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Combination of a low tube voltage technique with the hybrid iterative reconstruction (iDose) algorithm at coronary CT angiography

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Abstract

We compare the performance of low-tube voltage with the hybrid iterative reconstruction (iDose) with standard- and low-tube voltage with the filtered backprojection (FBP) using phantoms at CT coronary angiography (CTCA). At CTCA, application of the combined low-tube voltage with iDose resulted in significant image quality improvements compared to the low-tube voltage with FBP. Image quality was the same or better despite a reduction in the radiation dose by 76% compared with standard-tube voltage with FBP.

Keywords

Hybrid iterative reconstruction; computed tomography coronary angiography; low-tube voltage technique; filtered backprojection; radiation dose

INTRODUCTION

Multidetector computed tomography coronary angiography (CTCA) is an accurate method for the non-invasive diagnosis of coronary artery disease ^{1–5}. Potential applications of CTCA continue to emerge, leading to concerns regarding radiation exposure and to efforts to reduce exposure. Promising methods for low-dose CTCA include low tube voltage techniques ^{6, 7} that allow for a dose reduction of as much as 35% ^{8, 9} because the contrast between tissue and the contrast medium is enhanced at lower energy ^{10–13}. Although image noise is increased, low-contrast detectability is retained. To avoid the increase in image noise in clinical practice, the tube current-time product, and consequently, the radiation dose to the patient, is increased. Therefore, the technique is not applicable to large patients, because of the limited power of the high voltage generator; in our institution, we perform CTCA using the low-tube voltage technique in patients whose body mass index (BMI) is

less than 25.0. Alternately, the low-tube voltage technique has a potential to reduce the volume of the contrast media for patients who have impaired renal function.

A “hybrid iterative reconstruction algorithm (iDose)”¹⁴ has recently been developed, which consists of the following two de-noising components: (i) an iterative maximum likelihood-type sinogram restoration method based on Poisson noise distribution; and (ii) a local structure model fitting on image data that iteratively decreases the uncorrelated noise. When performed, iDose allows users to adjust the image noise level by inputting a parameter called “iDose level”. The larger the iDose level is, the larger the noise reduction is, allowing users to prospectively decrease the dose at the time of the scan (expecting that iDose will cancel the associated increased noise level during the reconstruction process).

We hypothesized that combining the low-tube voltage technique with the iDose-rather than the FBP algorithm would allow for a decrease in the radiation dose while yielding images whose quality and image noise would be comparable to or better than of images obtained with the FBP combination. In addition, we posited that the combined low-tube voltage with iDose technique may be useful for the acquisition of images in larger patients.

We used phantoms for our CTCA studies whose purpose was to compare the performance of the combined low-tube voltage with iDose technique with the performance of the combined standard voltage with FBP- and the combined low-tube voltage plus FBP technique. We paid particular attention to image noise- and radiation dose reduction and image quality improvement.

MATERIALS AND METHODS

Phantom

A Catphan phantom (Phantom Laboratory, Cambridge, NY) with a CTP486 module to calculate image noise and the noise power spectrum (NPS), and a CTP401 module was used^{15, 16}. The diameter and height (length along the z-axis) of the phantom were 20.0 cm. We assessed the contrast-to-noise ratio (CNR) and high contrast resolution.

Retrospectively-gated Helical Scanning and Image Reconstruction

We used retrospectively-gated helical scanning and image reconstruction on a 64 detector-row CT scanner (Brilliance-64, Philips Healthcare, Cleveland, OH). The scan parameters were detector configuration, 64×0.625 mm (detector collimation); slice thickness, 0.67 mm; gantry rotation time, 0.42 sec; beam pitch, 0.2; display field-of-view (FOV), 22.0 cm. An electrocardiogram (ECG) was acquired during the helical scan using the demo mode at a heart rate of 60 beats-per-minute (bpm). The standard tube voltage was 120 kV at 900 effective mAs without the ECG-based tube current modulation technique. The calculated CTDI_{vol} was 52.9 mGy. At low tube voltage (80 kV) the effective current-time product was 1400, 1260, 1120, 980, 840, and 700 effective mAs; the corresponding calculated CTDI_{vol} was 25.2, 22.7, 20.2, 17.6, 15.1, and 12.6 mGy, respectively. The dose reduction rate for each effective mAs at 80 kV was 52.4, 57.1, 61.8, 66.7, 71.5, and 76.2% compared to the CTDI_{vol} at 120 kV. The scan parameters for 900 effective mAs at 120 kV and for 1400 effective mAs at 80 kV were as used at our institution for routine coronary CT angiography scans. We acquired 2 scans for each setting.

The 120- and 80 kV images were reconstructed in synchronization with ECG data by using a standard FBP algorithm with a coronary standard kernel (CB). The center of the imaging window was set at 75% of the R-R interval. In addition, 80-kV images were reconstructed with the iDose algorithm (Philips Medical Systems) with the CB kernel. iDose is image reconstruction software, it allows the user to adjust the image noise level by inputting a

parameter called “iDose level” in percentages. The larger the iDose level, the larger the noise reduction; this allows the user to prospectively decrease the dose at the time of scanning with the expectation that iDose will cancel the increased noise level during the reconstruction process. Images were reconstructed from 80 kV data using the retrospectively ECG-gated method as above, but not with FBP but with iDose algorithm. When scanned with 80 kV, $CTDI_{vol}$ under various effective mAs conditions decreased from that with 120 kV, resulting in increased noise level (see Table 1). Thus, in these experiments, iDose levels were set to compensate for the increases in noise as follows: iDose level of 50% for 1400 effective mAs, 60% for both 1260 effective mAs and 1120 effective mAs, 70% for both 980 effective mAs and 840 effective mAs, and 80% for 700 effective mAs. When an iDose level of 50-, 60-, 70-, and 80% is applied, the compensating increase in noise (noise reduction) ideally corresponds to 29-, 37-, 45-, and 55%, respectively.

Measurement of Standard Deviations (SD) of the CT Number and of the Noise Power Spectrum (NPS)

For each scanning technique of 700 – 1400 effective mAs at 80 kV and 900 effective mAs at 120 kV, image noise was measured by one of the authors (Y. F.) at the uniformity module of the Catphan phantom; the noise index was the SD of pixel values within a 50-mm diameter circular region of interest (ROI).

Fifty consecutive images in the z direction were used for each of the helical scans and on each image the SD value at a central location was measured. The mean of the 50 SD values was used as the index of image noise at the center. Improvements in image noise were calculated by comparing the values obtained with the FBP and iDose algorithm. To examine the variance and spatial frequency characteristics of image noise, we measured the NPS on reconstructed images obtained with the FBP and iDose algorithm^{17, 18}. The basis for this analysis was 50 noise realizations, each of which was comprised of 256×256 pixels. We used CT data generated from the uniformity module of the Catphan phantom designed to simulate the attenuation of water. To remove low-frequency artifacts, 2 images acquired at the same location with the same scan settings but at different sessions were subtracted; this yielded the zero-mean signals (noises) of one realization. The NPS was then computed from the squared magnitude of the two-dimensional Fourier transform of each realization divided by the physical area of the realization. This procedure was performed for 50 slice locations. Finally the mean NPS was calculated from the mean of the NPS of the 50 images and expressed as the result. We also calculated the normalized NPS (NNPS); it was normalized by the max value of the noise power within a given spatial frequency.

Measurement of the Contrast-to-Noise Ratio

We calculated the contrast-to-noise ratio on images acquired at 120- and 80 kV with the FBP and the iDose algorithm using the CTP401 module of the Catphan phantom. One of the authors (Y.F.) measured the CT number of the teflon objects in a 13 mm-diameter area and of the background of the phantom using a circular ROI cursor. CNR values were calculated as follows: $(ROI_m - ROI_b)/SD_b$, where ROI_m and ROI_b are the CT number of the teflon objects and the background ROI, and SD_b is the standard deviation of the attenuation values of the background¹⁶. Mean CNR was calculated from the CNR measurements of 50 continuous images. In addition we calculated the figure of merit (FOM) for each effective current-time product; FOM was defined as $FOM = CNR^2/CTDI_{vol}$, where the CNR used is the mean CNR value¹⁰.

Measurement of Full Width at Half Maximum (FWHM)

We calculated FWHM to evaluate high contrast resolution on images obtained with the FBP and iDose algorithm. One author (Y.F.) measured the image profile curve for each effective

mAs using 5.0-mm spaced teflon objects (Fig. 1). FWHM was calculated from the half value between peaks and valleys of the profile curve. The measurements were performed 3 times and expressed as the mean value.

Visual Inspection of the Phantom Images

Images acquired at 120- and 80 kV and reconstructed with FBP and iDose were compared by 2 radiologists (S.O. and Y.Y.) for graininess, streak artifacts, nodule conspicuity, detectable nodule size, and overall image quality (Fig. 1). The evaluators inspected images acquired in 12 scan sessions (6 effective mAs $\times$ FBP or iDose) in an independent and blinded fashion. Each series consisted of 80 continuous images.

Graininess, streak artifacts, nodule conspicuity, and overall image quality were recorded on a 5-point scale where 1 = unacceptable with respect to diagnostic quality, 2 = suboptimal diagnostic quality and poor image quality compared to 120 kV, 3 = adequate diagnostic quality and slightly inferior image quality compared to 120 kV, 4 = good diagnostic quality and image quality comparable to 120 kV, 5 = excellent diagnostic quality and superior image quality compared to 120 kV. With respect to minimal detectable nodule size, the acrylic spheres we used ranged from 2 – 10 mm. Data were recorded as the mean of values reported by the 2 radiologists.

Statistical Analysis

We used the Cohen *k* test to assess the degree of agreement between the readers. A *k* value lower than 0.20 indicated poor-, 0.21–0.40 fair-, 0.41–0.60 moderate-, 0.61–0.80 good-, and 0.81–1.0 excellent agreement.

RESULTS

As shown in Table 1, image noise on images obtained at 80 kV with FBP was 25.33 HU at 1400- and 36.50 HU at 700 effective mAs; it increased as the effective current-time product decreased. With iDose, image noise on 80-kV images was 21.21 HU at 1400- and 19.57 HU at 700 effective mAs; at different iDose levels it remained constant regardless of the effective current-time product. Consequently, with iDose, image noise improved from 1.19 at 1400- to 1.87 at 700 effective mAs. The NPS reflected improvements in image noise; at 1400 effective mAs with iDose, but not with FBP, the NPS was decreased over the entire the frequency range (Fig. 2a). This trend was pronounced at 700 effective mAs and the NPS with iDose was reduced to a greater degree than with FBP (Fig. 2b). The normalized NPS for the FBP and iDose algorithm is shown in Figs. 3a and 3b. With iDose, the response for spatial frequency was identical to the response with FBP for the entire frequency range. In addition, although the noise reduction rate was increased at 700 effective mAs with iDose, there was no difference between FBP and iDose with respect to the response for spatial frequency (Fig. 3b).

At 120 kV, CNR was 79.16; on 80 kV images with FBP it ranged from 19.75 to 36.97 and was proportional to the increase in the effective current-time product (Table 2). With iDose, CNR ranged from 79.30 to 101.35 regardless of increases in the effective current-time product. With FBP, the value of FOM at 80 kV was less than half the FOM value at 120 kV (30.95 – 54.23 at 80- vs. 118.46 at 120 kV) (Table. 2). With iDose, the FOM at 80 kV was large and inversely proportional to increases in the effective tube-current time product, i.e. 714.41 at 700- and 352.02 at 1400 effective mAs.

FWHM values obtained at 80 kV were 2.55 – 2.65 with the FBP- and 2.50 – 2.55 with the iDose algorithm; these values are comparable to the 2.55 value obtained at 120 kV with FBP (Table 3).

The mean scores assigned by the 2 radiologists for the FBP and the iDose algorithm are shown in Table 4. The images they used in their evaluation of the 2 algorithms at different effective current-time products are presented in Fig. 4. With FBP, the minimal detectable nodule size was 6.0 mm at 1400-, 8.0 mm at 1260 – 1120-, and 10.0 mm at 980 effective mAs; there was no detection at 840 and 700 effective mAs. With iDose, the minimal detectable nodule size was smaller, i.e. 4.0 mm at 1400 – 980- and 6.0 mm at 840 – 700 effective mAs. The mean score for all effective mAs with iDose was greater than 3.0 and diagnostic quality was adequate compared to 120 kV scans. The mean score for 1260 – 700 effective mAs with the FBP algorithm was reflective of the poor image quality compared to images obtained at 120 kV.

There was good concordance between the 2 radiologists; the kappa coefficient was 0.79.

DISCUSSION

We investigated whether the low tube voltage (80 kV) with iDose technique allows for the acquisition of images at reduced radiation doses and image noise and whether their quality is comparable to images obtained at 80 kV or 120 kV with FBP. We posited that positive results would allow for the CTCA study of large patients irrespective of the BMI.

To evaluate changes in image noise at 80 kV, we changed the effective tube current-time product from 1400 to 700 effective mAs. We found that with FBP, image noise increased as the effective current time product decreased. On the other hand, with iDose it remained constant regardless of the effective current-time product. This is due to an increase in the iDose level with a decrease in the effective mAs to maintain an image noise level of 1400 effective mAs. Consequently FOM and CNR values are improved. Consequently FOM and CNR values are improved. The iDose algorithm makes it possible to reduce the noise on images acquired at 80 kV with FBP, approaching levels seen on 120 kV images. Our findings indicate that compared with the radiation dose at 120 kV, scanning at 80 kV with iDose allows for a dose reduction by 76% and that the FBP algorithm facilitates a 52.4% reduction at 80 kV scanning.

Excessive image noise reduction tends to result in image blurring and resolution degradation and reconstructed images appear “artificial”. Leipsic et al.^{19, 20} who used a different iterative reconstruction technique suggested that highly iterative reconstructions produce images that are significantly different in appearance from images acquired with FBP; their noise texture appears different and their borders manifest a significantly higher degree of smoothness. Similar observations were made by Silva et al.²¹ who noted that noise-freeness may manifest as an artifactual over-smoothing of the image. Unlike image-based adaptive filtering techniques, iDose use a combination of processing in the projection and image domain providing clinical advantages such as the ability to effectively remove photon starvation related artifacts e.g. streaks. Also, through the use of dynamic frequency noise removal techniques in such iterative reconstruction techniques it is possible to effectively remove noise across all frequencies. We used the iDose algorithm as an iterative reconstruction technique and neither of our readers reported a different noise texture or significantly smoothed borders. Image evaluations, the normalized NPS graph, and FWHM results also attest to the adequacy of our method. In their evaluation of overall image quality, both radiologists judged that the 80 kV with iDose method yielded images of the same or superior quality as the 120 kV with FBP technique. Even if two images have the same image noise, its frequency distribution can differ. In our study, the plots of the normalized NPS (Figs. 3a and 3b) show near-identity of the frequency responses of images reconstructed with iDose and FBP at 1400 and 700 effective mAs, respectively. This implies that the distribution of noise power remained the same for the entire spatial frequency.

Consequently, the noise texture of images reconstructed with iDose and FBP was not different although noise per se was significantly reduced by application of the iDose algorithm.

The challenge with “Iterative reconstruction” approach has been that iterating between the two domains (projection & image) is a time expensive operation, causing clinically “unacceptable” long reconstruction times. With iDose it has been possible to process/treat these domain independently, performing iterations in each domain separately. From a results endpoint, this approach helps address the same limitations of FBP (increased noise and artifacts) as achieved with iterative reconstruction, without the associated penalty of increased reconstruction time. The reconstruction speed for the iDose technique (taking into account all protocols), is approximately 16 IPS (images per second) on the enhanced reconstruction server compared with the 22 IPS of the FBP technique. Note that we are reporting numbers that take into account all the scan protocols as the reconstruction speeds vary significantly among the protocols as a function of the acquisition and reconstruction parameters (pitch, rotation time, matrix size, etc.). The reconstruction algorithm starts with projection data; first it identifies and corrects the noisiest CT measurements, i.e. those with a very poor signal-to-noise ratio or very low photon counts. Each projection is examined for points that are likely to have resulted from very noisy measurements using a model that includes the true photons statistics. Through an iterative diffusion process, the noisy data are penalized and edges are preserved. As this process ensures that the gradients of underlying structures are retained, resolution is retained and a significant noise reduction is allowed. The noise that remains after this stage of the algorithm is propagated to the image space; however, the propagated noise is now highly localized and can be effectively removed to support the desired level of dose reduction. The next major component of the iDose algorithm deals with subtraction of the image noise while preserving the underlying edges associated with true anatomy or pathology. This subtraction begins with an estimate of the noise distribution in the image volume and the estimate is used to reduce the noise while preserving the true structure.

Our study has some limitations. First, we calculated image noise improvements achieved with iDose from $CTDI_{vol}$ for scans acquired at 120- and 80 kV, although image noise was not the same on 120 kV and 80 kV images. Direct measurement of the center radiation dose and improvement calculations made at the iDose level may return results indicating that image noise at 80 kV is closer to that at 120 kV. Second, we used the CTP401 module that features 5 kinds of modules ranging from 2.0 to 10.0 mm to identify the minimal detectable nodule size. Although we had planned to use 9 kinds of low contrast module from 2.0 – 15.0 mm, the reconstructed slice thickness of 0.67 mm rendered detection impossible.

In conclusion, at CTCA, use of the low tube voltage with iDose technique significantly improves the quality of images compared to images acquired with the low tube voltage with FBP technique. In addition, with the former technique the radiation dose can be reduced by 76% compared with the 120 kV with FBP technique. Based on our findings we highly recommend the use of the low tube voltage (80 kV) technique with the iDose algorithm because it reduces image noise as well as the radiation dose of patients undergoing CTCA study.

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Figure 1.
Example of a phantom image used for FWHM measurements and image evaluation.

Figure 2.

a: Comparison of the noise power spectrum of the FBP and the iDose algorithm: 80 kV, 1400 effective mAs

b: Comparison of the noise power spectrum of the FBP and the iDose algorithm: 80 kV, 700 effective mAs

Figure 3.

a: Noise power spectrum normalized by the max value of the FBP and the iDose algorithm:
80 kV, 1400 effective mAs

b: Noise power spectrum normalized by the max value of the FBP and the iDose algorithm:
80 kV, 700 effective mAs

Figure 4. Phantom images acquired at 80 kV using the FBP and the iDose algorithm at effective mAs ranging from 1400 – 700. The calculated dose reduction rate, based on $CTDI_{vol}$ of 120 kV scans ranged from 52.4 – 76.2%.

Table 1

Image noise and noise improvement between FBP and iDose algorithm at 80 kV and image noise with FBP algorithm at 120 kV

Tube voltage (kV)	Effective mAs	CTDIvol (mGy)	Image noise		Noise improvement
			FBP	iDose (iDose level[%])	
80	1400	25.2	25.33	21.21 (50)	1.19
	1260	22.7	27.02	20.08 (60)	1.35
	1120	20.2	28.66	21.29 (60)	1.35
	980	17.6	30.84	19.84 (70)	1.55
	840	15.1	33.46	21.71 (70)	1.54
	700	12.6	36.50	19.57 (80)	1.87
120	900	52.9	16.28	—	—

Table 2

CNR and FOM of FBP and iDose algorithm obtained at 80 kV and 120 kV

Tube voltage (kV)	Effective mAs	CNR		FOM	
		FBP	iDose	FBP	iDose
80	1400	36.97	90.50	54.23	325.02
	1260	31.05	101.35	42.48	452.53
	1120	29.12	79.30	41.98	311.33
	980	28.8	89.18	47.15	451.85
	840	23.19	83.40	35.63	460.68
	700	19.75	94.88	30.95	714.41
<hr/>					
120	900	79.16	—	118.46	—

* Figure of merit (FOM)= $CNR^2/CTDI_{vol}$

Table 3

Full width at half maximum (FWHM) of FBP and iDose algorithm obtained at 80 kV and 120 kV

Tube voltage (kV)	Effective mAs	FWHM (mm)	
		FBP	iDose
80	1400	2.55	2.55
	1260	2.55	2.50
	1120	2.60	2.55
	980	2.55	2.55
	840	2.55	2.55
	700	2.65	2.55
120	900	2.55	—

Table 4

Mean score assigned by 2 radiologists evaluating phantom image with FBP and iDose

	Effective mAs											
	1400		1260		1120		980		840		700	
	FBP	iDose	FBP	iDose	FBP	iDose	FBP	iDose	FBP	iDose	FBP	iDose
Graininess	3	4	2	4	2	4	1.5	3.5	1	3.5	1	3.5
Streak artifact	3	5	3	5	3	4.5	2.5	4	2	4	1	4
Conspicuity of the nodule	2	5	2	5	2	4.5	1	4	1	4	1	3
Overall image quality	3	5	2	5	2	4.5	1	4	1	4	1	4
Detectable minimal nodule size (mm)	6	4	8	4	8	4	10	4	-	6	-	6

* represents the undetectable nodule