THE SOUTH AFRICAN RADIOGRAPHER volume 57 number 2 | NOVEMBER 2019
OPEN ACCESS online only

peer reviewed **ORIGINAL ARTICLE**

Preliminary review of paediatric coronary computed tomography angiography doses at a children's hospital in Johannesburg South Africa

B Van Wyk[1,2] *MSc* | **J Naidoo**[1] *FCRad (Diag Radiology) Dip Paed Radiology* | **A Mazibuko**[1] *BTech Radiography* | **P Leshalabe**[1] *Diploma Radiography*

[1] *Department of Radiology, Nelson Mandela Children's Hospital, Johannesburg, South Africa*
[2] *Sefako Makgatho University, Medical Physics Department, Pretoria, South Africa*

Abstract

Purpose. To share our experience of reviewing paediatric coronary computed tomography angiography (CCTA) dose using a GE Revolution EVO computed tomography scanner.

Methods and materials. This study had three phases. Phase 1 consisted of validating the scanner's DLP by measuring the GE Revolution EVO CT dose using a test phantom. Phase 2 retrospectively analysed the paediatric dose of the first two CCTA cases and was followed by a prospective analysis of the next six CCTA cases treated at Nelson Mandela Children's Hospital (NMCH). Phase 3 consisted of image quality analysis by three radiology consultants.

Results. The percentage differences between the displayed and estimated $CTDI_{vol}$ were well within the limits prescribed in literature. A dose reduction of 2.32 mSv was seen in the prospective cases when compared to the retrospective results. A 5-point radiologist rating revealed that the diagnostic radiologists were overall satisfied with the subjective image quality even after changes made to the scan parameters.

Conclusion. Small changes in scan parameters can reduce the dose to children significantly whilst keeping image quality for CCTA cases.

Keywords paediatric, optimising, computed tomography, coronary, dose length product, image quality

INTRODUCTION

Exposure to ionising radiation is of special concern in children because of the greater vulnerability to radiation effects of this population compared to adults. The organ doses to children are much higher than that of adults. Children are more sensitive to radiation-induced cancer than adults, especially since they have many years of life ahead.[1] Knowing this the radiology team at Nelson Mandela Children's Hospital (NMCH) created paediatric protocols for all computed tomography (CT) cases treated at the institution, in order to deliver radiation doses that are as low as reasonably achievable (ALARA).

Coronary computed tomography angiography (CCTA) is often used as a non-invasive imaging modality for the evaluation of coronary artery disease. Its advantages include high spatial resolution, fast patient throughput, and relatively low cost compared to other advanced cardiovascular modalities.[2] Protocols, based on mass (weight), were determined using nominal tube voltage settings and automatic tube current modulation for CCTA. Several authors have recommended reducing the tube current-time product or the tube potential or both as a function of patient size, with the goal of obtaining constant diagnostic quality and image noise at reduced radiation.[2-6] CCTA protocols, which are based on patient size, tube current-time (mAs), and tube voltage (kV), should be designed to deliver radiation doses that are ALARA.[7] A patient's radiation dose is directly proportional to mAs; decreasing mAs leads to lower doses. Image noise (graininess) is however inversely proportional to the square root of radiation dose.[8-9] Lower tube voltage leads to lower radiation dose and better contrast; however, the image noise will increase. Automatic exposure control (tube current modulation), compensates for beam attenuation at different sections of the body by changing the tube current to maintain constant image quality over a whole scan range.[10] Tube current can be changed both while the tube is rotating around the patient and as the table moves in the direction of patient's long axis (z-modulation). Automatic exposure control (AEC) software predicts a patient's attenuation beforehand from a topogram. Maintaining the same image quality is not always practical. In cases where there is more attenuating tissue at the end of the scan range, AEC tends to increase the tube current on that area, even if a noisier image would be adequate for a diagnosis, such as evaluating lung tissue at the liver level. Setting the maximum tube current value for modulation can reduce a patient's radiation exposure without compromising image quality Therefore a patient-centric approach should be adopted, whereby both the dose exposure and image quality are optimised.

For the General Electric (GE) Revolution EVO CT scanner used in our study, the tube current (mA) is controlled both while the rotation angle of the tube is changing (xy-modulation) and parallel to a patient's z-axis.[11] The latest planning image (scout), either posterior-anterior (PA) or anterior-posterior (AP) is used for calculating the required mA value. It is very important to centre a patient in the middle of a scanner aperture for proper mA modulation.[11] Wrong centring increases a patient's surface dose and negatively affects image quality. Even with these optimised scan parameters tailored to an individ-

ual patient, the image quality issues still remain;the greatest impact on interpretation of the CT being misalignment and image noise. High-resolution CT of the chest is generally performed using mAs settings between 100 and 200. Lowering the mAs value will result in increased image noise and a potential decrease in detecting low-contrast detail. Therefore care must be taken to ensure that image noise properties are appropriate for accurate lumen segmentation.

The CT dose volume index ($CTDI_{vol}$) describes the radiation dose on a scanned area; this is measured in a standard quality assurance phantom.[12] The latter is an acrylic cylinder with diameters of 16 cm (head) and 32 cm (body). The weighted dose length product (DLP_w); is the product of $CTDI_{vol}$ and the length of the scanned area. In the dose displays of most scanners both $CTDI_{vol}$ and DLP_w are given. Since the CT scanner dose display indicates the radiation dose to a cylindrical standard size phantom, it does not take a patient's size into consideration, and thus it does not indicate reliably the 'real radiation' dose received by a patient. For this reason, determination of a patient's organ dose, effective dose and risk requires evaluation of the DLP with the conversion factors obtained from the scanning parameters as per recommendations from the International Commission on Radiological Protection (ICRP).[13] The accuracy of a dose display reading is verified by regular measurements, as part of the quality assurance of scanners. The objective of this study was to preliminarily review paediatric CCTA doses at NMCH.

MATERIALS AND METHODS

Phase 1. Validation of the study's DLP

To ensure that the study's DLP that was used was correct, the researchers independently verified the scanner's $CTDI_{vol}$ through an experimental procedure. A 0.6 cc ion chamber, electrometer, thermometer, barometer, 16 cm diameter (head) and 32 cm diameter (body) phantoms were used. The phantoms and set-up are demonstrated in Figure 1. The scanning parameters used were 120 kV and 300 mAs for the body phantom; 100kV and 260 mAs for the head phantom. Both scans had a pitch of 1 and a beam width of 10 mm.

Using equations 1 and 2 the $CTDI_{vol}$ was calculated and compared to the displayed $CTDI_{vol}$ on the screen.

Figure 1. CTDI phantom.

(Equation 1)

$$CTDI_w = \frac{1}{3} CTDI_{100}^{centre} + \frac{2}{3} CTDI_{100}^{periphery}$$

(Equation 2)

$$CTDI_{vol} = \frac{CTDI_w}{Pitch}$$

Phase 2. Retrospective and prospective analysis of the CCTA paediatric cases

The DLP of the first two CCTA cases scanned in the radiology department at NMCH was retrospectively reviewed. This was followed by a prospective review of six CCTA cases. The data collected were not gender specific. There were no inclusion or exclusion criteria in this study as they would not have added any value.

All eight cases underwent a lateral (LAT) and anterior-posterior (AP) scout covering the heart and coronaries. This was followed by an ECG-gated axial data acquisition of the coronaries, whereby the 'smart preparation' automatically triggered a diagnostic scan. The scan parameters of all eight cases are demonstrated in Tables 1 and 2. The first two cases were scanned with a slice thickness of 1.25 mm; the next six had slice thickness of 2 mm; all of the cases (n=8) had a pitch of 1.5.

The AEC modulation for the diagnostic scan had a range of 80-210 for the first two cases and 60-120 for the next six cases. The estimated dose length product (DLP) as displayed on the scanner's screen was used to calculate the paediatric effective dose using equation 3.

(Equation 3)

$$e = k \times DLP$$

where,
e = effective dose
k = is the correction factors as per ICRP 103 for new born chests (Table 3)

Phase 3. Image evaluation by three radiologists

Image evaluation was performed on a standard 3D-enabled workstation, GE Advantage Sim with a standardised window level. Each subject was analysed independently by three paediatric radiologists in terms of image quality. The criteria for image quality were the subjective perception of image noise, soft-tissue contrast,

Table 1. Scan parameters of the retrospective cases

	SCOUT	SMART PREPARATION		DIAGNOSTIC SCAN	
		Pre-set time delay (s)	mA	mA Range	Pre-set kV
Case 1	AP and LAT	1	30	60-350	100
Case 2	AP and LAT	1	30	60-350	100

Table 2. Scan parameters of the prospective cases

	SCOUT	SMART PREPARATION		DIAGNOSTIC SCAN	
		Pre-set time delay (s)	mA	mA Range	Pre-set kV
Case 3	AP and LAT	2	30	40-200	80
Case 4	AP and LAT	2	20	40-200	80
Case 5	AP and LAT	2	10	40-200	80
Case 6	AP and LAT	2	10	40-200	80
Case 7	AP and LAT	2	10	40-200	80
Case 8	AP and LAT	2	10	40-200	80

Table 3. ICRP 103 correction factors for new-borns

Tube Voltage (kV)	Chest
80	0.0823 mSv.mGy^{-1}.cm^{-1}
100	0.0739 mSv.mGy^{-1}.cm^{-1}

and degree of image degradation by artifacts. All structures were assessed using a 5-point scale for diagnostic quality: 1=unacceptable; 2=suboptimal; 3=adequate; 4=good; and 5=excellent. On the basis of the individual scores an average quality score was calculated for each patient. The image quality mean ($\bar{x}$) was also calculated using the following formula:

(Equation 4)

$$\bar{x} = \frac{x_i}{n}$$

where,
x_i is the sum of all the scores
n is the number of samples

RESULTS

Phase 1. Validation of the study's DLP

The results of the independent CTDI measurements are given in Tables 4 and 5.

The DLP findings are presented in Tables 6 and 7.

Phase 2. Retrospective and prospective analysis of the CCTA paediatric cases

The results of the calculated effective doses for all paediatric cases are given in Tables 6 and 7.

Phase 3. Image evaluation by three radiologists

The results of the 5-point radiologists' subjective image quality, blinded to the details of the CT datasets and parameters, are given in Tables 8 and 9.

The image quality mean value for the first two cases was 3.9 ± 0.1; the mean value for the next three cases was 3.6 ± 0.3. The mean values overlap within the first standard deviation.

The images of the prospective paediatric cases are presented in Figures 2 to 7.

DISCUSSION

The percentage differences between the displayed and estimated $CTDI_{vol}$ were well below the limit prescribed by the ICRP of 20%.[13] The displayed results used in this study were therefore validated. Cases 1 and 2 were scanned using 100kV and a mA range of 60-350 mA for the diagnostic scan. This resulted in an average effective dose of 5.68 mSv. A reduction of mA range to 40-200mA, and the kV to 80kV, resulted in an average effective dose of 3.35 mSv. This was done for a range of patient masses.

The population however is quite small and therefore a larger similar study is recommended. This study therefore recommends reduced scan parameters for paediatric cases. It also recommends weight categorised scanning parameters for paediatric CCTA cases.

The image quality was maintained with these dose reduction methods; the three paediatric radiologists commented that they were overall satisfied with the subjective image quality of all eight images. The researchers found that increasing the pre-set time delay for bolus/contrast detection, and reducing the mA for these scans, advocated for the dose reductions discussed above. The radiologists however did highlight concerns related to patient positioning, which was not part of this study, and should be part of a larger similar study on CCTA optimisation.

A limitation of our study is that radiation dose was not directly measured. It was derived from the DLP. Furthermore, determining paediatric

Table 4. The independently measured and displayed $CTDI_{vol}$ for the body phantom

	$CTDI_{100}$ (rad)	Estimated $CTDI_{vol}$ (mGy)	Displayed $CTDI_{vol}$ (mGy)	Percentage deviation
Central	19.79	32.22	31.15	3.4%
Periphery (1)	40.51			
Periphery (2)	40.17			
Periphery (3)	33.78			
Periphery (4)	39.30			

Table 5. The independently measured and displayed $CTDI_{vol}$ for the head phantom

	$CTDI_{100}$ (rad)	Estimated $CTDI_{vol}$ (mGy)	Displayed $CTDI_{vol}$ (mGy)	Percentage deviation
Central	65.28	64.52	63.05	2.3%
Periphery (1)	68.15			
Periphery (2)	63.97			
Periphery (3)	58.09			
Periphery (4)	66.33			

Table 6. Doses received by the retrospective cases

	Mass (kg)	Age (days)	Scanned DLP (mGy.cm)	Effective Dose (mSv)
Case 1	5.0	90	80.71	5.96
Case 2	4.0	22	73.13	5.40

Table 7. Doses received by the prospective cases

	Mass (kg)	Age (days)	Scanned DLP (mGy.cm)	Effective Dose (mSv)
Case 3	4.0	13	62.79	5.18
Case 4	2.5	6	50.39	4.15
Case 5	2.0	5	45.18	3.72
Case 6	4	12	16.60	1.36
Case 7	20	330	26.93	2.22
Case 8	5	60	42.74	3.52

Table 8. Image quality on the retrospective cases

	Radiologist 1 Score	Radiologist 2 Score	Radiologist 3 Score
Case 1	4	3.5	3.5
Case 2	4	4	4

Table 9. Image quality on the prospective cases

	Radiologist 1 Score	Radiologist 2 Score	Radiologist 3 Score
Case 3	4	4	4
Case 4	3	3.5	3.5
Case 5	3	3.5	3.5
Case 6	3	3.5	3.5
Case 7	3	3.5	3.5
Case 8	3	3.5	3.5

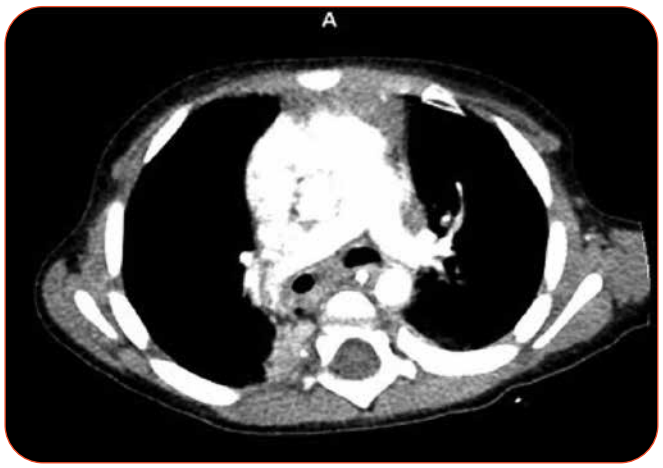

Figure 2. Static axial "screen grab" image of case 3.

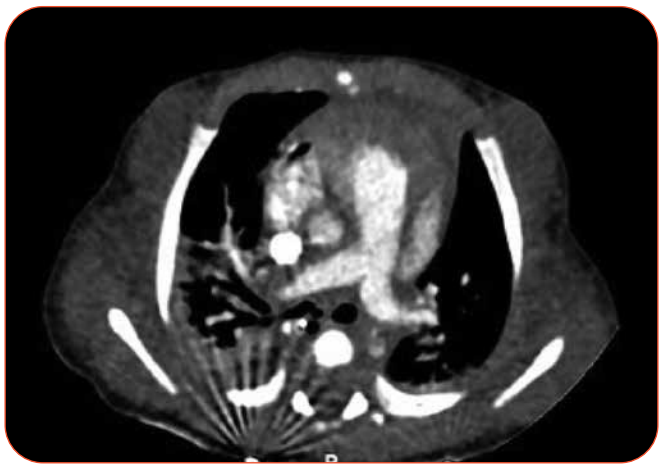

Figure 3. Static axial "screen grab" image of case 4.

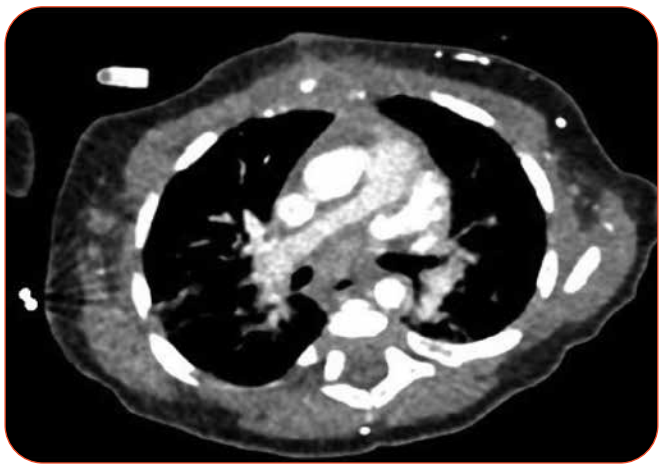

Figure 4. Static axial "screen grab" image of case 5.

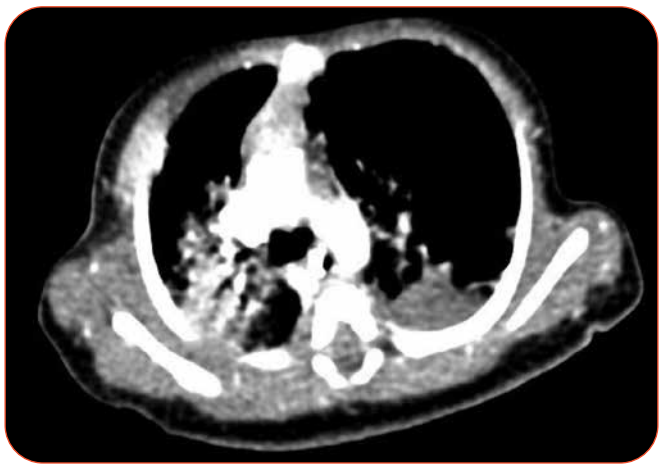

Figure 5. Static axial "screen grab" image of case 6.

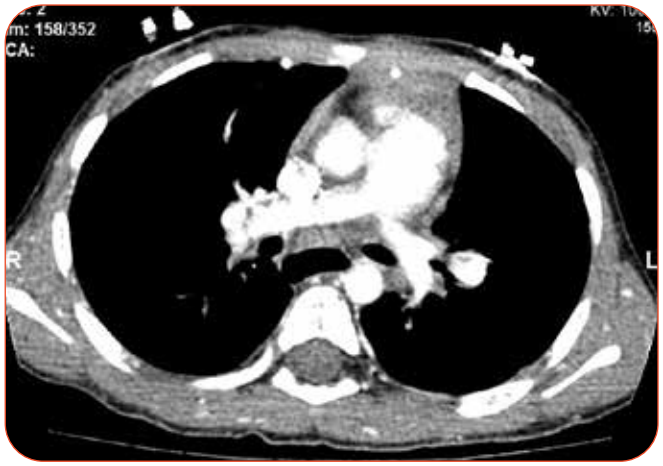

Figure 6. Static axial "screen grab" image of case 7.

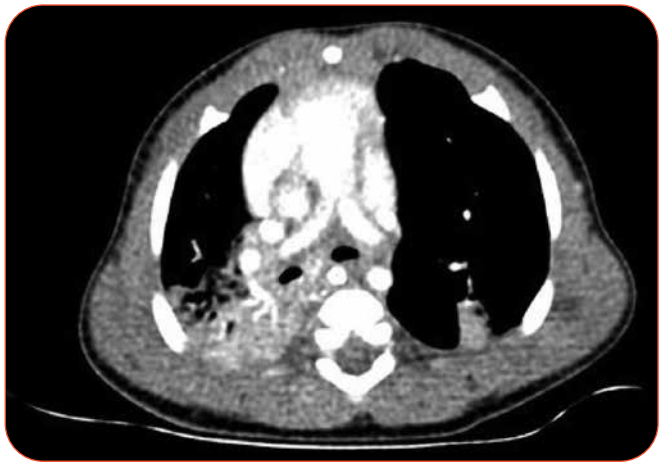

Figure 7. Static axial "screen grab" image of case 8.

radiation dose is less straightforward than in adults because the DLP is calculated on the basis of the $CTDI_{vol}$, and the protocol for the measurement of $CTDI_{vol}$ is based on only two sizes of a cylindrical acrylic phantoms: 16 cm (simulating an adult's head) and 32 cm (simulating an adult's body). Phantom studies show that the mean imparted section dose increases with smaller patient diameter because there is less tissue absorbing radiation.[8] Thus the larger diameter phantom used would underestimate the measured doses. The eight CCTA cases studied in this paper are barely a representative of the paediatric population at NMCH. This was a limitation of our study. Our study however shows how subjective image quality can be maintained whilst reducing paediatric dose.

CONCLUSION

All paediatric CT protocols/parameters should be reviewed according to paediatric mass. The results presented in this paper showed how slight changes in the tube current, slice thickness and tube voltage can reduce the paediatric DLP, which was not at the expense of the image quality.

ACKNOWLEDGEMENTS

We appreciate the three radiologists' participation in this study. Sincere thanks to the staff at Nelson Mandela Children's Hospital for allowing us to do the study, as well as Sefako Makgatho University for the ethics clearance.

A special word of thanks is expressed to Dr Thulani Nyathi for his "proof reading" of the paper and wise guidance.

COMPETING INTERESTS

The authors do not have financial or personal relationships which may have inappropriately influenced them in writing this article. They do not have any conflicts of interest either.

ETHICS APPROVAL

The Sefako Makgatho Health Sciences University Research Ethics Committee (SMUREC) approved this study, SMUREC Ethics Reference Number: SMUREC/M/330/2018: (J) Journal

AUTHORS' CONTRIBUTIONS

All authors contributed to writing the paper. Collection of data done by BVW, AM and PL.

REFERENCES

1. Brenner DJ, Elliston CD, Hall EJ, Berdon WE. Estimates of the cancer risks from pediatric CT radiation are not merely theoretical: comment on "point/counterpoint: in X-ray computed tomography, technique factors should be selected appropriate to patient size –against the proposition." (commentary) Med Phys 2001; 28:2387-2388.
2. Kamel IR, Hernandez RJ, Martin JE, Schlesinger AE, Niklason LT, Guire KE. Radiation dose reduction in CT of the pediatric pelvis. Radiology 1994; 190:683-687.
3. Brody AS. Thoracic CT technique in children. J Thorac Imaging 2001; 16:259-268.
4. Lucaya J, Piqueras J, Garcia-Pena P, Enriquez G, Garcia-Macias M, Sotil J. Low-dose high-resolution CT of the chest in children and young adults: dose, cooperation, artifact incidence, and image quality. AJR 2000; 175:985-992.
5. Mettler FA, Wiest PW, Locken JA. CT scanning: patterns of use and dose. J Radiol Prot 2000; 20:353-359.
6. Linton OW, Mettler FA Jr. National Conference on Dose Reduction in CT, with an emphasis on pediatric patients. AJR 2003; 181:321-329.
7. Yu L, Bruesewitz MR. Optimal tube potential for radiation dose reduction in pediatric CT: principles, clinical implementations, and pitfalls. RadioGraphics 2011; 31: 853-848.
8. Lucaya J, Piqueras J, Garcia-Pena P, Enriquez G, Garcia-Macias M, Sotil J. Low-dose high-resolution CT of the chest in children and young adults: dose, cooperation, artifact incidence, and image quality. AJR 2000; 175:985-992.
9. Boone JM, Geraghty EM, Seibert JA, Wootton- Gorges SL. Dose reduction in pediatric CT: a rational approach. Radiology 2003; 228:352-360.
10. Kalra MK, Dang P, Singh S, Saini S, Shepard JA. In-plane shielding for CT: Effect of off-centering, automatic exposure control and shield-to-surface distance. Korean Journal of Radiology 2009; 10 (2): 156-163.
11. Leipsic J. Acquisition and reconstruction techniques for coronary CT angiography. GE Healthcare Scanner Platforms, heartflow.com.
12. International Atomic Energy Agency Human Health Series Number 1, Quality assurance for PET and PET/CT systems, 2009.
13. Deak P, Smal Y and Kalender W. Multisection CT protocols: Sex and age specific conversion factors used to determine effective dose from dose length product. Radiology 2010; 257:1.