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Patient Vertical Centering and Correlation with Radiation Output in Adult Abdominopelvic CT

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Abstract The purpose of this study was to determine if there is a significant effect, independent of patient size, of patient vertical centering on the current-modulated CT scanner radiation output in adult abdominopelvic CT. A phantom was used to evaluate calculation of vertical positioning and effective diameter at five different table heights. In addition, 656 consecutive contrast-enhanced abdominopelvic scans using the same protocol and automatic tube current modulation settings on a Philips Brilliance 64 MDCT scanner were retrospectively evaluated. The vertical position of the patient center of mass and the average effective diameter of the scanned patient were computed using the reconstructed images. The average volume CT dose index ($CTDI_{vol}$) for each scan was recorded. The mean patient center of mass y coordinate ranged from −3.7 to 6.7 cm (mean ± SD, 2.8 ± 1.2 cm), indicating that patients were on average positioned slightly below the scanner isocenter. There was a slight tendency for smaller patients to be mis-centered lower than larger patients. Average $CTDI_{vol}$ closely fit a quadratic regression curve with respect to mean effective diameter. However, the value of the regression coefficient relating $CTDI_{vol}$ to the patient's vertical position was nearly zero, indicating only a very slight increase in $CTDI_{vol}$ with patient mis-centering for the scanner used in this study. The techniques used here may be useful both for automated evaluation of proper patient positioning in CT and for estimating the radiation dose effects of patient mis-centering for any CT scanner.

Keywords Computed tomography · Radiation dose · Body imaging · Quality control · Image analysis

Introduction

Automatic tube current modulation techniques are widely used in CT to provide predictable image noise in response to variation in patient size, thereby reducing the radiation dose for a desired level of image quality compared to a fixed current technique. These systems are typically dependent on initial localizer radiographs as a basis for modulating X-ray tube radiation output. It has been reported that inappropriate centering of the patient in the CT gantry influences the operation of the current modulation, due to greater magnification of body size when the patient is positioned closer to the tube during the acquisition of a frontal localizer radiograph [1, 2].

However, the effect of patient off-centering on scanner output is expected to vary among scanners, given that different scanner manufacturers use a variety of proprietary and incompletely documented methods for calculating the response of the scanner tube to localizer image data [3]. In addition, systematic evaluation of patient centering in clinical scans has been limited in previous reports. Centering data in prior studies has typically been reported on phantoms of fixed size [2], from localizer images of clinical scans [4, 5] or from table heights of clinical scans [1]. Furthermore, the effect of inaccurate patient positioning on the scanner radiation output in clinical scans is unclear. While one study indicated that volume CT dose index ($CTDI_{vol}$) does not vary significantly with off-centering of phantoms [6], a more recent study suggested

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that there was an increase in $\mathrm{CTDI_{vol}}$ and dose length product (DLP) with vertical mis-centering of patients [1].

The aims of this study were to demonstrate a method for estimating the center of mass from volumetric data in clinical CT scans, to retrospectively calculate both the effective diameter and the vertical centering of patients undergoing abdominopelvic CT studies, and to correlate the centering with the average $\mathrm{CTDI_{vol}}$ produced by the scanner's automated tube current modulation system.

Materials and Methods

Institutional Review Board approval was obtained for the data collection and analysis in this study. A retrospective data set of 656 consecutive contrast-enhanced abdominopelvic scans was obtained on a Philips Brilliance 64 scanner from January to April 2015. The scans were performed with patients in feet-first supine position using constant scanner parameters with a kV of 120 kV, current modulation along the z axis of the patient (Philips Z-DOM), pitch of 0.671, rotation time of 0.5 s, and collimation of 64 × 0.625 mm. The Z-DOM current modulation on the scanner modulates mA along the patient longitudinal axis and is based on a posteroanterior localizer scan, i.e., with the tube positioned beneath the scanner table; no lateral localizer scan was used. The image quality reference parameter on the Philips Brilliance 64 scanner is mAs/slice, specified as 105 mAs/slice for scans in this study. Studies were reconstructed at 2-mm slice thickness with iterative reconstruction (Philips iDose4 level 3). The table height, reconstruction diameter, and center of image reconstruction were at the discretion of the technologist. The scans were performed by seven different CT technologists. As this was a retrospective study, the technologists were not specifically instructed on patient positioning for the acquisition of the image series in this study.

The 656 scans were performed on 578 distinct patients (356 male and 222 female), with 66 patients undergoing two separate studies during the time interval and 6 patients undergoing three studies during the time interval. For patients who underwent more than one study, the time period between consecutive studies ranged from 11 to 112 days (mean ± SD, 64 ± 19 days). Patient age ranged from 18 to 95 years (mean ± SD, 61 ± 14 years).For each image slice of each study, an estimated effective diameter D of the patient was computed by

$$D = 2\sqrt{\frac{\sum_{\{HU(p)>t\}} p_x \cdot p_y}{\pi}} \tag{1}$$

where p_x and p_y are the x and y dimensions of a single pixel and $\{\mathrm{HU}(p) > t\}$ is the set of all pixels p where the Hounsfield unit value exceeds the threshold t. Based on previously reported work, the threshold was set to −500 Hounsfield units [7]. A mean effective diameter over all the slices for each study was calculated.

In addition, for each slice of each study, the y coordinate in millimeters of an in-plane center of mass c_y was calculated by

$$c_y = \frac{\sum_p \left(y_p \cdot f(p)\right)}{\sum_p f(p)} - y_{\mathrm{rot}} \tag{2}$$

where y_p is the y coordinate of pixel p (in mm, with origin at the center of the image and y increasing from the top to the bottom of the image), y_{rot} is the y coordinate of the axis of rotation for the scanner (with origin at the center of the reconstructed image), and f is a function of the Hounsfield unit value of the pixel p. A mean center of mass vertical position over all the slices for each study was calculated. By our convention, positive values for c_y indicate a patient center of mass below the axis of rotation. The geometric relationships among the center of the scanner, the center of the image, and the calculated center of the mass of the patient are illustrated in Fig. 1.

The specification of y_{rot} varies among different CT scanner models and manufacturers. For the Philips Brilliance 64 scanner, y_{rot} is given by the relation

$$y_{\mathrm{rot}} = 255 - \frac{d_{\mathrm{recon}}}{2} - y_{\mathrm{pos}} - y_{\mathrm{table}} \tag{3}$$

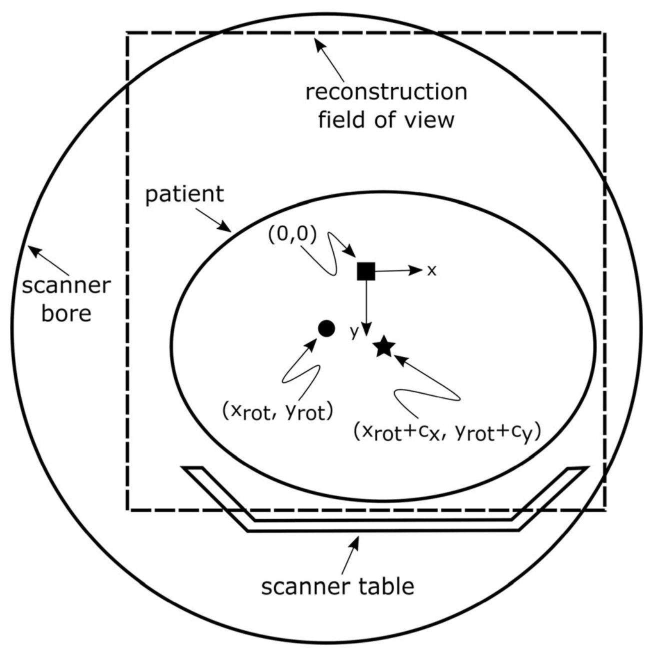


Fig. 1 Schematic of the patient, reconstruction field of view, and scanner bore. *Small square* with coordinates (0,0) represents the center of the reconstruction field of view. *Small circle* with coordinates (x_{rot}, y_{rot}) represents the center of the scanner bore. *Small star* with coordinates ($x_{\mathrm{rot}} + c_x$, $y_{\mathrm{rot}} + c_y$) represents the calculated center of mass of the patient. Note that (c_x, c_y) represents the displacement of the calculated center of mass with respect to the center of the scanner bore

where d_{recon} is the reconstruction diameter of the image given by the "ReconstructionDiameter" DICOM tag (0x0018, 0x1100), y_{pos} is the y coordinate of the patient position given by the "ImagePositionPatient" DICOM tag (0x0020, 0x0032), and y_{table} is the table height given by the "TableHeight" DICOM tag (0x0018, 0x1130). This relation was derived empirically from scanning a QA phantom at varying table heights and reconstructing the resulting images at varying offsets and fields of view.

For $f(p)$, we use as a baseline

$$f(p) = \mathrm{HU}(p) + 1000 \tag{4}$$

so that all pixels contribute positive weighting to the center of mass calculation. In order to assess whether high Hounsfield unit pixels (e.g., from metal or artifact) might disproportionately distort the position of the center of mass y coordinate c_y, we also recalculated the y coordinate using a continuous piecewise linear function with a ceiling parameter m:

$$f_m(p) = \begin{cases} \mathrm{HU}(p) + 1000, & \mathrm{HU}(p) < m \\ m \;\; + 1000, & \mathrm{HU}(p) \geq m \end{cases} \tag{5}$$

where m was assigned values of 0 and 1024 HU. We compared differences among the calculated center of mass y coordinate c_y, $c_y^{(1024)}$, and $c_y^{(0)}$, where the latter two values were computed using ceiling parameters m of 1024 and 0, respectively.

Evaluation of center of mass calculations from reconstructed images was also performed using a Philips QA CT phantom, specifically a 30-cm-diameter "body" component of the phantom composed of nylon with holes containing Teflon and water (Fig. 2). The phantom was scanned on the CT table with the center of the phantom centered visually at the isocenter of the scanner, using the same CT body protocol described above; a separate posteroanterior localizer radiograph was obtained prior to each scan to provide updated input for automatic tube current modulation. Subsequently, the phantom was scanned at table vertical displacements +5, +10, −5, and −10 cm below the estimated isocenter of the scanner. The geometric center of the phantom was then measured manually from the reconstructed images by noting the vertical position of the center of a circular region of interest fitted to the outer surface of the phantom in medical image viewing software [8]. Calculated center of mass y coordinate c_y for the phantom was compared to the measured geometric centers. Calculated effective diameters and $\mathrm{CTDI_{vol}}$ were also recorded.

We also calculated an approximate "corrected" patient center of mass independent of the CT table by manually segmenting the dense pixels of the table on a selected image of a reference study. We translate the set of table pixel y coordinates by the difference in table heights and centers of reconstruction for the patient image and the table reference image to obtain a transformed set of table pixel y coordinates y_t. Accounting for the difference between the pixel dimensions p_x and p_y of the patient image and the pixel dimensions t_x and t_y of the reference image, we subtract appropriately summed transformed Hounsfield unit values $f(t)$ for the table pixels t from the numerator and denominator of the calculation of the center of mass to obtain a corrected patient y position for each image:

$$c'_y = \frac{\sum_p \left(y_p \cdot f(p) \cdot p_x \cdot p_y\right) - \sum_t \left(y_t \cdot f(t) \cdot t_x \cdot t_y\right)}{\sum_p \left(f(p) \cdot p_x \cdot p_y\right) - \sum_t \left(f(t) \cdot t_x \cdot t_y\right)} - y_{\mathrm{rot}} \tag{6}$$

A corrected mean center of mass vertical position over all the slices for each study was calculated. Note that this correction does not account for truncation of the table from the reconstructed field of view for low table positions.

All computational image processing unless otherwise specified were performed using custom scripts written in Python 2.7.10 [9] with the numerical Python (NumPy) extension [10]. Regression analyses were performed using the R statistical computing environment, version 3.2.1 [11].

Results

Scans of the quality assurance phantom at five different table heights were used to assess the calculation of center of mass based on reconstructed CT images. The calculated center of mass position below the isocenter vs. the distance between the manually measured geometric center of the phantom and the isocenter for identical slice positions in each scan is given in Table 1. There is relatively close agreement of the calculated and measured distances, with increased discrepancy when the phantom is higher than the isocenter (negative positions in Table 1). The error is related both to the inclusion of the entire table in the center of mass calculation (Fig. 2c–e) and the partial exclusion of the top of the phantom from the field of view at the highest displacement (Fig. 2e). Correction of the center of mass position by the table subtraction method in Eq. (6) reduces the error for these table positions but increases error when the table is partially excluded from the field of view (Fig. 2a, b). Of note, the calculated effective diameters and reported $\mathrm{CTDI_{vol}}$ values at the assessed slices did not change significantly with table position (Table 1).

The calculated mean effective diameters of the patients ranged from 21.3 to 44.8 cm (mean ± SD, 29.9 ± 3.7 cm). The mean $\mathrm{CTDI_{vol}}$ values for the studies ranged from 3.4 to 24.2 mGy (mean ± SD, 8.2 ± 3.8 mGy).

The mean center of mass y coordinate c_y (including the table) ranged from −3.7 to 6.7 cm (mean ± SD, 2.8 ± 1.2 cm), where positive values indicate the center of mass below the axis of rotation of the scanner; the results indicate that on average, the center of mass was positioned slightly below the axis of rotation (Fig. 3). This result held when the

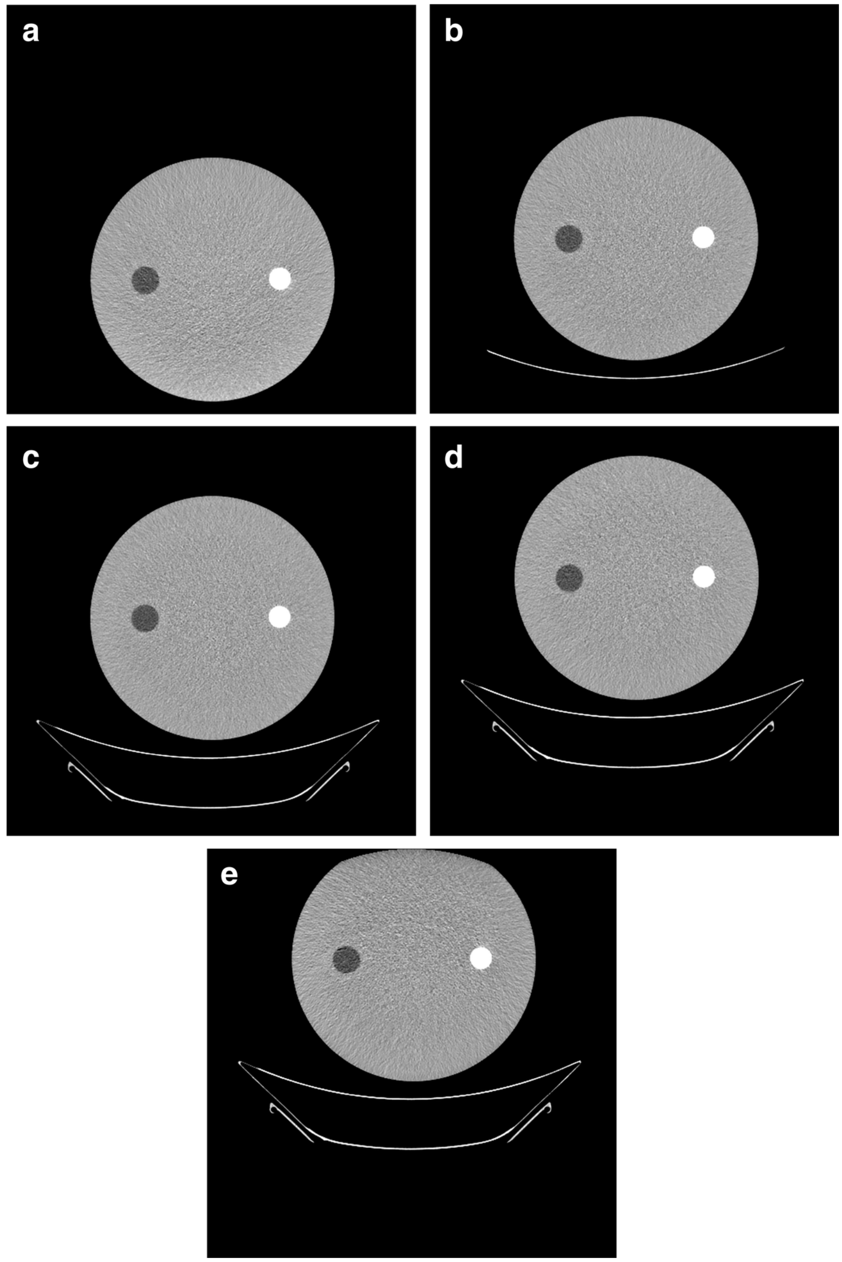


Fig. 2 Scans of the body component of a Philips QA phantom with the center of the phantom at the following five different heights with respect to the scanner isocenter: **a** 8.5 cm below scanner isocenter, **b** 3.5 cm below isocenter, **c** 1.7 cm above isocenter, **d** 6.5 cm above isocenter, and **e** 11.4 cm above isocenter. Note complete or partial omission of the table from the reconstructed field of view in **a** and **b** and partial omission of the top of the phantom in **e**

center of mass y coordinate was recalculated with ceiling parameters of 1024 and 0 HU as described in the "Materials and Methods" section, suggesting that high-attenuation materials or artifacts in the scan did not significantly affect the vertical position calculation. The mean difference $(c_y^{(1024)} - c_y)$ ranged from −0.2 to 3.3 mm (mean ± SD, 0.0 ± 0.0 mm). The mean difference $(c_y^{(0)} - c_y)$ ranged from −1.2 to 0.6 mm (mean ± SD, −0.3 ± 0.3 mm).

There was a slight tendency for smaller patients to be mis-centered lower than larger patients. Regression analysis of c_y (mm below center of rotation) vs. the mean effective diameter d (cm) supports this trend with a regression line of $c_y = 59.9 - 1.1d$ and $p < 0.001$, but the adjusted R^2 was only 0.10.

We also calculated a corrected center of mass y coordinate c_y' for each study using an approximate correction for the density of the table as described in the "Materials and Methods" section. The difference between corrected and uncorrected vertical positions is approximated by a simple affine transform, as shown in Fig. 4. A regression line gives $c_y' = 0.97c_y - 8.3$ mm with $p < 0.001$ and adjusted R^2 of 0.99; in other words, within the range of table heights for patients in this study, a correction for the table can be approximated by subtracting 8 mm from the calculated aggregate center of mass y coordinate (including the table). However, for the remainder of the calculations in this study, we use the "uncorrected" center of mass, since (to our

Table 1 Calculated center of mass (with table and after table subtraction), effective diameter, and $CTDI_{vol}$ as a function of vertical displacements of a 30-cm body phantom

Phantom vertical position (mm)	Calculated center of mass (mm)	Calculated center of mass after table subtraction (mm)	Effective diameter (cm)	$CTDI_{vol}$ (mGy)
−114	−91	−101	30.4	8.3
−65	−49	−58	30.9	8.3
−17	−3	−12	30.9	8.3
35	39	29	30.3	8.6
85	82	71	30.1	8.4

Positions are given in terms of distance in millimeters below the scanner isocenter; phantom vertical position is based on table height displacements from visual centering at 0 mm (scanner isocenter)

knowledge) the automated tube current modulation system does not correct for the presence or density of the table.

For the 72 patients who had at least two studies during the time period, the mean absolute difference in calculated mean effective diameters for the first two studies ranged from 0.01 to 2.02 cm (mean ± SD, 0.54 ± 0.47 cm) (Fig. 5a). The mean absolute difference in mean $CTDI_{vol}$ for the first two studies was generally small, ranging from 0.01 to 3.69 mGy (mean ± SD, 0.48 ± 0.64 mGy) (Fig. 5b). The mean absolute difference in vertical center of mass position ranged from 0 to 58 mm (mean ± SD, 13 ± 12 mm) (Fig. 5c). An example of a patient imaged at two different vertical positions is shown in Fig. 6.

A scatter plot demonstrates a clear curvilinear relationship between mean $CTDI_{vol}$ vs. mean effective diameter (Fig. 7). However, no clear relationship is seen between the mean $CTDI_{vol}$ and the center of mass y coordinate (Fig. 8). In fact, without accounting for patient body size, it would appear from the plot that there might be an inverse relationship between patient positioning below the scanner isocenter and patient dose.

A multiple regression analysis was performed to determine mean effective diameter and mean vertical positioning predicted mean $CTDI_{vol}$ for an abdominopelvic CT study. Based on the curvilinear form of the $CTDI_{vol}$ vs. mean effective diameter scatter plot, a quadratic term for mean effective diameter was added to the model. The results of the regression analysis are provided in Table 2; the value of the adjusted R^2 was 0.97. Although the regression coefficient for the center of mass vertical position was statistically significant, the value of the coefficient was nearly zero, indicating a practically negligible increase in $CTDI_{vol}$ with increasing distance of the patient center of mass below the center of rotation of the scanner.

Discussion

In this work, our primary aim was to assess the relationship between patient vertical position and the scanner radiation output as reported in the form of $CTDI_{vol}$. The main motivation for this work came from preliminary observations with both patient and phantom data that did not support a

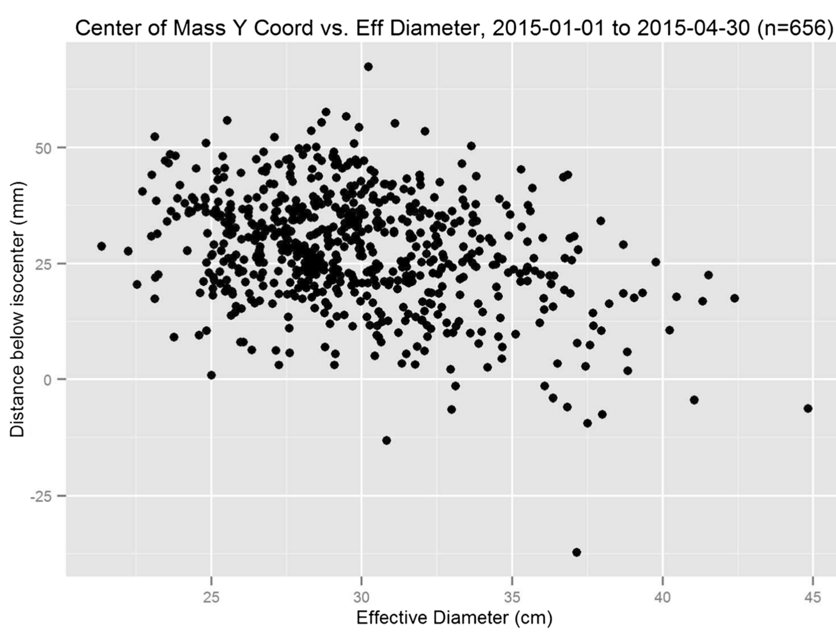


Fig. 3 Calculated distance of the patient's center of mass below the isocenter plotted against the calculated mean effective diameter of the patient, from 656 image series from CT scans of the abdomen and pelvis. On average, patients were positioned below isocenter, with a slight tendency toward greater mis-centering with smaller-diameter patients

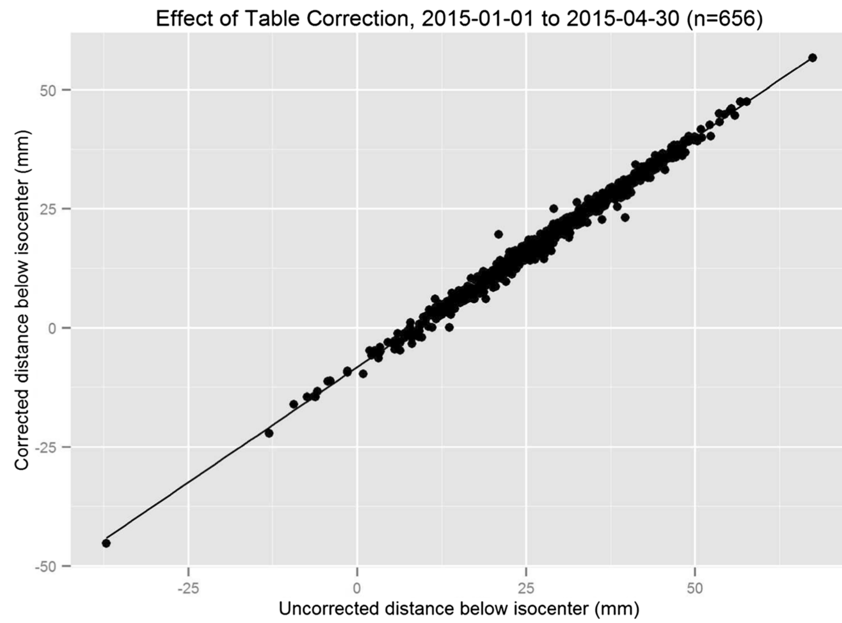


Fig. 4 Scatter plot of center of mass corrected by subtraction of the table vs. uncorrected center of mass. *Black line* represents the linear regression fit to the data

significant change in $CTDI_{vol}$ with changes in table vertical position on the Philips Brilliance 64 CT scanner. In order to more systematically evaluate patient vertical position in retrospective patient data, we calculated centers of mass from reconstructed images as detailed in the "Materials and Methods" section. We also retrospectively calculated the patient effective diameter using the reconstructed images, since we expect patient diameter to account for much of the variation in radiation output due to automatic tube current modulation.

Consistent with findings in previous reports [2, 4, 5], we found a tendency by our technologists to center the patient below the center of axis of the scanner in our study population. In addition, as has been reported previously [4], there is a slight tendency toward greater centering error in smaller patients.

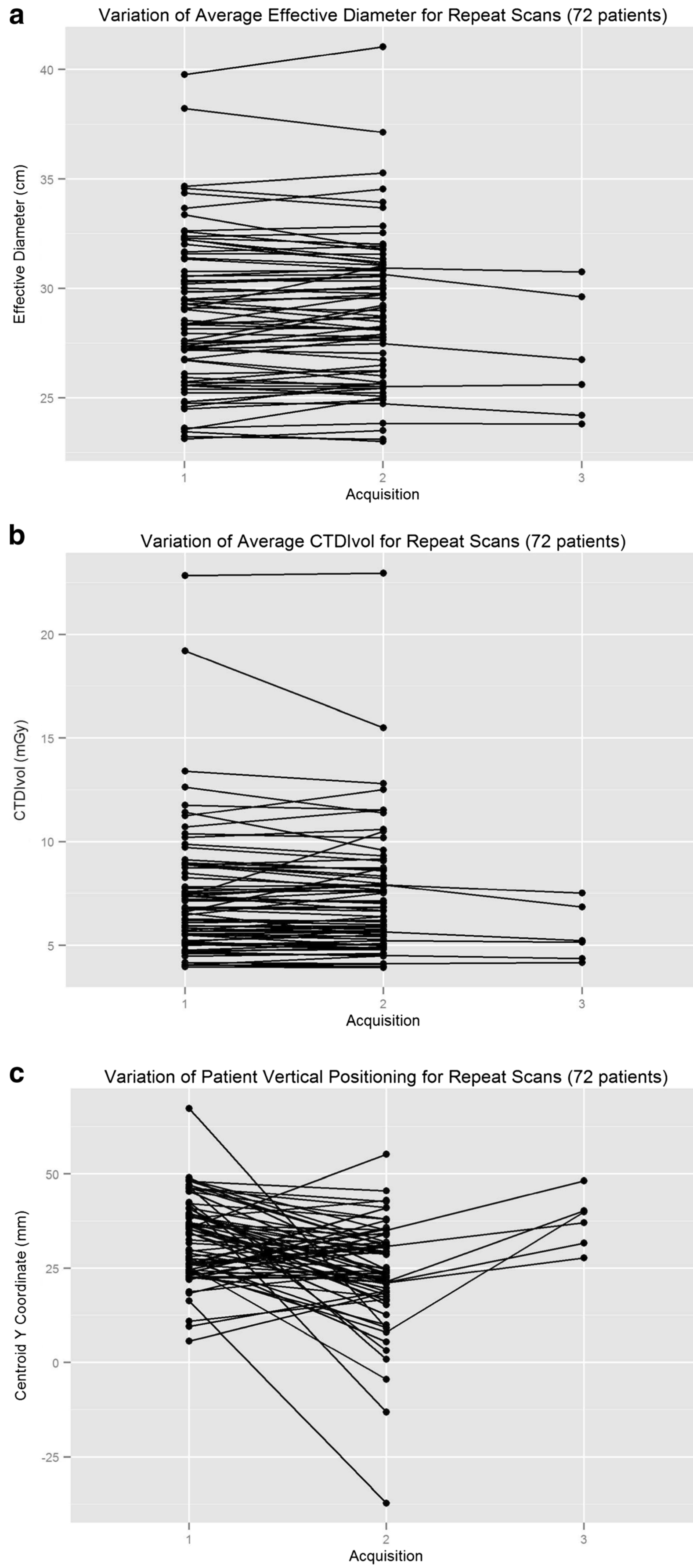
We found that although there was a statistically significant correlation between patient centering and $CTDI_{vol}$, the actual value of the regression coefficient was practically negligible, as most of the variance in $CTDI_{vol}$ produced by the automatic tube current modulation system could be explained by the effective diameter of the patient. In the subset of 72 patients who had more than one scan, there was a relatively small variation in $CTDI_{vol}$ compared to the variation of patient vertical positioning. This lack of significant variation of the average $CTDI_{vol}$ with patient mis-centering supports findings from phantom studies by Li et al. on a GE LightSpeed VCT scanner [6]. Although a more recent paper reported correlations of $CTDI_{vol}$ or DLP with patient mis-centering [1], the strength of these relationships was not reported through regression coefficients.

It is likely that our observations of insensitivity of $CTDI_{vol}$ with respect to patient centering are primarily attributable to the particular proprietary implementation of tube current modulation in the Philips Brilliance 64 CT scanner. Unpublished data suggests that the sensitivity of tube current modulation to patient vertical position varies significantly among vendors, with Philips scanners among the least sensitive (Rong et al., cited in [12]). It should be noted, however, that the lack of change of $CTDI_{vol}$ with patient vertical centering on the Philips Brilliance 64 CT scanner does not imply that patient centering is unrelated to patient absorbed dose. $CTDI_{vol}$ is a standardized measure of the radiation output of a CT scanner without reference to the patient's size, organ positions, and body composition [13]. Phantom studies have shown that, particularly with the use of prepatient bow tie X-ray filters in the CT scanner, patient mis-centering increases surface CTDI measurements in a manner that is not measured in volume CTDI measurements [6]. Since surface CTDI is not routinely measured in clinical CT, retrospective quantitation of patient mis-centering may provide an indirect measure of excess patient surface radiation not reflected in routine $CTDI_{vol}$ or DLP reporting.

Furthermore, several studies have used phantom data to demonstrate significant increases in image noise with mis-centering [2, 4–6, 14, 15]. Changes in image noise with patient position were not specifically evaluated in this study, as image noise is less amenable to systematic reproducible quantitation from clinical images. However, the inherent difficulty of automatically and reliably quantifying image noise in reconstructed CT images supports a role of quantitation of mis-centering for imaging quality control and improvement.

For most of the analysis performed in this study, we included the table in the center of mass calculations, since to our knowledge, the automated tube current modulation does not include a correction for table density. Prior work evaluating the patient centroid from the localizer radiograph suggests that

Fig. 5 Variation among repeated scans for the same patient, for 72 patients: **a** relatively small variation in patient effective diameter over repeated scans, **b** relatively small variation in average $CTDI_{vol}$ over repeated scans, and **c** larger variation in vertical position over repeated scans



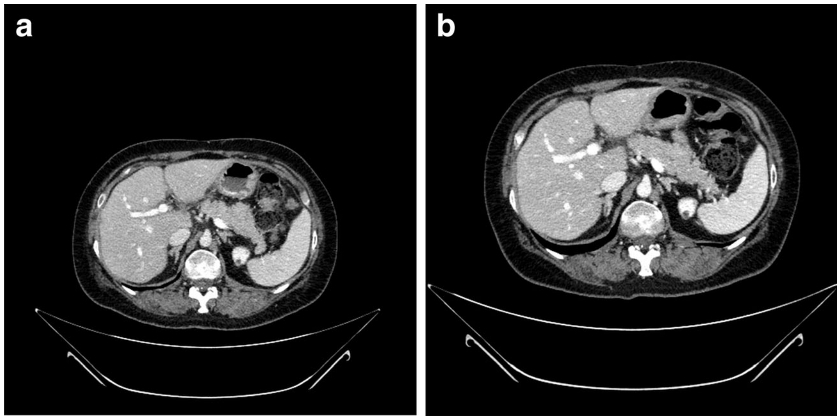


Fig. 6 Fifty-two-year-old female with two postcontrast CT images obtained 72 days apart. In each image, the center of rotation of the scanner is at the center of the reconstructed image. **a** Volumetric centroid was calculated as 4.2 cm below the center of rotation of the scanner, average $\mathrm{CTDI_{vol}}$ 8.75 mGy, average effective diameter 30.8 cm. Reconstructed field of view is 50 cm. **b** Volumetric centroid was calculated as 1.3 cm above the center of rotation of the scanner, average $\mathrm{CTDI_{vol}}$ 8.74 mGy, average effective diameter 30.8 cm. Reconstructed field of view is 42 cm

estimation errors without table correction would be larger with smaller-diameter patients [4]. We found that within the range of patient sizes of this study, subtraction of the table density could be approximated by a constant affine transform (not accounting for table truncation by the field of view). The transform would need to be recalibrated from one scanner table to another, but the overall magnitude of the offset of the transform is relatively small (~8 mm). Furthermore, as demonstrated in our phantom measurements, the transform does not improve accuracy of the patient center of mass calculation for low table positions, when the table is partially excluded from the field of view.

In practice, technologists center patients on the scanner based on an approximation of the geometric center of the patient. We evaluated center of mass of patients and phantoms in this work, rather than geometric centers; one reason for this is computational efficiency, as the center of mass calculation does not require segmentation of the patient from the scanner table. We believe that the center of mass and geometric center are similar in most patients, but this should be demonstrated more clearly in future work. A more important question to answer is whether center of mass or geometric center is more appropriate as a centering target in order to maximize image quality for a level of patient dose. The answer likely depends on the specific details of the tube modulation algorithm used in a given scanner.

Since the scanner used in this study does not use lateral localizer radiographs as input for the automated tube current modulation algorithm, we were unable to compare our results with prior calculations of mis-centering based on

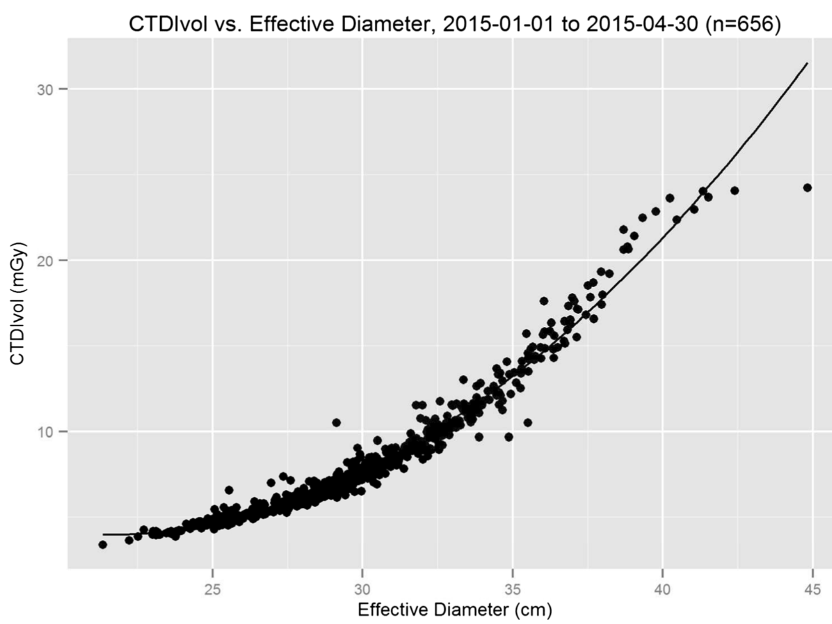


Fig. 7 Scatter plot of $\mathrm{CTDI_{vol}}$ vs. patient effective diameter. *Black line* represents the quadratic regression fit to the data

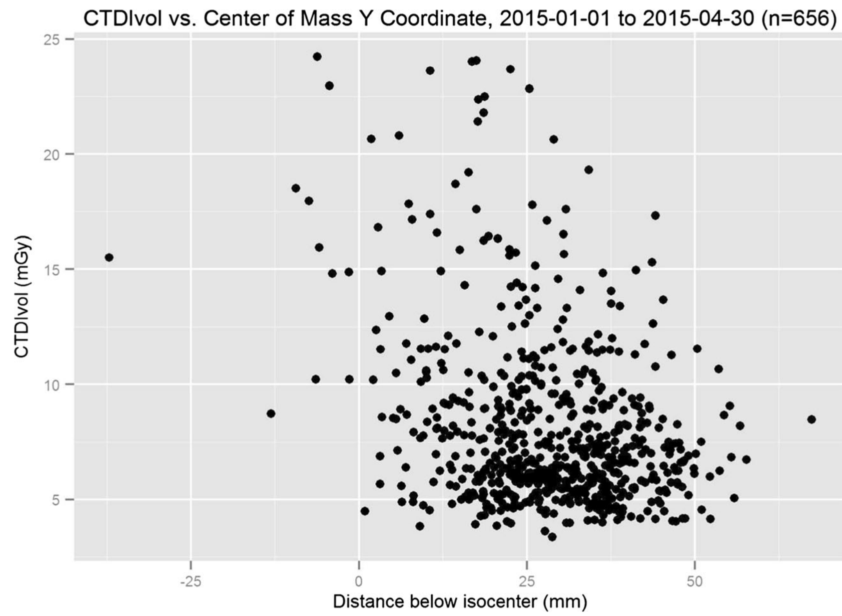


Fig. 8 Scatter plot of $CTDI_{vol}$ vs. patient vertical distance below the isocenter. The apparent slight trend of decreased $CTDI_{vol}$ with increased patient positioning below the isocenter in this plot is not seen after accounting for variation in patient effective diameter; specifically, multiple regression demonstrates a slight increase in $CTDI_{vol}$ with patient distance below isocenter when patient effective diameter is included in the model

polygonal regions of interest from the scout radiograph [5]. An advantage of the centering assessment method used in this study is that it should be generalizable to any scanner regardless of the availability of lateral localizer radiographs. We suspect that calculation of the center of mass from volumetric data would be more accurate than calculation from localizer radiographs, although definitive demonstration may require evaluation of asymmetric phantoms with known center of mass.

Although horizontal centering was not specifically reported in this study, it is straightforward to extend the technique for measurement of patient vertical position to also assess horizontal position of the patient. We have assessed horizontal centering as a potential quality improvement focus in our institution and confirmed that no significant bias in patient position either to the left or the right was present (data not shown). Again, as noted above, since no lateral localizer radiograph was used for the automated tube current modulation, we do not expect a horizontal shift in patient position to result in significantly altered magnification that might produce a change in the current modulation.

There were several limitations of this study. Since data for patient position was obtained from the reconstructed field of view, calculation of both effective diameter and patient centering is limited when the patient's body size is large enough to exceed the field of view. In such cases, the localizer radiograph may provide the best assessment of patient size and centering. However, we expect that the effect of patient positioning on radiation dose would be more pronounced in small-sized patients compared to large-sized patients.

Furthermore, since this was a retrospective study on adult patients, we were limited in the range of patient diameters and vertical positions available for analysis. With respect to patient sizes, we do not include patient effective diameters below 20 cm, a range that is clinically relevant in the pediatric population. Future evaluation should include this population, as it is possible that changes in $CTDI_{vol}$ with patient mis-centering are more pronounced with smaller patient sizes. In addition, even in our adult population, it is possible that the degree of mis-centering error was too small in this population to detect corresponding changes in $CTDI_{vol}$.

Table 2 Regression coefficients for prediction of $CTDI_{vol}$ as a function of patient effective diameter and vertical positioning (center of mass distance below isocenter)

Variable	Coefficient	Standard error	*t* value	*P* value
Effective diameter (cm)	−2.2	0.08	−26.5	<0.001
(Effective diameter)^2	0.05	0.001	38.2	<0.001
Center of mass *y* position (mm)	0.008	0.002	3.8	<0.001
Intercept	27	1.27	21.5	<0.001

Although both effective diameter and center of mass vertical position correlate with $CTDI_{vol}$, the regression coefficient for the vertical position is small in magnitude

A single circular QA phantom was used to evaluate the calculation of patient center of mass, as well as the dependency of $CTDI_{vol}$ with vertical positioning. Scans of noncircular phantoms such as anthropomorphic phantoms may be helpful for further verification of the appropriateness of the center of mass calculation. Furthermore, a more controlled evaluation of the dependency of scanner output with patient vertical positioning might be accomplished with systematic positioning of anthropomorphic phantoms, although such evaluation is limited by the available size and shape range of the phantoms.

A final but important limitation is that the study was only performed on a single CT scanner. As noted above, we expect the behavior of the automated tube current modulation system to vary among scanner manufacturers according to the specific details of each proprietary modulation algorithm. It is likely, given previously reported findings of variation of radiation output with vertical positioning, that other systems have a more pronounced variation in $CTDI_{vol}$ with changes in patient position. Evaluation of the radiation output of other CT scanners as a function of vertical positioning is a logical and necessary extension of the current work; the methods we used for calculation of patient effective diameter and center of mass are generic and should be readily generalizable across CT scanner models.

Conclusions

In summary, we present a simple means of calculating both mean patient effective diameter and vertical center of mass position of patients from reconstructed CT images. Although we saw an overall tendency to position the center of mass of the patient below the center of rotation of the scanner, the mean patient effective diameter accounted for most of the variation of mean $CTDI_{vol}$ among the scans in the study. Calculations of vertical patient position may nevertheless be of value by indirectly reflecting surface CTDI variability and image noise for the purposes of radiation dose reduction and image quality improvement.

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