, respectively. The analysis of patients' scout scans showed miscentering of 2.2 cm in average below the isocenter. An average increase of 23% and 7% was observed for patient dose and image noise, respectively. The maximum variation in patient miscentering derived from the comparison of imaging centers using the same scanner was 1.6 cm. Patient miscentering may substantially

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increase surface dose and image noise. Therefore, technologists are strongly encouraged to pay greater attention to patient centering.

Introduction

The progress in x-ray CT technology has been immense during the last two decades allowing the introduction of multidetector CT scanners with up to 320-slice capability [1] and many novel technologies such as dual-source CT, C-arm flat-panel-detector CT and micro-CT [2,3]. However, CT is a high dose procedure in comparison to conventional diagnostic radiology [4,5] which has raised concerns particularly to the pediatric population [6]. In 1989, the National Radiological Protection Board (NRPB) showed that although CT accounted for 2% of all examinations in the UK, it contributed about 20% of the collective dose to the UK population from diagnostic x-ray imaging [7]. More recent studies indicate that CT delivers 40% of the collective dose to the UK population from medical x-ray examinations [8,9] whereas latest reports indicate an increase of up to 50% [10]. These figures substantiate the increase of the rate of CT examinations and the important role of dose reduction techniques.

It is well established that the ability to measure any quantity in the real world is limited by statistical noise, which remains an important challenge in medical imaging. This is especially true in the case of radiographic imaging where x-ray photon statistics is always a limiting factor [11]. According to Brooks' formulation [12], the absorbed dose is inversely proportional to noise. This inverse relationship limits endeavors for dose reduction in clinical x-ray imaging. To achieve clinical CT images with valuable diagnostic quality having the lowest possible dose, an optimization of scanning parameters is necessary. For instance, on-line adjustment of tube current according to patient size might lead to establishing an appropriate balance between image noise and radiation exposure [13—15]. Therefore, technologists play a key role in terms of selection of scanning parameters to achieve this goal. Some investigations reported substantial variations of patient dose for common examinations, mainly caused by technologists' choice of exposure parameters [16].

The implementation of automatic tube current modulation (ATCM) on recent CT scanners opened a new window to achieve a desirable optimization by which the image quality sustained constantly among patients with various attenuation and size characteristics [17,18]. ATCM automatically tunes the tube current to maintain a user-specified quantum noise level in the acquired data by estimating an appropriate tube current to obtain images with a chosen noise level [17].

Patient size is an important dose influencing factor that needs to be looked at in CT imaging. A phantom study performed by Nickoloff et al. [19] showed that the radiation dose is much greater for small phantoms compared to large ones for the same tube current and voltage. Even the ATCM technique which modulates tube current with respect to patient asymmetry, patient size is an important parameter that should be considered. Abdominal scans on multislice CT with tube current modulation convey considerably higher doses to oversized patients than thin patients [20].

The beam shaper or bowtie filter is one of the components of a CT scanner which affects the absorbed dose. The latter modifies the spatial distribution of radiation emitted within the fan beam according to its shape and material. The thickness of this filter is minimal in the middle and increases toward the edges (Fig. 1). The role of the bowtie filter is to convey maximum radiation to the thickest part of the patient which attenuates the most x-rays and to reduce x-ray intensity in regions where patient attenuation decreases. Thus, it reduces the dose especially in peripheral areas. This component is used to optimize the dynamic range of the CT detection system [21,22]. The correct operation of the bowtie filter to achieve this objective requires that the object being scanned is properly centered in the scanner's field-of-view (FOV). If the object is miscentered as shown in Fig. 2, it would be exposed to more surface dose in the region that goes toward the less attenuating part of the bowtie filter and the noise would increase in the region that moves into the more attenuating part of the bowtie filter [23,24]. This is especially important for anterior and posterior organs because anterior organs receive a higher dose when a supine patient's position is under the isocenter whereas posterior organs receive

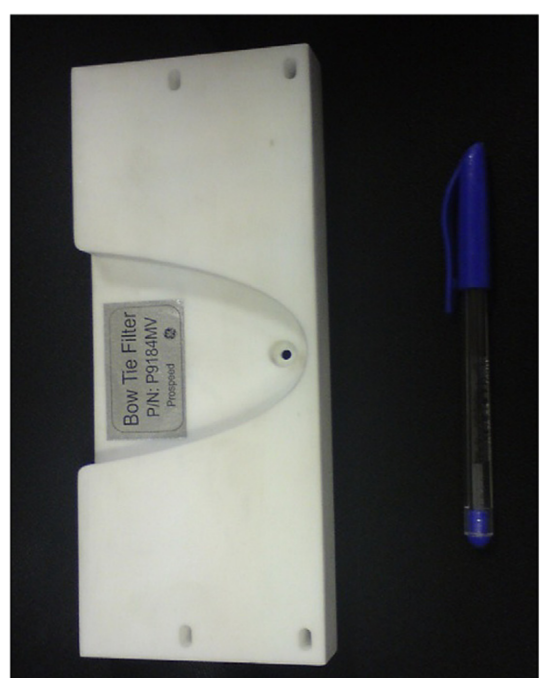

Figure 1 Photograph of typical Teflon made bowtie filter which is narrow at center and thick at the edges.

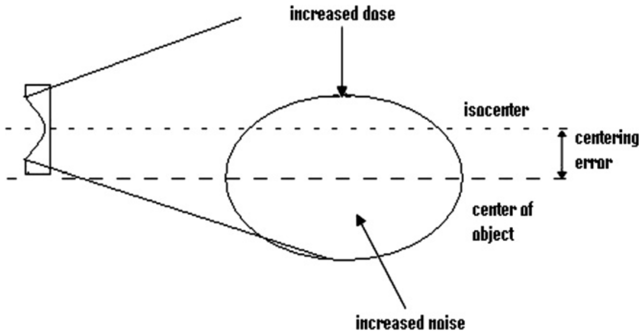

Figure 2 The scanned object is positioned below the isocenter. Because of the shape of the bowtie filter, the upper half is exposed to more radiation and receives a higher dose while the image noise of the lower half increases.

a higher dose when the patient is placed above the isocenter. Patient miscentering is an important factor that increases patient surface dose and image noise in busy imaging centers where technologists cannot afford to spend sufficient time for patient positioning. A study which considered a single imaging center reported that 95% of patients undergoing chest and abdominal CT examinations were miscentered in the vertical direction (y-axis) within the gantry [25].

In this multicentre study, we investigate and quantify the influence of patient miscentering on image noise and patient dose of 480 patients encompassing a wide range of examinations in seven imaging centers equipped with three different commercial CT scanners. We also highlight the impact on image quality and patient dose during x-ray CT imaging. It should be emphasized there is a lack of multicentre trials addressing the influence of patient miscentering on image noise and patient dose. Moreover, this study also highlighted the role of the technologist by including similar scanners installed in different imaging centers and thus involving different operators.

Materials and methods

CT scanners

Seven commercial CT scanners manufactured by GE Healthcare (General Electric Healthcare Technologies, Waukesha, WI, USA) belonging to one of the three different models described were used in this study.

64-Slice GE lightspeed VCT

This third generation CT scanner has a 540 mm source to isocenter and 950 mm source to detector distance. It is composed of 58,368 individual elements arranged in 64 rows of 0.625 mm thickness at isocenter, each containing 888 active patient elements and 24 reference elements with Highlight ($Y_2Gd_2O_3$:Eu) ceramic scintillator. The tube current can be set up to a new x-ray tube supported by a powerful generator delivering high peak mA up to 800 mA.

8-Slice GE brightspeed edge

The BrightSpeed series are compact systems having a small size designed using the LightSpeed VCT technology with maximum gantry speed of 0.5 s per rotation. However, the tube is different from VCT scanner's tube and delivers up to 440 mA.

4-Slice GE lightspeed QX/*i*

This four-slice scanner has a tube with 6.3 MHU of heat capacity. The fastest gantry rotation of this scanner is 0.8 s whereas the maximum system's output is 440 mA at 120 kV.

Experimental phantoms

Six cylindrical phantoms with various sizes and materials (Table 1) were designed and scanned on the above described CT scanners with different scanning parameters and levels of miscentering. These include four water phantoms, one polyethylene and one CTDI phantom. Water phantoms contained distilled water and are common phantoms used in the assessment of absorbed dose and image noise studies. The CTDI phantom is designed specifically for dose characterization in CT body imaging whereas the Polyethylene phantom was used to widen the range of materials and densities of phantoms.

Dosimetry system

The Barracuda dosimetry system (RTI Electronics AB, Sweden) was used for dose measurements in this study. The system was calibrated by the manufacturer before use in experimental measurements. This system can measure parameters of interest such as kVp, exposure time, absorbed dose, etc. for standard quality assurance (QA) procedures. Barracuda is connected to a handheld computer running the QA browser software which guides the examiner through the various tests and measurements of x-ray systems. The DCT10 pencil ionization chamber (10 cm length) dedicated for dose measurement in CT is one of the Barracuda system's accessories.

Experimental setup

To calculate the influence of miscentering on patient dose and image noise, the phantoms were positioned on the scanner's table where the center of phantoms is aligned at 0, 2, 4 and 6 cm below the center of rotation (isocenter). Axial scanning was performed for each phantom's position using scanning parameters of 120 kVp tube voltage, 200 mA tube current, 2 s gantry rotation speed and 4×5 mm axial slice collimation. A large body bowtie filter was chosen for

Table 1 List of phantoms used in this study and their corresponding calculated square root of projection area (*sqrtPA*).

Phantom	Material	Diameter (cm)	*sqrtPA*
W15	Water	14.8	22.7
W17	Water	17.0	25.0
W21	Water	21.0	31.3
W23	Water	22.5	32.8
CTDI32	PMMA	32.0	49.3
P26	Polyethylene	26.5	38.5

each scan given that a large scanning FOV was used. Anterior-posterior (AP) scout scans were also obtained for phantoms at the center while lateral scouts were obtained at positions below the center. Figure 3 shows the experimental setup used for dosimetry estimates.

Absorbed dose and noise characterization

The surface dose was measured using a standard and calibrated pencil chamber (DCT10) placed on the top surface of phantoms during scanning using specified scan parameters. Measurements were repeated for different miscenterings to calculate the influence of miscentering on patient dose.

Standard deviation (SD) of CT numbers in selected regions of interest (ROIs) on the image was considered as an indicator of image noise. SD measurements were made for ROIs representing approximately 60% of the area of the lower half of the phantoms (Fig. 4). For each scan, SD calculations were performed on images acquired in one rotation (four slices) and then averaged over the four axial slices.

Characterization of surface dose and noise

Regression models of the increase of surface dose and noise of the lower half of the image were generated as a function of phantom size and magnitude of miscentering. The increase of measured dose and noise due to centering error were used to define series of regression models. These models were generated for each scanner to assess the behavior of dose and noise due to miscentering. These regression models allow the prediction of surface dose increase and lower half image noise for each patient from the scout scans. Statistical analysis of the data to generate regression models was performed using SPSS version 14 (SPSS Inc, IL).

Estimation of patient size

Patient size was determined from AP scout scans of patients through calculation of projection area (PA). This

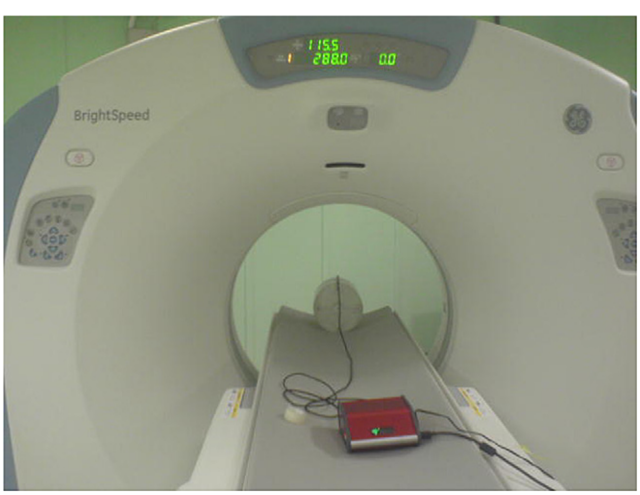

Figure 3 Dose measurement setup using the Barracuda dosimetry system.

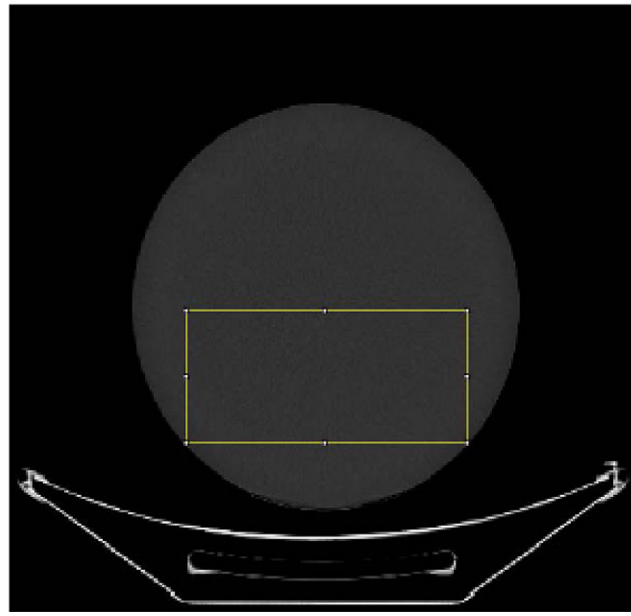

Figure 4 An ROI defined on a CT image of the Polyethylene phantom used for noise measurement through estimation of the standard deviation (SD).

parameter is used on GE CT scanners for automatic tube current modulation as an indicator of patient's size [17]. Projection area includes information about object's size and density and can be calculated from the summation of detector channel data values after corrections. This parameter is a good indicator of total attenuation of the object under study. Since the projection area is a quantity reflecting a surface, its unit is square meter (m^2), the square root of this parameter (*sqrtPA*) is a parameter that can be used for estimation of object size as reported by Toth et al. [24]. The correctness of using *sqrtPA* as an indicating factor of object size was investigated again for further validation. For this purpose, regression models for estimation of *sqrtPA* as a function of phantom size were calculated.

The method using scout scan for *sqrtPA* calculation relies on scout attenuation area (SAA) [26]. Figure 5 shows these parameters on a scout image. The relationship between these parameters is given by:

$$SAA = W \times ROI \times 0.001 \tag{1}$$

$$sqrtPA_{64} = SAA + 8.7 \tag{2}$$

$$sqrtPA_{4,8} = SAA + 10.7 \tag{3}$$

The anterior-posterior scout images were used to determine *sqrtPA* for each phantom. *W* in Eq. (1) is the width of the ROI defined on the scout image. It should be noted that Eq. (2) is valid for the 64-slice GE LightSpeed VCT whereas Eq. (3) is valid for the 4-slice GE LightSpeed and 8-slice GE BrightSpeed scanners. The same procedure was performed on patients' scout scans to determine *sqrtPA* as function of patient size.

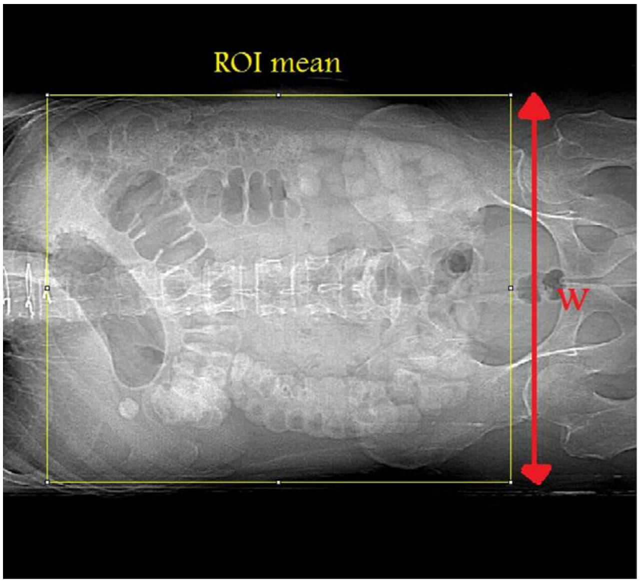

Figure 5 Illustration of the parameters used for calculation of the scout attenuation area. The ROI mean is the average CT number of the pixels within the ROI whereas *W* is the average lateral width.

Characterization of centering error

In physics, the word centroid means the geometric center of object's shape. The geometrical centroid of an object is defined as the intersection of allstraight linesfrom the apex or the middle of one side which divides opposite sides into two equal parts.

The centroid of patient images is required to determine the amount of patient miscentering. Lateral scout scans taken from side view (tube at 3:00 position) show offsets in the *y*-axis of the scanned object if it is miscentered. The centroid of a selected ROI in an image is calculated using freely available *ImageJ* software (National Institute of Mental Health, Bethesda, MA). Lateral scout scans of a phantom at 2, 4, and 6 cm below the isocenter were used to check whether the software's results match expected values. An ROI covering the height of a phantom located 2, 4, and 6 cm below the center was selected. The result obtained using *ImageJ* agreed well with the amount of miscentering.

A polygonal ROI fitting the shape of patient's body was defined on lateral scout scans to calculate the centroid of the patient. Figure 6 shows a sample of a ROI defined on a patient's lateral scout scan using *ImageJ*. The amount of miscentering was calculated by subtracting of Y coordinate of calculated centroid from the Y coordinate of the image center.

Clinical assessment

Clinical assessment of the impact of patient miscentering was performed on scout scans of patients scanned on the selected CT scanners. First, *sqrtPA* of patients were calculated from the AP scout scans and then, the amount of miscentering determined from the lateral scout scans. 960

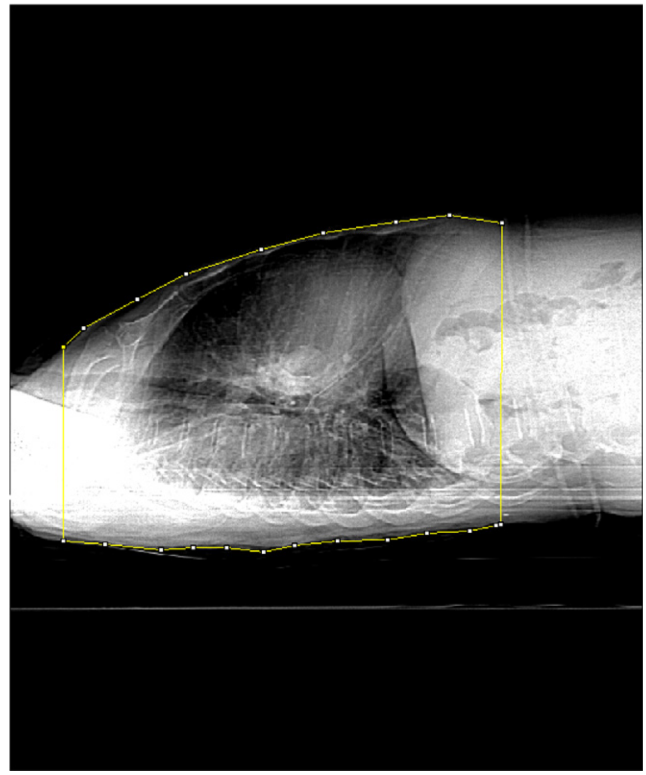

Figure 6 Typical ROI used for the calculation of miscentering from scout view scan.

lateral and AP scout images corresponding to 480 patients were collected from 7 imaging centers. Centers' IDs 1, 2 and 3 were used for those facilities equipped with a 64-slice scanner whereas IDs 4, 5 and 6 were used for centers using a 4-slice scanner. ID 7 was used for the only center equipped with a 8-slice scanner. Patient population consisted of 188 females and 292 males with ages ranging from 18 to 83 years old. Using the derived regression models of surface dose and image noise obtained from calculated *sqrtPA* and miscentering, the amount of increase in dose and image noise for each patient was estimated.

Results

Phantom studies

The results of *sqrtPA* calculations using the previously described method are shown in Table 1. There is a linear relationship between *sqrtPA* and phantom size which indicates that *sqrtPA* is an appropriate factor for representation of patient size.

As expected, if the scanning object is placed below the center of rotation, the top surface dose and noise of the lower half of the image increase. Increasing the miscentering increases the surface dose and noise. Figure 7 shows that the trend of the dose changes with respect to the value of centering error for 64-slice, 8-slice and 4-slice CT scanners. As an example, the increase in surface dose for the 64-slice scanner measured using the CTDI-32 phantom was 13.5%, 33.3% and 51.1% for miscentering of

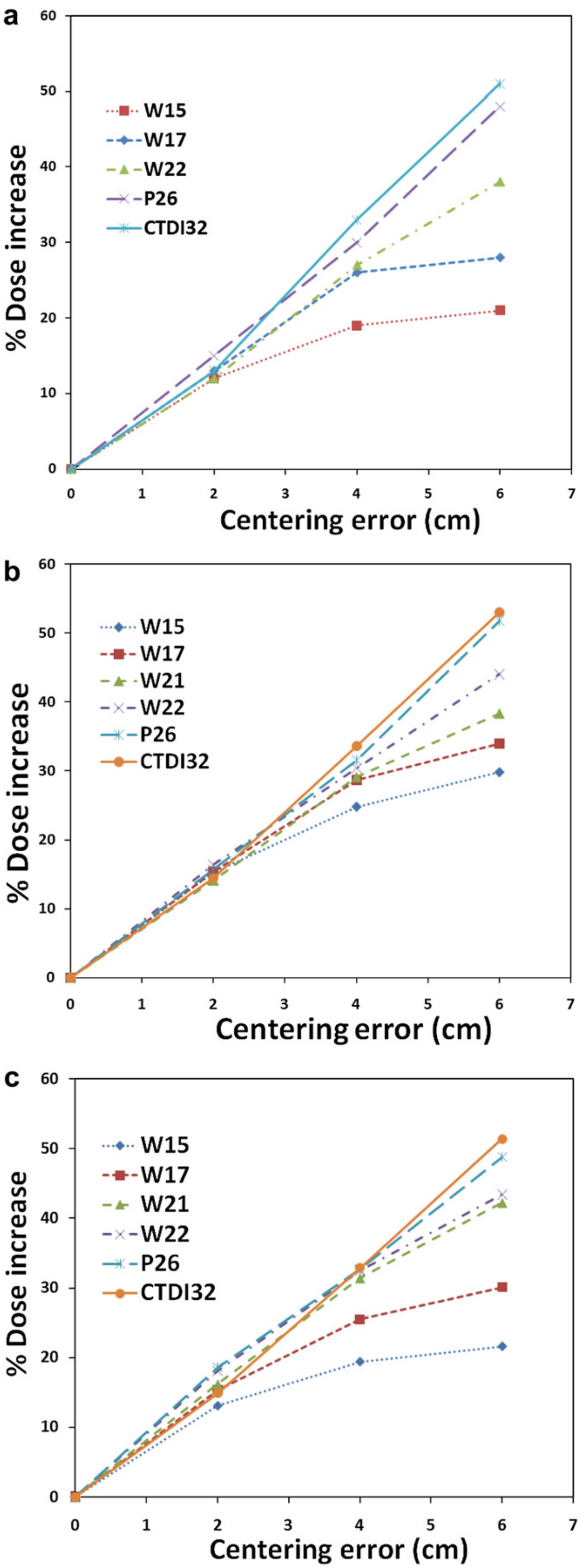

Figure 7 Relationship between surface dose and centering error for different phantoms scanned on (a) the 64-slice, (b) 8-slice, and (c) 4-slice CT scanners.

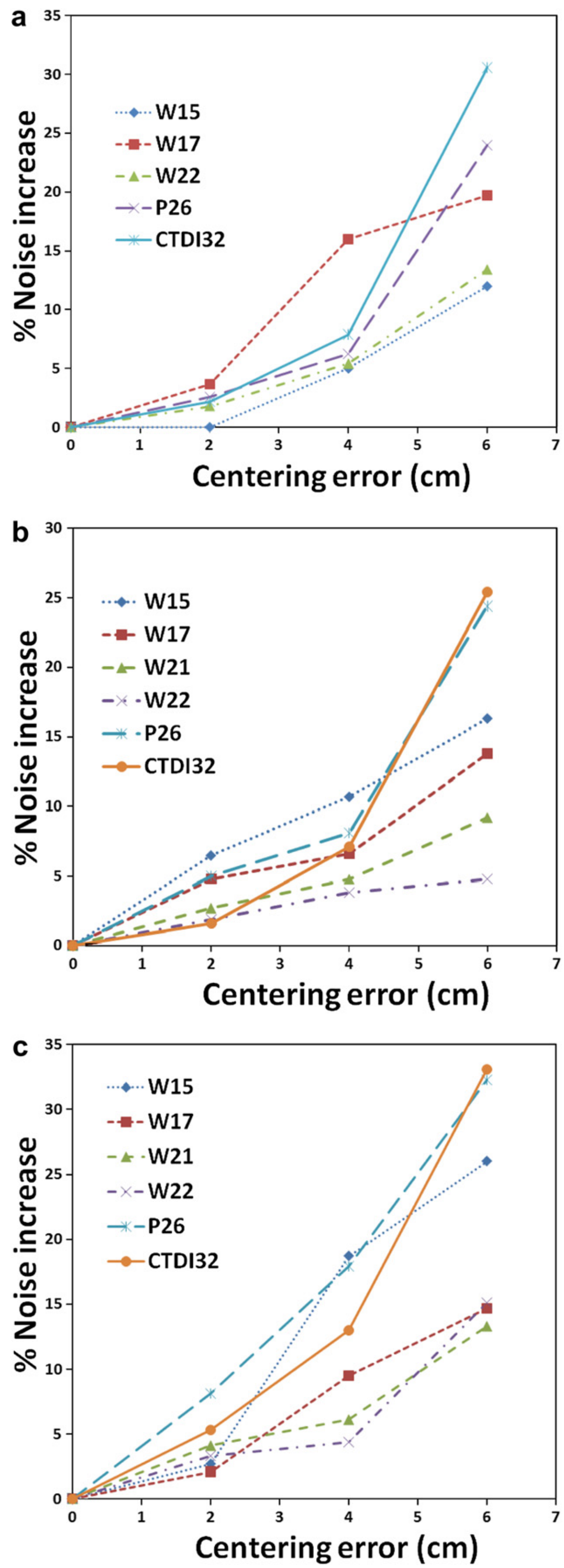

Figure 8 Relationship between image noise in the lower half of the image and centering error for different phantoms scanned on (a) the 64-slice, (b) 8-slice, and (c) 4-slice CT scanners.

2, 4 and 6 cm below the isocenter, respectively. The corresponding increase of surface dose was 14.4%, 33.6% and 53% whereas it was 14.9%, 32.9% and 51.4%, for the 8-slice and 4-slice CT scanners, respectively.

Figure 8 shows the percentage increase of image noise vs. the miscentering for different CT scanners. For example, the increase in image noise for the 64-slice CT scanner using the W23 phantom was 1.8%, 5.4% and 13.4% for miscenterings of 2, 4 and 6 cm below the isocenter, respectively. The corresponding increase of image noise was 1.9%, 3.8% and 8.4% whereas it was 3.3%, 4.4% and 15.1%, for the 8-slice and 4-slice CT scanners, respectively.

Clinical assessment

An average miscentering of 2.1 cm below the isocenter was calculated from the analysis of 80 patient's scout scans acquired at imaging center#1, thus leading to an average increase of patient surface dose and image noise of 21.9% and 6%, respectively. The same analysis was performed for the remaining imaging centers. The results are summarized in Table 2.

Discussion

Based on phantom studies, patient positioning out of the center of the FOV causes an increase in patient surface dose and image noise according to the shape and operation of the bowtie filter. Therefore, much worthwhile effort focused on the assessment of the effect of miscentering on patient surface dose and image noise in common clinical procedures. Moreover, the responsibility of technologists in patient positioning is undeniable and as such the comparison of imaging centers' results could be a good indicator of their skilled working habits.

Considerable dose reduction along with good quality images can be achieved through close cooperation between medical physicists, radiologists and radiographers. Technologists operating in various imaging centers have different experience and working habits and as such perform clinical examinations in a different way. In addition, the clinical activity of various imaging centers in terms of number of patients scanned per day is quite variable. In centers with high rate examinations, operators have the duty to perform the procedures in a relatively short time at the expense of the quality of clinical examinations. The comparison of results obtained from different imaging centers using the same CT scanners and scanning protocols demonstrates the role of technologists in clinical setting.

In this study, the technologists' faulty operation from imaging center#2 leads to an increase of 1.6 cm in average patient centering error leading to an additional average dose of 7.2% and noise of 4.9% compared to imaging center#1 (Table 2). Likewise, the working habits of imaging center#6 also resulted in 0.9 cm extra error in patient centering leading to an average additional dose of 6.8% and noise of 2.5% compared to imaging center#4.

The results obtained on the 64-slice CT scanners show that centering errors range from 4.4 above to 6 cm below the isocenter. It was observed that 85% of the patients were positioned below and 15% were placed above the isocenter. The results also indicate that if obese, medium and thin patients are separated according to their calculated *sqrtPA*, obese patients tend to be positioned below the center with a mean value of 0.45 cm while this value is 3.35 cm for slim patients.

For the center equipped with the 8-slice CT scanner, centering errors range from 2.8 cm above the center to 6.9 cm below the center where 84% of patients were positioned below the isocenter and 16% were placed above the isocenter. By separating obese and slim patients according to their *sqrtPA*, it is determined that obese patients are positioned below the center with mean value of 0.55 cm while this value is 2.7 cm for slim patients.

For the 4-slice CT scanners, centering errors range from 4.1 cm above to 6.2 cm below the isocenter. The analysis of patient data showed that 80.5% of the patients were placed below while 19.5% were positioned above the center. In centers equipped with this scanner, obese patients were positioned 0.5 cm on average below the center with a corresponding value of 2.4 cm for slim patients. Although these values are smaller those reported for the 64-slice scanner, there is a clear indication that slim patients are more commonly positioned below the center compared to obese patients.

Positioning errors above and below the center demonstrate that technologists usually have the tendency of placing the patients below the isocenter. In addition, it was demonstrated that the probability of miscentering is larger for slim patients than obese patients. Furthermore, the majority of patients were miscentered by more than 1 cm (85% for 64-slice, 75% for 8-slice, and 67% for 4-slice CT scanners). In most of the cases, technologists make significant mistakes during patient positioning. In some imaging

Table 2 Results of miscentering assessment using 480 patients' scout scans acquired in 7 imaging centers equipped with three different CT scanners.

Imaging center ID	Number of patients	CT scanner	Miscentering (cm)	Dose increase (%)	Noise increase (%)
1	80	64-slice	2.1 ± 2.0	21.9 ± 10.3	6.0 ± 6.0
2	33	64-slice	3.7 ± 1.8	29.1 ± 10.5	10.9 ± 6.7
3	89	64-slice	2.5 ± 1.9	25.0 ± 9.6	7.6 ± 7.0
4	78	4-slice	1.5 ± 1.5	17.6 ± 10.8	5.3 ± 5.2
5	72	4-slice	1.7 ± 1.7	21.9 ± 8.3	6.7 ± 5.7
6	45	4-slice	2.4 ± 2.0	24.4 ± 10.1	7.8 ± 7.4
7	83	8-slice	1.9 ± 1.8	19.8 ± 13.2	4.7 ± 4.7

centers such as centers #3 and #6 in this study, operators rely only on AP scout views for the majority of patients and use also the lateral scout views only in few situations. It should be emphasized that the addition of a lateral scout view allows the operators to more accurately center the patients.

Multislice CT scanners generally have two kinds of bowtie filters, namely small and large filters (some modern CT scanners also have medium size filter). In this study, only the large bowtie filter was used for phantom imaging owing to the size of the phantoms and hence the small filter was not studied. One should keep in mind that increasing radiation dose is a particular concern for pediatric patients. Future studies will assess the impact of patient miscentering on pediatric patients through the use of dedicated pediatric phantoms mimicking pediatric situations [27,28] combined with the small bowtie filter used for this kind of examinations.

Conclusion

We evaluated the influence of patient miscentering on patient dose and image noise using three different CT scanners installed in seven imaging centers. The results show that patient miscentering during positioning inside the CT gantry increases the surface doses delivered to patients. Besides, image noise increases but the impact of miscentering on patient dose is more relevant than image noise. To compensate image noise, increasing radiation exposure is usually employed which conveys more dose to patients and also reduces the effective lifetime of the x-ray tube.

The comparison of various imaging centers using the same model of CT scanner and scanning protocols demonstrates the important role of technologists in minimizing patient dose and image noise through careful and accurate patient positioning. Thus, they should be strongly encouraged to pay greater attention to patient centering and use the lateral scout view besides the AP scout view for more accurate positioning.

Acknowledgments

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References

[1] Klingebiel R, Siebert E, Diekmann S, Wiener E, Masuhr F, Wagner M, et al. 4-D imaging in cerebrovascular disorders by using 320-slice CT: feasibility and preliminary clinical experience. Acad Radiol 2009;16(2):123—9.
[2] Kalender WA. X-ray computed tomography. Phys Med Biol 2006;51(13):R29—43.
[3] Boone JM. Multidetector CT: opportunities, challenges, and concerns associated with scanners with 64 or more detector rows. Radiology 2006;241(2):334—7.
[4] Shrimpton PC, Hillier MC, Lewis MA. Data from computed tomography (CT) examinations in the UK. Chilton: National Radiological Protection Board; 2003.
[5] UNSCEAR. Sources and effects of ionizing radiation. United Nations, New York: United Nations, Scientific Committee on the Effects of Atomic Radiation Report to the General Assembly; 2008.
[6] Brenner DJ, Hall EJ. Computed tomography-an increasing source of radiation exposure. N Engl J Med 2007;357(22): 2277—784.
[7] Shrimpton PC, Jones DG, Hillier MC. Survey of CT practice in the UK: part 2; dosimetric aspects. National Radiological Protection Board; 1991.
[8] Shrimpton PC, Edyvean S. CT scanner dosimetry. Br J Radiol 1998;71(841):1—3.
[9] Mettler Jr FA, Wiest PW, Locken JA, Kelsey CA. CT scanning: patterns of use and dose. J Radiol Prot 2000;20(4):353—9.
[10] Hart D, Wall BF. UK population dose from medical X-ray examinations. Eur J Radiol 2004;50(3):285—91.
[11] Curry TS, Christensen EE, Murry RC, Dowdey JE. Christensen's physics of diagnostic radiology. 4th ed. Philadelphia: Lea & Febiger; 1990.
[12] Brooks RA, Di Chiro G. Statistical limitations in x-ray reconstructive tomography. Med Phys 1976;3(4):237—40.
[13] Kalender WA, Wolf H, Suess C, Gies M, Greess H, Bautz WA. Dose reduction in CT by on-line tube current control: principles and validation on phantoms and cadavers. Eur Radiol 1999;9(2):323—8.
[14] Kalra MK, Prasad S, Saini S, Blake MA, Varghese J, Halpern EF, et al. Clinical comparison of standard-dose and 50% reduced-dose abdominal CT: effect on image quality. AJR Am J Roentgenol 2002;179(5):1101—6.
[15] Donnelly LF, Emery KH, Brody AS, Laor T, Gylys-Morin VM, Anton CG, et al. Minimizing radiation dose for pediatric body applications of single-detector helical CT: strategies at a large children's hospital. AJR Am J Roentgenol 2001;176(2): 303—6.
[16] Koller CJ, Eatough JP, Bettridge A. Variations in radiation dose between the same model of multislice CT scanner at different hospitals. Br J Radiol 2003;76(911):798—802.
[17] Kalra MK, Maher MM, Toth TL, Schmidt B, Westerman BL, Morgan HT, et al. Techniques and applications of automatic tube current modulation for CT. Radiology 2004;233(3): 649—57.
[18] Keat N. CT scanner automatic exposure control systems. London: Medicines and Healthcare Products Regulatory Agency; 2005.
[19] Nickoloff EL, Dutta AK, Lu ZF. Influence of phantom diameter, kVp and scan mode upon computed tomography dose index. Med Phys 2003;30(3):395—402.
[20] Schindera ST, Nelson RC, Toth TL, Nguyen GT, Toncheva GI, DeLong DM, et al. Effect of patient size on radiation dose for abdominal MDCT with automatic tube current modulation: phantom study. AJR Am J Roentgenol 2008;190(2):W100—5.
[21] Toth TL, Cesmelia E, Ikhlef A. Image quality and dose optimization using novel X-ray source filters tailored to patient size. In: Proceedings of SPIE; 2005. p. 283—291.
[22 Tack D, Gevenois PA, Abada HT. Radiation dose from adult and pediatric multidetector computed tomography. Berlin; New York: Springer; 2007.
[23] Aviles Lucas P, Castellano IA, Dance DR, Vano Carruana E. Analysis of surface dose variation in CT procedures. Br J Radiol 2001;74(888):1128—36.
[24] Toth T, Ge Z, Daly MP. The influence of patient centering on CT dose and image noise. Med Phys 2007;34(7):3093—101.
[25] Namasivayam S, Kalra MK, Mittal P, Small WC. Can radiation exposure associated with abdominal and/or pelvic CT be minimized with better practice? Vancouver, canada: ARRS; 2006.

[26] Udayasankar U, Kalra M, Li J, Toth T, Seamans J. Multidetector scanning of the abdomen and pelvis: a study for evaluation of size compensated automatic tube current modulation technique in 100 subjects. In: RSNA 2006; 2006.
[27] Siegel MJ, Schmidt B, Bradley D, Suess C, Hildebolt C. Radiation dose and image quality in pediatric CT: effect of technical factors and phantom size and shape. Radiology 2004;233(2):515—22.
[28] Papadakis AE, Perisinakis K, Damilakis J. Automatic exposure control in pediatric and adult multidetector CT examinations: a phantom study on dose reduction and image quality. Med Phys 2008;35(10):4567—76.