DEVELOPMENT OF IMPROVED MEASUREMENT ALGORITHM FOR ACCURATE CT DOSE ESTIMATION IN TRUNCATED IMAGES

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DEVELOPMENT OF IMPROVED MEASUREMENT ALGORITHM FOR ACCURATE CT DOSE ESTIMATION IN TRUNCATED IMAGES

by

BARDE MUSTAPHA ALHAJI

Thesis submitted in fulfilment of the requirement for the degree of Doctor of Philosophy

September 2025

CERTIFICATE

This is to certify that the dissertation entitled **"DEVELOPMENT OF IMPROVED MEASUREMENT ALGORITHM FOR ACCURATE CT DOSE ESTIMATION IN TRUNCATED IMAGES"** is the bona fide record of research work done by Mr "BARDE MUSTAPHA ALHAJI" during the period of September 2021 to August 2024 under my supervision. I have read this dissertation and that in my opinion it conforms to acceptable standards of scholarly presentation and is fully adequate, in scope and quality, as a dissertation to be submitted in fulfilment of the requirement for the degree of Doctor of Philosophy.

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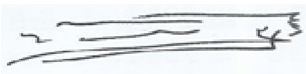
DECLARATION

I hereby declare that this dissertation is the result of my own investigations, except where otherwise stated and duly acknowledged. I also declare that it has not been previously or concurrently submitted as a whole for any degree at Universiti Sains Malaysia or other institutions. I grant Universiti Sains Malaysia the right to use the dissertation for teaching, research and promotional purpose.



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Mustapha Alhaji Barde

Date:

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All praise is due to Allah the master planner of the universe, the only God whom unique characteristics can never be all known to any creature or non-living, because is above the human sense. Surely, the last messenger conveys all Allah's messages, may Allah have mercy upon him.

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LIST OF SYMBOLS

Cm	Centimeter
kV	kilovolt
Kvp	peak kilo voltage
mA	milliampere
mAs	milliampere per second
mGy	milligray
mGy.cm	milligray centimeter
mm	millimeter
mSv	millisievert

LIST OF ABBREVIATIONS

AAPM	American Association of Physicist in Medicine
ACR	American College of Radiologist
AEC	Automatic Exposure Control
ATCM	automatic tube current modulation
CNR	Contrast to Noise Ratio
CT	Computed Tomography
CTDI	Computed Tomography Dose Index
CTDI100	CTDI using 100 mm Ionization Chamber
$CTDI_{air}$	Computed Tomography Dose Index Air
$CTDI_c$	Center Computed Tomography Dose Index
$CTDI_p$	Peripheral Computed Tomography Dose Index
$CTDI_{vol}$	Volume Computed Tomography Dose Index
CTDIw	Weighted Computed Tomography Dose Index
DICOM	Digital Imaging and Communication in Medicine
DLP	Dose Length Product
ED	Effective Dose
FBP	Filtered Back-Projection
HUSM	Hospital Universiti Sains Malaysia
ICRP	International Commission on Radiological Protection
JEPeM	Jawatankuasa Etika Penyelidikan Manusia USM
LAT	Lateral
MDCT	Multi Detector Computed Tomography
NABD improved	Normal Abdomen Improved Algorithm Measured
NDRL	National Diagnostic Reference Level
NTRX improved	Improved Normal Thorax Improved Algorithm Measured

NTRX Indo	Normal Thorax IndoseCT Measurement
PACS	Picture Archieving and Communication System
PMMA	Polymethyl-Methacrylate
ROI	Region of interest
SD	Standard Deviation
SPSS	Statistical Package for the Social Sciences
SSDEindoNABD	Normal Abdomen SSDE Based on indoseCT Measurement
SSDEindoNTRX	Normal Thorax SSDE Based on indoseCT Measurement
SSDENABD	Normal Abdomen SSDE Based on Manual Measurement
SSDENTRX	Normal Thorax SSDE Based on Manual Measurement
TABD improve	Truncated Abdomen Improved Algorithm Measured
TTRX exprt	Truncated Thorax Expert Measured
USM	Universiti Sains Malaysia

LIST OF APPENDICES

PEMBANGUNAN ALGORITMA PENGUKURAN YANG DITAMBAHBAIK BAGI ANGGARAN DOS CT YANG TEPAT DALAM IMEJ TERPOTONG

ABSTRAK

Pengimejan tomografi berkomputer (CT) menyumbang hampir 50–60% daripada jumlah dos radiasi yang diterima daripada pendedahan perubatan, sekali gus menimbulkan kebimbangan dalam kalangan profesional perubatan terhadap dos radiasi CT, mendorong kepada penambahbaikan dalam perlindungan radiasi dan pengoptimuman amalan CT. Anggaran dos mengikut saiz pesakit (SSDE) telah diperkenalkan bagi memberikan penilaian dos radiasi yang lebih tepat dengan mengambil kira variasi saiz pesakit. Walau bagaimanapun, faktor seperti pemotongan imej akibat saiz badan yang besar dan kedudukan pesakit yang tidak tepat boleh menjejaskan ketepatan pengiraan SSDE. Kajian ini bertujuan menilai kesan pemotongan imej terhadap pengiraan SSDE dan membangunkan algoritma automatik yang diperbaik bagi membetulkan kesan pemotongan, sekali gus menghasilkan anggaran dos spesifik saiz yang lebih tepat. Pengumpulan data melibatkan kajian menggunakan fantom dan imej klinikal. Bahagian pertama kajian melibatkan penilaian kesan pemotongan terhadap pengukuran saiz khusus dan SSDE. Fantom CTDI bersaiz 22 cm dan 32 cm diimbas pada pelbagai kedudukan bagi mensimulasikan peratusan pemotongan imej (Truncation Percentage, TP) sebanyak 5%, 10%, 15% dan 20%. Fantom yang diletakkan pada titik tengah (isocenter) digunakan sebagai imej rujukan dengan TP 0% (tidak dipotong). Diameter efektif (D_{eff}) ditentukan pada imej CT dan digunakan untuk menganggarkan SSDE. Seterusnya, algoritma automatik yang diperbaik dibangunkan menggunakan perisian MATLAB (versi R2023b), yang merangkumi pengukuran automatik bagi saiz spesifik dan SSDE, serta pembetulan terhadap artifak pemotongan. Algoritma yang dibangunkan ini dinilai menggunakan kaedah pengukuran manual dan automatik pada imej fantom serta 80 imej klinikal. Ketepatan dan keberkesanan algoritma diperbandingkan dengan perisian IndoseCT serta pengukuran manual oleh pakar. Hasil kajian menunjukkan terdapat hubungan yang sangat kuat antara peratusan pemotongan (TP) dan ralat peratusan (PE) dalam pengiraan SSDE bagi kaedah D_{eff} dan D_w (r = 0.9673 dan r = 0.9598). Hasil penilaian menunjukkan bahawa algoritma yang diperbaik mempunyai nilai ralat mutlak purata (MAE) yang lebih rendah bagi pengiraan D_{eff} (0.03), berbanding MAE bagi perisian IndoseCT (0.61 – 0.86) dan pengukuran manual (0.97 – 1.54). Selain itu, nilai pekali korelasi antara-kelas (ICC) untuk algoritma ini adalah 0.89 bagi imej abdomen dan 0.81 bagi imej toraks, menunjukkan tahap kebolehpercayaan yang tinggi. Bagi imej klinikal yang mengalami pemotongan, nilai MAE bagi SSDE berdasarkan Deff yang dikira menggunakan algoritma ini adalah lebih rendah (0.066 dan R^2 = 0.999 untuk abdomen; 0.043 dan R^2 = 0.997 untuk toraks), berbanding dengan perisian IndoseCT. Kesimpulannya, artifak pemotongan memberi kesan ketara terhadap penentuan saiz pesakit dan seterusnya menjejaskan ketepatan pengiraan SSDE. Algoritma yang dibangunkan dalam kajian ini menawarkan kaedah yang lebih tepat dan efisien untuk menentukan D_{eff} dan SSDE dalam pengimejan CT, dengan ralat antara pemerhati yang minimum.

DEVELOPMENT OF IMPROVED MEASUREMENT ALGORITHM FOR ACCURATE CT DOSE ESTIMATION IN TRUNCATED IMAGES

ABSTRACT

Computed tomography (CT) imaging contributes almost 50-60% of total radiation dose received from medical exposure, leading to growing concern among medical professional on CT radiation doses, driving improvements in radiation protection and CT practice optimisation. Size-specific dose estimation (SSDE) was introduced to provide a more accurate assessment of radiation dose, which consider patient variability. However, factors like truncation from large body sizes and poor positioning can hinder accurate SSDE estimation. This study aimed to evaluate the effect of truncation on SSDE calculation and develop an improved automated algorithm to correct the truncation effect for more accurate size-specific dose estimation. The data collection involved both phantom and clinical study. First part involved the evaluation of truncation effects on measurement of specific size and SSDE. CTDI phantoms, 22 cm and 32 cm in size, were scanned at different positions to simulate truncation percentages (TP) of 5%, 10%, 15%, and 20%. The phantom positioned at the isocenter served as the reference image with 0% TP (non-truncated). The effective diameter (D_{eff}) was determined on the CT images and used for estimating the SSDE. Then, an improved automated algorithm was developed using the MATLAB software (Version 28) that involve automated measurement of specific size and SSDE, and correction for truncation artefacts. The developed algorithm was validated using manual and automated measurement on the phantom and 80 clinical images. The accuracy and effectiveness of the improved algorithm was benchmarked with measurement by the IndoseCT software and manual measurements by experts.

The results showed strong positive correlations between truncation percentage (TP) and percentage error (PE) in SSDE calculation for both D_{eff} and D_w approaches (r = 0.9673 and r = 0.9598), respectively. The validation results showed lower value of mean absolute error (MAE) of D_{eff} measurements (0.03) for the improved algorithm, as compared to the MAE of IndoseCT (0.61 – 0.86) and manual measurement (0.97 – 1.54). Besides, the values of intraclass correlation coefficient (ICC) for the improved algorithm were 0.89 and 0.81 for abdomen and thorax images, indicating excellent reliability. For truncated clinical images, the MAE of the calculated $SSDE_{Deff}$ for the proposed algorithm was lower (0.066 and R^2 = 0.999; 0.043 and R^2 = 0.997) for abdomen and thorax images, respectively, as compared to IndoseCT sofware. In conclusion, the truncation artefacts significantly affect size determination, and hence, hinder the accuracy of SSDE calculations. The proposed algorithm offers robust, and more accurate method for determining D_{eff} and SSDE in CT imaging, with small observer variability.

CHAPTER 1

INTRODUCTION

1.1 Study Background

Computed Tomography (CT) is a popularly used to take images of the body because it can make high-quality cross-sectional images quickly. Despite making up only 16% of imaging procedures, CT accounts for 50–60% of the total radiation dose in medical practice. Medical practitioners are concerned about this, especially in light of the possible risk of radiation-induced cancer, especially in paediatric patients.

The "As Low as Reasonably Achievable" (ALARA) principle is used in CT dose optimisation to guarantee patient safety. The CT console shows the widely used dose indicators, the dose length product (DLP) and CT dose index volume ($CTDI_{vol}$). These indicators, however, overestimate or underestimate the actual dose received because they are based on standard phantom sizes (16 cm or 32 cm) and do not take individual patient size into consideration.

This was addressed with the introduction of the Size-Specific Dose Estimate (SSDE) method, which uses measurements like effective diameter (D_{eff}) or water-equivalent diameter (D_w) to modify dose estimates based on patient size. Although D_w provides greater accuracy by accounting for tissue attenuation, manually calculating D_{eff} is laborious and prone to mistakes, particularly when calculating dose across several CT slices.

Additionally, several factors can affect the accuracy of SSDE, including patient positioning errors, truncation artefacts (from large body size extending beyond the scan field), metal artefacts, and air gaps in certain anatomical areas. Among these, truncation artefacts significantly impact the accuracy of size measurements, thereby affecting dose estimation.

Therefore, this study aims to investigate the impact of truncation artefacts on size measurement and SSDE calculation using both phantom and clinical CT images. Furthermore, it aims to develop and validate an automated correction algorithm that improves the accuracy of D_{eff} measurement and SSDE, leading to more reliable patient dose estimation in CT procedures.

1.2 Problem statement

For large CT image datasets involving extensive populations, manual calculation of SSDE is labour-intensive and time-consuming. Besides, manual measurement of individual patient size may introduce variability and potential inaccuracies due to dependence on manual caliper placement and subjective interpretation. These limitations hinder consistency and efficiency in dose estimation, particularly in large-scale studies or clinical settings. Consequently, there is a growing need for reliable, automated SSDE algorithms to improve accuracy, reduce observer variability, and streamline the dose estimation process (Osman et al., 2024). However, the main problems with the current automated SSDE algorithms available in commercial software is their high cost, and making them unaffordable for many healthcare centres (Abdulkadir et al., 2022; AAPM, 2011). Furthermore, because these commercial tools are made for particular CT scanner manufacturers or models, they are frequently not universally compatible, which restricts their use in various institutions. Therefore, there is need to develop a system that is universal and implementation of automated SSDE analysis, particularly in settings with diverse scanner types. Therefore, there is a critical need to develop a cost-effective, vendorneutral automated system for SSDE calculation that can be applied universally, regardless of machine model or manufacturer.

Another limitation of the current automated software is it only allows accurate measurement only on a perfect image (which covers full anatomical structure or full body coverage). Few limitations can lead to inaccurate measurement of patient size (diameter) such as incomplete view of CT image due to truncation effect, superimposition or existence of other body parts within primary FOV (Anam et al., 2021), gantry tilt or angulations of x-ray beam during image acquisition in head CT scanning, metal artefacts inhomogeneity (dark and bright streaks), presence of air within lungs in thorax region, and also reformatted images (other than axial plane) such as multiplanar reconstruction (MPR), reconstructed thickness (thinner slices) (Abdulkadir et al., 2022). Thus, the effects of these problems to measurement accuracies need to further examine and a new method to overcome these problems should be developed. In clinical practice lateral part of images (x- axis direction) are the parts prone to truncation, because CT images are acquired by rotation of the tube around the object been imaged (Dio et al., 2022).

1.3 Study Objective

1.3.1 General objective

This study aims to develop and validate the effectiveness of an automated algorithm for accurate measurement of individual size with truncation corrections on CT images in estimation of size-specific dose.

1.3.2 Specific objectives

i) To investigate the effects of truncation artefacts on the degree of inaccuracy in individual size measurement.

ii) To develop an improved algorithm for automated individual size measurement, integrating a correction technique for truncation effects.

iii) To validate the effectiveness of the proposed algorithm for size specific dose estimation using phantom and clinical CT images.

1.4 Significance of study

This study is highly significant as it addresses a key limitation of the existing automated SSDE algorithm, which currently only works with full-view CT images and are unable to accurately process images affected by truncation. It improves the accuracy of size-specific dose estimation (SSDE) in CT imaging by developing an improve algorithm that accounts for truncation effects. Additionally, by offering a dependable, automated solution that minimises human error, decreases manual labour, and ensure more accurate and consistent patient dose assessments in standard clinical practice, it seeks to enhance the productivity and workflow of medical physicists and radiation personnel.

Moreover, this study's influence goes beyond the immediate setting. Accurate dose estimations for different body parts are made possible by the developed algorithm, which can be used for large-scale, multicenter assessments in a variety of geographic locations. This thorough approach improves the findings' generalisability and applicability to various clinical settings and populations.

Lastly, the system offers effective but also cost-efficient, with rapid processing capabilities. This study contributes significantly to the field of medical imaging and radiation safety, benefiting both healthcare professionals and patients alike.

1.5 Organisation of thesis

This dissertation is divided into five chapters each individual chapter present a component and subsections. Chapter one gives a general overview of the topic background, problem statement and highlights on the study objectives.

Chapter two elaborated on relevant literature related to the key aspects of the study, including SSDE and truncation artefacts, an overview of CT and radiation dose in medical imaging, and the historical development of dose estimation in CT. It also highlights the importance of accurate patient-specific dose assessment and provides a critical review of existing SSDE measurement techniques. Additionally, the chapter outlines current limitations and identifies research gaps in SSDE methodologies, which serve as a fundamental basis for the present study.

Chapter three outlines the research tools and methodology employed in this study. This chapter details the phantom study specifically designed to investigate the effect of truncation on SSDE, including a clear explanation of the study design, experimental setup, data collection procedures, and analysis methods. In addition, this chapter describes the methodology used for developing an automated algorithm aimed at correcting truncation artefacts to enable more accurate SSDE measurements.

Chapter four presents the results of this study in detail, encompassing both the phantom study and clinical image analyses. Descriptive and inferential statistical methods was employed to analyse the data for each component. Furthermore, validation of the improved algorithm was highlighted by comparing the result with the results obtained using IndoseCT software. Similarly, the effectiveness of the new developed algorithm was also presented by comparing the measurement performed by the experts, IndoseCT and the developed algorithm on the normal CT images.

Besides, the SSDE values obtained using D_{eff} approach were compared with SSDE D_w approached in the phantom study. The analysis of the impact of truncation on SSDE determination based on the phantom study data was presented. Finally, the discussion of the algorithm's accuracy and efficiency, the Implications of the research findings for CT dose estimation along with limitations of the study and potential sources of error were highlighted.

Chapter five constitutes conclusion and future directions of the study, key findings from both phantom study and algorithm development were delineated. The contributions of the study to the field of CT dose estimation and practical implications for medical imaging practices were stated. Furthermore, recommendations for future works and improvements to the algorithm were suggested. Finally, reflection on the significance of addressing truncation in CT imaging and conclusion were drowned.

CHAPTER 2

LITERATURE REVIEW

In this literature review, the critical aspects of patient-size specific dose estimation in computed tomography (CT) imaging were explored. This review covers important topics including size-specific dose estimates (SSDE) and the impact of truncation artefacts, provides an overview of CT technology and radiation dose, traces the historical evolution of dose estimation techniques, emphasises the importance of accurate patient-specific dose calculations, reviews on existing SSDE measuring methods, and identifies limitations and gaps in current approaches.

2.1 Overview of computed tomography (CT)

The utilisation of X-ray radiation in clinical practice began in 1895 when Wilhelm Conrad Roentgen discovered its imaging potential. Through his pioneering work, he demonstrated that radiation could be harnessed for diagnostic purposes, marking a significant milestone in medical history. Decades later, numerous technological advancements and research further unveiled the potential of electromagnetic radiation for advanced imaging. This progress led to the invention of the computed tomography (CT) machine, which has become one of the most widely used medical imaging modalities (Anam et al., 2017a; Brenner et al., 2007; Kayun et al., 2021; Terashima et al., 2019).

Since its debut in the 1970s, CT has progressively improved in its ability to generate high-quality images. CT is a diagnostic imaging modality that uses X-ray to examine internal organs and bony structures of the human body (Dio et al., 2022; Christianson et al.,2021b). This modality has a wider usage in clinical applications due to its high image contrast compared to other conventional radiographic modalities (Dio et al., 2022; Nitasari et al., 2021; Porzio & Anam, 2022). CT imaging enhances diagnostic accuracy and is considered beneficial for patients, despite the higher radiation doses involved. CT scans help with accurate and timely medical diagnosis. They are used in emergency situations, such as detecting neurological abnormalities or assessing severe headaches (Cheng, 2013; Hasan et al., 2022; Huda et al., 2008). A new generation of multidetector-row CT (MDCT) with higher X-ray tube power, shorter acquisition times, and dual-energy and even multi-energy x-ray capabilities is the result of advancements in CT hardware. A single axial image can now be obtained in 0.005 seconds, compared to over 300-second duration it takes for the first-generation CT scanner. (Dewi et al., 2021).

However, CT procedures pose substantial risk to patients, due to its relatively higher doses (Brenner et al., 2007; Chu et al., 2022; Podgorsak et al., 2021; Tahiri et al., 2021). Patients often undergo multiphases CT scans within a short period of time. To minimise radiation exposure, the practitioners should review their current practices and adopt optimised imaging protocols. Additionally, an accurate estimation of radiation dose is required to ensure dose optimization and proper documentation for future reference in determining radiation absorbed by patient(s) (Tzanis et al., 2024; Dio et al., 2022; Hwang et al., 2021). As such its vital to ascertain the radiation dose received by patients undergoing CT scan procedures.

Computed Tomography (CT) scanning produces a cross-sectional image of patient anatomy by series of X-ray projections from different angles. The advancement of technology has resulted in wide applications of CT examination due to its capability to produce improved quality images through fast imaging procedures (Dio et al., 2022; Fahmi et al., 2019; Tsujiguchi et al., 2018).

CT scanning has posed a major concern among medical practitioners as compared to other radiological modalities, due to higher dose delivery and high potential risk of radiation-induced cancer, especially in paediatrics (Kritsaneepaiboon et al., 2019). One of the greatest challenges due to its levels and variations in radiation doses between patients, institutions, and nations. Almost 50-60% of the medical radiation dose is attributed by CT examinations. However, this only constitutes 16% of imaging investigations (Ahmadifard et al., 2022; Brenner et al., 2007; Burton, 2020; Chu et al., 2022; Juszczyk et al., 2021; Khawaja et al., 2015; Xu et al., 2020).

The dose optimisation principle of 'As Low as Reasonably Achievable' (ALARA) is mandatory in CT procedures. This principle is fully utilised in CT imaging optimization due to the two routinely dose metrics used in CT, which are the volume CT dose index (CTDIvol) and dose length product (DLP) (Burton, 2020; Burton & Szczykutowicz, 2018; Chu et al., 2022; Gharbi et al., 2020; Moore et al., 2014; Tack, 2011; Xu et al., 2020).

According to International Electrotechnical Commission (IEC 2002) standard, both CTDIvol and DLP were recommended to be displayed on CT console as post-exam dose report (Bauhs et al., 2008; Juszczyk et al., 2021; Kayun et al., 2021; Pourjabbar, 2014; AAPM , 2011; Xu et al., 2020). The main limitation with the current dose metrics (CTDIvol and DLP) values is they did not present the actual patient's absorbed dose and tend to underestimate or overestimate the dose to patient based on their individual sizes (Bauhs et al., 2008; Brady et al., 2015; Christner et al., 2012; Gharbi et al., 2020; Khawaja et al., 2015). This is because a standard phantom size of 16 cm or 32 cm diameter of cylindrical phantom was used to measure and derive the CTDIvol. While DLP is the product of CTDIvol and scan length (in centimetre, cm)

(Dio, Anam, & Hidayanto, 2022; Huda, 2015; Juszczyk et al., 2021; Khawaja et al., 2015; Porzio & Anam, 2022; Xu et al., 2020).

The standard phantom used in dose measurement is made up of acrylic or polymethyl-methacrylate (PMMA) that has similar properties to human soft tissues and water. The limitation of this phantom and measurement method is it does not consider the morphological make up of subjects that consist of various anatomical structures and tissues with different attenuation properties (Adhianto et al., 2020; Ali et al., 2018; Kostou et al., 2019; Leng et al., 2015).

Anterior-posterior (AP) and lateral (LAT) dimension are measured on the axial or scout image and used for the computation of individual Deff. The recent recommendation by AAPM as reported in Report No. 220, the new size terminology known as water equivalent diameter, Dw concept is introduced for SSDE calculation (Anam et al., 2017b; Burton & Szczykutowicz, 2018; Iriuchijima et al., 2018; Juszczyk et al., 2021; AAPM, 2019). A conversion factor was determined based on reference value (Table 1) provided in the reports to normalize CTDIvol which define patient size based on effective diameter (Deff) or water-equivalent diameter (Dw), presented in centimetres (cm). However, Dw yield more accurate SSDE measurement by considering the CT attenuation on each slice and mapping area of interest on the axial images (Juszczyk et al., 2021; Xu et al., 2020). For Deff determination, there is no specific protocol for the number of images to be used, however, the middle slice of patient image is preferable, besides accurate calculation of Deff can be done from all images, but it is a tedious process (Fahmi et al., 2019).

The approach of estimating of SSDE was termed to be more time consuming due to manual measurement of individual patient size and obtaining the mean SSDE over all slices. It was reported that digital analysis minimises human error in visual evaluation rather than relying just on subjective human visualization and assessment (Cheng, 2013; Omer et al., 2021). Recently, various researchers and manufacturers gave attention to SSDE for accurate dose measurement (Barde et al., 2022; Barde et al., 2024; El Mansouri et al., 2022; Payne & Badawy, 2023; Christianson et al., 2012a). It was proved in a study that, taking the whole scan CTDIvol and Dw of the central slice will provide accuracy in SSDE determination. However, the limitation was only torso examination was analysed. The overall key factor to consider for accurate patient dose estimation in CT procedures is the patient's size. Theoretically, CT dose will decrease as the patient size increased. thicker body size patient will absorb lower dose than a small body size patient irrespective of same CTDIvol from the machine (Anam, Haryanto, Widita, Arif, & Dougherty, 2017b; Dio, Anam, & Hidayanto, 2022; Fahmi et al., 2019).

Besides, other associated factors need to be also investigated with respect to inaccuracy of dose estimation, despite taking the individual size such as X-ray tube angling, truncation artefact (cut off) which might be as a result of large patient body habitus, presence of air in certain anatomical region, metal artefact interference and presence of other body parts in the field of view (FOV) (Abdulkadir et al., 2022; Anam et al., 2021; Barde et al., 2024; Hsieh et al., 2014; Maltz et al., 2007; Terashima et al., 2019) . The CT scans were enhanced by applying several filters to eliminate noise, which allowed for the effective display of crucial details without posing an increased risk to the patients from higher doses of radiation (Omer et al., 2021).

2.2 CT dosimetry

CT imaging has led to increased population exposure to ionising radiation. However, concerns have been raised about the variability of radiation doses across patients, institutions, and countries. Dose optimisation is necessary. For dose reduction strategies, few components of CT scanner are useful for this purpose such as beam shaping-filters, collimators, and detectors, together with techniques (Brink & Morin, 2012; Chu et al., 2022; Iriuchijima et al., 2018; Terashima et al., 2019; Yu et al., 2009). Software developments have produced tube current modulation (TCM), which enables low-dose CT, and iterative reconstruction (IR) approaches. These advancements qualify CT as a dependable and preferred imaging technique. (Dewi et al., 2021). Tube current modulation (TCM) is one of the techniques employ for patient dose reduction, which depend on pre-selected noise level. Such that, the tube current is reduced when a lower attenuate is detected within a scan region. TCM technique proves decrease in patient dose compared to non-TCM (Anam et al., 2016a; Anam, Haryanto, et al., 2018). Dose reduction in CT is enhanced by use of noise reduction filters, tube current modulation, hardware optimization such as detectors and the use of some recent algorithms likes iterative (Gariani et al., 2018).

In early commercial CT scanners, filtered backprojection (FBP) was initially used for over four decades in CT image reconstruction, due to its faster and easier usage, but it has a drawback of inability to improve resolution and reduce radiation. Later, iterative reconstruction (IR) technique was developed which has the capability of reducing CT dose by lowering the tube current or tube potential, several iterative techniques exist in various scanners such as Adaptive Statistical Iterative Reconstruction (ASIR), Iterative Reconstruction in Image Space (IRIS), Model-Based Iterative Reconstruction (MBIR) and Sinogram-Affirmed Iterative Reconstruction

(SAFIRE) (Padole et al., 2015). Scanning is done at hight pitch in dual source CT scanners with 128 detectors. Studies reveal that high pitch CT is vital in radiation dose reduction in special and general CT examination. Image quality can be achieved without compromise with high pitch and iterative reconstruction (Gariani et al., 2018)

A previous study has demonstrated that the displayed dose, specifically the $CTDI_{vol}$ based on a 32cm phantom for chest and abdomen imaging, tends to underestimate the actual dose received by the patient. This discrepancy arises because the typical sizes of the chest and abdomen for most patients are smaller than the 32cm diameter of the phantom used for calibration (Ohene-Botwe et al., 2023; Xu et al., 2020). It is noteworthy that using a 16 cm phantom for chest and abdomen procedures in paediatric patients resulted in DLP values that were twice as high as those reported in studies using a standard body phantom. (Chu et al., 2022). The current CTDI phantoms underestimate the dose to average-sized adult and paediatric torsos by 40-70%. However, Combining CTDI with patient size estimations and anatomical knowledge can improve patient size-specific dose accuracy to more than 10% (Brady & Kaufman, 2012).

Besides, both $CTDI_{vol}$ and DLP present the average output of CT scanners which is subject to technical parameters of CT scanner like, pitch, tube voltage, rotation time, beamwidth, and tube current (Anam et al., 2021; Kostou et al., 2019; McCollough et al., 2011; Brady & Kaufman, 2012; Steuwe et al., 2020).

However, dose estimation in CT based on $CTDI_{vol}$ and DLP was proven to be not effective, due its inability to ascertains varying patient morphology. To address this, SSDE was introduced by AAPM (Brink & Morin, 2012; Dio et al., 2022; Porzio & Anam, 2022; Tahiri et al., 2021).

2.2.1 $CTDI_{100}$

The quantification of CT output was proposed to enhance multiple scan average dose (MSAD) by CTDI. Later, to ensure CT dosimetry supervision CTDI determination was modified to $CTDI_{100}$, $CTDI_{w}$ and $CTDI_{vol}$ subsequently. (Anam et al., 2021) In essence, CTDI measurement is achieved by dividing the result by the beam collimation after integrating a single scan dose profile along the longitudinal axis (z-axis). The effectiveness of $CTDI_{100}$ in wide beam CT has sparked questions because it is evaluated using a 100 mm pencil ionisation chamber. This arises from concerns that a 100 mm length might not adequately capture the scattered dose distribution. (Bauhs et al., 2008; Iriuchijima et al., 2018; McCollough et al., 2011).

To solve this issue, the phantom length should be increased to at least 45 cm, and dosage equilibrium should be achieved using a small-volume ion chamber (Figure 2.1). Throughout the field of view (FOV), the $CTDI_{100}$ varies. For example, in body CT imaging, the surface $CTDI_{100}$ ($CTDI_{100,p}$) is usually one or two times higher than the centre $CTDI_{100,c}$. The weighted CTDI ($CTDI_{w}$) represents the average $CTDI_{100}$ over the FOV.

2.2.2 $CTDI_{vol}$

When representing output dose in a series of scans, it's critical to account for any overlaps or spaces between x-ray beams caused by the x-ray source's subsequent rotations. Volume CTDI is used to characterize this ($CTDI_{vol}$). A conventional polymethyl methacrylate (PMMA) phantom is used to measure the $CTDI_{vol}$. It can have a fixed diameter of 16 cm to represent the patient's head or a diameter of 32 cm and a length of 15 cm to represent the patient's body. Both phantoms, however, do not accurately depict the range of patient sizes. A newborn child's geometrical size in a clinical setting can be less than 10 cm, whereas an obese adult patient can be more than 35 cm. Furthermore, the homogenous PMMA phantom was designed homogenous in nature. which is not a true representation of the heterogeneous human body. Due to these factors, $CTDI_{vol}$ only shows an accurate and consistent index of scanner output. (Anam et al., 2021; Brink & Morin, 2012; Chu et al., 2022; Haba et al., 2019; Steuwe et al., 2020)

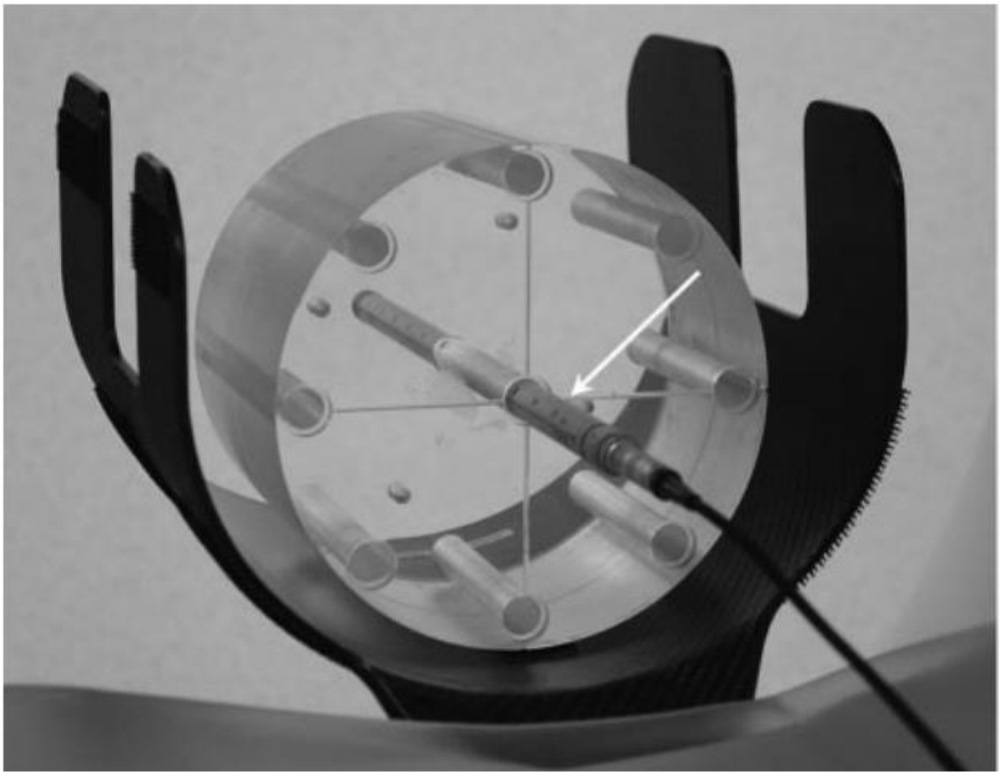

Figure 2.1 The experimental setup demonstrating $CTDI_{100}$ measurement using commercial PMMA phantom (CTDI phantom) and pencil dosimeter (Bauhs et al., 2008)

2.3 Introduction to size-specific dose estimation (SSDE)

American Association of Physicists in Medicine (AAPM) introduced the concept of size specific dose estimation (SSDE) to accurately presenting the CT dose with respect to individual body size of patient using effective diameter (D_{eff}), as recommended in AAPM Report No. 204 (Fahmi et al., 2019; Juszczyk et al., 2021; Kritsaneepaiboon et al., 2019; Parikh et al., 2018; Xu et al., 2020).

2.4 Size-specific dose (SSDE) existing techniques

SSDE consider both output dose and morphology of patient in determining patient dose. As recommended by AAPM in the Reports No. 204 and 220, the individual patient sizes were determined using two approaches, (AAPM, 2014; AAPM, 2011). Figure 2.3 described both methods to determine the individual specific size. (A) effective diameter, D_{eff} using LAT + AP diameters and (B) water equivalent diameter, D_w on axial CT images. Measurement of patient's physical dimension of the CT scout (topogram) or axial images can be obtained using electronic measurements tool available (AAPM, 2014; Wang et al., 2012).

2.4.1 Effective Diameter (D_{eff})

The first method is effective diameter (D_{eff}). D_{eff} is the square root product of anteroposterior (AP) and lateral (LAT) plane of scout or axial image of the subject, as shown in Figure 2.2 (Anam et al., 2021; Anam, Fujibuchi, et al., 2018b; Juszczyk et al., 2021). D_{eff} demonstrates patient size as a simple geometric measurement (Anam et al., 2016b).

It was reported that localizer radiographs overestimate patient size, leading to an underestimating of SSDE. Thus, AP and LAT diameters should be determined using transverse CT scans to avoid underestimate of SSDE (Pourjabbar, 2014).

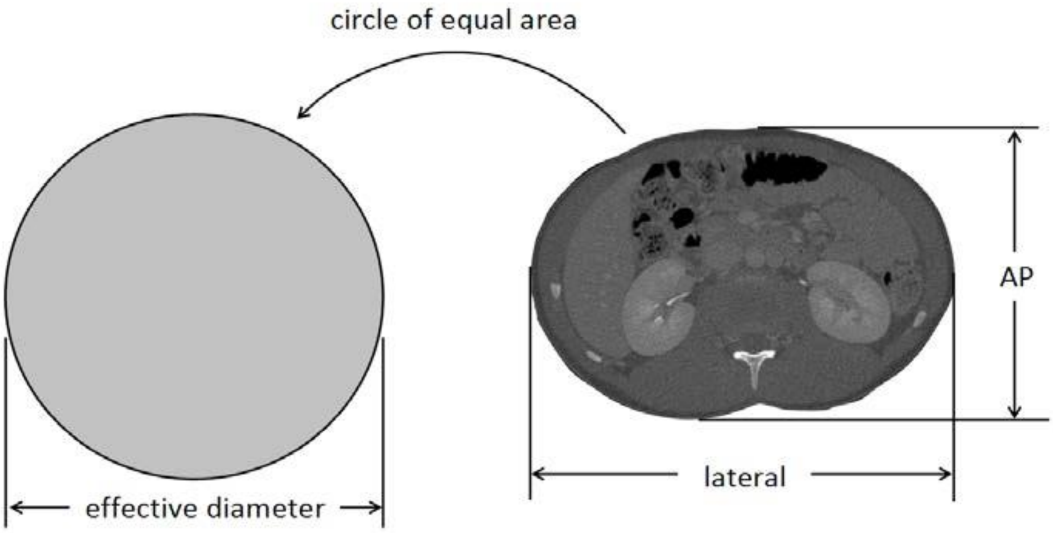

Figure 2.2 Measurement method of effective diameter, D_{eff} (AAPM, 2011)

The D_{eff} approach uses two physical dimensions, anteroposterior (AP) and lateral (LAT) lengths that were measured on axial CT images (central slice), as shown in Figure 2.3 (A). Deff was derived from the measured AP and LAT dimensions, as described in Equation 2.1 (Report of AAPM task Group 204, 2011).

Effective diameter (D_{eff})

$$D_{eff} = \sqrt{AP \times LAT} \tag{2.1}$$

While D_w is calculated as the entire area of the CT slice and number (Hounsfield units) measured at the ninth central slice or central image of the scan range, it can be calculated on either axial images or CT scanogram. Similarly, D_{eff} measurement was proved to be not statistically significantly different, irrespective of considering the slice area, middle or central point (Anamet al., 2017a; Daudelin et al., 2018; Juszczyk et al., 2021). The calculation of D_{eff} and D_w is considered accurate on either scanogram or axial CT images and is measured in centimetre (Barde et al., 2024; Fahmi et al., 2019; Leng et al., 2015).

2.4.2 Water equivalent Diameter (D_w)

The second approach is the water equivalent diameter, D_w. It uses the attenuation values (described as CT numbers in Hounsfield Unit, HU) within the region-of-interest (ROI) and the water equivalent area (A), as described in Equation 2.2 (AAPM, 2014).

Water equivalent diameter (D_w)

$$D_w = 2 \cdot \sqrt{\left(\frac{CT_{ROI}}{1000} + 1\right) \cdot \frac{A_{ROI}}{\pi}} \tag{2.2}$$

Figure 2.3 described both methods to determine the individual specific size. Two approaches used for specific size measurement: (A) effective diameter, D_{eff} using LAT + AP diameters and (B) water equivalent diameter, D_w on axial CT images.

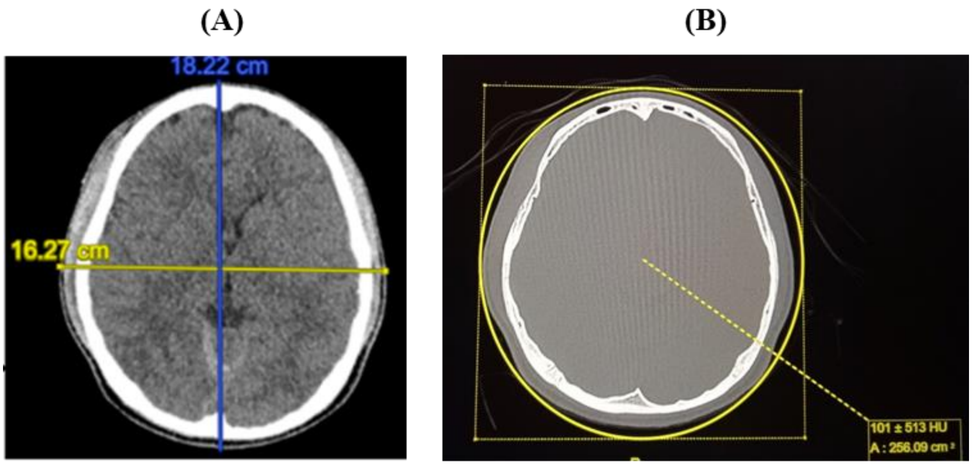

Figure 2.3 Two approaches recommended for individual size measurement, (A) effective diameter, D_{eff} using AP + LAT diameters, and (B) water equivalent diameter, D_w (Kabir et al., 2022).

2.4.3 Size specifications dose estimations computation

The measured patient sizes, D_{eff} and D_w on CT images were then used to determine the SSDE conversion factor (Table 2.1). The SSDE is derived by multiplying a conversion factor value from AAPM look up table by a concept termed effective diameter (D_{eff}), calculated from patient cross sectional CT image, as geometric representation (Anam et al., 2016b, 2021; Anam et al., 2017a; Juszczyk et al., 2021). SSDE was derived by multiplying scanner displayed dose, $CTDI_{vol}$ by a $CTDI_{vol}$-to-SSDE conversion factor, *f* corresponding to estimated patient size (D_{eff} or D_w) (AAPM, 2014; Report of AAPM task Group 204, 2011). the computation of D_{eff} and D_w is possible by either manual or automated method (Porzio & Anam, 2022)

Table 2.1 The correction factor values for the D_{eff} obtained from AAPM Task Group Report 204.

D_{eff} (cm)	Conversion factor	D_{eff} (cm)	Conversion factor
8	2.76	27	1.37
9	2.66	28	1.32
10	2.57	29	1.28
11	2.47	30	1.23
12	2.38	31	1.19
13	2.3	32	1.14
14	2.22	33	1.1
15	2.14	34	1.06
16	2.06	35	1.02
17	1.98	36	0.99
18	1.91	37	0.95
19	1.84	38	0.92
20	1.78	39	0.88
21	1.71	40	0.85
22	1.65	41	0.82
23	1.59	42	0.7
24	1.53	43	0.76
25	1.48	44	0.74
26	1.43	45	0.71

In AAPM Report 204, the conversion factor for normalisation of patient size is based on CT images of abdomen and pelvic regions. Moreover, it can be applicable for thorax CT because the presumed errors reported were less than 20% (Daudelin et al., 2018; AAPM, 2019). Localizer based D_w estimation is limited due to challenge of calibrating localizer pixel value with respect to D_w (Zhang et al., 2018).

The nomenclature of AAPM Report 204 was changed to include the body part examine and phantom size use in expressing the SSDE by inputting the nomenclature in the conversion factor and $CTDI_{vol}$ as superscript and subscript respectively (AAPM, 2019). The equation as thus "*B*" represent body while "*H*" represents head.

$$SSDE = f^{B32}.CTDI_{vol,32} \tag{2.3}$$

$$SSDE = f^{B16}.CTDI_{vol,16} \text{ (pediatric body metrics)} \tag{2.4}$$

$$SSDE = f^{H16}.CTDI_{vol,16} \tag{2.5}$$

where f is conversion factor which equates to the measured sized (D_{eff} or D_w) obtained from the AAPM table. The superscript alpha numericals indicates the body part and phantom size used in the conversion. While $CTDI_{vol}$ is the average dose output of the x-ray tube along the scan range of the CT scanner. Similarly, the subscript numerical indicates the phantom size used in the conversion. The SSDE is the final estimated dose value based on the specific size, that is obtained from the product of the conversion faction based on either D_{eff} or D_w measurement and the displayed $CTDI_{vol}$ obtained from the CT scanner.

2.4.4 Truncation effect on Size specific dose estimation

A previous study examining the effects of truncated axial CT images, characterised by truncation percentage (TP). The study reported correction factor to accurately predict D_w and SSDE. The correction factor increased exponentially with increased TP. The highest number of truncated images was observed in thoracic CT scans, while the least occurrence was noted in head CT scans. The thorax region showed significant corrections in D_w and SSDE, whereas the head and abdomen regions did not demonstrate significant changes. Truncation of axial CT images leads to overestimation of SSDE and underestimation of D_w (Anamet al., 2017b). Head tilting or gantry tilt is recommended by AAPM for head and brain CT scanning as a strategy to reduce the radiation dose to the eye lens (Hardy et al., 2019). However, the gantry tilt may produce inaccurate AP and LAT diameter due to tilted view of the head region. It was reported that truncated (cropped) CT images can lead to underestimation of patient's size (AP and LAT diameter) values and overestimation of SSDE in look-up tables, unlike localiser radiographs, which underestimate SSDE (Pourjabbar, 2014).

2.5 Reported practices for size specific dose estimation.

The SSDE estimation based on both D_{eff} and D_w were derived from the displayed dose descriptors, the volume CT dose index ($CTDI_{vol}$) and dose length product (DLP). It is suggested that the $CTDI_{vol}$ and SSDE methods should not be used interchangeably in clinical practice. It was reported that, to calculate the mean SSDE for CT exams of the torso in adults, a single image in the centre of the scan range and the average $CTDI_{vol}$ from the full scan were adequate (Burton & Szczykutowicz, 2018; Leng et al., 2015).

However, the International Commission on Radiological Protection (ICRP) suggested that SSDE method is preferred for patient with significant variant. It is recommended that both parameters should be used especially for paediatric patients. A phantom of 16 cm and 32 cm is the standard reference size used to calculate $CTDI_{vol}$ for small and large size representing standard head and body. However, manufacturers of CT scanner tend to use incorrect reference size (larger size) in paediatric CT protocol due to lack of smaller size of reference phantom used (Juszczyk et al., 2021). Body and head PMMA phantom comparison revealed the differences of ±20% in the estimated SSDE. It was reported that in non-contrast CT studies, considerable number of patients (250) were scanned at five different tube potentials other than 120 kV, published size-dependent conversion factors, reveals correctly calculated size-specific dose estimates (Abdulkadir et al., 2022; Viggiano et al., 2021). It was reported that, there is insignificant variation within ±10% in SSDE measurement using phantom, with IndoseCT software this is within acceptable range ±20% (Dio, Anam, & Hidayanto, 2022).

The use of D_{eff} is not conclusive in determining patient characteristics due to its inability to analyse body composition based on parts involve. Therefore, D_w was employed to address this limitation that encompass patient attenuation in SSDE determination (Anam et al., 2016a). The attenuation properties are the basic parameter that should be considered in determining SSDE, as it is more relevant and accurate than considering geometric size alone. The thorax and abdomen regions will demonstrate different $CTDI_{vol}$ even their geometric dimensions look similar since lungs have less attenuation and therefore abdomen demonstrate a higher absorbed dose (AAPM, 2014). The estimation of SSDE or D_w analyses the slice location during data acquisition. The mean value of nine central slices was used to estimate the water equivalent diameter, D_w and SSDE.

It was reported that there is better correlation in D_w than D_{eff} in dose estimation based on size of thorax and abdomen region. This was supported by some theoretical assumptions in previous work, which stated patient size considering attenuation (D_w) gives more accurate dose estimation (AAPM, 2014; Kusuma Dewi et al., 2021). In children aged below 15 years old, there is a correlation between the body diameter and body weight. As weight can be simply obtained, thus it may serve as surrogate in paediatric SSDE (Khawaja et al., 2015). Previous studies also reported that tube current modulation (TCM) is a recommended technique for patient dose reduction in CT imaging. SSDE and $CTDI_{vol}$ were estimated based on the effective tube current effect. The rotation of tube for each complete revolution was used to calculate the modulated tube current, which also affects the $CTDI_{vol}$, SSDE and water equivalent diameter, D_w. The $CTDI_{vol}$ estimation relies on the mean tube current in milliampere (mA). It was also reported that there was a strong correlation between SSDE and average tube current (Juszczyk et al., 2021).

$CTDI_{vol}$ demonstrates a different relationship between D_w and D_{eff} when TCM is off. Reducing the patient diameter leads to constant $CTDI_{vol}$ in head CT protocol. However, SSDE increases when head diameter is decreased. While in thorax CT protocol with TCM on mode will lead to reduction in $CTDI_{vol}$ due to reduced patient diameter. In thorax, reduction in the patient diameter leads to SSDE reduction. The patient dose is drastically reduced by using TCM. It ensures constant noise level and reduce patient radiation dose, by continuous change in tube current (Anam et al., 2016a). A study was conducted by Sitti et al. (2022) using D_w approach for SSDE to ascertain the variation exist based on the type of phantom used for CT localiser calibration. Radiation dose can be measured by simulation of the human body using a phantom. In the study two different phantoms of varying shapes and make ups were used (CTDI and step-wedge). The same exposure setting was used with a 64 Slice Siemens CT machine. The pixel values and water equivalent thickness relationship between the analysed CT localiser images of the phantoms were calculated. Twenty patients CT localizer images were calculated by D_W approach to measure the SSDE, the result reveals that there is no significant difference recorded for D_w and SSDE measurements between the two phantoms used (Sitti et al., 2022).

Another study was carried out by Payne and Badawy (2023). The study compared the average Water Equivalent Diameter (D_w) values derived from CT Contour (an open-source programme) to vendor-specific data provided by Philips scanners. A random sample of 50 adult and 50 paediatric abdomen-pelvis CT images from Philips scanners were analysed with CT Contour, and $D_w$values were retrieved from the DICOM headers. CT Contour's average D_w values were marginally lower than those obtained using vendor-specific estimates. The average disagreement was -5.62% for adults and -2.88% for paediatric datasets. Both techniques produced clinically acceptable estimates of the average D_w. However, CT Contour is not validated for any CT protocols other than abdomen-pelvis examinations. CT Contour can produce comparable average D_w readings to vendor specific SSDE and D_w computations in CT dosimetry. Additional study is required to validate these findings for additional CT methods (Payne & Badawy, 2023). In another vein, it was reported, in a study of 250 adult and paediatric patients, tube potential had no significant impact on water-equivalent diameter at chest and abdomen CT, and only a modest effect (0.7 cm) at 70-kV head CT. The tube potential had a minor impact on the size-specific dose