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Effects of Patient Dose Reduction Efforts on Image Quality for Thoracic CT in A Moroccan Hospital

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ABSTRACT

Thoracic CT imaging is a strongly recommended means of medical imaging. It is accompanied by a high exposure to radiation due to the number of examinations carried out, which requires efforts to reduce the CT dose index (CTDI) while trying to preserve image quality. To this end, this study proposed the possibility of introducing two new imaging protocols for chest tomography. A 16-slice HITACHI SUPIA CT scanner and two phantoms were used to investigate CTDIvol and image quality; the first phantom was made of PMMA and the other of AAPM model 610. Three tube voltages were studied by varying the intensity of the tube current (mA): 120 kVp (120-160-210-230) mA, 100 kVp (160-200-240-290) mA, and 80 kVp (230-260-300-350) mA. The values for noise uniformity and accuracy, contrast-to-noise ratio (CNR), and spatial resolution (SR) were determined using IndoQCT c22a. 92 software. The first thoracic protocol proposed with 100 kVp compared to 120 kVp resulted in a 27.51 % reduction in CTDIvol, a 20 % increase in mA, and a 19.50 % increase in noise. The CNR showed a slight regression of 23.08 %. For the second scan procedure at 80 kVp, the CTDIvol was reduced by 53.32 %, while noise was increased by 53.95 %. There was no statistically significant difference in CNR and SR ($p > 0.05$) when kVp and mA were reduced compared to the routine protocol. It is suggested that it is possible to adopt two new acquisition protocols at 100 kVp and 80 kVp while reducing the patient exposure dose (CTDIvol) by 28 % and 54 % and taking into account the effect of varying these parameters on image quality. Their choices must be made by integrating and considering clinical issues and a good understanding of the pathophysiology and imaging results of the suspected condition. Consequently, radiologists and technicians should always take a part in improving imaging practices in such a way as to make more effective use of radiation.

INTRODUCTION

CT imaging is an important part of medical imaging that uses X-rays to obtain details of body parts [1]. This diagnostic equipment emits high doses compared with other conventional imaging methods [2,3] A previous study showed that the cancer risk caused by radiation increases following exposure to an effective dose of more than 100 mSv [4]. Following Advanced Trauma Life Support guidelines, thoracic imaging (thoracic X-ray and thoracic CT scan) are the most widely used for investigating patients with chest pathologies [5].

Dose selection for CT scans is an extremely complex procedure. The principal indicator of performance in CT dose quantification was the Computed Tomography Dose Index (CTDI) [6,7]. which determines the dose received for the region scanned for each CT slice, enabling the average

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dose to be assessed. The CTDI is measured with a polymethyl methacrylate (PMMA) phantom of 16 or 32 cm diameter and corresponds to the calculated average radiation dose in the scan segment [8].

The physical performance of the image quality obtained by the scanner is evaluated, including noise uniformity and accuracy, contrast/noise ratio (CNR), linearity, slice thickness, as well as spatial resolution (SR) [9,10].

Morocco aims to increase the number of computed tomography [11]. Most of these medical imaging centers use protocols predefined by the manufacturer that are not always clinically appropriate [12,13]. A poorly selected protocol may increase the exposure dose as it may affect image quality. A higher dose exposure can be deployed to compensate for image quality in thoracic CT imaging [14-16].

Several studies have shown that it is possible to reduce the intensity of the electric current (mA) while reducing the CTDIvol and conserving image quality [17,18]. Other studies have reported that it is possible to adjust the voltage (kVp) tube without degrading image quality by reducing the radiation dose [19,20].

The absence of studies in our country has obliged us to conduct this type of research. The objective is to suggest two new protocols for scanning the thoracic by reducing the (CTDIvol), tube voltage (kVp), and increasing the current intensity (mA) while maintaining image quality (uniformity and accuracy of noise, CNR and SR).

MATERIALS AND METHODS

In this study a Hitachi Supria 16-bar scanner was used to accommodate with new MSCT technology, installed since 2018 in a Moroccan radiology department with an average workload of 110 scans per week. Air calibration was adopted the same day before starting the scans. The phantoms were scanned three times for every tube current regulation [21]. The kVp and mA were manually adjusted from the lower to the upper limit between 120 and 350 (mA), the scan parameters for the adult thoracic protocol were described as 120 kVp (120-160-210-230) mA, 100 kVp (160-200-240-290) mA, and 80 kVp (230-260-300-350) mA, 0.75 second rotation time, 16 (detectors) × 1.25 mm (detector collimation), FOV 364 mm, pitch 1.0625, section thickness 3.75 mm, and matrix size is 512 × 512 pixels. Dose validation was done in a Phantom body (32 cm diameter). IntelliEC mode pressed on standard deviation (SD) which defines the mode of modulation intensity of the electric current has been deactivated during all experiments [22].

The measurements were obtained using a PMMA phantom, 32 cm in diameter simulating the thorax of an adult of cylindrical shape. Its length is 15 cm and a density of 1.19 g/cm^3, pierced with 5 holes, the central position evokes the mediastinum, the other positions 12 and 06 are used to represent the anatomy of the anterior region of the thoracic that includes the sternal bone and the dorsal spine respectively, the two holes 03 and 09 simulate the axillary regions. Note that these holes are offset by 90 degrees [21].

The PMMA has been placed at the end of the gantry table, these openings have been aligned by means of light marks (Fig. 1). The proper centering of the phantom has been ensured by its fixing with the blockers, removable inserts have been placed in these openings with the exception of the one in the center that housed a calibrated ionization chamber (model 10X6-3CT) with 100 mm of slowness and a measurement precision of 4 % and a dose measurement zone of 200nGy-1kGy [8], the latter has been connected to an electrometer of type RADCAL CORPORATION (California, USA) and a software of Accu - Gold+ interface for the display of the parameters of output, the remaining empty holes were plugged by removable rods not used during the measurement. When the phantom was correctly positioned, an image of the topogram of the PMMA was scanned. The scout was placed and the FOV was 364 mm.

The $CTDI_{100}$ values were obtained by reading the values recorded on the electrometer. These measurements were repeated three times for each position of the ionization chamber to obtain three readings. These experimental measurements were similar to measuring procedures of the International Atomic Energy Agency (IAEA) Code of Practice, TRS 457 series.

Fig. 1. PMMA phantom with a diameter of 32 cm.

Fig. 2. AAPM CT phantom model 610.

To determine the volume CTDIvol, the weighted CT Dose Index (CTDIw) was determined using the center and periphery values Eqs. (1) and (2), which are in mGy [23].

$$CTDIw = \frac{1}{3} CTDI_{100,center} + \frac{2}{3} CTDI_{100,periphery} \quad (1)$$

$$CTDIvol = \frac{CTDI_w}{\text{Pitch}} \quad (2)$$

The phantom used to evaluate image quality performance was a 610 model from the American Association of Physicists in Medicine (AAPM) group [24]. Positioning was achieved by matching the phantom height axis with the coronal, axial and sagittal lasers of the CT scanner (Fig. 2). No laser modifications were made during this study.

Scanning was performed using the same parameters used to scan the PMMA phantom with a FOV of 266 mm. Specific performance parameters investigated when testing the CT phantom include the following:

Noise was determined as the SD of the Hounsfield numbers in a region of interest (ROI), the noise level must be similar, the fifth section of the phantom has been exploited. The ROI (Region of Interest) was measured by plotting circles covering 40 of the homogeneous diameters of the phantom at 5 positions, in the center, at 3 o'clock, 6 o'clock, 9 o'clock and 12 o'clock (Fig. 3a). This process was measured for each voltage change and of the current intensity [25], the central value corresponds to the precision of the noise (Fig. 3b). The SD between the maximum and minimum determines the uniform noise value of the image. As the scan was not performed to BAPETEN's standard parameters (i.e., 120 kV, 300 mAs and 8 mm slice thickness), the noise values were translated to the four edges (standard deviation at the center minus standard deviation at each edge). This value is called the measured noise value (σm). The σm was then converted to a normalised noise value (σs) or a noise value in the BAPETEN standard (σs) using Eq. (3).

$$\sigma_s = \sigma_m \frac{kV_m}{120\ \text{kV}} \sqrt{\frac{\text{mAs}_\text{m} \times \text{slicethickness}_\text{m}}{300\ \text{mAs x } 8\ \text{mm}}} \quad (3)$$

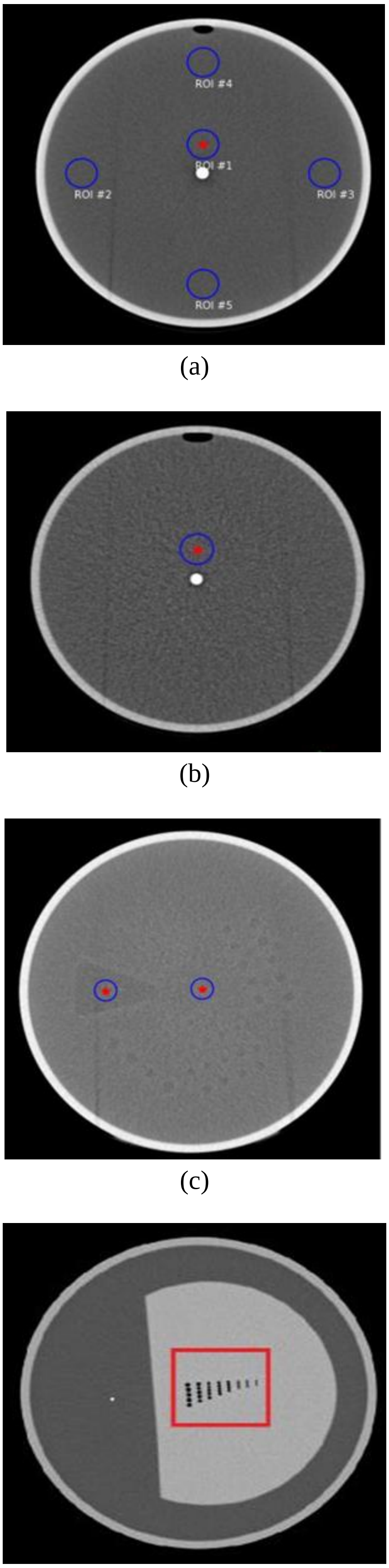
CNR was evaluated in a solid acrylic equivalent cylinder with 12 cavities (610-06 module) using Eq. (4). Pairs of spheres with a diameter of 25.4, 19.1, 12.7, 9.5, 6.4, and 3.2 mm were spaced from center-to-center twice their diameter along a centerline, filled with external orifices with distilled water and iodized contrast solution. The ratio of the sphere to the bottom for the water CT number was about 1 % (10 HU):

$$\text{CNR} = \frac{\text{numberCT}_\text{int} - \text{numberCT}_\text{ext}}{SD} \quad (4)$$

Or CT Number$_{int}$ et Number$_{ext}$ is the difference between the average CT number inside and outside the ROI with a 30-pixels diameter of the homogeneous module of the phantom inside the 25.4 mm diameter sphere and outside the sphere in an area where the attenuation is uniform (Fig. 3c) [10].

Spatial resolution is the measure of the ability to differentiate adjacent small objects in an image, it has been evaluated for the thorax protocol by selecting a rectangular ROI of dimension (80×60) pixels so that it encompasses the 40 holes. The pairs of holes in the fourth section with a diameter of 1.00mm must be fully visible (4 pairs of holes/mm). The object (module 610-03) has eight series of air holes with five holes per series (Fig. 3d). The diameters of the holes in the eight series are 1.75, 1.5, 1.25, 1.00, 0.75, 0.61, 0.50, and 0.40 mm. Each gnawed is spaced 5mm apart, the distance between each gnawed hole is equal to the diameter of the hole, it is measured by counting the number of even holes/mm [26].

The images saved in DICOM format were reviewed automatically by IndoQCT v22a.92 software installed on an Intel i7 processor at 2.9 GHz, in which the analysis processes were performed by selecting the automatic method. These observers were not aware of the changes in the kVp and mA of the thorax protocol. The first observer is a medical physicist and radiation protection in one of Morocco's hospitals, experienced in the visualization and evaluation of quality assurance images produced from quality assurance phantoms. The second observer is a radiologist in one of the Hospitals in Morocco with more than 6 years of experience in the field. Note that the spatial resolution was assessed based on visual investigation by observers.



(a) (b) (c) (d)

Fig. 3. (a) Noise uniformity, (b) Noise accuracy for the AAPM phantom, (c) Automatic placement of ROI CNR measurements for the AAPM phantom, (d) Hole pair objects in the AAPM phantom (SR).

The measurements results were utilised for statistical analysis using a simple linear regression test to investigate the effect of tube current on image noise uniformity values. We applied a test of equality of expectations: paired observations comparing means to investigate differences in CTDIvol. The differences between the mean values of the CNR and SR were evaluated with the Mann-Whitney test and the independent t-test by comparing the scans performed with the 120 kVp and 100 kVp groups on the one hand and the scans acquired at 120 kVp and 80 kVp on the other. IBM SPSS version 21 software was deployed for statistical analysis. Statistical significance was set at $p < 0.05$.

RESULTS AND DISCUSSION

The ease perform of rapid scanning with improved CT technology can result in increased radiation dose to patients. It is therefore necessary to give particular importance to quality assurance and dose assessment. In low- and middle-income countries, quality control equipment is not available in all establishments. Performing dose reduction efforts is therefore essential during CT scans of the thoracic.

In this study, 36 acquisitions were performed by selecting the standard protocol of the adult thorax and manually varying kVp and mA. Three series were discussed; One of the current protocols 120 kVp and two other series for two proposed protocols 100 kVp and 80 kVp, taking into account the effect of the variation of these parameters on image quality. To do this, we measured the CTDIvol, which is a good indicator for studying the behaviour of acquisition parameters.

Table 1 shows the results of the thoracic exam. Four physical quantities of image quality such as uniformity and noise accuracy, contrast/noise ratio, spatial resolution, and the CTDIvol of the CT examination, were investigated to determine two new scanning protocols.

The results in Table 1 show that the mA values increase as the radiation dose received by the object increases. For acquisitions carried out at 120 kVp, a manual setting of the mA (120-160-210-230) showing variations between 7.1 and 13.56 mGy are higher than those carried out at 100 kVp a mA (160-200-240-290), with a mean CTDIvol of 10.49 ± 2.8 mGy (for scans at 120 kVp) compared with 8 ± 1,988 mGy consider as average at 100 kVp ($p < 0.01$). The dose was halved when acquired at 80 kVp at mA (230-260-300-350) compared with 120 KVp, with a mean CTDIvol of 5.24 ± 0.93 mGy ($p < 0.01$).

Table 1. $CTDIvol$, noise accuracy and uniformity, CNR and SR values obtained for all scan configurations.

Tube voltage (kVp)	Tube current (mA)	CTDIvol (mGy)	Noise uniformity (HU)	Noise Accuracy	CNR	SR
120	120	7.1	1.2	9.9	0.8	5
	160	9.45	1	8.92	1.1	5
	210	11.85	0.9	8	1.38	6
	230	13.56	0.8	7.63	1.5	7
100	160	5.79	1.5	13.9	0.7	5
	200	7.18	1.3	11.2	0.81	6
	240	8.59	1	9.7	1	6
	290	10.44	0.9	8.47	1.1	7
80	230	4.16	1.9	13.9	0.5	4
	260	4.89	1.7	13.3	0.66	5
	300	5.6	1.5	12.8	0.79	5
	350	6.33	1.3	12	1	6

The Moroccan Agency for Nuclear and Radiological Safety and Security (AMSSNuR) is responsible for the regulatory framework that organises these procedures under Law 142-12. To date quality control tests have never been carried out due to the absence of a medical physicist in radiology departments to carry out and develop quality assurance protocols for radiology equipment. The European (Council of the European Union 1997) and French regulatory frameworks require that equipment and acquisition protocols be adjusted to the size of the patient and adapted to the use of radiology equipment. In fact, the absence of standard norms in our country encouraged us to adopt the BAPETEN standard norms as reference values, which are presented in Table 2.

The variations in noise uniformity based on tube current intensity for the three voltages increase with the increase in tube current intensity (Table 1). The lowest mean value was 0.8 HU, obtained at 120 kVp with 230 mA, while the highest was 1.9 HU during the scan acquired at 80 kVp with 230 mA. These values remain below 2 HU, not exceeding the tolerance limit, and stay within the acceptable limits of the BAPETEN standard regulation (Table 2).

The noise accuracy is the second measure of image quality that was evaluated. When transitioning from 120 kVp to 100 kVp, the rates of variation in mean values increase from 11.84 % to 24.37 %, and when changing the tube voltage from 120 kVp to 80 kVp, these rates increase from 41 % to 56.27 %. The measured mean noise was higher for lower tube currents (mA).

The lowest CNR mean value was 0.5 to 80 kVp, 230 mA and the highest was 1.5 detected on images acquired by 120 kVp 230 mA. The rate of change from 120 kVp to 100 kVp increased by 12.50 % (0.90 ± 0.18, $p = 0.19$) around 26.67 % and from 33.33 % to 40 % when the voltage was reduced from 120 KVp to 80 KVp (0.73 ± 0.21, $p = 0.06$).

Table 2. BAPETEN Standard used in CIRS 610 AAPM Phantom Image Tests.

Parameter	BAPETEN Standard
Noise uniformity	≤ 2
Noise accuracy	Within 8 HU
CNR	≥ 1
SR	Discernible, less than 1.00 mm

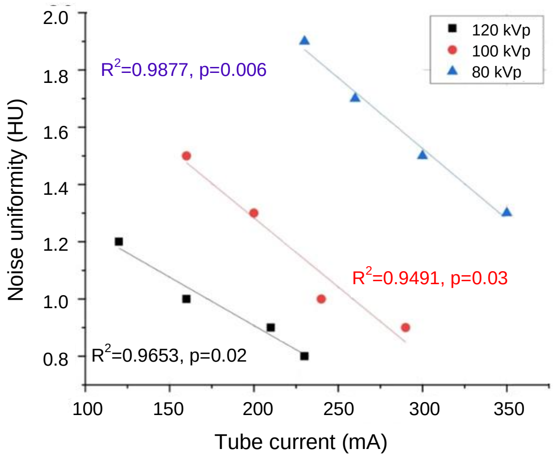


Fig. 4. Correlation between noise uniformity and mA.

The mean measurement values obtained for the old protocol and for the two new protocols proposed at 100 kVp (5.6 ± 0.57, $p = 0.117$) and 80 kVp (5 ± 0.81, $p = 0.23$) passed the quality control test (Table 2), with values ranging from 4 pairs of holes/mm, which was the striking value for 80 kVp 230 mA, to 7 pairs of holes/mm with a hole diameter of 0.5 mm detected by 120 kVp 230 mA and 100 kVp 290 mA.

The values of 120 kVp ($R^2 = 0.9651$, $p = 0.02$), 100 kVp ($R^2 = 0.9476$, $p = 0.03$) and 80 kVp ($R^2 = 0.9838$, $p = 0.008$) were obtained by the simple linear regression test. These results demonstrated that variations in tube voltage and current affected image noise uniformity values by 96.51 %, 94.76 %, and 98.38 %. This test also shows the current in the tube has a significant effect on the uniformity of the noise ($p < 0.05$). The study reported that the kVp and mA significantly affect the uniformity of the noise image ($R^2 = 0.9768$, $p < 0.05$). According to the study [27], the noise uniformity value image is significantly influenced by kVp and mA ($R^2 = 0.9768$, $p < 0.05$). These results show that reducing mA can lower image noise uniformity (Fig. 4). All values are still within the standard and the exposure dose received by the patients is uniformly distributed, a decrease in the image noise uniformity value is due to an increase in the number of photons generated by the tube and received by the sensor to produce a CT image.

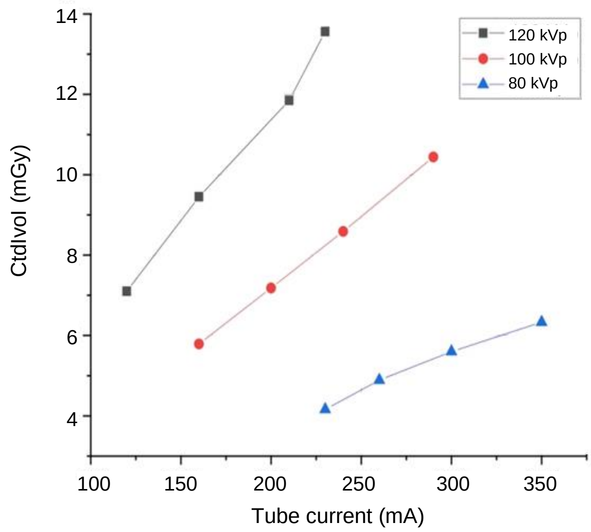


Fig. 5. $CTDI_{vol}$ at 120 kVp,100 kVp and 80 kVp.

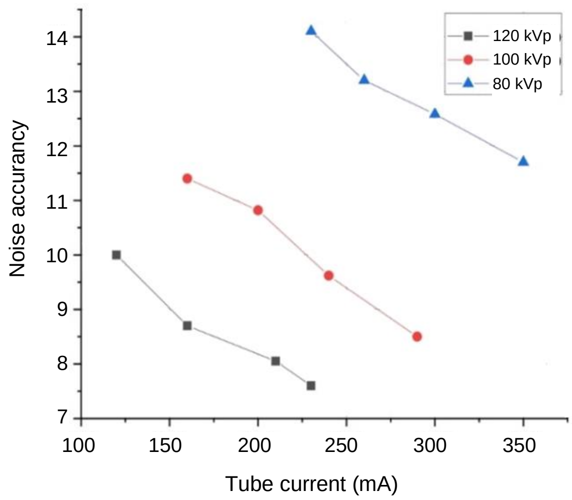
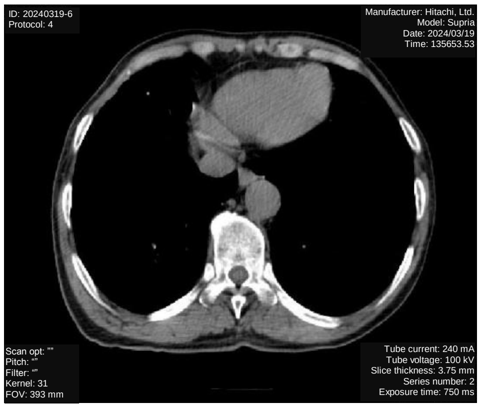
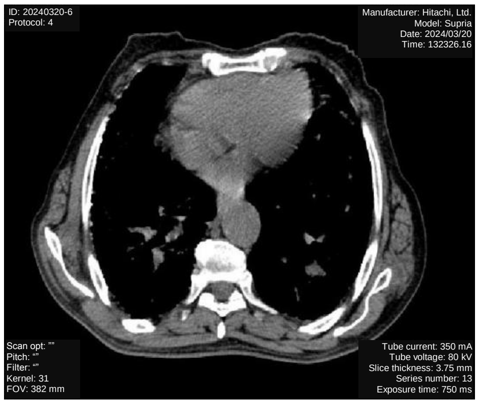


Fig. 6. Noise accuracy at 120 kVp,100 kVp and 80 kVp.

The first suggested protocol of the thoracic CT at 100 kVp compared to the old protocol of 120 kVp made it possible to reduce the CTDIvol from 27.51 % to 18.45 % (Fig. 5). This was accompanied by an increase in noise from 11.84 % to 24.37 % (Fig. 6) The higher the tube current, the lower the noise reducing the kVp by 16.66 % and increasing the mAs from 13.92 % to 33.33 %. This study is consistent with the conclusions of El Mansouri et al., who was able to reduce the kVp to 16 % and increase the mAs from 39 % to 58 %, thereby reducing the dose from 39.70 % to 14.66 % with an increase in noise from 11.82 % to 38 % [19]. Other authors reduced kVp from 14 % to 17 %, resulting in a dose reduction from 38 % to 32 %, while significantly increasing noise from 16 % to 29 % [28]. CNR increases as mA levels increase**.** The higher the mA, the higher the CNR (Fig. 7), up from 0.7 to 1.1.



(a). 100 kVp, 240 mA, (CTDIvol = 8.5 mGy).



(b). 80 kVp, 350 mA, (CTDIvol = 6.3 mGy).

Fig. 7. Effect of tube voltage variation (kVp) and current intensity (mA) on image quality. Two chest CT images of two patients, the first was obtained with a 100 kVp, 240 mA scan and the second was acquired with 80 kVp, 350 mA (Fig. 7a). The other analysis parameters are the same for both patients (Fig. 7b).

The rate of change shows a slight regression from 12.50 % to 26.67 %, unaffected by the change in tube voltage ($p > 0.05$). For the SR, all the values obtained at 100 kVp and from the visual test, the mean measurement values were slightly modified between 4 and 6 holes/mm, there was no statistically significant difference when switching from 120 kVp to 100 kVp ($p > 0.05$). This result is consistent with previous studies by Kun Tang et al., [29] whose results were derived from a CATHPAN 500 and 600 image quality phantom**.**

We postulate that thoracic CT with a 17 % reduction in tube voltage (120 to 100 kVp) is possible without loss of diagnostic accuracy when the tube current is greater than 240 mA (Fig. 7a) to replace the reduction in photon fluence and allow a reduction in radiation dose from 18.45 % to 27.51 %. The study by Heyer et al., reported that a 100 kVp procedure does not lead to a reduction in contrast/noise ratio or visual image quality, although it does lead to a 44 % decrease in exposure dose [30]**.**

Regarding the second proposed scan protocol of the thoracic imaging at 80 kVp compared to the old protocol by lowering the voltage to 33.33 %, the mA progression from 42.41 % to 93.33 % could decrease the 41.41 % to 53.32 % (Fig. 5). In return, the noise was increased from 41 % to 56.27 % (Fig. 6). Results from Francis Zarb et al., reported that the 49 % decrease in the with a lower increase in noise (50 %) was achieved by combining a reduction in mA and kVp [28]. The CNR values were between 0.5 and 1, it had no statistically significant difference when changing the tube voltage ($p < 0.05$). Mean SR values are between 4 and 6 while remaining within BAPETEN standards.

Estimating that the 80 kVp, 350mA protocol is feasible as new protocol while reducing the CTDI in half to 53.32 % with an increase in noise by 53.95 % during a thoracic CT (Fig. 7b). The application of this protocol necessitates a compensatory increase in tube current to decrease image noise. Nevertheless, procedures at 80 kVp have been reported to contribute to a dose reduction of 36 % to 52 % when contrast to noise ratio is not sacrificed [31].

The use of kVp and lower mA also have a considerable advantage for pediatric and small adult patients. The improvement of image production, in particular the introduction of methods of iterative reconstruction, should significantly reduce radiation dose.

CONCLUSION

Over the last few decades, strategies to investigate ways of reducing the exposure to radiation during CT imaging of the thorax have been developed. It is suggested that it is possible to adopt two new acquisition protocols at 100 kVp and 80 kVp while reducing the CTDIvol to 28 % and 54 % and taking into account the effect of the variation of these parameters on image quality. Their choices must be made by integrating clinical issues and a sound knowledge of the pathophysiology and results of imaging of the suspected pathology. Consequently, staff from the medical profession, in particular radiologists and technicians, should always be involved in efforts to improve imaging practices in order to increase the efficacy of the application of radiation.

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The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

AUTHOR CONTRIBUTION

Z. Saga, A. Rezzouk and A. Rahmouni were equally involved in the writing of this document. All authors read and validated the final version of the article.

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