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RESEARCH

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Studies on the radiation dose, image quality and low contrast detectability from MSCT abdomen by using low tube voltage

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Abstract

Background: To study radiation dose, image quality and low-contrast cylinder detectability from multislice CT (MSCT) abdomen by using low tube voltage using the American College of Radiology (ACR) phantom. The ACR phantom (low-contrast module) was scanned with 64 MSCT scanner (Brilliance, Philips Medical System, Eindhoven, Netherlands) with 80 and 120 KVP, utilizing different tube current time product (mAs) range from 50 to 380 mAs. The image noise (SD), signal to noise ratio, contrast-to-noise ratio (CNR), and scores of low contrast detectability were assessed for every image respectively.

Results: From images analyses, the noise essentially increased with the use of low tube voltage. The CNR was 0.94 ± 0.27 at 120 KVP, and CNR was 0.43 ± 0.22 at 80 KVP. However, with the same dose, there were no differences of statistical significance in scores of low-contrast detectability between 120 KVP at 300mAs and 80 KVP at (200–380) mAs ($p > 0.05$). At 300 mAs, the $CTDI_{vol}$ obtained at 80 KVP was about 29% of that at 120 KVP. The $CTDI_{vol}$ obtained at 80 KVP were decreased from 5% at 50 mAs, to 37% at 380 mAs.

Conclusions: There is a possibility to decrease exposure of radiation virtually by reducing KVP from 120 to 80 KVP in examination of abdominal CT when the high tube current is used, though increasing image noise at low tube voltage.

Keywords: Multislice CT, Radiation dose, Image qualityCT ACR phantom

Background

There is a marked increment in using multislice computed tomography (MSCT) from its presenting. Yearly scan of computed tomography (CT) has been increased rapidly from (2–72) million since 1980 to 2007, approximately [1–4]. MSCT has major diagnostic ability and enables expanded medical applications. However, it may has ability to lead up to increasing radiation dose due to using of the thinnest sections routinely, the expanded volume of irradiation, and multiple-phase irradiations that was routinely completed in patients who are anticipated of having hepatocellular tumors. The patient has been exposed in CT imaging to ionizing radiation [5]. Depending on the National Council on Radiation Protection and Measurements (NCRP) [6], the radiation exposure from CT is half of all the radiation exposure from medical processes to the United State (U.S.) population approximately. According to the literature, actually, CT accounts for approximately 7% of all radiological tests in earth but participates more than 40% of the assembled effective dose [7]. The theoretic risk of irradiation in fact cancer due to CT scanning to patients cannot be neglected [8–11]. Assess the risk of dying from cancer for those subjecting the process of CT per pass of CT scan across the abdomen was 12.5/10,000 population [1]. Subsequently, interests relating a reducing of radiation dose lately have been excited through abdomen CT examinations.

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Though reducing tube current has the more means for a reduction in radiation dose of CT [4, 12], such modification as well decreases the contrast to noise ratio CNR, that already impact in the diagnostic result of the test. This is mostly true in abdominal examination [13], wherever low-contrast areas have hardly and badly impacted via the CNR. Several studies [14–16] propose that examination at low tube voltage can be decrease dose without remarkably effecting image quality. In present work aims to study the radiation dose, the image quality and the low-contrast cylinder detectability from MSCT abdomen by using low tube voltage using the American College of Radiology (ACR) phantom.

Methods

Description of phantom

We used Gammex 464 ACR CT Accreditation Phantom ACR phantom model: 464 CT Phantom Serial Numder: 804882-4996, manufacturers: Gammex Inc USA.

The ACR CT accreditation phantom consists of four independent parts which can measure the required image quality parameters [17–20]. In our study, we use Module 2 Fig. 1.

Scanning of CT

The ACR phantom has been scanned two times per protocol (120-80) KVP with a 64-section MSCT scanner (Brilliance, Philips Medical System, Eindhoven, Netherlands). Scanning was performed with the standard tube voltage (120 KVP) and with the low tube voltage (80 KVP), with tube current time product settings arrangement of 50, 100, 150, 200, 250, 300, and 380 mAs, respectively for slice thickness 2.5 mm,pitch 1.172 mm/rotation, rotation time 0.75 s, collimation 64 × 0.625, matrix 512 and standard filter (Fig. 2).

Radiation dose measurement

We utilized the Computed Tomography dose index volume ($CTDI_{vol}$) according to the output's data for radiation dose appreciation. We are record the obtained CT dose index volume ($CTDI_{vol}$) of each irradiation conditions from the picture archiving and communication system (PACS) (Fig. 3).

We make a comparison between the $CTDI_{vol}$ that we are obtained at protocol of standard tube voltage with that protocol we are obtained at low tube voltage.

CNR and SNR measurements

We measured the object of low-contrast in 25 mm diameter CT number (HU) and that of the background, for each scanning protocol (120 KVP-80 KVP). The region of interest (ROI) utilized to execute the measurements be maintain at 100 mm^2.

CNRs were calculated as follows:

$$CNR = (HU_M - HU_B)/SD_B,$$

SNRs were calculated as follows:

$$SNR_M = (HU_M)/SD_M$$

where HU_M is the object of low-contrast (a 25 mm diameter) average CT numbers, HU_B is the background average CT numbers, SD_B is the standard deviation of the attenuation values of the background and SD_M is the

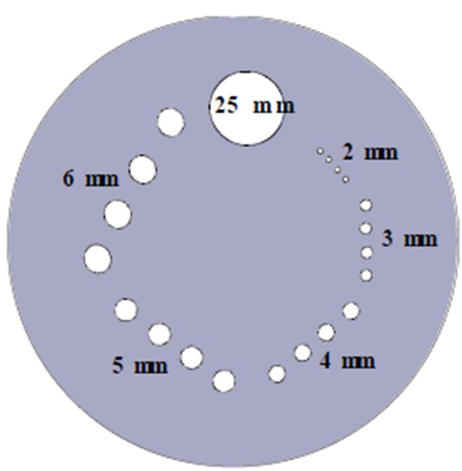

Fig. 1 The low-contrast module of ACR phantom. It contains six groups of cylinders with different diameters (25, 6, 5, 4, 3 and 2 mm). In this study, a 25 mm diameter object only was chosen to be analyzed [17–20]

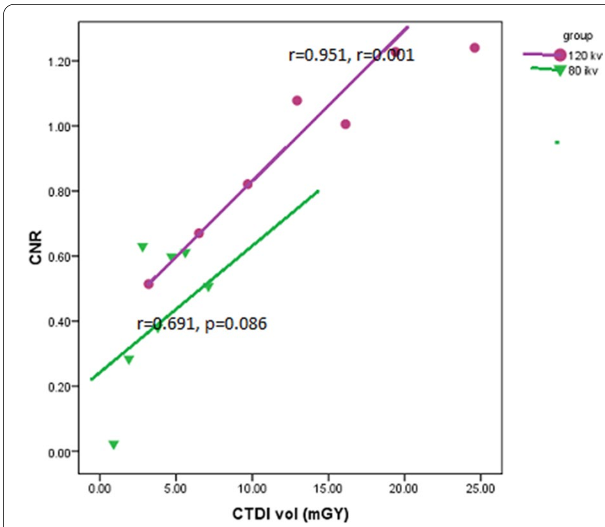

Fig. 2 Scatter diagram for positive correlation between $CTDI_{vol}$ and CNR

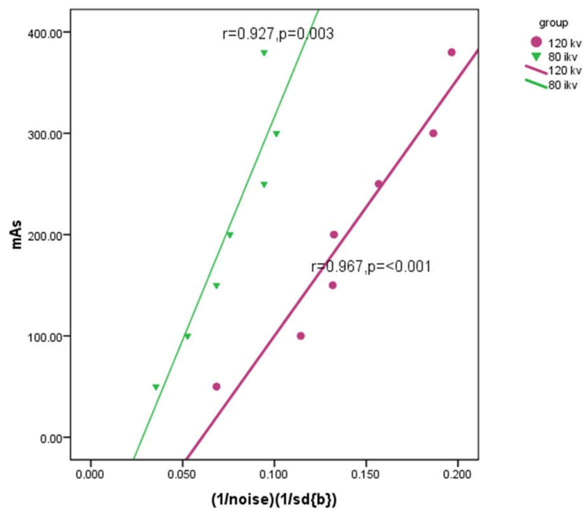

Fig. 3 Scatter diagram for positive correlation between (1/noise) and mAs. At same mAs, image noise acquired with 120 KVP is less than image noise acquired with 80 KVP

standard deviation of the low-contrast object (a 25 mm diameter) attenuation values.

Assessment of LCD

Requested by two proficient viewers who were sightless for each group of survey parameters, check the images independently. Each object visualization was progressive by each observer on scoring scale of a 3-point: if the object was appeared and visible clearly as perfect cylinder, a 3.0 score was acquired, if the object do not visible clearly a 2.0 score was acquired, and a 1.0 score was acquired if the object not appeared (cannot be detected). All images were evaluated respectively by each observer.

Statistical analysis

Data were analyzed with SPSS version 21.

Results

Radiation dose

Results show that $CTDI_{vol}$ range was from 3.2 to 24.6 mGy at 120 KVP, also the mean was 13.21 mGy and stander deviation was ± 7.46 mGy. The result shows that $CTDI_{vol}$ range was from 0.90 to 7.12 mGy when 80 KVP was used, the mean was 3.83 mGy and stander deviation was ± 2.16 mGy. The $CTDI_{vol}$ acquired of each group of irradiation conditions (120–80) KVP is listed in Table 1. At the same mAs settings, the $CTDI_{vol}$ acquired at 80 KVP was about 29% of the $CTDI_{vol}$ acquired when 120 KVP is utilized. By make a comparison with $CTDI_{vol}$ acquired when120 KVP and 300 mAs is used, the relative $CTDI_{vols}$ acquired with 80 KVP are 5% at 50 mAs, 10% at 100 mAs, 14% at 150 mAs, 20% at 200 mAs, 24% at 250 mAs, 29% at 300 mAs, and 37% at 380 mAs.

Table 1 The $CTDI_{vols}$ values obtained at 120 KVP and 80 KVP for each set of acquisition conditions

Tube current–time product (mAs)	CTDIvols (mGY)	
	120 KVP	80 KVP
50	3.2	0.9
100	6.5	1.9
150	9.7	2.8
200	12.94	3.8
250	16.12	4.7
300	19.4	5.6
380	24.6	7.12

Table 2 Comparison of $CTDI_{vol}$ (mGy) between 120 and 80 KVP

	CTDIvol (mGY)		Student t-test	*p* value
	120 KVP	80 KVP		
Mean ± SD	13.21 ± 7.46	3.83 ± 2.16	3.19	0.008*
Min–Max	3.20–24.60	0.90–7.12		

*means $p < 0.05$

Table 3 Comparison of CT number between 120 and 80 KVP

	CT number		Student t-test	*p* value
	120 KVP	80 KVP		
Mean ± SD	90.47 ± 0.62	64.97 ± 1.29	47.04	< 0.001*
Min–Max	89.64–91.45	63.14–66.50		

*means $p < 0.05$

Table 2 shows that, there is significant difference between 120 and 80 KVP regarding $CTDI_{vol}$ (mGY). Mean (SD) $CTDI_{vol}$ is higher at 120 KVP 13.21 (7.46) compared to 3.83 (2.16) at 80 KVP with p value < 0.05.

Image quality

For each scanning technique, the CT numbers, stander deviation of image noise, SNR, also CNR results are registered in the below tables. The result shows that CT number range was from 89.64 HU to 91.45 HU when used 120 KVP, the mean was 90.47 HU and stander deviation was ± 0.62 HU. The result shows that CT number range was from 63.14 HU to 66.50 HU when used 80 KVP, the mean was 64.97 HU and stander deviation was ± 1.29 HU.

Table 3 shows that, there is significant difference between 120 and 80 KVP regarding CT number. Mean

Table 4 Comparison of image noise (SD) between 120 and 80 KVP

	Image noise(SD)		Student t-test	*p* value
	120 KVP	80 KVP		
Mean ± SD	7.90 ± 3.23	15.15 ± 6.56	2.62	0.022*
Min–Max	5.09–14.61	9.89–28.21		

*means $p < 0.05$

Table 5 Comparison of CNR between 120 and 80 KVP

	CNR		Student t-test	*p* value
	120 KVP	80 KVP		
Mean ± SD	0.94 ± 0.27	0.43 ± 0.22	3.73	0.003*
Min–Max	0.51–1.24	0.02–0.63		

*means $p < 0.05$

Table 6 Comparison of SNR between 120 and 80 KVP

	SNR		Student t-test	*p* value
	120 KVP	80 KVP		
Mean ± SD	13.20 ± 4.32	4.88 ± 1.57	6.77	< 0.001*
Min–Max	6.14–21.36	2.01–6.79		

*means $p < 0.05$

(SD) CT number is higher at 120 KVP 90.47 (0.62) compared to 64.97 (1.29) at 80 KVP with *p* value < 0.001.

Image noise (SD) range was from 5.09 HU to 14.61 HU when used 120 KVP, the mean was 7.90 HU and stander deviation was ± 3.23 HU. The result show that image noise (SD) range was from 9.89 HU to 28.21 HU when used 80 KVP, the mean was 15.15 HU and stander deviation was ± 6.56 HU (Table 4).

Table 4 shows that, there is significant difference between 120 and 80 KVP regarding image noise (SD). Mean (SD) image noise is lower at 120 KVP 7.90 (3.23) compared to 15.15 (6.56) at 80 KVP with *p* value < 0.05.

The CNR range was from 0.51 to 1.24 when used 120 KVP, the mean was 0.94 and stander deviation was ± 0.27. The result shows that CNR range was from 0.02 to 0.63 when used 80 KVP, the mean was 0.43 and stander deviation was ± 0.22 (Table 5).

Table 5 shows that, there is significant difference between 120 and 80 KVP regarding CNR. Mean (SD) CNR is higher at 120 KVP 0.94 (0.27) compared to 0.43 (0.22) at 80 KVP with *p* value < 0.05.

SNR range was from 6.14 HU to 21.36 HU when used 120 KVP, the mean was 13.20 HU and stander deviation was ± 4.32 HU. The result shows that SNR range was from 2.01 HU to 6.79 HU when used 80 KVP, the mean was 4.88 HU and stander deviation was ± 1.57 HU (Table 6).

Table 6 shows that, there is significant difference between 120 and 80 KVP regarding SNR. Mean (SD) SNR is higher at 120 KVP 13.20 (4.32) compared to 4.88 (1.57) at 80 KVP with *p* value < 0.001.

Table 7 shows that, there is positive significant correlation between $CTDI_{vol}$ (mGY) and CNR at 120 KVP ($r = 0.951$, $p = 0.001$). On the other hand no significant correlation is observed at 80 KVP *p* value > 0.05.

Table 8 shows that, there is positive significant correlation between (1/noise) and mAs at 120 KVP ($r = 0.967$, $p = < 0.001$) and at 80 KVP ($r = 0.927$, $p = 0.0$).

Table 7 Correlation between $CTDI_{vol}$ (mGY) and CNR

	CTDIvol (120KVP)		CTDIvol (80KVP)	
	Pearson correlation r	*p*	Pearson correlation r	*p*
CNR	0.951	0.001*	0.691	0.086

*means $p < 0.05$

Table 8 Correlation between tube current (mAs) and image noise

	(1/noise) (120 KVP)		(1/noise) (80 KVP)	
	r	*p*	*r*	*p*
mAs	0.967	< 0.001**	0.927	0.003*

Table 9 LCD subjective score

KVP/mAs	LCD subjective score		Mean	*p* value
	observer A	observer B		
80 KVP/50 mAs	3.0 (2.0,1.0)	3.0 (2.0,1.0)	1.50 ± 0.71	0.014*
80 KVP/100 mAs	3.0 (2.0,1.0)	3.0 (2.0,1.0)	1.50 ± 0.71	0.014*
80 KVP/150 mAs	4.0 (3.0,1.0)	4.0 (3.0,1.0)	2.00 ± 1.41	0.182
80 KVP/200 mAs	4.0 (3.0,1.0)	3.0 (2.0,1.0)	2.00 ± 1.41	0.182
80 KVP/250 mAs	5.0 (3.0,2.0)	5.0 (3.0,2.0)	2.50 ± 0.71	0.5
80 KVP/300 mAs	5.0 (3.0,2.0)	6.0 (3.0,3.0)	2.50 ± 0.71	0.5
80 KVP/380 mAs	6.0 (3.0,3.0)	6.0 (3.0,3.0)	3.0 ± 0.0	1
120 KVP/300 mAs	6.0 (3.0,3.0)	6.0 (3.0,3.0)	3.0 ± 0.0	(*r*)

[*] Significant $p < 0.05$, (*r*) reference group, the *p* values are those obtained with 80 KVP Comparative with results acquired with 120 KVP/300 mAs. There is a good agreement among the two observers (*A*, *B*) regarding the personal evaluation of the LCD ($\kappa = 0.667$)

LCD results

The results of subjective scores of LCD assigned by two proficient observers are listed in Table 9.

At 120 KVP and 300 mAs, the average score of the images was 2.50 ± 0.71. At 80 KVP, the average score was 1.50 ± 0.71 at (50–100) mAs which was significantly much less than the average score at 120 KVP and 300 mAs with p value = 0.014 at (50–100 mAs) (Table 9). The score was 2.00 ± 1.41 at (150–200 mAs), and the score was 2.50 ± 0.71 at (250–300 mAs) there were no differences of statistical significance between average score at 120 KVP and 300 mAs and average score acquired at 80 KVP and other tube current settings (mAs) that were investigated (p value = 0.182 at 150–200 mAs; p value = 0.5at 250–300 mAs and p value = 1.0 at 380 mAs).

Discussion

Now, advancement in MSCT technique permits CT examinations to be fast and easily implementation, and this leads to a potential increasing of radiation dose for patients. particularly, risk of cancer death and the radiation exposure from hepatic CT examinations have increased significantly due to dynamic-enhanced of multiple-phase computed tomography examination was performed constantly. Management patient dose is subsequently a main interest in MSCT abdominal examinations.

We find in this study that, the relation between CNR and CT dose index volume ($CTDI_{vol}$) was a directly relation, which was compatible with Waaijer et al. [21, 22], they found that SNR^2 was proportional to effective tube current and CT dose index volume. Though the average CNR was reduced whenever CT examination was executed at 80 KVP tube voltages with the same tube current adjusting, CNR ameliorated basically when selfsame $CTDI_{vol}$ was utilized.

Also we find in this study that CNR range was from 0.02 to 0.63 when used 80 KVP. Furthermore, the relative radiation dose obtained at 80 KVP and 380mAs was 37% of the relative radiation dose obtained at 120 KVP and 300 mAs. Thus, we assume that low KVP scanning as low as 80 KVP is available for a pediatric Abdominal computed tomography scanning without diagnostic accuracy loss when mAs is maximum than 300 mAs allowing radiation dose reduction by 29% to 37%. This results have agreement with American ACR abdominal CT (CNR for a pediatric was more than 0.5) [16].

One of the greatest substantial factors for abdominal computed tomography is LCD, specially when we are searching for little lesions of organs of Abdomen like, pancreas, liver, kidneys or spleen.

LCD also is related for contrast improved series, not only related for not improved series, because contrast between abnormal and normal tissue could only increasing lightly by the iodine [23].

The low contrast detectability subjective scores obtained with 80 KVP and 150, 200, 250, 300 and 380 mAs were not varying significantly from scores obtained with 120 KVP. Moreover, the low contrast detectability average score obtained with 80 KVP and 380 mAs was similar to the low contrast detectability average score obtained with 120 KVP and 300 mAs. From our results, reducing KVP from 120 to 80 KVP also may result in up to 37% dose reduction without degeneration low contrast resolution. Funama et al. [21] reported that a 35% reducing of radiation dose might be obtained with examination was assessed at 90 KVP instead of at 120 KVP without degeneration of low contrast detectability. Our results have agreement with Funama also suppose that the capability of using lower tube voltage in abdominal computed tomography thereby achieving decreasing of radiation dose while image quality is maintaining acceptable.

Verdun et al. [24] shown that there is a strong correlation significantly between the average measurements of CNR and the low contrast detectability subjective scores (r = 0.95, p < 0.05).

Increasing in image noise and decreasing in SNR which resulted from the decreasing of photon flux were considered the basic obstacle of low tube voltage technology.

In our study, we found that there is significant difference between 120 and 80 KVP regarding image noise (SD). Also, there is positive significant correlation between (1/noise) and mAs at 120 KVP (r = 0.967, p = < 0.001) and at 80 KVP (r = 0.927, p = 0.003), such as Waaijer et al. [13, 22] we showed that the tube current has correlative relationship inversely with image noise. In other expression, when the lower tube voltage strategy or lower tube current is performed, the increased noise will be obtained.

However, image noise, has a good effect on abdominal images quality as the region of abdomen is lower contrast inherently.

So, low tube voltage computed tomography scanning required settings of higher tube current to recompense for the photon slower number.

As well, new technologies must be evolved to obtained image noise reduction. Over last years, Sundry articles showed that filter back projection (noise decreasing filters) (FBP) [25–28], also reconstruction methods, like adaptation statistical iterative reconstruction technique (ASIRT) [29–31] can virtually help to decrease the Computed Tomography images noise with reducing radiation dose without degradation of image quality.

Study limitations

First, this results must be further confirmed for clinical using as that CT examination at low KVP (80 KVP) was only implemented in ACR phantom, but this solid phantom was not consider body composition variability, nevertheless, Marin et al. [32] reported that a technology of low tube voltage with 80 KVP might be performed to ameliorate the conspicuity of liver tumors of malignant hyper vascular while significantly reduction of radiation dose of patient.

Second, the incident X-ray beam attenuation in computed tomography depends on the body portion size that wants to be evaluated but our studies do not take into consideration body sizes differences; that is, most great exposure is wanted for fat patients to achieve the same image quality to that for thinner humans [33]. However, preceding studies with a phantom propose that the technique of low KVP is efficient for dose reducing of abdominal computed tomography for comparatively patients having body weight below 80 kg (light weight patient) [34].

Another part is that more humans existing with implants of high-attenuation, that can significantly reduce image quality by applying low KVP protocols continuously.

Third, the contentious usage of maximum high tube current shortens the life of the tube so it has a bad effect on the machine. In our present work we use high mAs until 380 mAs but we did not get to the maximum tube current.

Lastly, we only utilize the computed tomography dose index volume ($CTDI_{vol}$) acquired from the picture archiving and communication system (PACS) via manufacturer to assess the radiation dose.

As a final of limitations we recommended to do this study with a protocol 80 KVP on patient for more assessment.

Conclusions

Results of our study detected that there is a possibility to decrease exposure of radiation virtually by reducing KVP from 120 to 80 KVP in examination of abdominal CT when the high tube current is used, though increasing image noise at low tube voltage. As an effective technique to reduce the dose of CT scan at low KVP safety and help patients of light weight relatively, specially patients who may want to subject MSCT tests for high-risk examination or long-term follow-up.

Abbreviation

MSCT: Multislice computed tomography; CT: Computed tomography; NCRP: National Council on Radiation Protection and Measurements; U.S.: United State; CNR: Contrast to noise ratio; SNR: Signal to noise ratio; ACR: American College of Radiology; $CTDI_{vol}$: Computed tomography dose index volume; PACS: Picture archiving and communication system; ROI: Region of interest; LCD: Low contrast detectability.

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Authors' contributions

Prof. MTA and Dr. AM gave idea and ZAA collected the data and analyze them. Dr. AM put study design and ZAA, Dr. AS and Prof HMG wrote the paper with revision. All author read and approved the final manuscript.

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Availability of data and material

The datasets used and/or analyzed during the current study are available from the corresponding author on reasonable request.

Declarations

Ethics approval and consent to participate

Not applicable.

Consent for publication

Not applicable.

Competing interests

The authors declare that they have no competing interests.

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References

1. Tsapaki V, Rehani M, Saini S (2010) Radiation safety in abdominal computed tomography. Semin Ultrasound CT MRI 31:29–38
2. Alsafi KG (2016) Radiation protection in X-ray computed tomography: literature review. Int J Radiol Imaging Technol. https://doi.org/10.23937/2572-3235.1510016
3. Mansour Z, Mokhtar A, Sarhan A, Ahmed MT, El-Diasty T (2016) Quality control of CT image using American College of Radiology (ACR) phantom. Egypt J Radiol Nucl Med 47:1665–1671
4. Baskan O, Erol C, Ozbek H, Paksoy Y (2015) Effect of radiation dose reduction on image quality in adult head CT with noise-suppressing reconstruction system with a 256 slice MDCT. J Appl Clin Medi Phys 16:285–296
5. Stoffey RD (2011) MDCT physics. The basic technology, image quality and radiation dose. Am J Roentgenol 196:484–484
6. National Council on Radiation Protections and Measurements (NCRP): ionizing radiation exposure of the population of the United States, NCRP Report No. 160. Bethesda, Md,2009
7. Mettler FA Jr, Bhargavan M, Faulkner K et al (2009) Radiologic and nuclear medicine studies in the United States and worldwide: frequency, radiation dose, and comparison with other radiation sources—1950–2007. Radiology 253:520–531
8. Pearce MS, Salotti JA, Little MP et al (2012) Radiation exposure from CT scans in childhood and subsequent risk of leukaemia and brain tumours: a retrospective cohort study. Lancet 380:499–505
9. Strauss KJ, Goske MJ (2011) Estimated pediatric radiation dose during CT. Pediatr Radiol 41(suppl2):S472-482
10. Ulsh AB (2015) Are risks from medical imaging still too small to be observed or nonexistent? Dose Response 13:14–030
11. DeMaio DN (2017) CT radiation dose and risk: fact vs fiction. Radiol Technol 89:199–205

12. Anam C, Haryanto F, Widita R, Arif I, Dougherty G, McLean D (2018) Volume computed tomography dose index (CTDIvol) and size-specific dose estimate (SSDE) for tube current modulation (TCM) in CT scanning. Int J Radiat Res 16:289–297
13. Task Group on Control of Radiation Dose in Computed Tomography. Managing patient dose in computed tomography. A report of the International Commission on Radiological Protection. Ann ICRP. 2000;30:7–45.
14. Chian TC, Nassir NM, Ibrahim MI, Yusof AKM, Sabarudin A (2017) Quantitative assessment on coronary computed tomography angiography (CCTA) image quality: comparisons between genders and different tube voltage settings. Quant Imaging Med Surg 7(2017):48–58
15. Hausleiter J, Martinoff S, Hadamitzky M et al (2010) Image quality and radiation exposure with a low tube voltage protocol for coronary CT angiography: results of the PROTECTION II Trial. Cardiovasc Imaging 3:1113–1123
16. Bischoff B, Hein F, Meyer T et al (2009) Impact of a reduced tube voltage on CT angiography and radiation dose. Results of the PROTECTION I Study. Cardiovasc Imaging 2:940–946
17. American College of Radiology (2013) CT accreditation phantom instructions. http://www.acr.org/~/media/ACR/Documents/Accreditation/CT/PhantomTestingInstruction.pdf
18. ACR CT Accreditation phantom, Gammex 464: Testing the ACR phantom,instruction manual. 2019. https://www.gammex.com
19. Mccllough CH, Bruesewitz MR, Mcnitt-gray MF, Bush K, Ruckdeschel T, Payne JT, Brink JA, Zeman RK (2004) The phantom portion of the American College of Radiology (ACR) Computed Tomography (CT) accreditation program: Practical tips, artifacts examples, and pitfalls to avoid. Med Phys 31:2423–2442
20. ACR 2017 American College of Radiology CT Accreditation Program Testing Instructions Revised 1–06 (2017)
21. Funama Y, Awai K, Nakayama Y et al (2005) Radiation dose reduction without degradation of low-contrast detectability at abdominal multisection CT with a low-tube voltage technique: phantom study. Radiology 237:905–910
22. Waaijer A, Prokop M, Velthuis BK, Bakker CJG, De Kort GAP, Van Leeuwen MS (2007) Circle of Willis at CT angiography: dose reduction and image quality—reducing tube voltage and increasing tube current settings. Radiology 242:832–839
23. Viry A, Aberle C, Racine D, Knebel J-F, Schindera ST, Schmidt S, Beccef F, Verdun FR (2018) Effects of various generations of iterative CT reconstruction algorithms on low-contrast detectability as a function of the effective abdominal diameter: a quantitative task-based phantom study. Eur J Med Phys 48(2018):111–118
24. Verdun FR, Denys A, Valley JF, Schnyder P, Meuli RA (2002) Detection of low-contrast objects: experimental comparison of single- and multi-detector row CT with a phantom1. Radiology 223:426–431
25. Tomographic reconstruction: a guided tour, Irène Buvat (2016)
26. Zeng GL (2014) Comparison of a noise-weighted filtered backprojection algorithm with the standard MLEM algorithm for poisson noise. J Nucl Med Technol 41:1–6
27. Kalra MK, Maher MM, Sahani DV et al (2003) Low-dose CT of the abdomen: evaluation of image improvement with use of noise reduction filters—pilot study. Radiology 228:251–256
28. Kalra MK, Maher MM, Blake MA et al (2004) Detection and characterization of lesions on low-radiation-dose abdominal CT images post-processed with noise reduction filters. Radiology 232:791–797
29. Prezzi D, Goh V, Virdi S, Mallett S, Grierson C, Breen DJ (2017) Adaptive statistical iterative reconstruction improves image quality without affecting perfusion CT quantitation in primary colorectal cancer. Eur J Radiol Open 4:69–74
30. Bellesi L, Gaudino D, Wyttenbach R, Colleoni P (2017) A simple method for low-contrast detectability, image quality and dose optimisation with CT iterative reconstruction algorithms and model observers. Eur Radiol Exp 1:18
31. Jensen K, Andersen HK, Smedby Ö, Østerås BH, Aarsnes A, Tingberg A, Fosse E, Martinsen AC (2018) Quantitative measurements versus receiver operating characteristics and visual grading regression in CT images reconstructed with iterative reconstruction. Acad Radiol 25:509–518
32. Marin D, Nelson RC, Samei E et al (2009) Hypervascular liver tumors: low tube voltage, high tube current multidetector CT during late hepatic arterial phase for detection—initial clinical experience. Radiology 251:771–779
33. Haaga JR (2001) Radiation dose management: weighing risk versus benefit. Am J Roentgenol 177:289–291
34. Nakayama Y, Awai K, Funama Y et al (2005) Abdominal CT with low tube voltage: preliminary observations about radiation dose, contrast enhancement, image quality, and noise. Radiology 237:945–951

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