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ORIGINAL ARTICLE

IMAGE QUALITY AND RADIATION DOSE WITH LOW TUBE VOLTAGE IN CORONARY CT ANGIOGRAPHY: AN EXPERIMENTAL STUDY WITH NORMAL TYPE AND SOFT PLAQUE PHANTOM

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Abstract We evaluated the effect of tube voltage and tube current on image quality and radiation dose in coronary computed tomography angiography (CCTA) with phantom. The phantom was constituted with ball and tubes, both filled with diluted contrast medium (ball: 120HU, tube: 300, 350 and 400HU). The phantom was scanned at 120kV, 100kV and 80kV with beating at 70bpm. The tube currents were changed from 300 to 400mA with every 25mA. Contrast-to-noise ratio (CNR) was calculated. We also performed assessment of image quality, especially noise and sharpness. The radiation dose was also evaluated. The CT numbers at 100kV and 80kV were 1.18~1.37 and 1.49~1.71-times higher than those at 120kV with normal type phantom. There was no significant difference in CNR between 120kV and 100kV. At 80kV both CNR and image score were lower than others. The contrast of tube and soft plaque were also evaluated, and those at 100kV and 120kV were almost same. Compared with 120 kV, radiation dose decreased to about 60% and 30% at 100kV and 80kV respectively. In conclusion, with 100kV setting the radiation dose could be reduced without degradation of image quality, and it is suggested that low tube voltage technique is also useful in CCTA.

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Key words: coronary CT angiography; low tube voltage technique; radiation dose reduction; tube voltage; tube current.

原　著

冠状動脈 CT における低電圧撮影の画質と被曝線量 ～正常モデルおよびソフトプラークモデルにおける検討～

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抄録　冠状動脈 CT における低電圧撮影法の画質や被曝線量を模擬動態ファントム実験で検討した．球体，チューブで心臓と冠状動脈を模し，各内腔を希釈造影剤で満たした．管電圧：120（通常），100, 80kV，管電流：300-400mA で撮影し，CT 値，ノイズを計測してコントラスト比（CNR）を算出した．視覚評価および被曝線量の評価も行った．管電圧低下に伴いノイズが増加したが，チューブ内腔の CT 値も上昇した．CNR は120 kV と100 kV で同等であったが80 kV では低下した．視覚評価でも同様の傾向だった．ソフトプラークの描出に関する検討も行い，120 kV と100 kV で画質は同等であった．被曝線量は120 kV と比較して100 kV では約60%，80 kV では約30%に減少した．管電圧100 kV の撮影では画質の低下なしに被曝低減が可能で，冠状動脈 CT における低電圧撮影の有用性が示唆された．

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キーワード：冠状動脈 CT；低電圧撮影；被曝低減；管電圧；管電流．

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1 Introduction

The development of multi-detector row computed tomography (MDCT) contributed to a wide variety of clinical applications. In the early stage computed tomography (CT) examination had difficulty in evaluating coronary arteries, but due to evolution of CT equipment it is currently widely used for coronary CT angiography (CCTA). Because CCTA is less invasive examination than coronary angiography, it has become widely used. The diagnostic images have been obtained with CCTA[1], but due to widespread availability of MDCT, the increase of radiation dose is becoming a big issue[2]. In Japan, large numbers of diagnostic radiological equipments are used, especially CT scanner[3]. Recently the interest in medical radiation exposure is growing also among the general public, and therefore reduction of radiation dose is a very important issue. Because of the small and complicated structure of coronary artery and heart beats, the radiation dose of CCTA is higher compared with that of other CT examinations[4]. Equipment which irradiates only during the stationary phase of the cardiac cycle, by adjusting tube current depending on cardiac phase, has recently been used to reduce radiation dose at many facilities[5, 6], but there is room for a study of methods which enable further dose reduction.

While fixed tube voltage (120 kV) has been generally used in routine examination, the CT examination technique with low tube voltage setting is also being studied to achieve dose reduction and contrast improvement. The low-tube voltage technique has been reported for CT examinations of abdomen, pulmonary artery and vein of lower extremities[7-11], but there are few reports on the relationship between scan parameters (i.e. tube voltage and tube current) and image quality with low tube voltage technique in CCTA[12].

This study investigated the effect of tube voltage and tube current on image quality and radiation dose, and estimated the usefulness of low tube voltage technique in CCTA.

2 Materials and Methods

2.1. MDCT scanner and workstation

A 16-MDCT scanner (Discovery ST, GE Healthcare) and electrocardiogram simulator (ECG SIMULATOR, MARQUETTE ELECTRONICS, INC.) were used. Image processing was performed with a dedicated workstation (Advantage Workstation ver4.2, GE Healthcare).

2.2. Phantom

The schematic view of the phantom is shown in Figure 1. The phantom was constituted with a ball (diameter: about 120mm) and tubes (inner diameter: 2mm, 3mm and 4mm), which were made by vinyl chloride and polypropylene respectively. The ball and tubes were filled with diluted contrast medium (ball: 120HU, tubes: 300, 350 and 400HU; the values of CT number are at 120kV). Iodinated contrast medium (Urografin® 76%; 370mg of iodine per milliliter, Bayer) diluted with distilled water was used for the phantom. We estimated two vessel types: normal type and soft plaque phantom. The phantom was compressed along the body axis direction to simulate heart beat. The phantom was set in the center of the gantry.

2.3. Scan protocols

The phantom was scanned at 120kV, 100kV and 80kV. The tube currents were 300, 325, 350, 375 and 400mA. The other scanning parameters were rotation time of 0.5sec, beam pitch of 0.3, slice thickness and interval of 0.625mm, field of view 22cm, and pixel matrix size of 512 × 512. The scan was performed with beating at 70 beats per minute (bpm). The phantom was

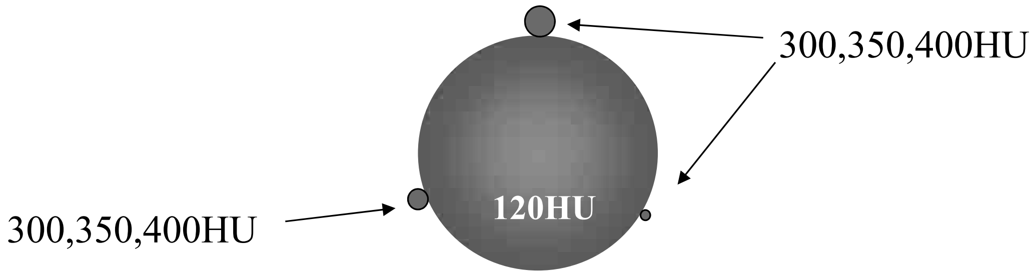


Figure 1 Schematic view of phantom.
The tubes made of polypropylene (inner diameter: 2mm, 3mm and 4mm) were loaded on the ball made of vinyl chloride (diameter: about 120mm). The ball and tubes were filled with diluted contrast medium. The CT number of the ball was 120HU while the tubes were changed from 300 to 400HU every 50HU

also scanned without beating as a rest model (at 350mA). Image reconstruction was performed with SnapShot Burst. The data acquisition was performed three times at each scanning parameter.

2.4. Image reconstruction and assessment for image quality

2.4.1. Image reconstruction

We defined R-R interval of electrocardiograms as 100%, and reconstructed axial image of 0~95% (increment of 5%). Two reviewers evaluated the reconstructed axial images, and determined the most motion-less phase by consensus. These most motion-less images were used in image analysis. Volume rendering (VR) and curved planar reconstruction (CPR) images were also reconstructed for image quality evaluation.

2.4.2. Assessment of image quality

To obtain objective indices of image quality, we measured the CT number at lumen of tube (CT number (T)), the CT number at inside of the ball (CT number (B)). The CT number was defined as mean value. Image noise was derived from the standard deviation of the CT number at inside of the ball (noise (B)). Then contrast-to-noise ratio (CNR) was calculated. CNR at normal type phantom was defined as follows:

CNR= (CT number (T) - CT number (B)) / noise (B).

In addition, the CT number at plaque (CT number (P)) was measured with plaque phantom. CNR at plaque phantom was defined as follows:

CNR= (CT number (T) - CT number (P)) / noise (B).

Measurement of the CT number and noise was performed at the center slice of the phantom. The CT number of the tube lumen was determined only in a tube of 4mm inner diameter. Measurement of the CT number and noise at inside of ball was carried out at 4 locations; one region of interest (ROI) and 3 ROIs were set at center and peripheral of ball respectively.

For qualitative analysis, two reviewers (radiological technologist) with experiences in coronary CT independently evaluated the image quality of axial, VR and CPR images. For assessment of normal type phantom, noise at inside of ball and sharpness of tube surface were estimated. The noise at inside of ball was assessed using the following three point scale: 3= good; 2= moderate; 1= poor. The sharpness of tube surface was also evaluated

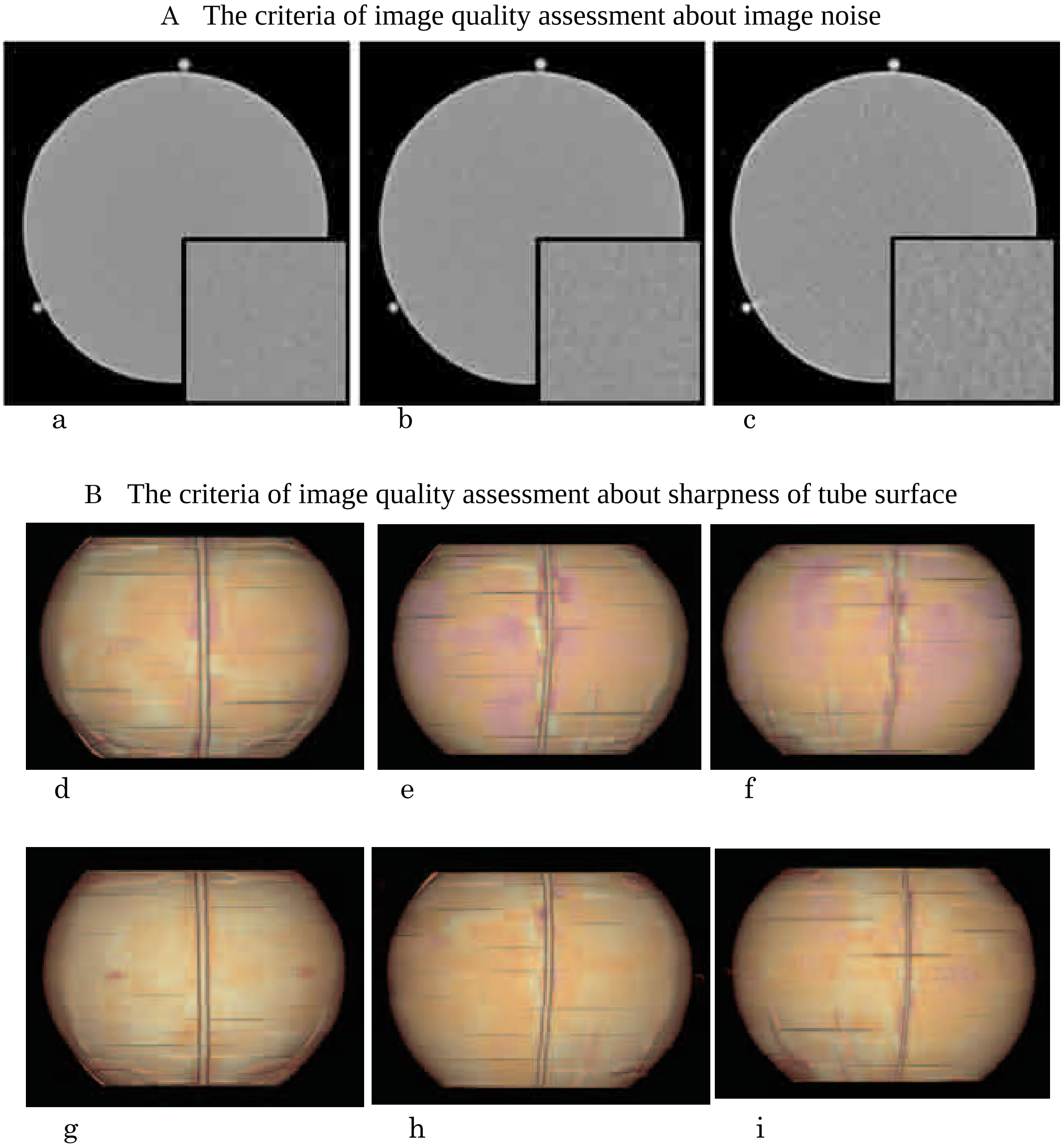
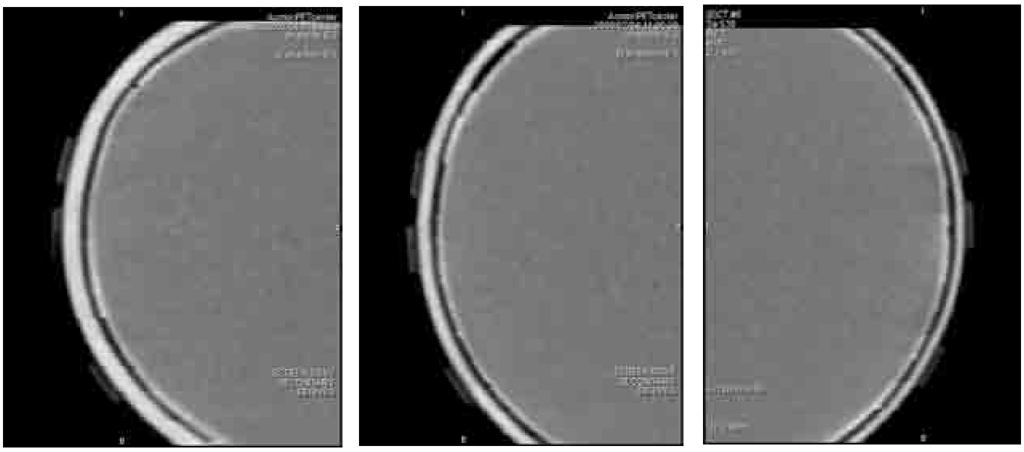


Figure 2 Examples of image quality scoring about normal type phantom.
(A) The criteria of image quality assessment about image noise
(a) good: score 3 (b) moderate: score 2 (c) poor: score 1
(B) The criteria of image quality assessment about sharpness of tube surface
(d) ~ (f): good; score 3 (g) ~ (i): moderate; score 2
d, g: ϕ 4mm, e, h: ϕ 3mm, f, i: ϕ 2mm.

using the following three point scale: 3= good; 2= moderate; 1= poor. Examples of image quality scoring are shown in Figure 2. In this study, images considered as score 1 were not obtained at the assessment about sharpness of tube surface, so that the images of score 1 are not shown in the scoring list. The assessment of image quality at normal type phantom was carried out with fixed window level (WL) and window width (WW) at each tube voltage.

For assessment of plaque phantom, the contrast between the lumen of tube and plaque was evaluated with CPR images by consensus. The images obtained at 120kV were used as a control, and the contrast of 100kV and 80kV were compared with 120kV. The contrast



a b c

Figure 3 Control images obtained at 120kV, which was used for evaluation of the contrast between the lumen of tube and plaque.
The contrast was evaluated using the following three point scale: 1: slightly higher, 2: equivalence, 3: slightly lower compared with 120kV. a: inner diameter 4mm, b: inner diameter 3 mm, c: inner diameter 2mm.

compared with 120kV at each tube voltage (i.e. 100kV or 120kV) was evaluated using the following three point scale: 1: slightly higher, 2: similar, 3: slightly lower. The control images obtained at 120kV are shown in Figure 3. The WL and WW were adjusted for each tube voltage and each contrast media concentration so that the degree of contrast enhancement displayed on the monitor was equivalent.

2.5. Dose estimation

Radiation dose was estimated by volume CT dose index (CTDIvol) which was obtained from the CT scanner console.

2.6. Statistical analysis

Statistical analyses were performed with commercially available software (Statcel 2). CT number, image noise and CNR were analyzed with *t* test, Bonferroni-Dunn's test and regression analysis. For a qualitative analysis, Mann-Whitney U-test was used to analyze image quality scores. Interobserver variability was analyzed by using the κ test.

For all studies, a difference with a *P* value of less than 0.05 was considered as significant.

3 Results

3.1 Image quality

3.1.1 Quantitative analysis

3.1.1.1 CT number

Figure 4a shows the relationship between tube voltage and CT number at each contrast media concentration with normal type phantom. The CT number of tube lumen increased with lowering of tube voltage. The CT numbers at 100kV and 80kV were 1.18~1.37 and 1.49~1.71-times higher than those at 120kV and there was statistically significant difference. Polypropylene bottles (diameter: about 75mm) filled with contrast materials were scanned as standard phantom. There were large fluctuations of the CT number of tube lumen compared with standard phantom (Figure 4b).

Figure 5 shows the relationship between tube voltage and CT number of plaque at each contrast media concentration at plaque phantom. The CT number of plaque at 80kV was significantly higher than that at 120k by

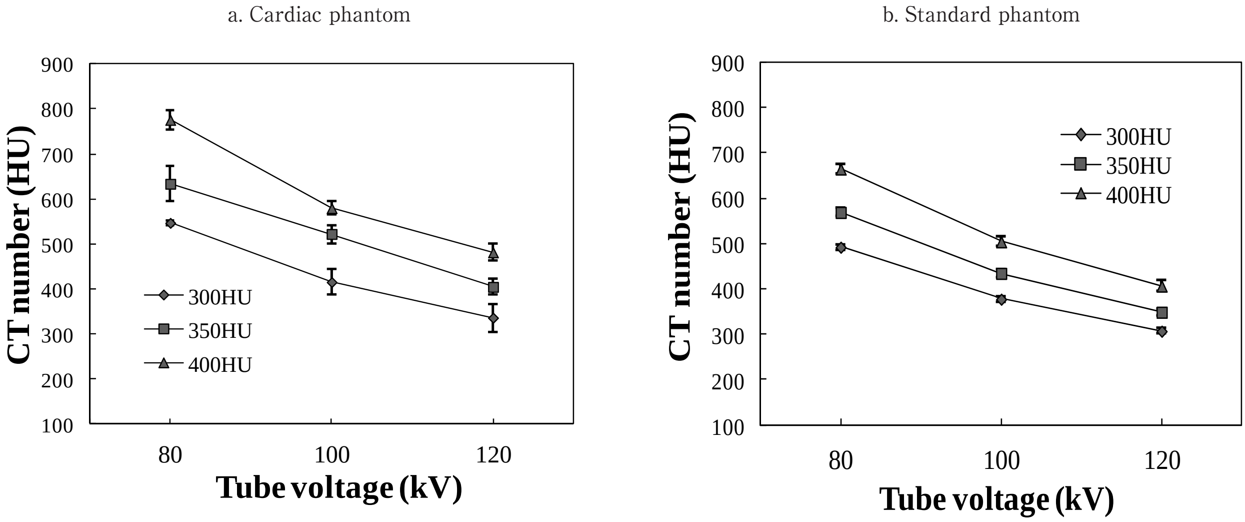


Figure 4 Graph shows the relationship between tube voltage and CT number at each contrast media concentration with normal type phantom. The CT number of tube lumen increased with lowering of tube voltage. The CT numbers at100 kV and 80kV were 1.18~1.37 and 1.41~1.81-times higher than those at 120kV and there was statistically significant difference (a). Polypropylene bottles filled with contrast materials were scanned as standard phantom (b). There were large fluctuations of the CT number of tube lumen compared with standard phantom.

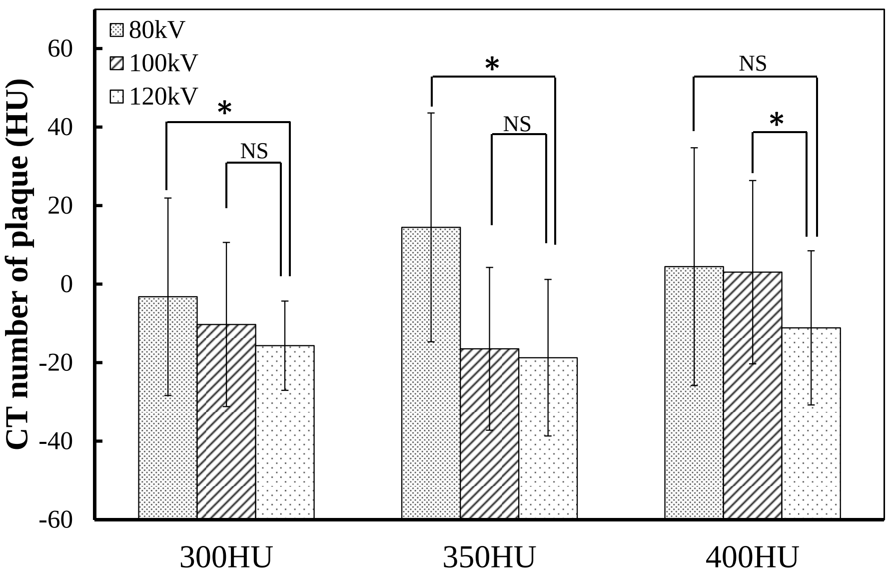


Figure 5 Graph shows the relationship between tube voltage and CT numbers of plaque at each contrast media concentration at plaque phantom. The CT number of plaque at 80kV was significantly higher than that at 120k by approximately 13HU at 300HU and by 33HU at 350HU. At 400HU, the CT number of plaque at 100kV was significantly higher by approximately 14HU than 120kV. The error bars represent the standard deviation. *P<0.05.

approximately 13HU at 300HU and by 33HU at 350HU. At 400HU, the CT number of plaque at 100kV was significantly higher than at 120kV. The mean values of CT number of plaque at low tube voltage were higher than those at 120kV in the other scans, but there was no significant difference. Because measurement might be inaccurate due to the small size of

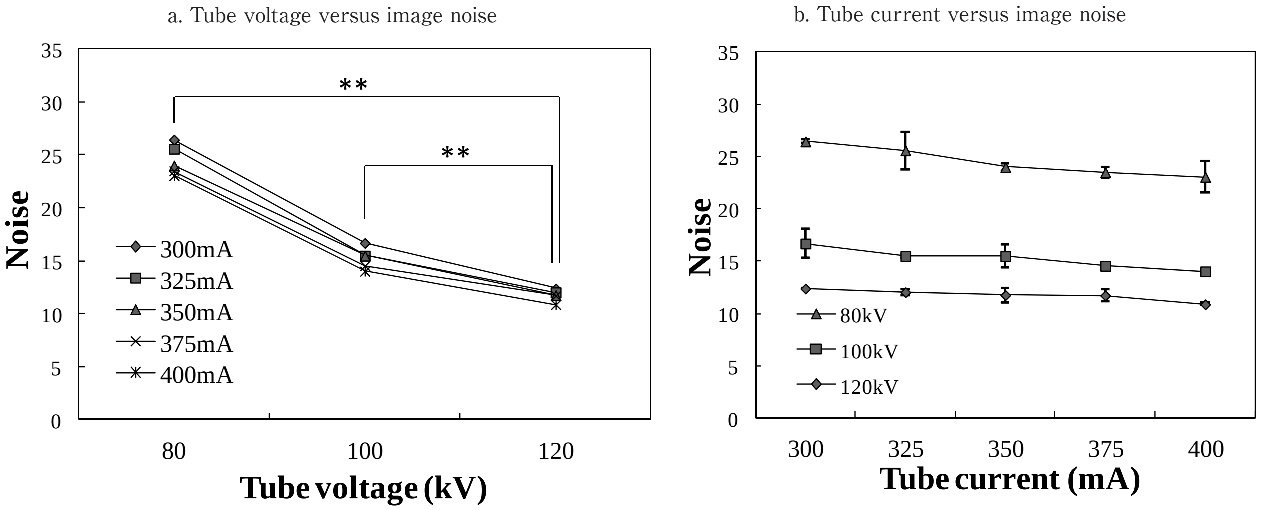


Figure 6 Graph shows the relationship between scan parameter (i.e. tube voltage or tube current) and image noise. The image noise of cardiac phantom had significantly increased by 26~29% and 102~112% at 100kV and 80kV respectively at each tube current (a). There was a negative correlation between tube current and image noise (R=-0.97~-0.75) (b). The error bars represent the standard deviation. **P<0.01.

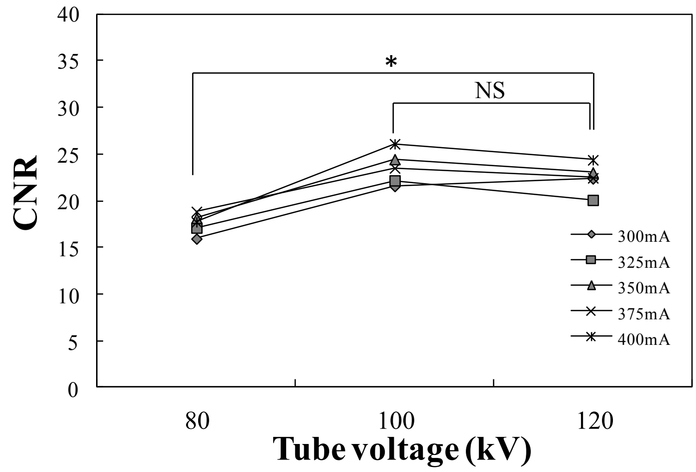


Figure 7 Graph shows the relationship between tube voltage and CNR at 350HU. CNR at 100kV was similar to one at 120kV, while at 80kV CNR was lower than 120kV in about 87% of all scans. *P<0.05.

plaque phantom, the substantial volume phantom which was made of the same materials as plaque phantom (put on the bottle filled with water) was scanned as preliminary study. Then the CT number increased with decrease of tube voltage. The CT number at 100kV and 80kV were higher by approximately 6HU and 20HU respectively than 120kV. The cardiac phantom was scanned also without beating (rest model), and then the CT number of plaque increased with decrease of tube voltage at all contrast media concentrations (i.e. 300, 350 and 400HU). The CT number at 100kV and 80kV was about 3~14HU and 23~31HU higher than those at 120kV respectively.

The CT numbers of plaque at each tube voltage were also evaluated by multiple comparison. Contrast medium concentration

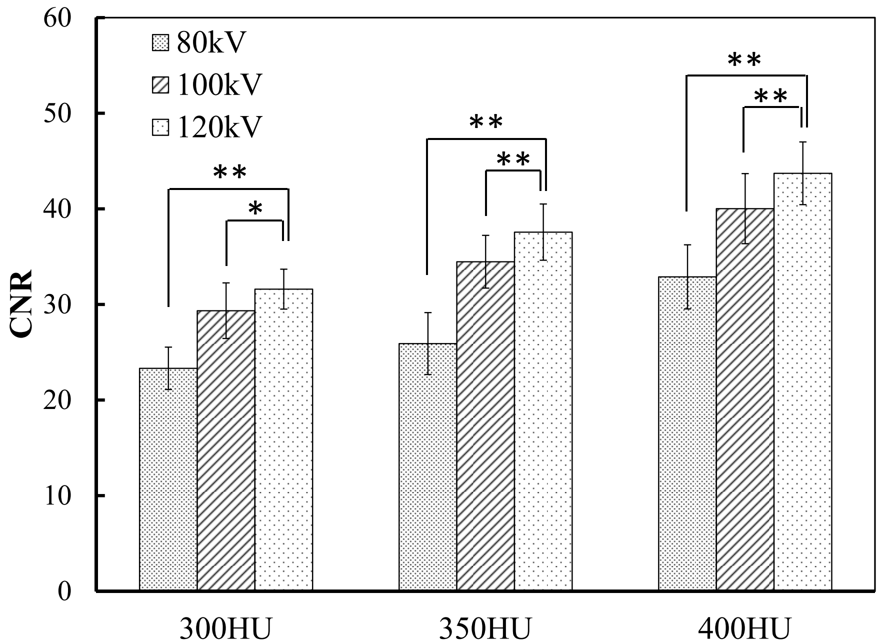


Figure 8 Graph shows the relationship between tube voltage and CNR at each contrast media concentration at plaque phantom. CNR was about 90% and 70% lower at 100kV and 80kV respectively than120 kV in every contrast media concentration. The error bars represent the standard deviation. * P<0.05, ** P<0.01.

did not have an effect on CT number of plaque at any tube voltages. There was no correlation between contrast media concentration and CT number of plaque at both cardiac and rest model.

3.1.1.2 Noise at inside of the ball

The image noise of cardiac phantom had significantly increased by 26~29% and 102~112% at 100kV and 80kV respectively at each tube current (P<0.01) (Figure 6a). There was a negative correlation between tube current and image noise (R=-0.80~-0.75) (Figure 6b). With rest model, increased rates of noise with lowering of tube voltage were similar.

3.1.1.3 CNR

With normal type phantom, CNR at 100kV was similar to one at 120kV, while at 80kV the CNR was lower than 120kV in about 87% of all scans (P<0.05). The relationship between tube voltage and CNR at 350HU is shown in Figure 7 as an example.

Figure 8 shows the relationship between tube voltage and CNR at each contrast media concentration at plaque phantom. CNR decreased to about 90% and 70% at 100kV and 80kV respectively compared with 120kV in every contrast media concentration. At rest model (plaque phantom), CNR at low tube voltage was similarly lower than 120kV.

3.1.2 Qualitative analysis

3.1.2.1 Normal type phantom

At the evaluation about noise inside of the ball, two reviewer's scores were all the same. The noise score of 120kV, 100kV and 80kV were all evaluated as 3, 2 and 1 respectively. The image quality score on sharpness of tube surface is shown in Table 1. The score about sharpness of tube surface at 100kV (tube inner diameter: 4, 3mm) was similar to 120kV (mean score: 3.0). At 80kV (all diameters), mean score was significantly lower than at 120kV (mean score: 2.0~2.2) (P<0.05). Only about tubes with

Table 1 Average score of qualitative analysis about sharpness of tube surface

	4mm			3mm			2mm		
	120kV	100kV	80kV	120kV	100kV	80kV	120kV	100kV	80kV
Reviewer 1	3	3	2.2*	3	3	2.2*	2.7	3.0**	2.2*
Reviewer 2	3	3	2.0*	3	3	2.0*	2.7	2.9	2.0*

*: significantly lower than 120 kV
**: significantly higher than 120 kV

Table 2 Results of qualitative analysis about contrast between the lumen of tube and plaque (control: 120 kV)

		Slightly higher		Similar		Slightly lower	
		Number of assess	rate (%)	Number of assess	rate (%)	Number of assess	rate (%)
4mm	100kV	0	0	45	50	0	0
	80kV	0	0	42	47	3	3
3mm	100kV	0	0	37	41	8	9
	80kV	0	0	31	34	14	16
2mm	100kV	0	0	34	38	11	12
	80kV	0	0	26	29	19	21

inner diameter of 2mm, the score of reviewer 1 at 100kV was significantly higher than that at 120kV. Interobserver agreement was good (κ=0.55~0.80).

3.1.2.2 Soft plaque phantom

Table 2 shows the results of evaluation of contrast between the lumen of tube and plaque. There was no “slightly higher” score at any tube voltage (i.e. 100kV or 80kV) and any tube diameter. As shown in Table 2, the rate of score showing that the contrast was similar compared with 120kV was higher at 100kV than 80kV at all tube diameters. At both 100kV and 80kV, the rate of score showing that the contrast was similar was highest at 4mm. As the tube diameter is smaller, the rate of score that the contrast was slightly lower was higher. The score that the contrast was slightly lower was not found at 100kV, 4mm.

3.2 Radiation dose

Figure 9 shows CTDIvol at each tube voltage and tube current. Compared with 120kV, the CTDIvol decreased to 60% and 29~32% at 100kV and 80kV respectively. The CTDIvol changed in a linear fashion with changes of tube current, and changed exponentially with changes of tube voltage.

4 Discussion

In CCTA, due to the small size and tortuous structure of coronary artery and heart beats, it is not always easy to obtain useful images. Therefore the radiation dose of CCTA is higher compared with other CT examinations. CT scan is generally performed with adjustment of tube current depending on patient size. However tube voltage has been generally fixed. Recently, the CT examination technique with low tube voltage setting has been receiving attention. The low tube voltage technique has some benefits, including improvement of image contrast by lowering of X-ray energy and reduction of radiation dose by decreasing X-ray intensity.

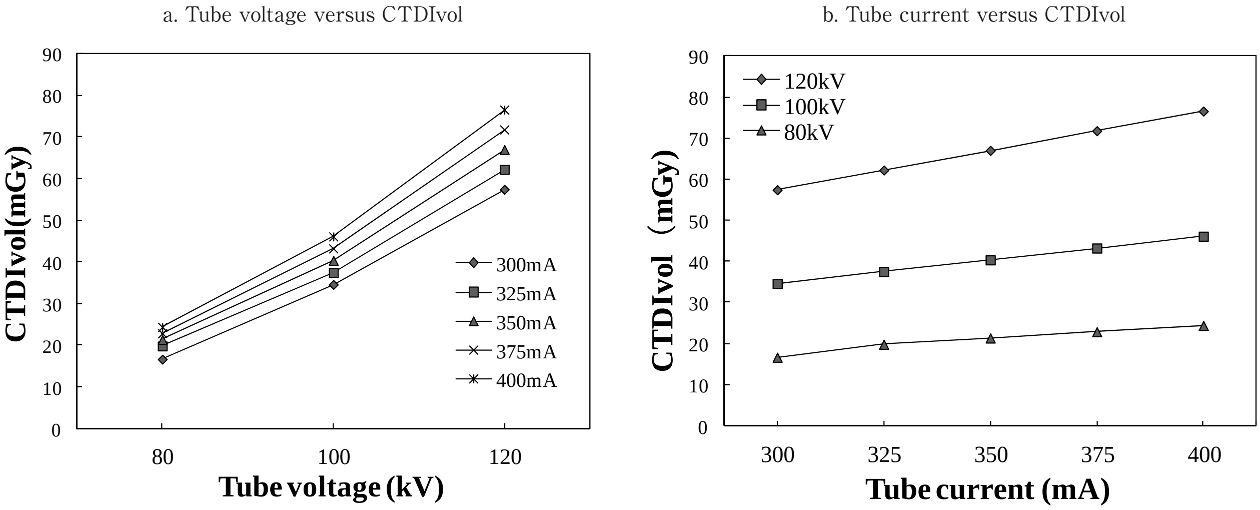


Figure 9 Graph shows CTDIvol at each tube voltage and tube current. Compared with 120kV, the CTDIvol decreased to 60% and 29~32% at 100kV and 80kV respectively. The CTDIvol changed in a linear fashion with changes of tube current, and changed exponentially with changes of tube voltage.

X-ray intensity is proportional to mA × kV^2, which affects radiation dose and image noise. X-ray intensity is proportional to tube current (with no spectrum change). On the other hand, intensity of X-ray that enters the detector is not proportional to generated X-ray intensity due to the change of radiation quality. At low tube voltage technique, image noise generally increases due to the heavy reduction of photon number, so that the trade-off between image quality and radiation dose must be considered. In this study, we investigated the effect of scan parameters (i.e. tube voltage and tube current) on image quality and radiation dose. This study was also designed to take radiation dose reduction into consideration.

In this study, the CT numbers increased with lowering of tube voltage. Because the mass attenuation coefficient of iodine for lower energy photon is high (Figure 10), the increase of CT numbers was attributable to increase of X-ray attenuation by contrast medium. By using low tube voltage technique, therefore, image contrast can be expected to improve at contrast enhancement examination. There were large fluctuations of the CT numbers of tube lumen compared with standard phantom. These fluctuations may be attributable to the small size of tube, beat and structure of phantom. A large difference of attenuation may also cause artifact because the tubes were surrounded by air.

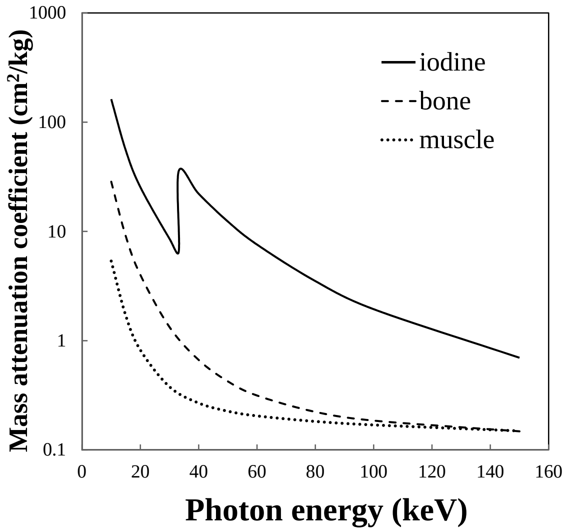


Figure 10 The figure shows the relationship between photon energy and mass attenuation coefficient of each material. The mass attenuation coefficient for lower energy photon is generally high, especially that of iodine. The figure was made with the national institute of standards and technology (NIST) data[13].

The increase rate of noise with lowering of tube voltage was similar to previous studies[14, 15]. Image noise generally increases inversely with the square root of radiation dose which enters the detector[16]. The amount of X-rays which entered detectors at low tube voltage were reduced for the following reasons: a) generated X-ray intensity was substantially decreased due to lowering of tube voltage, because X-ray intensity generated from X-ray tube is proportional to the square of tube voltage; b) attenuation rate of low energy X-ray by adding a filter is relatively high at low tube voltage; c) transmission strength of X-ray is weak at low energy X-ray.

Recently, iterative reconstruction which is used in nuclear medicine is becoming widely used as a reconstruction technique in CT examination, which makes it possible to reduce image noise. Increase of noise therefore might not be an important issue in the future.

CNR at 100kV was similar to one at 120kV because the degree of noise increase and increase of the CT numbers was similar at 100kV. While at 80kV, CNR was lower than 120kV because at 80kV the degree of noise increase is higher than the degree of the CT numbers increase.

The maximum tube current we set in this study was 400mA. If a CT scanner with higher tube current capability is used, the scan parameter without decrease of CNR at 80kV may be possible. At 100kV, if higher tube current is used, CNR may be similar compared with 120kV even if the contrast medium concentration is low. It might result in a reduction of contrast medium injection volume.

At a qualitative assessment, the score of noise was low at low tube voltage, whereas the score about sharpness of tube surface at 100kV was similar to 120kV. The finding that CNR didn't decrease at 100kV was compatible to the score that sharpness of tube at 100kV was similar to one at 120kV. The score of reviewer 1 at 100kV (tube diameter of 2mm) being significantly higher than 120kV could be attributed to the increase of the CT numbers at 100kV. At 80kV the mean score was significantly lower compared with 120kV at all scans. This decrease of score could be attributed to degradation of image quality (i.e. sharpness) of tube due to the increase of noise. Within this study, it is suggested that the increase of noise at 100kV does not affect the qualitative assessment.

The CT numbers of plaque significantly increased with lowering of tube voltage. For one thing, this increase of the CT numbers could be due to increase of X-ray attenuation by the plaque phantom made from synthetic resin. But mass attenuation coefficient of resin is lower than other materials (ex. iodine), so the increase of the CT numbers of plaque can be also attributed to the blooming artifact due to increase of the CT numbers of tube. At plaque phantom abutted on contrast medium (rest model), increase degree of the CT numbers with lowering of tube voltage was higher than those at the substantial volume phantom scanned on the bottle filled with water. This supported the conjecture that the increase of CT numbers of plaque phantom was also caused by the blooming artifact of contrast medium.

In some scans, there was no significant difference in the CT numbers of plaque at low tube voltage compared with 120kV. It was suspected that the CT numbers of plaque phantom didn't always increase because measurement might be inaccurate due to the small size of plaque phantom and the beat of phantom.

Contrast medium concentration did not have an effect on the CT numbers of the plaque at any tube voltages. This finding is not in accord with Cademartiri's report that coronary plaque attenuation values are significantly modified by differences in lumen contrast densities[17]. In our study, it may be suggested that the concentration did not have much effect on the CT numbers of the plaque because of the small range of contrast medium concentration (i.e. 300~400HU). The degree of the CT numbers increase of tube lumen seen with the increase of contrast medium concentration was smaller than the degree of those seen with lowering of tube voltage. The CT numbers at 350HU and 400HU were 1.17 and 1.33-times higher than those at 300HU. The CT numbers at 100kV and 80kV were 1.18~1.37 and 1.49~1.71-times higher than those at 120 kV (normal type phantom). From the previous discussions, the CT numbers of plaque was affected more by tube voltage than by contrast medium concentration. The CT numbers of plaque may be higher with low tube voltage. This should be taken into account when assessing the characteristics of plaque.

Additionally, the small size of plaque and beat of phantom were also considered as contributing cause of these results that suggest there was no correlation between CT numbers of plaque and contrast medium concentration. But also at rest model there was no correlation between contrast media concentration and CT number of plaque. So this finding could be attributed to the small size of plaque phantom rather than beat of phantom.

The detection of plaque which causes acute coronary syndromes (ACS) is important[18, 19], so the contrast between plaque and arterial blood must be maintained in any scanning protocols. In our study, CNR decreased to about 90% and 70% at 100kV and 80kV respectively compared with 120kV. This decrease of CNR was attributed to increase of plaque CT numbers and increase of noise with lowering of tube voltage. At qualitative analysis, the score that the contrast was slightly lower was not found at 100kV with 4mm tube. Decrease of CNR to about 90% compared with 120kV may not have an effect on image quality. At small tube diameter (i.e. 3mm and 2mm), the contrast was so slightly lower in some scans that this decrease of contrast may not be a big issue.

In summary about image quality of plaque phantom, the CT numbers of plaque increased with lowering of tube voltage. The contrast between the tube lumen and plaque at 100kV was similar to that at 120kV, but it decreased at 80kV.

The CTDIvol changed in a linear fashion with changes of tube current, and changed exponentially with changes of tube voltage. As mentioned before, X-ray intensity generated from X-ray tube is proportional to mA $\times$ kV^2 in theory. In our study, CTDIvol was proportional to tube current. But decrease rate of CTDIvol with lowering of tube voltage was higher than the square root of tube voltage. This finding can be attributed to attenuation of low energy X-ray by the added filter.

There were some limitations in this study. One is that the cardiac phantom used in this study simulated heart only. Thoracic wall, ribs and vertebrae were not considered. If these structures are considered, it may be difficult to apply low tube voltage technique in a clinical setting for the obese patient. Another is the size of the cardiac phantom is about 12cm, which is larger than average heart size of 10cm. When low tube voltage technique will be applied in a clinical setting the relationship between tube voltage and image contrast may be out of accord with those of this study, because X-ray attenuation by ball might be higher due to the large size. Third, because the phantom (ball filled with diluted contrast medium) was set in the center of the gantry, tubes were set off-center. Therefore the measurement of CT number might be inaccurate. However it seems that this layout isn't greatly different from one at clinical application because coronary artery is not at isocenter in clinical CCTA. Moreover, in this study, scan was performed with only 70bpm, the effect of heart rate and arrhythmia on image quality was not evaluated. Finally, the cardiac phantom was beaten along the body axis direction, which doesn't completely simulate contraction of the heart. In spite of these limitations in our study, the data suggests that low tube voltage technique is also useful for the small tube with beating.

This study investigated the effect of tube voltage and tube current on image quality and radiation dose, and estimated the usefulness of low tube voltage technique in CCTA by phantom study. Within this study, image quality at 80kV decreased but that at 100kV was similar compared with 120kV in both quantitative and qualitative analysis. At 100kV CTDIvol decreased to about 60% compared with 120kV. In conclusion, with 100kV setting, radiation dose could be substantially reduced without degradation of image quality, and it is suggested that low tube voltage technique may also be useful in CCTA especially for children and non-obese adults without cardiac arrhythmia.

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