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FULL PAPER

Establishing how patient size and degree of miscentring affect $CTDI_{vol}$, using patient data from a dose tracking system

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Objectives: The work aimed to investigate and model the relationship between CT dose index ($CTDI_{vol}$), patient size and miscentring metrics using data extracted from a dose tracking software tool. Further, using the established model we aimed to extract and estimate further AEC metrics and, finally, demonstrate how the $CTDI_{vol}$ may vary across a range of imaging exams and with variations in patient size, positional metrics and with scan parameters.

Methods: CT scan parameters, patient dose metrics, patient positioning information and patient and water equivalent diameter information was extracted from abdomen, thorax-abdomen-pelvis and thorax imaging exams acquired on a 128-slice Siemens Somatom Definition AS + CT system using dose tracking software over a 1 year period. A miscentring factor, accounting for the magnification of the patient due to miscentring

was derived. Significant predictors (p -value < 0.001) of the $CTDI_{vol}$ were considered as inputs into the model following regression analysis.

Results: The model was capable of describing the $CTDI_{vol}$ and its variation with patient miscentring and patient size. AEC parameters, such as the reference $CTDI_{vol}$, reference diameter and the AEC strength were estimated. Further, the model allowed for comparisons of how changes in scan settings, such as kVp, pitch and slice thickness affected $CTDI_{vol}$.

Conclusions: We demonstrate the use of clinical data, extracted from dose tracking software, to investigate and monitor AEC behaviour and CT output.

Advances in knowledge: The presented model allows for the remote investigation of AEC behaviour using patient data.

INTRODUCTION

The practice of modulating tube current in CT to optimize patient dose and image quality has been in widespread use since its introduction in the early 2000s.¹ The technique operates by adjusting tube current output along the z -axis and through the x - and y -axes according to estimates of patient attenuation.² A localizer radiograph, a two-dimensional X-ray image acquired in the anteroposterior (AP), postero-anterior (PA) or lateral tube positions, is typically initially acquired and used to estimate the subject attenuation and/or size which then informs on the tube current modulation level for the subsequent CT scan (12). It has previously been established that patient miscentring during localizer radiograph acquisition, due to suboptimal positioning of the patient distant from the isocentre, results in the misinterpretation of the patient size and attenuation properties and subsequently leads to degradation in image quality and poorer radiation dose optimization (3, 4, 7, 13, 12). A previous study modelling surface radiation dose and image noise for a 320 mm CT

dosimetry phantom has demonstrated that phantom off-centring by 30 and 60 mm resulted in an 18 and 41% increase in surface dose, respectively, with image noise also increasing by 6 and 22%.³ Miscentring an anthropomorphic phantom 40 mm below isocentre has been shown to result in a 16 and 24% increase in breast and thyroid dose, respectively.⁴ In this case, elevated radiation doses to the breast and thyroid gland were attributed to an increase in absorbed dose when the projected X-rays passed through the thinnest and least attenuating part of the bowtie filter during the scan rotation.⁴ Moreover, an investigation utilizing a 320 mm CT dosimetry phantom in a 64-slice CT scanner reported the evaluated maximum increases in surface dose to be 13.5%, 33.3% and 51.1% when miscentering the phantom by 20, 40 and 60 mm, in the direction closer to the X-ray tube.⁵ While numerous studies have demonstrated the effect of subject miscentering under standardized phantom conditions, the extent to which miscentring affects CT radiation output in the clinical setting has not been evaluated.³⁻⁷

Table 1. CT imaging parameters for each protocol, with the regression coefficients of the relationship between the measured and modelled CTDI_{vol}

Protocol	TAP with contrast			Abdomen		Thorax	
	A	B	C	A	B	A	B
Number of CT scans (n)	149	520	972	211	584	240	211
kVp	100	100	120	100	120	120	120
Slice thickness (mm)	5	1	1	1	1	1	1
Pitch	0.6	0.6	0.6	0.6	0.6	0.6	1.2
R ²	0.79	0.83	0.79	0.73	0.74	0.88	0.69
Model CTDI _{vol} (D _{ref}) (mGy)	16.8	18.4	27.5	20.5	14.7	27.6	30.9
Model D _{ref} (mm)	315	313	308	313	272	305	327
Model strength (S)	0.013	0.014	0.016	0.015	0.015	0.016	0.015

CTDI, CT dose index; TAP, thorax–abdomen–pelvis.

Radiation dose tracking software packages have recently become available from a number of vendors offering the ability to collate large quantities of imaging and radiation dose-related data, and to facilitate their analyses (9, 10). Accordingly, using clinical data collated using a dose tracking package, this study sought to investigate the relationship between the CT dose index (CTDI_{vol}) and the patient size and miscentring metrics. This study aimed to model this relationship, extract and estimate further AEC metrics and, finally, demonstrate how CTDI_{vol} may vary across a range of exams and with variations in patient size, positional metrics and with scan parameter.

METHODS

CT examinations were carried out using a 128-slice CT system (Somatom Definition AS+, Siemens Healthcare, Forchheim, Germany) in a single tertiary referral centre. AP CT localizer radiographs were acquired at 120 kV and 35 mA prior to the CT, which was then acquired in helical mode using the CARE-Dose 4D AEC system (CARE Dose 4D, Siemens Healthcare) and a flat filter. Table 1 details the acquisition parameters used for abdomen, thorax–abdomen–pelvis and thorax imaging exams.

All patients who had undergone CT scans between June 2014 and June 2015 as per the protocols outlined in Table 1 were included in this study. CT scan parameters, patient dose metrics and patient positioning information such as CTDI_{vol} (mGy) and dose–length product (mGy.cm), table height (mm), and patient centroid to isocentre distance (mm) in the *x* and *y* directions were extracted from the CT scans using the Bayer Radimetrics™ dose tracking software. The system performs analysis on all axial slices in each CT scan to calculate the patient centroid, patient effective diameter (PED) and water equivalent diameter (WED). Patient centroid in this context is analogous to the centre of mass of the axial slice. The Hounsfield unit of each pixel in a slice is converted into a material density which is used to calculate the centre of mass of the body cross-section. The dose tracking system calculates the WED as per the method outlined in AAPM report 204 and 220.^{8,9}

In order to validate the data recorded by the dose tracking software, an empirical study was carried out using a 320 mm CTDI dosimetry phantom to establish the relationship between subject miscentring and CT radiation output, and that recorded by the dose tracking software. CTDI_{vol} was calculated by taking measurements at four peripheral and one centre location in the phantom following the acquisition of a 1 cm axial CT scan. All measurements were taken using an X2 CT 100 mm pencil chamber (Raysafe Billdal, Sweden). CTDI_{vol} measurements were carried out at isocentre and then repeated at 30 and 60 mm above and below isocentre, achieved by varying the bed height. A measurement of the phantom's diameter, recorded from the topogram image was also measured using the digital callipers measuring tool available on the system console. This was carried out for each table position and compared with the actual physical phantom diameter dimension of 320 mm, in order to quantify the magnitude of magnification or minification as visualized on the CT localizer radiograph as a result of miscentring.

The CareDose 4D AEC system estimates the subject size and attenuation from the localizer radiograph, compares this with a standard sized 75 kg reference patient and subsequently estimates an effective mAs for the scan. During the CT scan, the tube current (and therefore the effective mAs) is automatically adjusted further to compensate for variations in patient size and attenuation. This system provides, for different sized patients, image quality consistent with that obtained using the quality reference mAs level for a standardized patient. The magnitude of the adjustment (the mAs correction factor) for different sized patients has previously been derived as (14):

$$\frac{\text{Effective mAs}}{\text{Quality Reference mAs}} = e^{(D - D_{\text{Ref}}) \cdot S} \quad (1)$$

where *D* is the calculated patient-effective diameter, *D_{Ref}* is the diameter of the reference phantom and *S* is the CARE Dose4D strength, with the last two parameters both assumed to be constant for this work. The CARE Dose4D strength refers to the extent of change of mAs and can be set to very weak, weak, average, strong, or very strong. PED and WED values for each

respective patient were available from the dose tracking software. Given that CTDI_{vol} is linearly related to effective mAs, an expression for CTDI_{vol} can also be derived as (14):

$$\frac{CTDI(D)}{CTDI(D_{Ref})} = e^{(D-D_{Ref}) \cdot S} \quad 2$$

A derived miscentring factor (M) was defined as the ratio of the estimated patient projection angle (A) onto the image receptor to the projection angle that would occur were the subject to be positioned at isocentre. The distance between the X-ray tube (x_{tube} , y_{tube}) and the patient centroid (x_{mis} , y_{mis}), as reported by the dose tracking software and assumed for derivation purposes to be miscentred, can be calculated by using the distance formula between two points given in Eqn 3:

$$D_{tube \text{ to centroid}} = \sqrt{(x_{mis} - x_{tube})^2 + (y_{mis} - y_{tube})^2} \quad 3$$

For the Siemens Somatom AS+, the coordinates of the X-ray tube for the AP localizer were set as (0, 595 mm), while isocentre as reported by the dose tracking software is (0, 0). The estimated patient projection angle A can then be calculated using Eqn 4:

$$A = 2 \cdot \sin^{-1} \left(\frac{\frac{WED}{2}}{\sqrt{(x_{mis} - x_{tube})^2 + (y_{mis} - y_{tube})^2}} \right) \quad 4$$

with the projected angle, where the patient is positioned at isocentre, given as:

$$A_{centred} = 2 \cdot \sin^{-1} \left(\frac{\frac{WED}{2}}{D_{ti}} \right) \quad (5)$$

where D_{ti} is the X-ray tube to isocentre distance. M , as detailed in Figure 1, is then the ratio of the projected patient angle where miscentred, as compared with what it would be if the patient were centred correctly.

It was hypothesized that any miscentring of the patient would lead to a change of the diameter as interpreted by Caredose 4D during localizer acquisition. Accordingly, a logarithmic regression analysis was carried out to establish the relationship between the natural log of CTDI_{vol} and a number of parameters including the water equivalent diameter (WED), centroid to isocenter distance, patient equivalent diameter, the derived miscentring factor (M) and also the product of the derived M parameter with both WED and PED, respectively. The latter products were investigated to establish whether the magnification of the WED or PED due to miscentring were significant predictors of the CTDI_{vol}. Regression analysis was carried out using the data analysis and regression function in Microsoft Excel (2010), with a significant predictor of $\ln(CTDI_{vol})$ considered for a p -value < 0.001. Parameters such as kVp, reference mAs, rotation time, pitch and slice thickness were kept constant for each imaging protocol investigation, as detailed in Table 1, as any variation in these would result in a change in the CTDI_{vol} unrelated to the changes in patient diameter or centroid to isocentre distance.

Figure 1. Miscentring factor estimates as the patient projection angle (A) onto the detector, relative to that of the subject positioned at isocentre, where D_{ti} is the X-ray tube to isocentre distance, (x_{mis} , y_{mis}) are the coordinates of the patient centroid as reported by the dose tracking software and (x_{tube} , y_{tube}) are the coordinates of the X-ray tube during the localiser acquisition. WED, water equivalent diameter.

$$M = \frac{\sin^{-1} \left(\frac{\frac{WED}{2}}{\sqrt{(x_{mis} - x_{tube})^2 + (y_{mis} - y_{tube})^2}} \right)}{\sin^{-1} \left(\frac{\frac{WED}{2}}{D_{ti}} \right)}$$

Figure 2. Relative change in measured and system-estimated $CTDI_{vol}$, in addition to relative change in the measured diameter and miscentring factor, with changes in miscentring. CTDI, CT dose index.

RESULTS

The results of the phantom data are shown in Figure 2, where relative changes in the measured $CTDI_{vol}$, system-estimated $CTDI_{vol}$, measured phantom diameter and miscentring factor are plotted against miscentring distance. The absolute measured and system-estimated $CTDI_{vol}$ showed excellent correlation ($R^2 = 0.995$) and a percentage difference of <6.8% across the measured range, while similarly the derived miscentring factor and the measured relative change in phantom width also showed excellent correlation ($R^2 = 0.993$) with a percentage difference of <6%

across the measured range. The centroid to isocentre distance, as measured by the dose tracking software, was also validated.

The data collated by the CT Dose tracking system offered insight into patient positioning practices. Figure 3 shows a histogram analysis of miscentring in the case of the "Thorax A" imaging protocol, as detailed in Table 1, exhibiting patients to be positioned off-centre by an average of -146 mm. Indeed, Figure 4 illustrates for the same "Thorax A" protocol how a linear relationship between patient size (WED) and vertical

Figure 3. Relative frequency of miscentring in 10 mm bins for the "Thorax A" imaging protocol.

Figure 4. Vertical patient miscentring plotted against the water equivalent diameter, for "Thorax A" imaging protocol.

miscentred distance was related to patient WED. A similar linear trend was observed for all other exams investigated in this work.

For each imaging protocol examined, the regression analysis identified the most significant predictor of $\ln(\text{CTDI}_{\text{vol}})$ to be the product of the M correction factor and the WED, leading to the relation shown in eqn. 6, based on eqn. 2.

$$\text{CTDI}_{\text{vol}} = \text{CTDI}_{\text{vol}}(D_{\text{Ref}}) e^{(M \cdot \text{WED} - D_{\text{Ref}}) \cdot S} \quad (6)$$

The solver tool in Microsoft® Excel was employed to estimate the $\text{CTDI}_{\text{vol}}(D_{\text{Ref}})$, D_{Ref} and S values that led to a minimum Pearson's χ^2 Chi-squared goodness of fit value for each respective imaging protocol (Table 1).

Examples of the modelled CTDI_{vol} (mGy) values and actual data for the "Thorax A", "Abdomen B" and "TAP C" protocol against the WED are shown in Figure 5(a),(b) and (c). The patient data represent an average and standard deviation (error bars) for 20 mm WED bins along with the Modelled CTDI_{vol} against WED. The model also accounted for the trend of an increased degree of patient miscentring with increased patient size.

Having established the optimum model fit parameters for each imaging protocol and also the miscentred range, it was possible to simulate how changes in miscentring, patient size and AEC strength affected CTDI_{vol} . For instance, Figure 6 illustrates for a standard patient of WED 320 mm how miscentring for the various protocols affected CTDI_{vol} .

Figure 7 demonstrates, for a range of patient sizes, the extent to which miscentring affects CTDI_{vol} for the "Thorax A" protocol data. Figure 8 illustrates how the modelled CTDI_{vol} varies with a simulated variety of strength (S) values of 0.06, 0.1, 0.148 and 0.17 representing weak, average, the average modelled in this study and also constant noise.¹⁰ In this study, a standard-sized patient (WED = 320 mm) was modelled with a $\text{CTDI}_{\text{vol}}(D_{\text{Ref}})$ of 12 mGy at isocentre across a miscentring range of 100 to -100 mm, for illustrative purposes.

DISCUSSION

The literature to date has largely focussed on phantom studies when examining how CT AEC systems interpret patient size as a result of miscentring and, ultimately, how this affects the radiation output.³⁻⁵ In this study, we have carried out some phantom studies to validate the dose tracking software data and our models, followed by the presentation of clinical data offering an insight into CT patient positioning practices within a busy scanning environment. Of particular note is the pattern of patient positioning below the isocentre. While this trend has been observed previously,^{5,11} the results presented in this work suggest that across all scans there was a direct linear relationship between patient size and the extent to which the patient was miscentred below the isocentre. To the authors' knowledge, this has not been highlighted previously. Indeed, as a counterpoint the literature suggests there to be an increased probability of miscentring for smaller rather than larger patients,⁵ albeit, patient miscentring had typically been observed below the isocentre, in agreement with this study. For larger patients an increased miscentring would manifest as a slightly decreased magnification in an AP localizer^{6,7} leading to a reduced CTDI_{vol} . This may appear

Figure 5. (a-c) Average and standard deviation (error bars) of patient data for 10 mm WED bins, and Modelled $CTDI_{vol}$ for the "Thorax A" (a), "Abdomen B" (b) and TAP C (c) protocol against WED. CTDI, CT dose index; TAP, thorax-abdomen-pelvis; WED, water equivalent diameter.

Figure 6. Comparison of change in CTDI_{vol} with patient miscentring for the different protocols. CTDI, CT dose index.

advantageous; however, positioning a subject distant from the isocentre generally affects an increase in image noise, leading to poorer image quality and poorer radiation optimization. Further, organ doses may change significantly with miscentring, with increases in breast and thyroid dose previously described.⁴

Given the significant effect that patient positioning may have on the CT radiation output,³⁻⁵ this work has put forward a simple miscentring correction which was incorporated into a previously derived model for estimating CTDI_{vol} based on patient size.¹⁰ The miscentring correction takes into account the degree to which a patient has been miscentred and returns an estimate of the degree to which the patient would appear magnified in the localizer radiograph. While this is a basic correction accounting

for the effective magnification of the patient WED, it does not account for the changes in the attenuation of the subject as may be interpreted through analysis of the localizer. Nevertheless, the model, as per the graphical presentations in Figures 7 and 8, offers as an excellent demonstrative tool for exhibiting how an increased degree of miscentring may affect patient radiation dose and, indeed, how for larger patients the effect is further amplified.

The inclusion of the miscentring correction factor, in addition to a term for relating patient size and the extent of miscentring, allowed for a more accurate model for estimating CTDI_{vol}. The model also offered as a tool for comparing how different CT imaging parameters may affect radiation output, while

Figure 7. Simulated "Thorax A" CTDI_{vol} for range of miscentring values and a WED range. CTDI, CT dose index; WED, water equivalent diameter.

Figure 8. Simulated $CTDI_{vol}$ for a 320 mm WED value, value of 12 mGy and a range of miscentred and strength values (weak, average, average and constant noise). CTDI, CT dose index; WED, water equivalent diameter.

controlling for patient size and positioning. For instance, when comparing the $CTDI_{vol}$ values for the TAP C vs TAP B protocol models in the case of a standard-sized patient at isocentre, the change in energy from 120 to 100 kV resulted in a 39% decrease in $CTDI_{vol}$, in line with an approximate 33% value reported in the literature.¹² Similarly for the abdomen protocol, a 25% reduction was evident when changing the peak energy from 120 to 100 kVp. With the presented model, it was also possible to estimate the resultant reduction in $CTDI_{vol}$ for all patient sizes and for different degrees of miscentring. In the thorax scans, we noted a 22% reduction in $CTDI_{vol}$ when changing the pitch from 0.6 to 1.2. The literature suggests that the effective mAs in the Siemens AEC system remains constant or exhibits only a slight decrease for such changes in pitch.¹³ For other manufacturer, AEC systems doubling the pitch has been reported to lead to a decrease in $CTDI_{vol}$ by as much as 66–77%.¹³ There appears to be limited data available relating the changes in slice thickness with CT output when using AEC; however, for the TAP protocols in this work, a 15% reduction in output was observed when comparing the 5 mm to the 1 mm slice thickness protocol. It should be noted that image quality metrics have not been evaluated in this work, but it has been assumed that the image quality metric demands of the AEC system have been met leading to the observed and modelled $CTDI_{vol}$ data for clinical data.

Dose tracking packages have, to date, been used to establish diagnostic reference levels (DRLs) across different centres and countries offering potential for an automated means of establishing and updating existing diagnostic reference levels for large data sets.¹⁴ Such dose tracking software packages have also been used to perform analysis of imaging parameters towards achieving optimised radiation doses.¹⁵ This study also extols the overall potential offered by using such software packages. We have highlighted how patient positioning practices can be investigated and also demonstrated how, through the use of clinical data, the response of an AEC system to patient size and degree of miscentring can be investigated. With the capability of controlling for patient size, investigating how imaging parameter changes may affect output offers insight into the behaviour of AEC systems. Indeed, while DRLs are typically examined for standard-sized patients,¹⁴ it is evident from our data that $CTDI_{vol}$ data of standard-sized patients would certainly not represent the doses received to those that fall outside the standard range. Accordingly, with the use of our model analysis for the complete imaged population, as collated from dose tracking package, there is potential to examine and generate DRL trendlines for the full patient size range.

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