The Influence of Patient Miscentering on Patient Dose and Image Noise in Two Commercial ct Scanners

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Abstract— **The clinical influences of patient miscentering on patient dose and image noise were investigated for two models of commercial CT scanners. Several phantoms were scanned on 4-slice GE Lightspeed and 64-slice GE Lightspeed VCT. Regression models of surface dose and image noise were generated as a function of phantom size and the value of miscentering. 64 scout images of patients from the first scanner and 113 from the second one were analyzed to assess the possible amount of increasing in dose and noise. For the first scanner the average amount of miscentering was 3 cm below the isocenter which leads to 25.8% increase in dose and 8.3% increase in noise. These values for the second scanner were 1.6 cm below the center, 19.8% and 6.2%, respectively. The results clearly demonstrate that patient miscentering may substantially increase dose and image noise. Therefore, technologists are strongly encouraged to pay greater attention to patient centering.**

Keywords— **Bowtie Filter, Computed Tomography, Dose, Noise.**

I. INTRODUCTION

Bowtie beam shaping filter is an important element in the CT image formation chain that is used both for dose reduction and optimization of detector dynamic range. 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 where patient attenuation decreases [1]. The operation of the bowtie filter is based on the assumption that the object being scanned is properly centered in the scanner's field-of-view (FOV). If the object is miscentered as it is shown in figure 1, 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 [2]. In this study, we have investigated the effect of patient miscentering on image noise and patient dose in two different commercial CT scanners. In addition, in order to determine the role of technologists, results of two imaging centers for each of implied scanners were compared.

II. MATHERIALS AND METHODS

A. *Quantifying the Bowtie Filter Effect*

Six cylindrical phantoms (Table 1) with various size and materials were scanned on the 64-slice GE Lightspeed VCT and the 4-slice GE Lightspeed (General Electric Healthcare Technologies, Waukesha, WI, USA) scanner (5 phantoms were used for the 64-slice GE VCT). The phantoms include four water phantoms, one polyethylene and one CTDI phantom with different size and density in order to emulate various patient sizes. Phantom centers were positioned at 0, 2, 4 and 6 cm below the center of rotation. Scout scans were also obtained from the phantoms at anterior-posterior view. The scanning parameters were 120 kVp, 4×5 mm axial slice collimation, 200 mA with 2 sec gantry rotation speed. Large body bowtie filter was chosen for each scan given that a large scanning FOV was used. Dose was measured using a standard 10 cm pencil chamber placed on the top surface of the phantoms. The Barracuda dosimetry system (RTI Electronics AB, Flöjelbergsgatan 8 C, SE-431 37 Mölndal Sweden) was used for dose measurements with its associated accessories (DCT10 chamber). Figure 2 shows the experimental setup used for dosimetry estimates. Standard deviation (SD) in selected regions of interest (ROI) 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' images. For each scan, SD measurements were performed on all images acquired in one rotation (four images) and were then averaged over the four axial images acquired (figure 3).

Regression models of surface dose and noise of the lower half of the image were generated as a function of phantom size (the amount of sqrt PA [explained later]) and miscentering. Using these regression models, it was possible to

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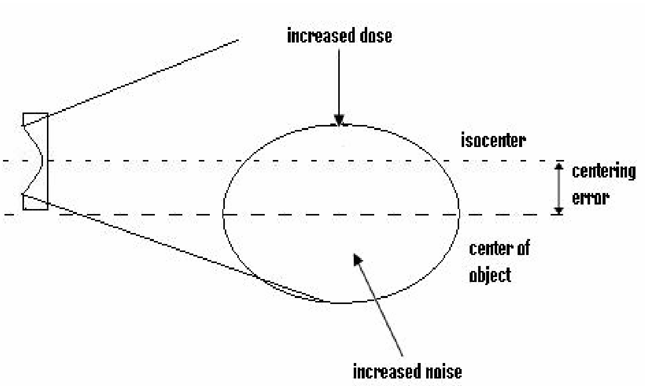


Fig. 1 The scanning object is positioned below the isocenter, so because of the shape of bowtie filter, higher half is exposed to more dose and the image noise of the lower half is increased [1]

predict the increase of the surface dose and lower half image noise for each patient from the scout images. Statistical analysis of the data to generate regression models was performed using SPSS software version 14.

B. Object Size Estimation

Since in this study, scout images of patients were used to determine the values of patient miscentering and on the other hand, the size of the patient is a parameter which affects the dose and image noise, it was needed to use a factor that could be calculated from the scout images and also represent the patient size. For this reason projection area was used. This parameter is already used in GE scanners for automatic tube current modulation [3]. Projection area is included the information of object's density and size and is the summation of detector channel data values after corrections. This factor represents total attenuation of the object. Because the dimension of projection area is area, thus the square root of it (sqrt PA) is a parameter which was used for representing the size [2].

Fig. 2 Experimental setup for Dose measurement using the Barracuda dosimetry system

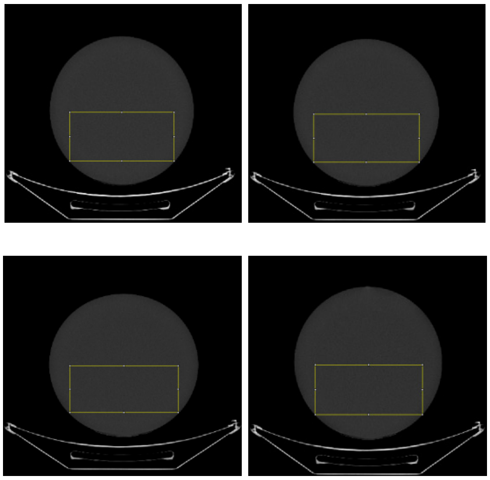


Fig. 3 Noise measurement through estimation of the standard deviation (SD) in each image slice of the Polyethylene phantom

One of the methods for calculation of sqrt PA is to use scout images [4]. In this method sqrt PA is calculated from scout attenuation area. Figure 4 demonstrate these parameters in a scout image. The relationship between these two parameters is [5]:

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

$$sqrtPA = SAA + 8.7 \tag{2}$$

$$sqrtPA = SAA + 10.7 \tag{3}$$

The anterior-posterior scout images were used to determine the square root of projection area (sqrt PA) for each phantom. First a rectangular ROI was selected on scout image, then ROI MEAN was calculated by Imagej software and at last sqrt PA was determined by use of implied equations. In the above equations, ROI is the summation of ROI MEAN and 1000 and W is the average lateral width. The relation (2) and (3) is true for 64-slice GE Lightspeed VCT and 4-slice GE Lightspeed, respectively. The same process was performed on patients' scout images to determine the sqrt PA as patient size.

Table 1 Phantoms used in this study with the corresponding values of square root of projection area (sqrt PA)

Phantom name	Material	Exact diameter (cm)	sqrt PA
W15	Water	14.8	22.7
W17	Water	17	25
W21	Water	21	31.3
W23	Water	22.5	32.8
CTDI 32	PMMA	32	49.3
P26	Polyethylene	26.5	38.5

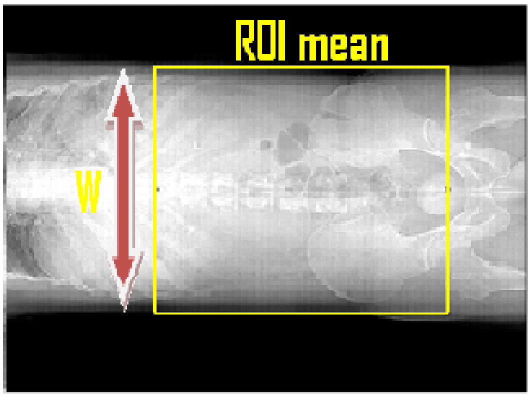


Fig. 4 This figure shows the parameters which are used for scout attenuation area calculation. ROI mean is the average CT number of the pixels within the ROI and W is the average lateral width

C. *Assessment Strategy*

To evaluate the influence of miscenterings, scout images of patients who were scanned with these models of scanners were used. First the sqrt PA of patients were calculated from the anterior-posterior scout and then the amounts of miscenterings were determined from the lateral scouts.

The amount of miscentering was calculated by use of *Imagej* software. This software is capable to determine the geometrical center of patient body's shape. By choosing a polygonal ROI of patient body in lateral scout image and determination of center of this ROI and subtracting the value of Y from the Y value of image's center, the amount of miscentering was found out. Figure 5 shows that a selection of an ROI in lateral scout image by *Imagej*.

From sqrt PA and miscentering calculated values and use of regression models of dose and image noise, the amounts of dose and image noise increase were assessed for each person due to miscentering. For each scanner, scout images were gathered from 2 imaging centers. The average centering error, dose increase percentage and image noise increase percentage was calculated for each center and finally the results of centers were compared.

III. RESULTS

Figure 6 demonstrate that how the dose changes with regards to the value of centering error for 64-slice GE Lightspeed VCT and 4-slice GE Lightspeed. For example, the increase of surface doses using the CTDI-32 phantom were 15.5%, 33.3% and 51.1% for miscentering of 2, 4 and 6 cm below the isocenter, respectively for 64-slice GE Lightspeed VCT. For the GE Light Speed 4-slice scanner, the corresponding increase of surface doses were 18.6%, 32.9%, 51.4%, respectively.

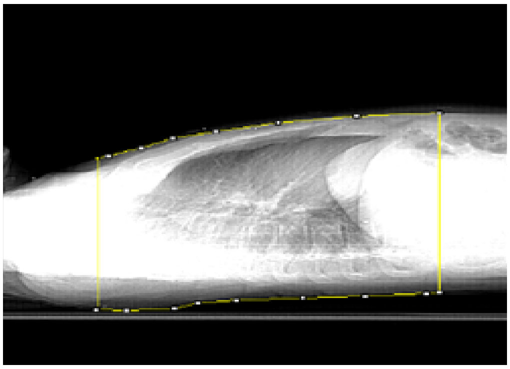


Fig. 5 A ROI is chosen regards to the shape of the patient's body

Figure 7 shows the percentage of image noise changes relative to the value of miscentering for the two implied scanners. For example, the image noise increases using the W23 phantom were 1.8%, 5.4% and 13.4% for miscentering of 2, 4 and 6 cm below the isocenter, respectively for 64-slice GE Lightspeed VCT. For the GE Light Speed 4-slice scanner, the corresponding increases of image noise were 3.3%, 4.4% and 15.1%, respectively.

As it is already implied, for each scanner, the scout images of patients from two imaging sites were used. The names 1 and 2 were used for the sites with VCT scanner and the names 3 and 4 were used for the sites with 4 slice scanner.

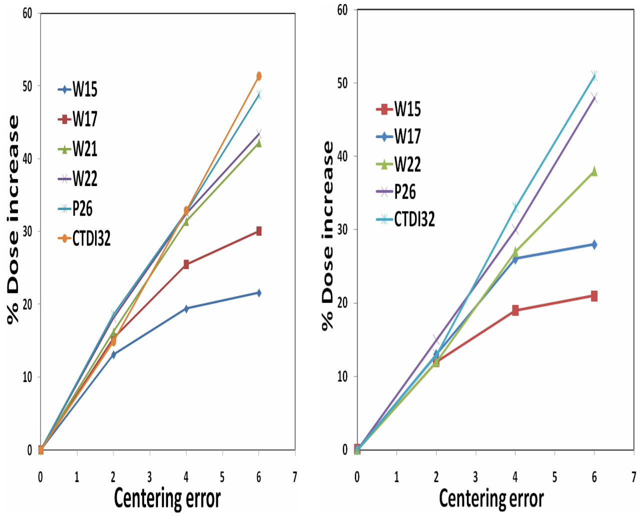
From site 1, 80 patient's scouts were analyzed which the average miscentering was 2.1 cm below the isocenter that leads to the average 21.9% increase of dose and 6% increase of noise. The same analyzes were done for 33 patients from aite 2 which is included in table 2.

From site 3, 41 patient's scouts were analyzed that the average increase of dose was 17.6% and the average increase of noise was 5.3% while the average miscentering value was 1.5 cm below the isocenter. The same analyzes were done for site 4 which is included in table 3.

IV. DISCUSSION

Technologists' faults from site 2 leads to 1.6 cm more error in patients centering and thus the average 7.2% more dose and 4.9% more noise than site 1. These faults from site 4 also leads to 0.2 cm more error in patients centering and therefore the average 4.3% more dose and 1.4% more noise than site 3.

Looking at the calculated values of miscentering for all the patients in each center, makes it clear that the high percentage (67-85 %) of patients was miscentered more than 1 cm. These amounts show that in the most of the cases tech

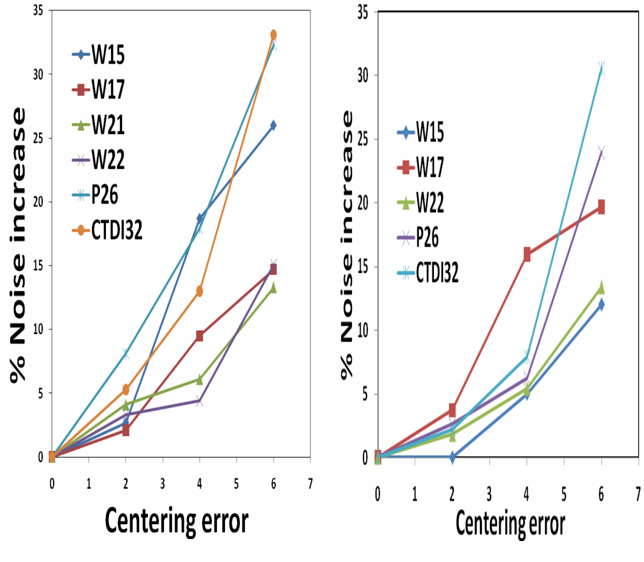


(a) (b)

Fig. 6 (a) The relation between surface dose and centering error for different phantoms on the 4-slice Light Speed and (b) Light Speed 64-slice CT scanners

nologists make unignorable mistakes. Of course the numbers of patients differ from one center to another. This is a factor which does not relate to technologists but affects their job quality.

Analyzing the results of large body and small body patients individually shows that small patients are miscentered by more probability than large patients.



(a) (b)

Fig. 7 (a) The relation between image noise in the lower half of the image and centering error for different phantoms in the 4-slice Light Speed and (b) 64-slice Light Speed CT scanners

Table 2 The results of patients' scout images investigation of site 1 and 2

Imaging site	Average miscentering (cm)	Average percentage of dose increase	Average percentage of noise increase
Site 1	2.1	21.9	6
Site 2	3.7	29.1	10.9

Table 3 The results of patients' scout images investigation of site 3 and 4

Imaging site	Average miscentering (cm)	Average percentage of dose increase	Average percentage of noise increase
Site 3	1.5	17.6	5.3
Site 4	1.7	21.9	6.7

V. CONCLUSIONS

Results show that surface doses are increased, this means that especially anterior organs take more doses. Also the image noise is increased but the effect of miscentering on dose is higher than noise. The comparison of two centers of each scanner shows the difference between the operations of technologists and the signification of their role in imaging process. So they should be strongly encouraged to pay greater care to patient centering.

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