Assessment of the Radiation Dose during 16 Slices CT Examinations

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Abstract: *In this work. Two studies were presented. The first one is an experimental study of the influence of different scan parameters (potential tube. current tube and pitch) on the radiation dose is presented. It has been demonstrated that the radiation dose increases if we increase the tube current or the potential current. Contrary to the pitch. the radiation dose is reduced when the pitch increases. The second study is a Monte Carlo validation of a CT named SOMATION emotion from Siemens using GATE. Results were carried out for different voltage 80.110.130 kVp. Results of the simulation are presented and good agreements are observed (less than 2.6% for head phantom. and less than 4.6% for body phantom for all applied voltages). The influence of changing the phantom diameter on the CTDIw was also presented. the CTDIw decreased nonlinearly with increasing the phantom diameter. Moreover. the CTDIw differences decreased if the phantom diameter increased. It was demonstrated that the CTDI100 values can be characterized as functions of the kVp. mAs and diameter of phantoms. Based on these results. it is possible to optimize the CT parameters in clinical applications. Finally the effective dose was calculated using the method dose length product and the conversion factors for four clinical protocols.*

Keywords *: CT, CTD,. GATE, SOMATOM Emotion CT, potential tube, current tube and pitch.*

I. INTRODUCTION

Nowadays medical physics is one of the important research domains being developed in Morocco[1][2][3]. The revolution of the medical imaging modalities made the Computed Tomography (CT) a major technique in diagnostic. However. CT is the most irradiating if compared with other technique. The current CT scanners have an extensive choice of exposure factors and employ techniques that can significantly influence the radiation dose given to the patient [3]. It can deliver a dose of 50 to 500 times more than a standard radiological examination [4]. This dose level in CT may be responsible for a long-term carcinogenesis process (stochastic effect) [5-6]. For that reason. The exact calculation of radiation doses is a key factor to optimize CT protocols, and to better understand the potential radiation risks of CT examinations. To evaluate the radiation dose in CT. The computed tomography dose index (CTDI) is estimated. It represents the absorbed dose along the longitudinal CT scanner axis during a single X-ray tube rotation [7].In this work , a CT system Siemens SOMATOM Emotion scanner (16 slices) was simulated using the Geant4 Application for Tomography Emission (GATE) platform [8-9] which allows simulating accurately a large range of medical devices using the Monte Carlo techniques. Presently, GATE is playing a key role in the design of new medical imaging devices. in the optimization of acquisition protocols and in the development and assessment of image reconstruction algorithms and correction techniques. [10] In this paper. We aimed to estimate the CTDIs. Using GATE and validate the results against the experimental ones. The experimental estimation of the CTDI was made in the national institute of oncology of Rabat in Morocco. The results of the GATE model validation, the simulated CTDI for various phantom sizes and the estimation of effective dose are presented.

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II. MATERIALS AND METHODS

A. GATE and high performance computing (HPC)

All simulations were performed using GATE [11]. It is a simulation platform designed specifically for emission tomography. Geant4 toolkit [12] is used as the basis layer for Monte-Carlo simulations of medical systems in GATE. GATE allows the user to realistically model experiments using accurate physics models and time synchronization for detector movement through a script language contained in a macro file. Various systems have been simulated with GATE. However. The Monte Carlo simulations using GATE are limited by long execution times and then alternative solutions have been studied [13]. To handle a big number of events and reduce the computing time of the simulation. Simulation task is split into several jobs using the HPC of the High Institute of Health at Settat. The execution time of the task was almost 18 hours per simulation (with ~10^{10} simulated particles). The output of the simulations was 3D dose-maps. Using the option Dose Actor in GATE.

B. Quantities in CT dosimetry

Since the 1980s. The computed tomography dose index (CTDI) has been used as a standard dose descriptor in CT dosimetry [14-15]. CTDI100 denote the estimated CTDI with a 100mm long pencil ionization chamber placed in the CT head and body phantom [6]. It is defined as:

$$CTDI_{100} = \frac{1}{nT}\int_{-50\ mm}^{-50\ mm} D(z)dz \qquad (1)$$

where n represents the number of slices. T the slice thickness and D(z) the dose profile along the axis of rotation over 100 mm.

The ion chamber used in this work is called Radcal 10x6-3CT and the corresponding electrometer is Radcal Radiation Monitor. Model 1515 (Radcal Corporation. Monrovia. California) The chamber is surrounded by a polyacetal cover. It has a volume of 0.3cm^3 with 10cm length. (Figure. 1).

Figure 1. The ion chamber for CTDI: Radcal 10x6-3CT.

In this work., the ion chamber was modelled as two concentric cylinders. The chamber wall was modeled as a cylinder with an inner diameter of 6.70mm and outer diameter of 10 mm filled with the material C552. The electrode is modeled as a C552 cylinder with 1.3mm diameter. The space inner wall-electrode is filled with air.

Additionally. there are two standard phantoms (head and body) to estimate the computed tomography dose index (CTDI) [16]. To validate our Gate-modeled CT. we have used the CTDI head and body phantom. Its represented two cylindrical phantoms with 15 cm in length and with a diameter of 16 cm and 32 cm for the head and the body phantoms respectively. They are made of polymethyl methacrylate (PMMA) and containing five holes to implement an insert or the pencil ion chamber (Figure. 2).

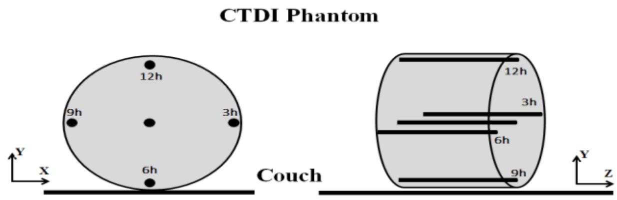

Figure.2 CTDI phantom geometry

The weighted computed tomography dose index $CTDI_W$ represents a single estimation of radiation dose to the phantom. It is calculated by combining the doses ($CTDI_{100}$) measured at the central hole (CTDIc) with the average value of $CTDI_{100}$ measured at the peripheral holes ($CTDI_{12}$. $CTDI_3$. $CTDI_6$ and $CTDI_9$; where the number subscripts refers to hole positions: 12h. 3h. 6h and 9h).

$$CTDI_w = \frac{1}{3} CTDI_c + \frac{2}{3}(CTDI_{12} + CTDI_3 + CTDI_6 + CTDI_9) \quad (2)$$

The $CTDI_{vol}$ is a standardized measure of the radiation output of a CT system. that enables users to gauge the amount of emitted radiation [7]. It is defined as:

$$CTDI_{vol} = \frac{CTDI_w}{Pitch} \quad (3)$$

Where the pitch represents the table increment per CT rotation divided by slice thickness.

C. CT scanner

All measurements and simulations were performed using a SOMATOM EMOTION 16 CT (Siemens Medical Solutions. Erlangen. Germany). the distance from the focal spot to the isocenter and the detector was 535 mm and 940 mm respectively. the scanner operated in both axial and helical modes at available tubes voltage 80kV. 110kV and 130kV. the tube current can be selected from 20 to 345 mA in a 1 mA steps. This 16 CT scanner operates at different beam collimations (2.4. 7.2. 9.6. 10. 14.4. 16 and 19.2 mm). Rotation times available are 0.5. 0.6. 1 and 1.5 s. Siemens CT scanner has a system called Care Dose 4D which can reduce dose by 66 %.. **(**Figure. 3)

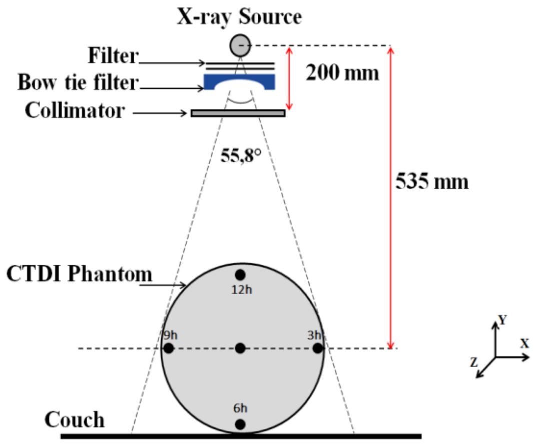

Figure. 3 The geometry of the SOMATOM Emotion CT.

D. Experiment

1. Influence of different CT parameter on CTDIw.

Three sets of measurements were affected on the CT scanner using the pencil ion chamber called Radcal. The first measurement was used to compute the CTDIw as function of kVp. One axial scan (one full rotation) of 10 mm slice thickness were acquired in CTDI phantom for beam energies of 80. 110 and 130 kVp. The tube current was set to 100 mA and the rotation time was fixed at 1s. The measurements were obtained using the body and the head CTDI phantoms. All obtained results were converted to computed tomography dose index using the 100 mm pencil ion chamber (CTDI100). The second measurement was used to compute the CTDIw as function of mA. The measurement was done using a fixed energy (110kVp). for tube currents of 100. 200 and 300 mA. In this case the CTDI head phantom was used. The other parameters were the same as the first measurement. The Third measurement was used to compute the CTDIw as function of Pitch. The measurement was done using 110 kVp energy and 100 mA current for pitchs of 0.5. 1 and 1.5. In this case the CTDI head phantom was used. The other parameters were the same as the first measurement.

2. Simulation of SOMATOM Emotion 16 slices CT.

2.1. X-ray source

The aim of this part was to set up an X-ray source (Figure. 2) using GATE. where we define the energy. type of particles. angles. …etc. This platform provides an option to implement an X-ray spectrum using extra code as SRS-78[16].

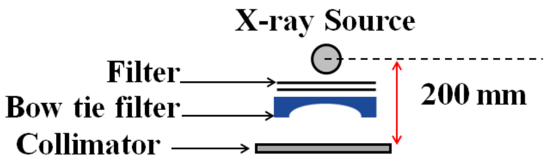

Figure. 4 The geometry of the X-ray source.

The source energy spectrum was generated using the software SRS78 [16].
and then inserted in GATE against General Particle Generator (GPS).
It was defined as a gamma particle point source with an isotropic angular distribution equal to:
$88.72^{o} \leq \varphi \leq 91.28$ and $62.1^{o} \leq \theta \leq 117.9^{o}$.
where φ and θ exhibit the spherical angle coordinates (Figure. 5).

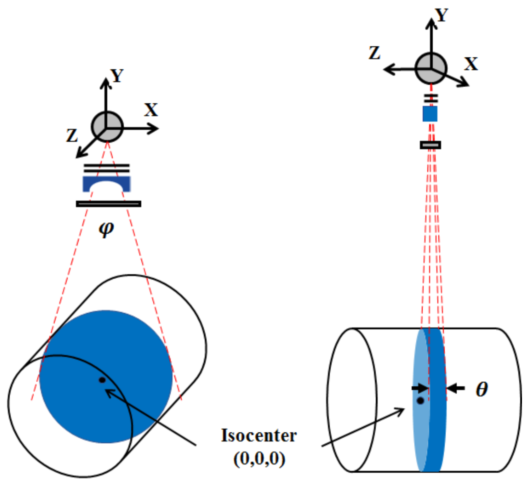

Figure. 5 The definition of the angles θ and φ of the X-ray distribution[3].

The energy photon spectrum was characterized by an energies histogram (in keV) and their intensity value in each respective histogram point (energy). To accelerate the simulations. we have rebinned the energy spectrum form 0.5keV to 5keV [16] (Figure. 6).

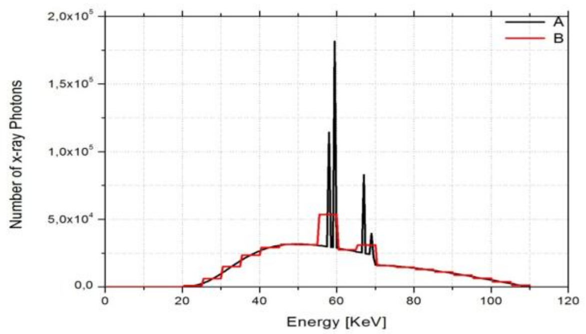

Figure. 6 the X-ray spectrum for a tungsten target at 110keV generated using the SRS-78 program. The curves A and B shows respectively the relative intensity using 0.5keV and 5keV step.

To model the rotational movement of the CT X-ray beam in our simulation. we have rotated the phantom-bed instead of rotating the X-ray tube. In this rotation. we have divided the 360o in 360 steps of 1o angle rotation in each step (2.7ms per step. for a full rotation of 1s). Geant4 standard model of physical process without cuts and variance reduction techniques (VRTs) has been used in this simulation. no variance reduction techniques (VRTs) were used in the modeling of the physical processes. nor cuts on the simulated physical volumes.

2.2. . CTDIw and $CTDI_{100}$ simulation

To compare the Monte Carlo simulation results to the experimental ones made above. the equivalent simulations were affected. To validate our simulations. we have followed the same conditions used during the experimental measurements. The simulations were achieved for both body and the head CTDI phantoms. The couch was modelled as rectangular parallelepiped made of carbon. The absorbed dose (CTDI) by the phantoms using GATE was obtained directly from the simulations through “DoseActor” option.

C. effective dose

The effective dose was estimated using the conversion factors from AAPM 96 report and the dose length product DLP. Through the effective dose we can compare the radiological risk from different imaging modalities.

$$DLP = CTDI_{VOL} \times \text{scan length (mGy*cm)}$$
$$\text{Effective dose} = DLP \times \text{conversion factor k (mSv)}$$

Region of body	Conversion factor k (mSv/mGy*cm)
Head	0.0021
Neck	0.0059
Chest	0.014
Abdomen-pelvis	0.015

Table 1:Conversion factors for adult (AAPM report N°96)

III. RESULTS AND DISCUSSION

A. Influence of different CT parameter on CTDIw

The measured CTDIw at various energies is found in figure 6:

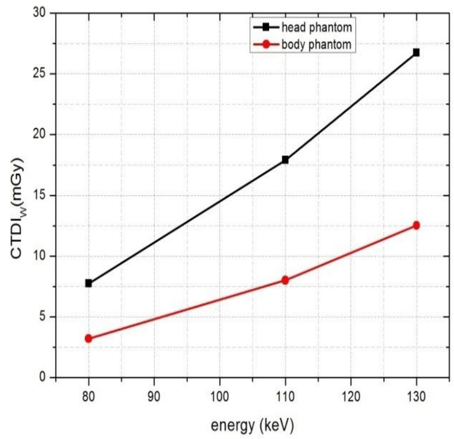

Figure 7. Measured CTDIw in the head and body phantoms vs of the tube potential for 100 mAs

Figure 7 shows that Reducing kVp from 130 to 80. dose drops by factor of 3.45 in the case of head phantom. In the case of body phantom this factor is equal 3.90.

The measured CTDIw for different tube currents and pitchs are presented respectively in Figure 8 and Figure 9.

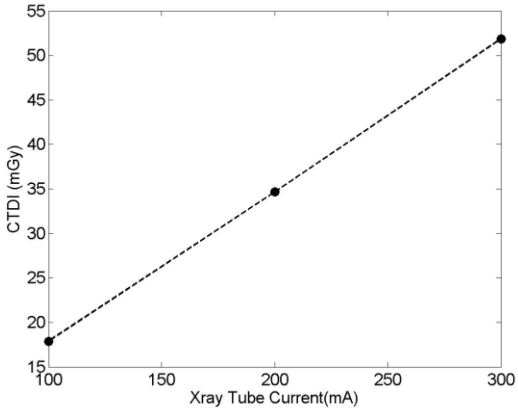

Figure 8. Measured CTDIw in the head phantoms vs the tube current.

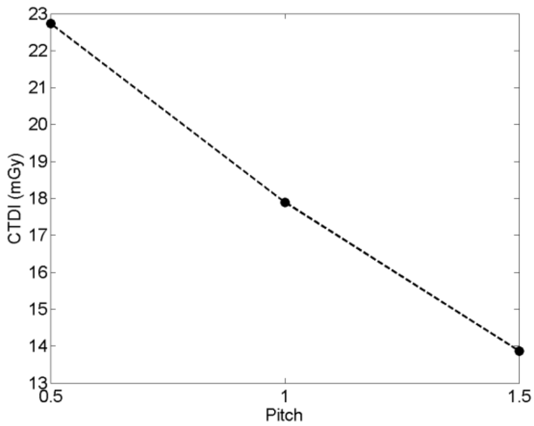

Figure 9 Measured CTDIw in the head phantoms vs the the Pitch.

Figure 8 exhibit that when we reduce tube current by half. the dose is reduced by half also.

Based on figure 9. We observe that if the pitch is equal to 1.5 the dose is reduced by a factor of 0.28. It the pitch is equal to 0.5 the dose is increased by a factor of 0.26. We conclude that in one hand. Increasing the tube current or the potential current increase radiation dose if the other parameters were unchanged. In the other hand. Increasing the pitch reduce radiation dose if the other parameters were unchanged.

From these results. We conclude that the dose scales almost linearly with tube current. Tube potential and pitch respectively.

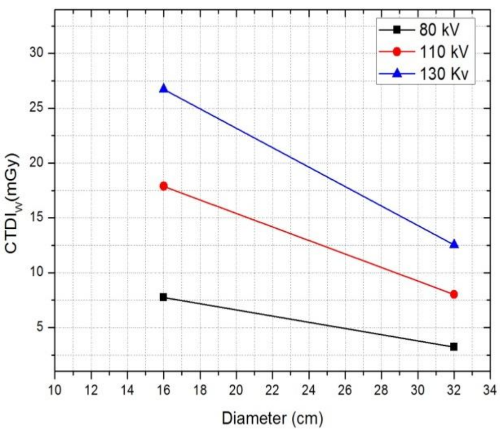

Figure. 10: $CTDI_w$ as a function of diameter for 110 kVp.

Figure 10 shows that at different tube potentials. when reducing the phantom diameter. The radiation dose is decreasing

B. Simulation result:

To have accurate and realistic simulation. a very high number of particle (~ 10^{10} in our case). Has been generated. The radiation doses obtained from the simulations are presented in this section. Comparison with experimental data is discussed.

kVp	$CTDI_W$ (mGy)	
	Head phantom	Body phantom
80	7.59 ± 0.30	3.17 ± 0.19
110	17.57 ± 0.45	8.41 ± 0.32
130	26.02 ± 0.54	11.98 ± 0.36

Table 2. simulation results $CTDI_w$ for head and body phantom.

The simulation of our GATE-modeled SOMATOM Emotion CT was performed using the geometrical parameters of the scanner and validated through the simulation of CTDIw. The comparison of simulated and measured CTDIw has been carried out using the equation 4.

$$\% \, difference = \left(\frac{SC-MC}{MC}\right) \times 100 \qquad (4)$$

Where SC is simulated $CTDI_w$. MC denote the measured $CTDI_w$.

Table 4 exhibits the comparison of the obtained CTDIw with the measured ones.

kVp	Head phantom		% difference	Body phantom		% difference
	Measured $CTDI_w$ (mGy)	Simulated $CTDI_w$ (mGy)		Measured $CTDI_w$ (mGy)	Simulated $CTDI_w$ (mGy)	
80	7.74	7.59	1.9	3.21	3.17	1.2
110	17.89	17.57	1.8	8.02	8.41	4.6
130	26.73	26.02	2.6	12.53	11.98	4.3

Table 3. Comparison between simulated and measured data for head and body phantom at 100mAs for different kVp

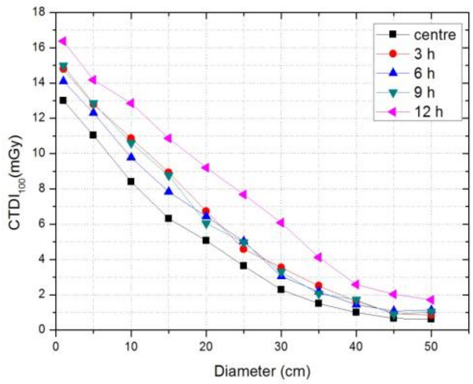

Figure 11. $CTDI_{100}$ in PMMA as a function of diameter for 80 kVp. 100 mAs at various positions (centre. 3h. 6h. 9h. 12h)

Concerning the head phantom. the agreement observed between the simulation and the experiments represents 1.9%. 1.8% and 2.6%. respectively for 80kVp. 110kVp and 130kVp. The comparison of the simulated $CTDI_w$ with the experimental ones.

shows an agreement within 1.2%. 4.6%. and 4.3%. for 80kVp. 110kVp and 130kVp. respectively. This comparison demonstrates a good agreement between measured and simulated $CTDI_W$ with maximum differences of 4.6 % (body) and 2.6% (head) for all kVp levels applied. These deviations are mainly due to the difficulties to model a precise geometry of the scanners. in addition to the fact of the uncertainties in the x-ray spectrum (filtration) in the simulation.

Figure 11 illustrates the diameter dependence of dose in PMMA phantom. for both center and peripheral locations. The $CTDI_{100}$ in all locations decreases with diameters greater due to the increasing attenuation path.

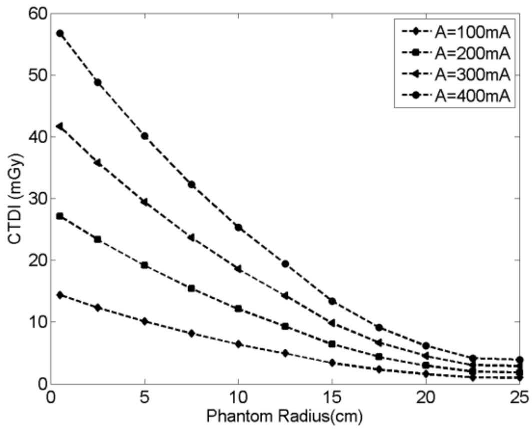

Figure 12. show $CTDI_W$ as a function diameter for 80 kVp and 10 mm slice thickness for various mAs.

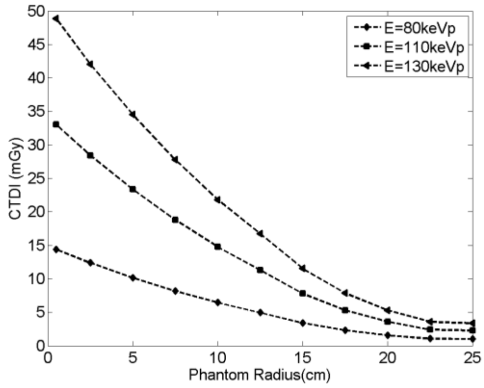

Figure 13 . show $CTDI_W$ as a function diameter for 100 mAs and 10 mm slice thickness for various kVp

Figure 13 illustrates the phantom diameter dependence of dose ($CTDI_w$). for different kVp. The obtained results showed that the CTDIw decreased nonlinearly with increasing the phantom diameter. For all applied kVp. it has been observed that if the phantom diameters increase. the difference between CTDIw becomes small [19-20].

Figure 12 shows the phantom diameter dependence of dose ($CTDI_w$). for different mAs. Like the relation between $CTDI_w$ values and kVp. the CTDIw decreased nonlinearly with increasing the phantom diameter. Moreover. the CTDIw differences decreased if the phantom diameter increased [19-20].

C. Effective dose

The following table shows the results of the effective dose from the DLP and the conversion factors:

Region of body	Gender	kVp	mAs	slices thickness	Scan lenght (cm)[21]	CTDI vol	DLP	k-factor	effective dose
Head	Male	130	100	10 mm	13.1	26.02	340.862	0.0021	0.71
	female	130	100	10 mm	14	26.02	364.28	0.0021	0.76
Neck	Male	130	100	10 mm	18	26.02	468.36	0.0059	2.76
	female	130	100	10 mm	17.6	26.02	457.952	0.0059	2.7
Chest	Male	130	100	10 mm	25	11.98	299.5	0.014	4.19
	female	130	100	10 mm	28	11.98	335.44	0.014	4.69
abdomen-pelvis	Male	130	100	10 mm	46.4	11.98	555.872	0.015	8.33
	female	130	100	10 mm	47.4	11.98	567.852	0.015	8.51

Table 5 effective dose from DLP

The effective dose was slightly higher for female than the male for the four clinical protocols; the maximum effective dose was reported for the abdomen-pelvis region about 8.33 mSv for male and 8.51 mSv for female. The obtained effective doses were compared with one study conducted in Tehran [21], we found that for the four protocols our effective doses were higher than the study [21].

IV. CONCLUSION

In this work. We have presented to studies. The first one is an experimental study of the influence of different scan parameters (potential tube. current tube and pitch) on the radiation dose is presented. It has been demonstrated that the radiation dose increases if we increase the tube current or the potential current. Contrary to the pitch. the radiation dose is reduced when the pitch increases. The second study is a Monte Carlo validation of a CT named SOMATION Emotion from Siemens using GATE. The determination of the radiation doses. from the simulations performed with the designed model of Siemens SOMATION Emotion 16 slices CT. using GATE. shows good agreements with the experimental data. The comparison between the simulated $CTDI_W$ and the measured ones show a good agreement < 2.6% for head phantom and <4.6 % for body phantom. The influence of changing the phantom diameter on the $CTDI_w$ was also presented. the CTDIw decreased nonlinearly with increasing the phantom diameter.

Moreover. the CTDIw differences decreased if the phantom diameter increased. These results demonstrated that the CTDI might be characterized as a function of tube potential. tube current and phantom diameter that can help to improve CT parameters for clinical applications. The effective dose obtained in this study was slightly higher than other studies.

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