Analysis of effective and organ dose estimation in CT when using mA modulation:

A single scanner pilot study

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Abstract

Effective dose (ED) estimation in CT examinations can be obtained by combining dose length product (DLP) with published ED per DLP coefficients or performed using software. These methods do not account for tube current (mA) modulation which is influenced by patient size.

Aim

To compare different methods of organ and ED estimation to measured values when using mA modulation in CT chest, abdomen and pelvis examinations.

Method

Organ doses from CT of the chest, abdomen and pelvis were measured using digital dosimeters and a dosimetry phantom. ED was calculated. Six methods of estimating ED accounting for mA modulation were performed using ImPACT CTDosimetry and Dose Length Product to ED coefficients. Corrections for the phantom mass were applied resulting in 12 estimation methods. Estimated organ doses from ImPACT CTDosimtery were compared to measured values.

Results

Calculated EDs were; chest 12.35 mSv (±1.48 mSv); abdomen 8.74 mSv (±1.36 mSv) and pelvis 4.68 mSv (±0.75 mSv). There was over estimation in all three anatomical regions. Correcting for phantom mass improved agreement between measured and estimated ED. Organ doses showed overestimation of dose inside the scan range and underestimation outside the scan range.

Conclusion

Reasonable estimation of effective dose for CT of the chest and abdomen can be obtained using ImPACT CTDosimetry software or k-coefficients. Further work is required to improve the accuracy of ED estimation from CT of the pelvis. Accuracy of organ dose estimation has been shown to depend on the inclusion or exclusion of the organ from the scan range.

Introduction

Advances in technology have facilitated Computed Tomography's (CT) expansion in providing rapid complex submillimetre imaging allowing more accurate diagnoses to be made [1-3]. In many instances CT is the first line investigation, however this has resulted in it becoming the dominant source of radiation risk/dose in medical imaging [3]. As with any medical imaging procedure involving radiation, there is a need for all involved to be aware of and monitor dose to patients; this is achieved by dose estimation as direct measurement of organ dose is not possible in the clinical environment.

Effective dose is often calculated by combining dose length product [DLP], (the product of the CT dose index volume ($CTDi_{vol}$) and scan length) with published coefficients (*k*-coefficient) [4-7]. This approach uses data published by the American Association of Physicists in Medicine (AAPM) [4]. Critics of this method state that the use of tissue weighting factors from ICRP 60 and scanner data from as early as 1990 means that the coefficients lack relevance to modern multidetector scanners in use today [7]. Updated figures have been published and used by Elbakri and Kirkpatrick [5] and Huda et al [6]. Both these articles argued that due to the updated tissue weighting factors published by the ICRP the conversion factors require updating too. Huda et al [6] provide figures that are independent of the make and model of scanner whilst Elbakri and Kirkpatrick [5] argued that accuracy can be improved further by taking into account scanner-specific results. In Elbakri and Kirkpatrick's paper figures for a range of scanner types are provided [5].

An alternative method of effective dose estimation can be performed by combining $CTDi_{vol}$ with data produced using Monte Carlo mathematical simulation. Data provided by the UK's National Radiological Protection Board (NRPB) can be used to calculate effective dose by utilising ImPACT's CT dosimetry tool (CTDosimetry V1.0.4) [8]. Although convenient, the data used in the Monte Carlo simulation was generated from early CT systems (not multidetector) therefore effective dose estimation has to be performed by fitting the characteristics of newer scanners to older designs [9]. This has the potential to introduce error [10].

Tube current modulation (mA modulation) is not taken into account when dose estimations are performed using the software or k-coefficients. mA modulation is standard on modern CT imaging equipment and it has the ability to manipulate the exposure and therefore dose as the patient is imaged. The ability to accurately estimate dose using fixed tube current has been shown but organ dose generally decreases with the use of tube current–modulated acquisition and this should be taken into account in any estimation method, but patient size can directly affect the dose reduction achieved [11].

The Medical Internal Radiation Dose (MIRD) mathematical phantom used in the development of the k-coefficients by the AAPM, Elbakri and Kirkpatrick and Huda et al and the stylised/mathematical phantom used in ImPACT's dosimetry tool represents a patient mass of 70 kg which is regarded as a low in comparison to modern demographics [4-6, 8, 12]. Research by Castellano stated that there is a change in effective dose for a change in mass with effective dose lower in larger patients for the same imaging parameters [13]. Castellano provides ratios for scaling effective dose using 70 kg as the reference value. The scaling factors indicate a 13% decrease in effective dose per 20 kg increase in mass for chest and abdomen CT acquisitions and a 9% decrease in effective dose per 20 kg

increase in mass for pelvis acquisitions. For fixed exposure parameters, effective dose decreases as patient mass increases suggesting that dose is likely to be overestimated [13].

This initial work utilised a single scanner type and phantom size and with a focuses on organ and effective dose calculation accuracy using mA modulation in CT of the chest, abdomen and pelvis. This paper examines different methods of dose estimation and compares these to direct measurements of organ doses made using an anthropomorphic phantom and MOSFET dosimeters. Effective dose calculations were compared against values generated using k-coefficients and ImPACT software. To account for mA modulation mass weighted corrections were applied in an attempt to improve accuracy for effective dose calculations.

Method

For CT of the chest, abdomen and pelvis twelve methods of estimating effective dose were used (Table 1) CT dosimetry software published by the ImPACT CT scanner evaluation group (ImPACT CTDosimetry spreadsheet v 1.0.4, ImPACT, London, UK) was used to estimate organ and effective dose. For each anatomical region, the DLP was recorded. Dose per DLP figures published by the American Association of Physicists in Medicine (AAPM) [4], Huda et al [6] and Elbakri and Kirkpatrick [5] were used to calculate effective dose (Table 2Table 2). One method of calculating effective dose from directly measured organ doses using MOSFETs was undertaken for CT of the chest, abdomen and pelvis using MOSFET dosimeters (Best Medical Canada, Kanata, Canada) and a male ATOM dosimetry phantom model 701-D (CIRS Inc. Virginia, USA). In all cases a Toshiba Aquillion 16 Multidetector CT scanner was used (Toshiba medical systems, Otawara-Shi, Japan). This system utilises filtered back projection reconstruction and for the purpose of the data collection manufacturer recommended reconstruction algorithms and a mA standard deviation of 5. The CT system uses Toshiba's SUREExposure3D method of tube current modulation during exposure. This uses the anterior-posterior and lateral scan plan radiograph to ascertain the optimum exposure. The system modulates the tube current in the z-axis and during rotation [14].

Direct dose measurements (MOSFET) were taken as the gold standard against which estimation methods were compared [15, 16].

Table 1 Methods of dose estimation

1. ImPACT effective mA
2. ImPACT average mA
3. ImPACT mA modulation
4. AAPM k-coefficient
5. Huda et al k-coefficient
6. Elbakri and Kirkpatrick k-coefficient
7. Mass corrected ImPACT effective mA
8. Mass corrected ImPACT average mA
9. Mass corrected ImPACT mA modulation
10. Mass corrected AAPM k-coefficient
11. Mass corrected Huda et al k-coefficient
12. Mass corrected Elbakri and Kirkpatrick k-coefficient

Table 2 Coefficients for calculation of effective dose from DLP

	k-coefficient mSv/mGy.cm		
	AAPM [4]	Huda et al [6]	Elbakri and Kirkpatrick [5]
Chest	0.014	0.017	0.020
Abdomen	0.015	0.016	0.017
Pelvis	0.015	0.018	0.017

Dose measurement

MOSFET dosimeters provide an accurate and reproducible method of collecting organ dosimetry data [17]. In this work four banks of five dosimeters were used (n=20). Calibration was performed as per manufacturer instructions at a tube voltage of 120 kV using the supplied calibration jig and a calibrated RaySafe X2 with R/F sensor (Unfors RaySafe AB, Bildal, Sweden). The error of these was 2.01%.

Indelible ink and radiopaque markers were used on an adult ATOM dosimetry phantom to ensure reproducible and accurate positioning and scanning of the phantom. MOSFET detectors were located within the phantom in the positions corresponding to the critical organs required for effective dose calculations [18-20] (see Table 3). Dose measurement was performed in 20 locations at a time as a total of 20 MOSFET sensors were available (Figure 1). In total 269 locations were used to measure organ dose and compute effective dose. For each MOSFET position, three exposures were made and a mean and standard deviation calculated to minimise random error.

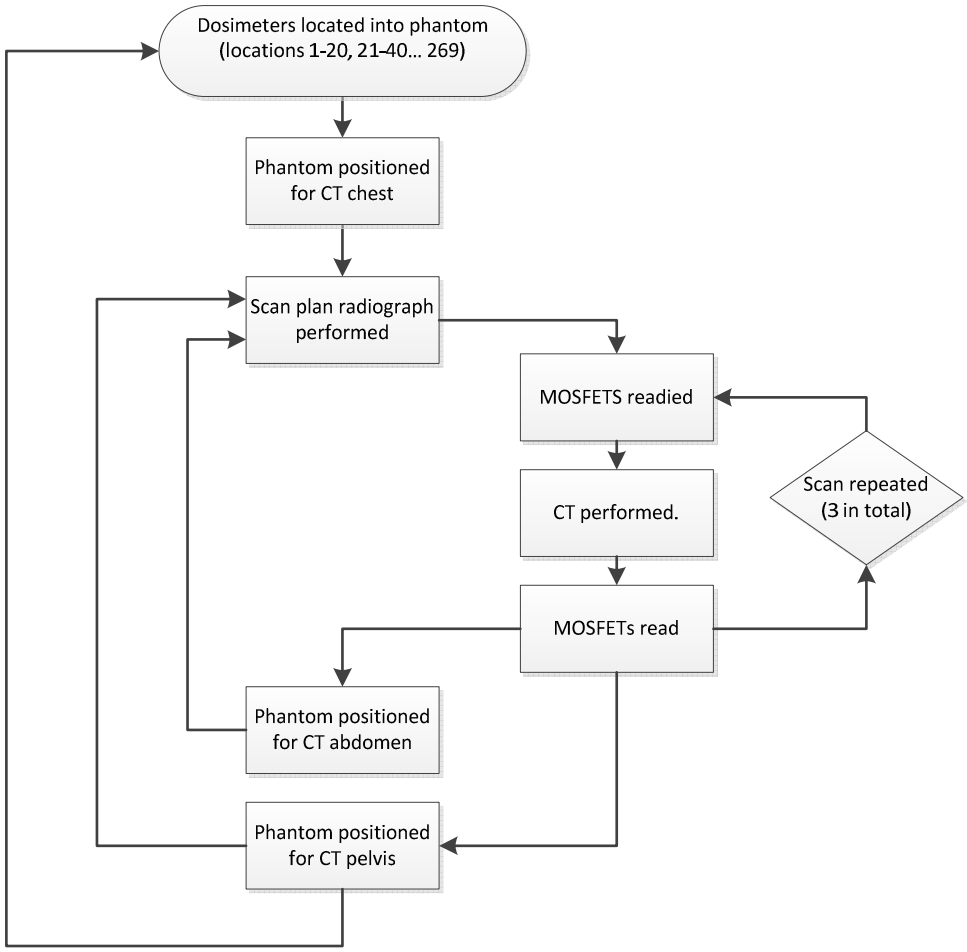

Figure 1 Flow diagram illustrating the data collection process for dose measurement

Table 3 Table stating the number of MOSFET locations used for organ dose measurement

Organ	Number of dosimeter locations
Adrenals	2
Bladder	16
Brain	11
Breast	2
Active bone Marrow	85
	Clavicle 20,
	Cranium 4
	Cervical Spine[†] 2
	Femora 4
	Mandible[◊×] 6
	Pelvis 18
	Ribs 18
	Scapula[o] 9
	Sternum 4
	Thoraco-lumbar Spine 9
Gall Bladder	5
Heart	2
Intestine (Small and large)	16
	Colon 11
	Small intestine 5
Kidneys	16
Liver	30
Lungs	36
Oesophagus	3
Pancreas	5
Prostate	3
Spleen	14
Stomach	11
Testes	2
Thyroid	10
Thymus	4

[†] locations in the anterior of C2 and upper oesophagus were used to calculate extra thoracic organ dose

[◊] locations in the left and right lingula of the mandible and to the left and right of the sublingual fossa were used to calculate salivary gland organ dose

[×] locations in the left and right lingula of the mandible were used to calculate oral mucosa organ dose

[o] locations in close proximity to the left and right glenoid fossa were used for dose to the upper humeri

For each scan, the dose length product (DLP) and the recorded effective mAs were noted. The mean DLP was calculated. Axial images were reconstructed at 1 mm slices to match the acquisition's collimation. The mAs values for each axial image were recorded and an average mAs was calculated (Table 4).

Table 4 Imaging parameters

	Chest	Abdomen	Pelvis
kV	120	120	120
Auto mA standard deviation	5	5	5
Rotation time (s)	0.5	0.5	0.5
mAs_{eff}	235	235	235
$mAs_{\bar{X}}$	203.21	201.41	211.34
Acquired slice thickness (detector length in z-axis)	16 x 1 mm	16 x 1 mm	16 x 1 mm
Pitch	0.938	0.938	0.938
$DLP_{\bar{X}}$	904.49	792.72	791.76
$CTDI_{\bar{X}}$	33.2	33.2	33.2

Comparison of ATOM and ImPACT standard phantoms

The ATOM phantom and the standard phantom within the ImPACT CT dosimetry software are of different dimensions. To compensate, scaling of the ImPACT phantom was performed. The ATOM phantom from the apex of the skull to the upper border of the symphysis pubis was measured at 830 mm. The length of the ImPACT standard phantom between the same reference points was 890 mm. Therefore a 1 mm slice in the ATOM phantom equated to a 1.07 mm slice in the standard phantom ImPACT CT dosimetry software. The proportions of the chest, abdomen and pelvis were compared to ensure that accurate comparisons were being made; Figure 2 illustrates good agreement between the two phantoms' proportions for these.

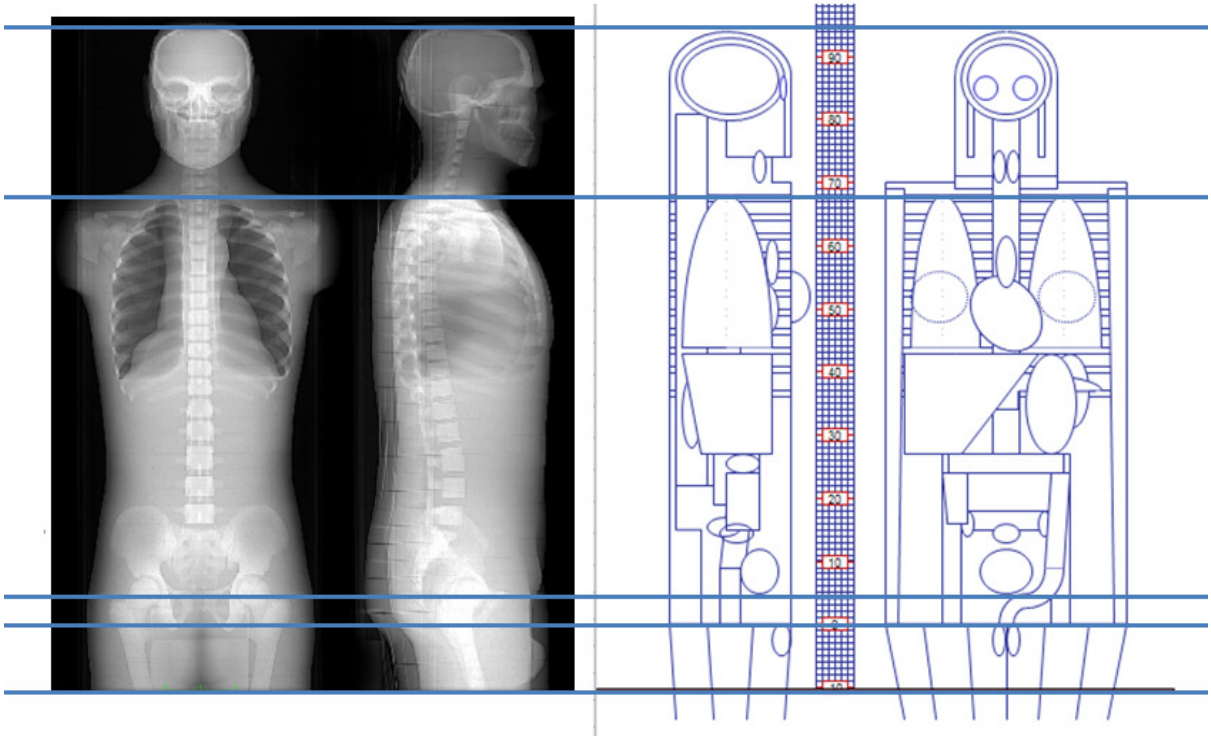

Figure 2 Comparison of the proportions of the chest, abdomen and pelvis of the two phantoms. The dosimetry phantom (left) has been scaled by a factor of 1.07 relative to the software phantom

The phantom used in ImPACT's CT dosimetry software that was used in the development of the DLP for effective dose calculations is based on a whole body mass of 70 kg [5, 6, 8, 21]. The ATOM phantom has a mass of 73 kg and consists of the head and torso only. According to Tozeren this accounts for only 55.4 % of total body mass [22] (Table 5).

Table 4 Relative weight of body segments for an adult male [22]

Region/limb	Percentage contribution to total body mass (%)
Torso	48.3
Head and Neck	7.1
Upper legs	21.0
Lower legs	9.0
Feet	3.0
Upper arms	6.6
Lower arms	3.8
Hands	1.2
TOTAL	100

Using the data, a total body mass of the ATOM phantom was calculated. The resulting mass was much greater at 131.8 kg; an increase in mass of 88.29%. Using the work of Castellano [13] Figure 3 was created to calculate the correction factor for mass. An assumption was made that this relationship remained constant (R^2=1).

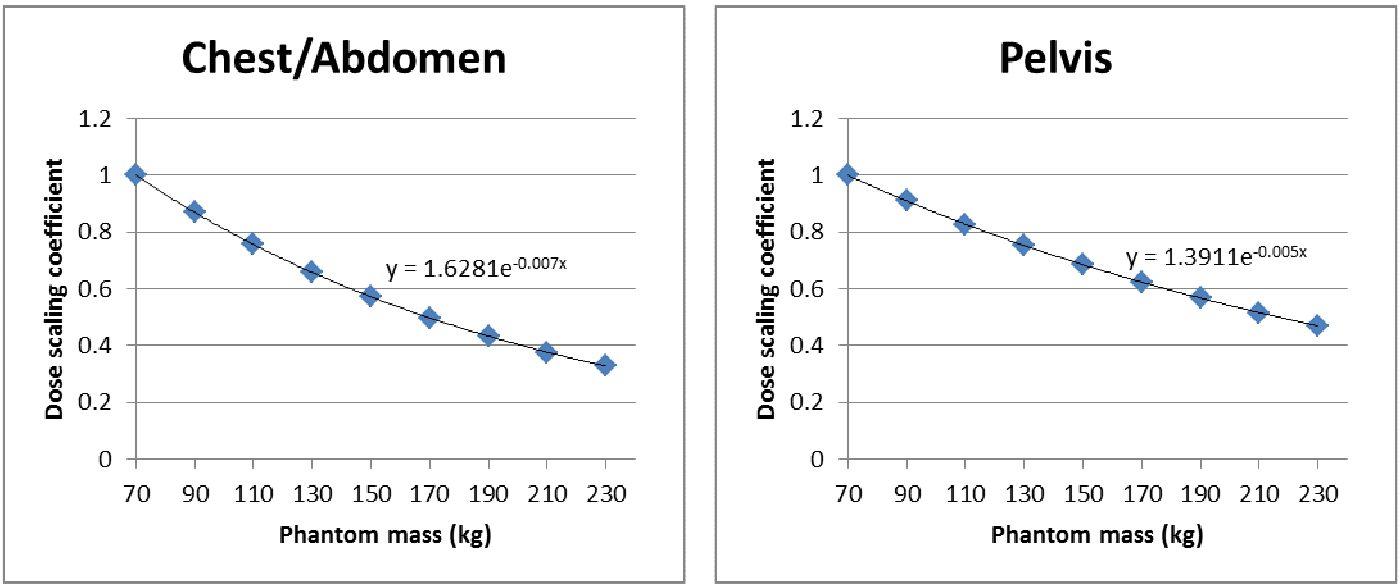

Figure 3 Graphs showing effective dose scaling coefficient calculations for the chest/abdomen and pelvis.

Dose calculations using ImPACT CT Dosimetry software was performed in three ways;

(i) Using the effective mAs quoted by the scanner for the full range of the acquisition,
(ii) Using the average mAs for the full range of the acquisition,
(iii) Using the mAs for each axial slice and summing the organ and effective doses to give final figures.

For method 3 in Table 1 a macro was created to be used within the ImPACT Excel spreadsheet that calculated the start and finish position for each slice and the corresponding mAs to calculate the effective dose per slice, which was then summated for the whole scan. The Toshiba Aquillion 16 has an overbeaming requirement of 2 rotations (1 at each end of the scan) [23]. With a collimation of 16x1 mm and a pitch of 0.938 an additional 15.0 mm at each end of the scan is required. When scaled, this is equivalent to 16.05 mm at the start and end of the acquisition in the simulation. The mA for the first and last slices was used as the mA for the respective upper and lower overbeamed sections of the scan for the mA modulation calculation. Comparisons between mass corrected and non-corrected figures using data from Table 5Table 4 were made.

Comparison of Organ doses

Effective dose is calculated by summing the weighted equivalent organ doses. Any difference in measured and estimated organ doses would be carried through into the final effective dose estimations. Comparison of estimated and measured organ doses was carried out to establish any sources of error. Unlike methods using DLP and conversion factors, dose estimation using ImPACT CT Dosimetry software allows figures for organ dose to be collated so this comparison could only analyse dose estimations using ImPACT CT Dosimetry software. It was also not feasible to correct for mass as accurate estimation of the distribution of intrathoracic and visceral fat was not possible. The difference between simulated and measured organ doses was calculated for each method (i. ImPACT

effective mA. ii. ImPACT average mA. iii. ImPACT mA modulation). These values were compared statistically using a single factor ANOVA.

Results

Effective doses calculated from the MOSFET organ dose measurements (Figure 4) were 12.35 mSv (±1.48 mSv) for CT of the chest; 8.74 mSv (±1.36 mSv) for CT of the abdomen and 4.68 mSv (±0.75 mSv) for CT of the pelvis.

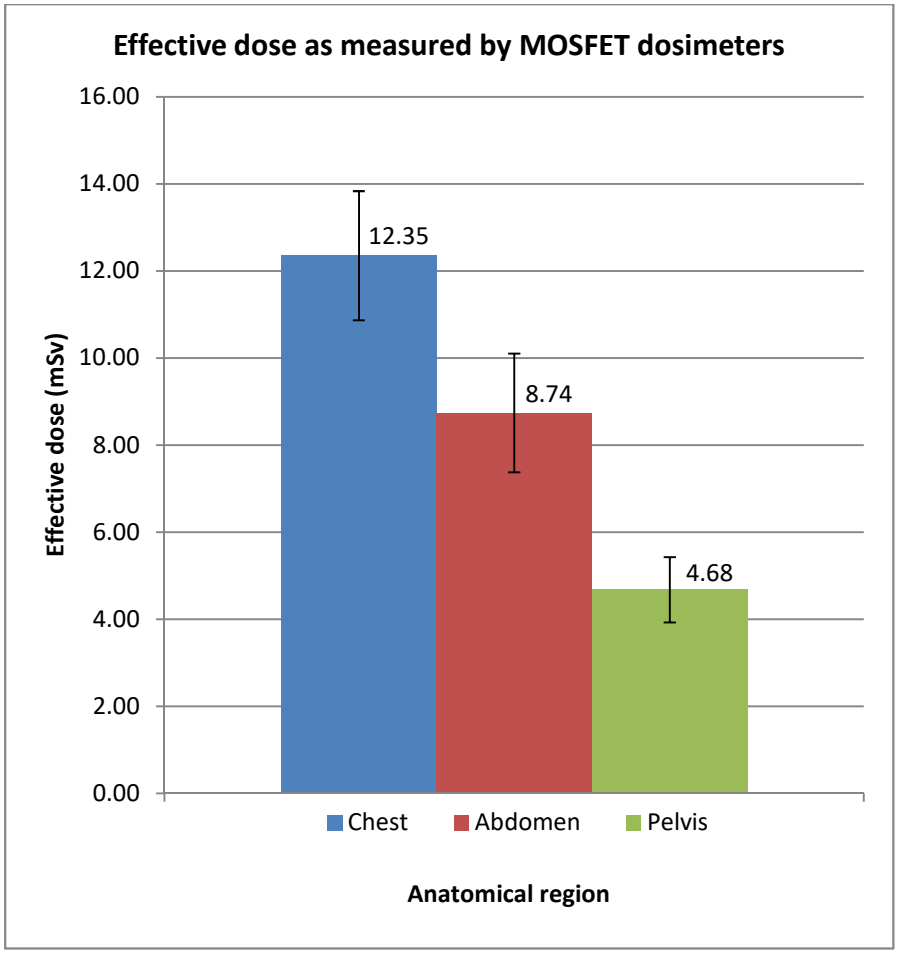

Figure 4 Calculated effective dose

Table 6 and Figure 5 illustrate the comparison of effective dose between measured and calculated values, with and without correction of phantom mass. Figure 5 demonstrates that using mass corrected values leads to greater accuracy for the calculated effective dose in comparison to the measured values.

Table 5 Effective dose measurement and calculation methods.

Method	Effective Dose (mSv) Chest	Abdomen	Pelvis
Calculated (MOSFET)	12.35 (±1.48)	8.74 (±1.36)	4.68 (±0.75)
Estimated (Uncorrected for mass)			
ImPACT effective mA	19.00	15.00	10.00
ImPACT average mA	17.00	14.00	9.90
ImPACT mA modulation	17.08	13.86	9.93
AAPM conversion factors [4]	12.66	11.89	11.88
Huda et al conversion factors [6]	15.38	12.68	14.25
Elbakri and Kirkpatrick conversion factors [5]	17.91	13.24	13.22
Estimated (Corrected for mass)			
Mass corrected ImPACT effective mA	12.30	9.71	7.20
Mass corrected ImPACT average mA	11.00	9.06	7.13
Mass corrected ImPACT mA modulation	11.05	8.97	7.15
Mass corrected AAPM conversion factors [4]	8.19	7.69	8.55
Mass corrected Huda et al conversion factor [6]	9.95	8.21	10.26
Mass corrected Elbakri and Kirkpatrick conversion factors [5]	11.59	8.57	9.51

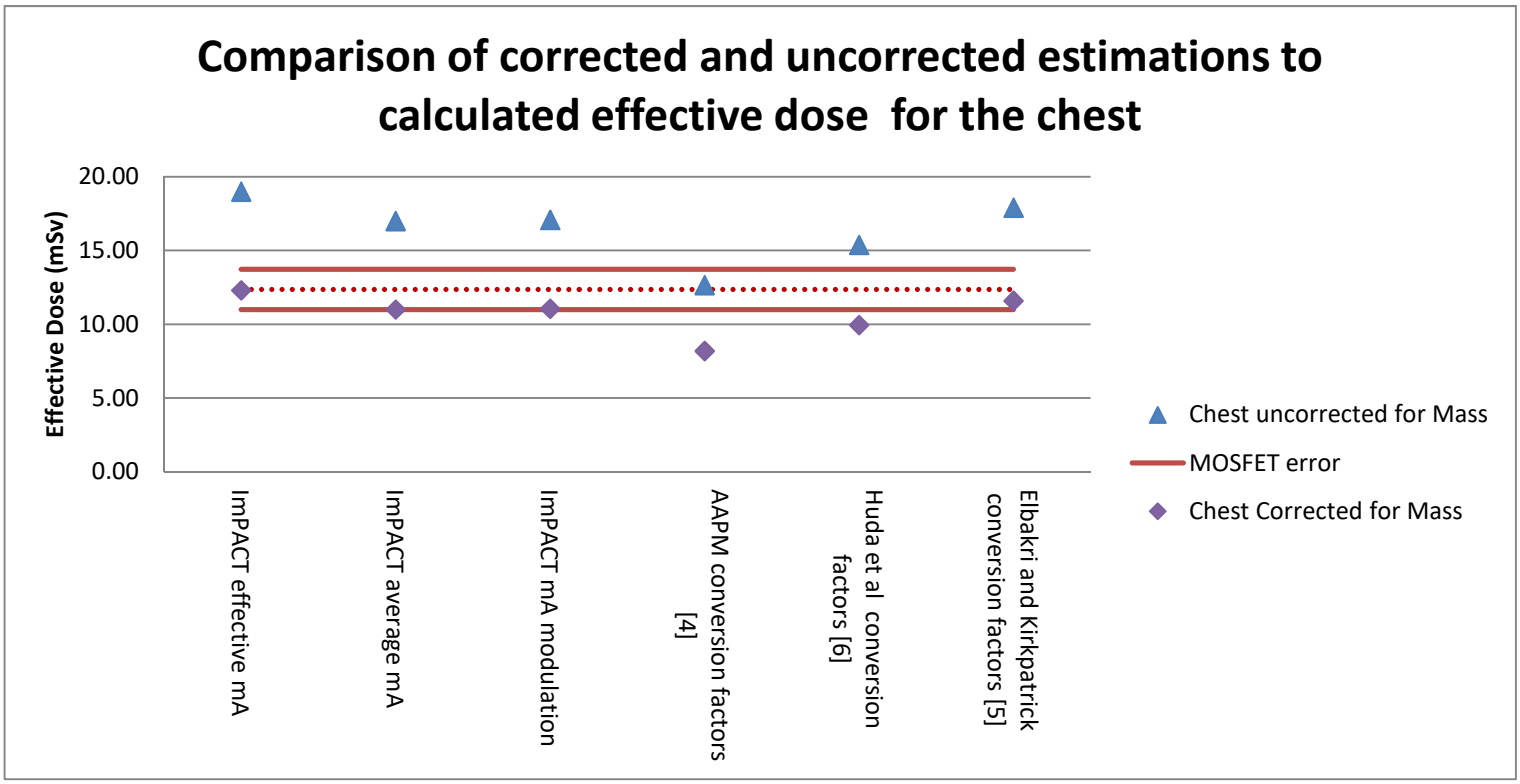

Figure 5a

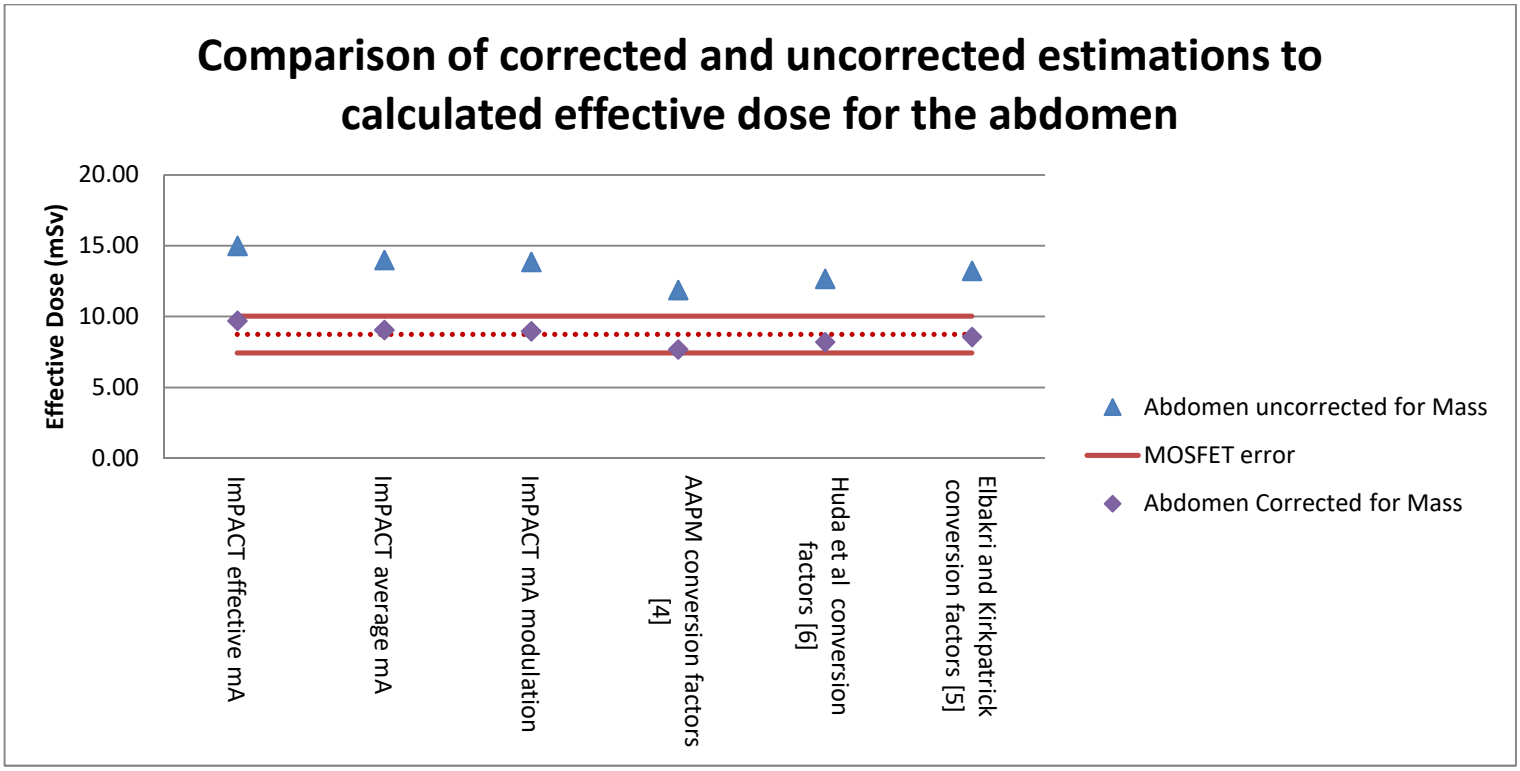

Figure 5b

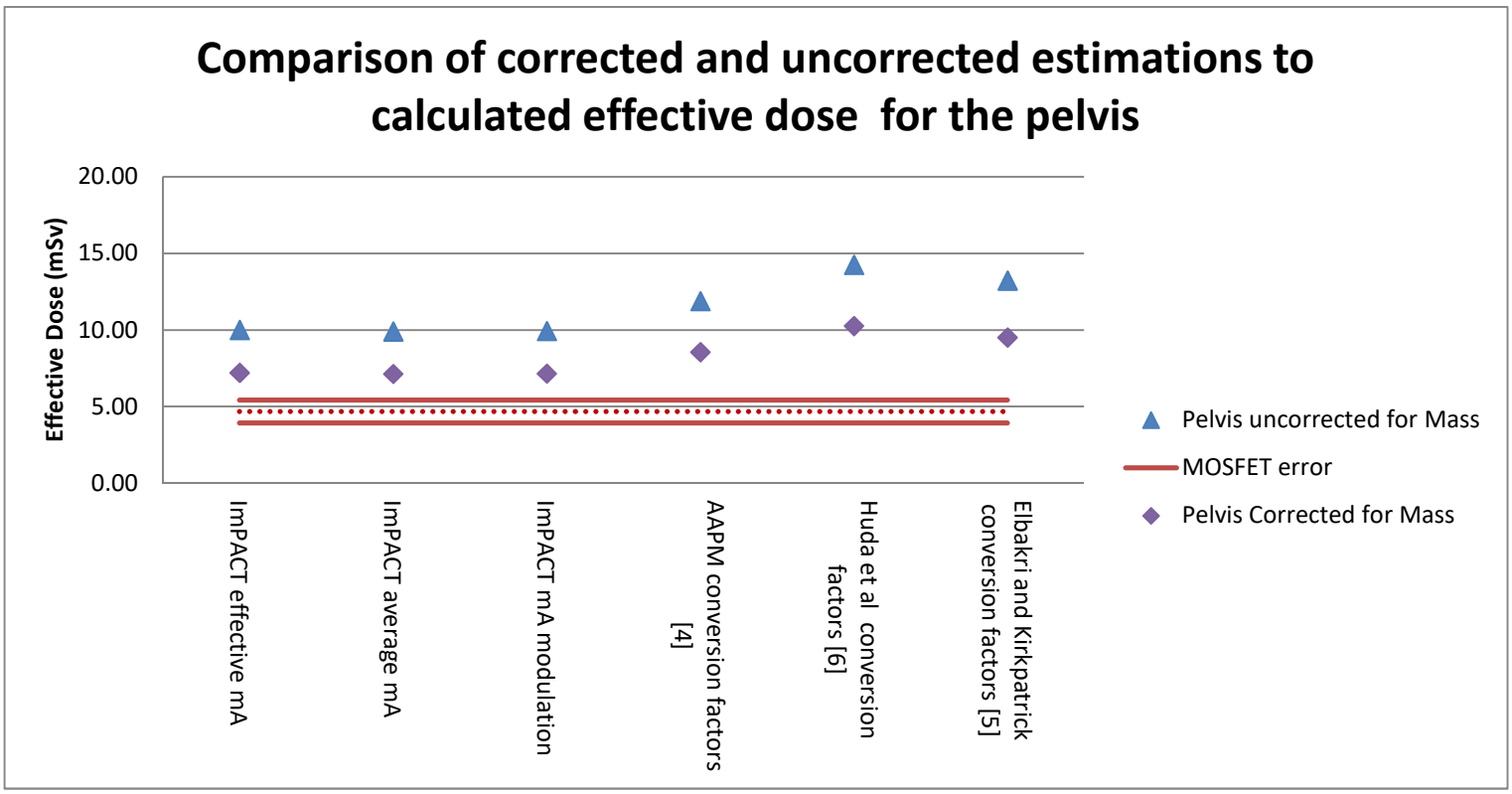

Figure 5c

Figure 5 Comparison of corrected and uncorrected effective dose estimations of the (a) chest (b) abdomen and (c) pelvis to calculated effective dose

Comparison of estimated organ for CT of the chest, abdomen and pelvis is shown in Table 7. A very strong positive correlation between the three estimation methods ($r>0.99$) and no significantly statistical difference is shown (Single Factor ANOVA, $p>0.9$).

Table 6 Comparison of estimated organ doses using mean, effective and modulated mAs for CT of the chest, abdomen and Pelvis.

	Organ dose (mGy)											
	Chest				Abdomen				Pelvis			
		ImPACT CTDosimetry				ImPACT CTDosimetry				ImPACT CTDosimetry		
Organ	Measured	$mAs_{\bar{X}}$	mAs_{eff}	mAs_{mod}	Measured	$mAs_{\bar{X}}$	mAs_{eff}	mAs_{mod}	Measured	$mAs_{\bar{X}}$	mAs_{eff}	mAs_{mod}
Gonads	0.04	0.00	0.00	0.00	0.16	0.07	0.07	0.07	5.58	37.00	39.00	36.00
Bone Marrow	8.10	12.00	14.00	12.22	4.55	8.50	9.30	8.59	5.45	10.00	10.00	10.01
Colon	0.51	0.35	0.40	0.32	14.3	19.00	20.00	18.76	16.6	23.00	24.00	22.81
Lung	24.7	45.00	52.00	45.27	5.48	9.50	10.00	9.44	0.16	0.04	0.05	0.04
Stomach	11.3	8.80	10.00	7.87	26.0	41.00	45.00	40.78	2.02	1.00	1.10	1.04
Bladder	0.09	0.02	0.02	0.02	1.93	1.10	1.20	1.16	19.8	47.00	49.00	47.93
Breast	27.7	35.00	40.00	36.79	1.47	1.80	2.00	1.79	0.08	0.04	0.04	0.04
Liver	18.9	14.00	17.00	12.71	22.4	38.00	41.00	37.70	0.71	0.62	0.65	0.62
Oesophagus	22.7	53.00	61.00	53.35	3.02	1.40	1.60	1.43	0.07	0.01	0.01	0.01
Thyroid	16.3	8.00	9.30	8.39	0.37	0.14	0.16	0.14	0.05	0.01	0.01	0.01
Brain	0.32	0.30	0.35	0.31	0.01	0.01	0.01	0.01	0.03	0.00	0.00	0.00
Salivary Glands	2.41	0.30	0.35	0.31	0.39	0.01	0.01	0.01	0.07	0.00	0.00	0.00
Remainder	11.0	13.00	15.00	12.84	11.7	19.00	21.00	19.27	4.09	6.90	7.20	7.00

Discussion

The options of mA value that are used within the ImPACT CTDosimetry software values (effective, average or modulated) has an insignificant effect on the estimated effective dose with the coefficient of variation of 6.4%, 4.4% and 0.5% for the chest, abdomen and pelvis respectively. Establishing the mAs per slice is a time consuming process and for convenience, the effective mAs can be used when estimating effective dose using ImPACT CTDosimetry software. This value is easily obtained from CT imaging equipment.

With the exception of the AAPM k-coefficient, uncorrected effective dose was over estimated (Figure 5). There was closer agreement for the CT of the chest (Figure 5a) with over estimation ranging from 2.48% to 42.4% (0.31 mSv to 6.65 mSv). There was poorer agreement in the abdomen (Figure 5b) and pelvis (Figure 5c) with over estimation of 30.54% to 52.74% (3.15 mSv to 6.26 mSv) and 72.48% to 101% (6.26 mSv to 9.57 mSv) respectively.

Tube current modulation takes into account patient size within set parameters. The phantom used in this study is larger than the phantom used within ImPACT's software and in the development of the k-coefficients therefore effective dose should be lower [4-6, 13]. Correcting for mass improved agreement between the effective dose estimations and calculations. Differences were -40.51% to -0.41% (-4.16 mSv to 0.05 mSv) in the chest, -12.78% to 3.60% (-1.05 mSv to 0.32 mSv) in the abdomen. It can be seen from Figure 5a,b and c that the majority of the mass corrected values fall within the error of the MOSFET dosimeters indicating no significant differences between the calculated and estimated values. Effective dose for the pelvis (Figure 5c) showed the greatest disagreement after correcting for mass with differences to MOSFET ranging from 41.76% to 74.70% (2.47 mSv to 5.58 mSv). The disagreement between effective dose of the pelvis suggests that the correction factor used is requires further research utilising phantoms of different sizes.

ANOVA showed no statistical difference in the estimation of organ dose using the average, effective or modulated mA ($p=0.9$). Using the mean of these three methods a comparison of estimated and measured organ doses shown in Table 7 highlights a pattern. It is apparent that organs within the scan range have an average estimate that is higher than measured values and those organs outside the scan range i.e. those organs whose dose comes from scattered radiation, have estimates that are lower than the measured values. To explore the effect this would have on effective dose estimations and calculations the tissue weighting factors were applied and the percentage contribution to effective dose of organs within and outside the scan range was calculated (Table 8). It is recognised that certain organs would be part in and part out of the scan range but for the purpose of this analysis, organs that were mostly in the scan range were classified as 'in' and vice versa.

Table 7 Percentage contribution of organs inside and outside the scan range to effective dose calculations

	Percentage contribution to calculated effective dose (%)	
	inside	outside
Chest	71.5	28.5
Abdomen	58.5	41.5
Pelvis	91.4	8.6

The chest and abdomen show better balance between contributions of organs inside and outside the scan range which would explain why these estimations are in closer agreement when compared to the chest and pelvis. The pelvis, however, has an imbalance with the greatest contribution to the effective dose calculation coming from organs inside the scan range- specifically the bladder, gonads and colon. With the suggested tendency of ImPACT dosimetry software to over-estimate organ dose inside the primary beam the reason for the large difference in calculated effective dose using measured organ dose to estimated effective dose is apparent. Reasons for these errors require further investigation and should focus on the suitability of the Monte Carlo data sets used in ImPACT's CTDosimetry software, the "best-fitting" of newer scanners to data in the ImPACT CTDosimetry software.

Limitations

It is recognised that this work is not without limitations. Only one scanner type and phantom was used. The tube current modulation parameters remained constant through the data collection and only filtered back projection reconstruction was used. Investigation into the mass correction for CT of the pelvis is required as this work has shown that over estimation occurs even after correction for mass. Should accurate organ dose estimations be required, clinicians should be aware of the under and over estimation of dose for organs inside and outside the scan range.

This work has shown that in this context, there is the potential to improve the accuracy of effective dose estimations by accounting for patient mass. Further work is required improve the accuracy of the mass correction factor of the pelvis and externally validate the factors for the chest and abdomen. Experimentation using phantoms of different sizes, imaging parameters and CT scanners from different manufacturers is planned.

Conclusion

This work has shown that a for this scanner type and exposure parameters a reasonable estimation of effective dose for CT of the chest and abdomen can be obtained using ImPACT CTDosimetry software or the k-coefficients referenced. The use of the k-coefficients is the quicker method compared to using ImPACT software but these do not give an indication of organ doses. This work has shown that there is a pattern for overestimation of organ dose inside the scan range and underestimation outside the scan range. Additional investigations using other scanners are required to establish if this is a consistent pattern. Further work is required to improve the accuracy of the mass correction factor for the pelvis and to test the external validity of the method varying the mass of the phantom and across different makes and models of CT scanner.

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