

# Reduced-Dose Low-Voltage Chest CT Angiography with Sinogram-affirmed Iterative Reconstruction versus Standard-Dose Filtered Back Projection<sup>1</sup>

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## Purpose:

To evaluate image quality of low-voltage chest computed tomographic (CT) angiography with raw data-based iterative reconstruction (sinogram-affirmed iterative reconstruction) in comparison with image quality of standard-dose standard-voltage filtered back projection (FBP) CT.

## Materials and Methods:

This prospective study was approved by the institutional review board, and the informed consent requirement was waived. Eighty consecutive patients who were referred for follow-up chest CT angiography underwent reduced-dose CT (hereafter, T2 examination) under technical conditions similar to those of the initial examination (hereafter, T1 examination), except the voltage selection was reduced by 20 kV with adaptation of the tube current to ensure a 50% reduction in CT dose index, and regular FBP was replaced by iterative reconstruction with sinogram-affirmed iterative reconstruction. The two techniques were compared by using paired tests (Student *t* test, Wilcoxon test, or McNemar test, according to the nature of variables).

## Results:

When compared with standard-dose T1 studies, reduced-dose T2 images showed: (a) significantly less objective noise at the level of the trachea on mediastinal and lung parenchymal images ( $P < .001$ ) and no significant difference in objective noise at the level of the aorta on mediastinal images ( $P = .507$ ); (b) significantly higher signal-to-noise and contrast-to-noise ( $P < .001$ ) ratios; (c) similar visual perception of noise on mediastinal ( $P = .132$ ) and lung ( $P = .366$ ) images, mainly rated as moderate; and (d) similar overall subjective image quality ( $P = .405$ ).

## Conclusion:

Raw data-based iterative reconstruction yielded equivalent subjective and improved objective image quality of low-voltage half-dose CT angiograms compared with standard-dose FBP CT images for an average dose-length product of less than 80 mGy·cm in this population.

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When faced with the challenge of providing diagnostic image quality at the lowest possible radiation dose, the radiologic community demonstrated the usefulness of several practical methods for routine chest computed tomography (CT), in particular, the individually adapted selection of scanning parameters that can be associated with automated tube current modulation systems (1,2). However, because dose and radiation exposure vary approximately with the square of voltage in the setting of a constant tube current (3), lowering the voltage is expected to have a greater effect on patient dose than reducing the tube current. This approach was investigated in several studies of adult populations, from which three main conclusions were drawn (3–9): First, substantial dose reduction can be achieved with low-voltage protocols for pulmonary CT angiography in adult patients, with an average dose reduction of 40% when lowering the setting from 120 to 80 kVp. This is probably a consequence of the improved vascular enhancement as a result of the increased attenuation of iodinated contrast media at low tube voltage, which improves the contrast-to-noise ratio of the images. Second, by lowering tube voltage, we improve vascular enhancement, as the attenuation of iodinated contrast media increases at low tube voltage. This was found to

improve the analyzability of central and peripheral pulmonary arteries (10). Third, all studies reported a common drawback of low-voltage protocols, namely increased image noise. The increased noise varied from minimal to marked depending on the patient's body mass index (BMI), running the risk of altering the diagnostic value of images. The recent possibility to reduce image noise with iterative reconstruction offers a new option, namely the possibility to achieve diagnostic image quality with reduced-dose data sets. To date, this approach has been investigated successfully exclusively in protocols in which the radiation dose is decreased by means of lowered tube currents (11–14). The purpose of the present study was to evaluate image quality of low-voltage chest CT angiography with raw-data-based iterative reconstruction in comparison with image quality of standard-dose standard-voltage filtered back projection (FBP) CT.

### Materials and Methods

One author (T.F.) is an employee of Siemens Healthcare. All other authors had control of any data that might have presented a conflict of interest.

### Patient Population

Over a 6-month period (April 2010 through September 2010), we used a modified chest CT angiographic protocol that was applicable to any patient who was previously examined with our CT unit and referred for a follow-up examination. This prospective study that

uses a second generation of iterative reconstruction was approved by our institutional review board, and the informed consent requirement was waived. In the present investigation, the follow-up CT examination had to be performed under strictly similar conditions to those at the time of the initial examination except for (a) voltage selection, which was reduced by 20 kV with adaptation of the tube current, and (b) replacement of FBP with iterative reconstruction. In our clinical routine, oncologic patients were the target population for this protocol, and changes in the patient's body weight, extent of tumoral disease, or both might have interfered with the overall image quality analysis. Thus, care was taken to exclude patients with a difference in body weight of more than 3 kg, as well as those with a change in overall tumoral extent of more than 10%. Whereas the former criterion could be checked before CT, the latter could be analyzed only after completion of follow-up CT. Eighty-nine consecutive patients were enrolled in this prospective investigation; however, nine were excluded because of regression ( $n = 4$ ) or progression ( $n = 5$ ) of the underlying tumoral disease.

The study population included patients referred for follow-up of thoracic malignancy ( $n = 66$ ) or indeterminate

### Advance in Knowledge

- When compared with the standard convolution filtered back projection reconstruction algorithm, the raw data-based iterative reconstruction algorithm yielded significantly less noise (18.33 HU vs 21.48 HU,  $P < .001$ ) and improved signal-to-noise (12.92 vs 10.86,  $P < .001$ ) and contrast-to-noise (11.68 vs 9.42,  $P < .001$ ) ratios at low-tube-voltage (80 kVp and 100 kVp) CT pulmonary angiography; the additional benefit of low-voltage scanning is the concurrent reduction in the radiation dose delivered to patients.

### Implications for Patient Care

- Enhanced chest CT images reconstructed with the raw data-based iterative technique can be of diagnostic image quality and can be obtained with a reduced dose ( $<1.5$  mSv).
- With raw data-based iterative reconstruction, it is possible to reduce dose while improving vascular enhancement at reduced-dose chest CT angiography in routine clinical practice.

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### Abbreviations:

BMI = body mass index  
 CNR = contrast-to-noise ratio  
 CTDI = CT dose index  
 DLP = dose-length product  
 FBP = filtered back projection  
 SNR = signal-to-noise ratio

### Author contributions:

Guarantors of integrity of entire study, F.P., J.R., M.R.; study concepts/study design or data acquisition or data analysis/interpretation, all authors; manuscript drafting or manuscript revision for important intellectual content, all authors; approval of final version of submitted manuscript, all authors; literature research, F.P., C.M., J.R., M.R.; clinical studies, F.P., J.P., J.B.F., J.R., M.R.; statistical analysis, F.P., A.D.; and manuscript editing, F.P., T.F., C.M., J.R., M.R.

Conflicts of interest are listed at the end of this article.

lung nodules ( $n = 14$ ). Because the lowest tube voltage available on the CT unit was 80 kVp, patients examined at 80 kVp in the initial examination could not be included in the present study. Hereafter, initial and follow-up CT angiographic examinations will be described as T1 and T2 examinations, respectively.

### CT Evaluation

**CT examinations.**—T1 and T2 examinations were performed with the same dual-source 128-section CT system (Somatom Definition Flash; Siemens Healthcare, Forchheim, Germany). Apart from voltage selection, which was reduced by 20 kV, and reference tube current-time product chosen to reduce the predictive CT dose index (CTDI) by 50%, CT parameters for T1 and T2 examinations were kept constant, including: (a) the number, projection (anteroposterior or lateral), and voltage selection for the topograms to ensure a similar contribution of the automatic exposure control (Care Dose 4D; Siemens Healthcare) at initial and follow-up CT and (b) the examination parameters (collimation,  $64.0 \times 2.0 \times 0.6$  mm, with z-flying focal spot for the simultaneous acquisition of 128 overlapping 0.6-mm sections with both measurement systems; rotation time, 0.28 second with 75-msec temporal resolution of the images; pitch, 3.0; four-dimensional dose modulation with an average curve). With regard to voltage selection, at the initial examination, there was a systematic adaptation of the voltage according to the patient's body weight (100 kVp for body weight of 50–80 kg, 120 kVp for body weight >80 kg). At the follow-up examination, the voltage was reduced by 20 kV, with adaptation of the tube current to reduce the predictive CTDI by 50%. The injection protocols were strictly similar in T1 and T2 examinations, consisting of the administration of 100 mL of a 30% iodinated contrast agent (iobitridol, Xenetix 300; Guerbet, Villepinte, France) at a flow rate of 4 mL/sec via a dual-headed pump injector (Stellant; Medrad France, Rungis, France) without saline flush. The

threshold of the bolus tracking system (Care Bolus; Siemens Healthcare) was set at 150 HU, with the region of interest positioned within the ascending aorta. In our clinical practice, T1 examinations represented standard-dose chest CT examinations, and T2 examinations represented reduced-dose chest CT examinations.

**CT image reconstruction.**—T1 images consisted of 1-mm-thick contiguous images reconstructed with a standard FBP algorithm, as implemented on the CT scanner with a high-spatial-resolution kernel (B50, lung images) and a soft-tissue kernel (B20, mediastinal images). Images were generated by using the built-in reconstruction computer of the CT system at the time of each patient's initial referral, and they were available in CD-ROM format at the time of follow-up CT to ensure strict parameter selection. T2 images consisted of 1-mm-thick contiguous images reconstructed with raw data-based iterative reconstruction (Sino-gram-affirmed Iterative Reconstruction [SAFIRE]; Siemens Healthcare) (Appendix E1 [online]). T2 images were reconstructed by using a high-spatial-resolution kernel (I50, lung images) and a soft-tissue-resolution kernel (I20, mediastinal images). The iterative reconstruction kernels I50 and I20 are designed to closely match the spatial resolution (in terms of the 50% and 10% values of the modulation transfer function for high-contrast objects) of the corresponding FBP kernels B50 and B20, respectively. The strength of the SAFIRE software was set at 3. At the beginning of this evaluation, the commercial version of SAFIRE software had not been released, leading the investigators to reconstruct images with a prototype workstation; the overall process of the off-line reconstruction process required 50–60 minutes. T1 and T2 images were viewed by using standard mediastinal (window width, 400 HU; window center, 40 HU) and lung parenchymal (window width, 1600 HU; window center, –600 HU) window settings. The SAFIRE software used in this study was similar to the version released commercially.

### CT Parameters Analyzed

Assessment of objective and subjective image noise and overall image quality of T1 and T2 images followed the method previously described by Pontana et al (12,13). Objective assessment of noise was obtained by measuring the standard deviation of pixel values in homogeneous regions of interest on mediastinal (B20 and I20) images (at two anatomic levels, the tracheal lumen above the aortic arch [the usual level of noise measurement]; the descending aorta at the level of ventricular cavities [chosen to analyze the level of noise in a region with highly attenuating anatomic structures]) and on lung (B50 and I50) images (at one anatomic level, the tracheal lumen above the aortic arch). The visual perception of noise, defined by the grainy appearance of the CT images, was evaluated on mediastinal and lung images of T1 and T2 examinations and rated as minimal (score of 1), moderate (score of 2), or marked (score of 3); the latter altered identification of normal structures, abnormal structures, or both. The overall image quality of lung and mediastinal images of each group of reconstructions was rated by using a three-point scale, as follows: a score of 1 indicated excellent image quality; a score of 2, good image quality; and a score of 3, a nondiagnostic examination. The signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR) were calculated by using the method described by Szucs-Farkas et al (9). No attempt was made to analyze image quality of the liver, upper abdomen, or both, as the studies consisted of chest CT images. An estimated effective dose was calculated for each CT examination by multiplying the dose-length product (DLP) by the normalized effective dose conversion coefficient for the chest ( $0.017$  mSv/[mGy·cm]).

### Conditions of Image Analysis

Two chest radiologists (J.P., F.P.; 4 and 8 years of CT experience, respectively) independently evaluated subjective image quality (6-month interval between the two sets of readings for each reader), while quantitative image parameters were obtained once, after the

Table 1

## Comparison of Objective and Subjective Image Quality between T1 and T2 Examinations

## A: Objective Image Quality

Image Quality	T1 Examinations (n = 80)	T2 Examinations (n = 80)	Difference between T2 and T1 Examinations*	P Value
Image noise				
At the level of the trachea on mediastinal images (HU)	21.48 ± 9.41	18.33 ± 5.89	-3.15 (-4.29, -2.01)	<.001
At the level of the aorta on mediastinal images (HU)	33.69 ± 8.86	34.20 ± 9.72	0.51 (-1.02, -2.04)	.507
At the level of the trachea on lung images (HU)	48.77 ± 11.73	41.28 ± 10.39	-7.49 (-9.31, -5.67)	<.001
SNR	10.86 ± 3.15	12.92 ± 3.80	2.06 (1.43, 2.69)	<.001
CNR	9.42 ± 3.13	11.68 ± 3.64	2.26 (1.66, 2.87)	<.001

## B: Subjective Image Quality

Image Quality	T1 Examinations (n = 80)	T2 Examinations (n = 80)	P Value
Image noise			
Mediastinal images	...	...	.132
Minimal (score 1)	23 (28.8)	18 (22.5)	...
Moderate (score 2)	57 (71.2)	62 (77.5)	...
Marked (score 3)	0	0	...
Lung images	...	...	.366
Minimal (score 1)	30 (37.5)	27 (33.8)	...
Moderate (score 2)	50 (62.5)	53 (66.2)	...
Marked (score 3)	0	0	...
Overall image quality	...	...	.405
Excellent (score 1)	21 (26.2)	18 (22.5)	...
Good (score 2)	59 (73.8)	62 (77.5)	...
Nondiagnostic (score 3)	0	0	...

Note.—In part A, unless otherwise indicated, data are means ± standard deviations, and statistical analysis was performed with the two-sided Student *t* test. In part B, unless otherwise indicated, data are numbers of patients, data in parentheses are percentages, and statistical analysis was performed with the McNemar test.

\* Data are means, and data in parentheses are 95% confidence intervals.

regions of interest were positioned in consensus by the two readers. To standardize subjective evaluation, images obtained in five patients not included in the study were read by consensus prior to qualitative assessment. In case of discordant scores in subjective image analysis of the study group, the corresponding series of images (the lung images, mediastinal images, or both) with discordant scores was reanalyzed, and consensus between the two readers was reached. Results of subjective image quality reported in Tables 1, 2, and 3 represent concordant readings.

Image analysis was performed at a clinical workstation (Syngo CT Workplace; Siemens Healthcare), where lung and mediastinal images obtained at T1 and T2 examinations had been transferred systematically. Because images reconstructed with iterative reconstruction have a typical appearance that is

easily recognized on CT images, the radiologists were not blinded to the differences in technique. Nevertheless, the images were presented in random order in each group of studies.

### Statistical Analysis

**Sample size.**—We chose image noise at the level of the trachea on mediastinal images as the primary outcome for sample size computation because measurement at this anatomic site on mediastinal images is the usual method used to assess objective image noise on chest CT images. We assumed that in the worst-case scenario, the dose reduction induced by the reduced voltage could increase the image noise despite use of the raw data-based iterative reconstruction technique. We used a noninferiority hypothesis to compute an estimated sample size and estimated that a small increase in image noise (the

worst-case scenario) would be acceptable with respect to the benefit of raw data-based iterative reconstructions in terms of dose reduction. For standard FBP reconstructions, the mean image noise measured at the level of the trachea on mediastinal images had been previously estimated at 22.6 HU ± 6 (standard deviation) (15). We hypothesized that an increase in noise between 5% and 10% could be acceptable with the raw data-based iterative reconstruction technique. Consequently, the noninferiority margin was fixed at 7.5% of the mean level of image noise, namely 1.7. We assumed a correlation value of 0.5 between the two measures of noise according to the two reconstruction techniques, and an estimated 80 patients were necessary for 80% power, with significance at 5%.

**Data analysis.**—Results were expressed as means and standard devia-

Table 2

## Comparison of Objective and Subjective Image Quality in Patients Initially Scanned at 120 kVp and Followed Up at 100 kVp

## A: Objective Image Quality

Image Quality	Standard-Dose T1 CT Images Acquired at 120 kVp and Reconstructed with FBP (n = 53)	Reduced-Dose T2 CT Images Acquired at 100 kVp and Reconstructed with the Raw Data-based Iterative Technique (n = 53)	Difference between T2 and T1 Examinations*	P Value
<b>Image noise</b>				
At the level of the trachea on mediastinal images (HU)	19.48 ± 5.14	17.52 ± 3.36	-1.96 (-3.0, -0.92)	<.001
At the level of the aorta on mediastinal images (HU)	29.97 ± 5.81	29.85 ± 5.90	-0.12 (-1.70, 1.46)	.875
At the level of the trachea on lung images (HU)	45.98 ± 9.96	39.53 ± 9.33	-6.46 (-8.71, -4.21)	<.001
SNR	10.92 ± 3.00	13.25 ± 4.26	2.33 (1.62, 3.04)	<.001
CNR	9.36 ± 2.96	11.92 ± 4.07	2.56 (1.89, 3.23)	<.001

## B: Subjective Image Quality

Image Quality	Standard-Dose T1 CT Images Acquired at 120 kVp and Reconstructed with FBP (n = 53)	Reduced-Dose T2 CT Images Acquired at 100 kVp and Reconstructed with the Raw Data-based Iterative Technique (n = 53)	P Value
<b>Image noise</b>			
Mediastinal images	...	...	.706
Minimal (score 1)	18 (34.0)	17 (32.1)	...
Moderate (score 2)	35 (66.0)	36 (68.9)	...
Marked (score 3)	0	0	...
Lung images	...	...	.706
Minimal (score 1)	25 (47.2)	26 (49.1)	...
Moderate (score 2)	28 (52.8)	27 (50.9)	...
Marked (score 3)	0	0	...
Overall image quality	...	...	.739
Excellent (score 1)	16 (30.19)	17 (32.08)	...
Good (score 2)	37 (69.8)	36 (68.9)	...
Nondiagnostic (score 3)	0	0	...

Note.—In part A, unless otherwise indicated, data are means ± standard deviations, and statistical analysis was performed with the two-sided Student *t* test. In part B, unless otherwise indicated, data are numbers of patients, data in parentheses are percentages, and statistical analysis was performed with the McNemar test.

\* Data are means, and data in parentheses are 95% confidence intervals.

tions for continuous variables and as frequencies and percentages for categorical variables. The noninferiority hypothesis was assessed by using a two-sided 95% confidence interval for the true difference between the two means of objective image noise at the level of the trachea on mediastinal images (raw data-based iterative reconstructions minus standard FBP reconstructions). Noninferiority could be acknowledged if the upper limit of this confidence interval was less than 1.7 (the noninferiority margin). After the test run of noninferiority, it is possible to conclude to a superiority if the two following conditions hold: the assumption of noninferiority is accepted, and the two-sided test of

difference between the means enables one to reject equality of means and conclude in the sense of superiority (16). This procedure was used for objective image noise at the level of the trachea on mediastinal images. For the other parameters, we used the following procedure: For continuous variables, comparisons between the techniques were performed by using a paired Student *t* test and a paired Wilcoxon test. For categorical parameters, comparisons were performed by using the McNemar test. For each categorical variable, the McNemar test was based on the contingency tables crossing the results of full-dose images reconstructed with FBP and reduced-dose images reconstructed with raw data-based iterative

reconstruction. Statistical significance was defined as  $P < .05$ . Interobserver reliability was assessed by using the  $\kappa$  coefficient; a value greater than 0.8 was considered very good. Statistical analyses were performed by using SAS software (SAS Institute, Cary, NC).

## Results

## Patient Characteristics

The mean age of the study population at the time of patient enrollment was 58.8 years ± 10.4 (median age, 58.0 years); 58 male patients (mean age, 60.6 years ± 8.7) and 22 female patients (mean age, 53.8 years ± 12.9) were included. The mean BMI of our study group was 25.5

Table 3

## Comparison of Objective and Subjective Image Quality in Patients Initially Scanned at 100 kVp and Followed Up at 80 kVp

## A: Objective Image Quality

Image Quality	Standard-Dose T1 CT Images Acquired at 100 kVp and Reconstructed with FBP (n = 27)	Reduced-Dose T2 CT Images Acquired at 80 kVp and Reconstructed with the Raw Data-based Iterative Technique (n = 27)	Difference between T2 and T1 Examinations*	P Value
<b>Objective image noise</b>				
At the level of the trachea on mediastinal images (HU)	25.40 ± 13.85	19.92 ± 8.87	-5.2 (-7.9, -1.0)	<.001
At the level of the aorta on mediastinal images (HU)	40.99 ± 9.34	42.76 ± 10.14	0.1 (-3.2, 5.5)	.391
At the level of the trachea on lung images (HU)	54.22 ± 13.14	44.71 ± 11.65	-7.6 (-12.4, -3.9)	<.001
SNR	10.73 ± 3.48	12.26 ± 2.63	2.29 (-0.32; 4)	.016
CNR	9.52 ± 3.51	11.21 ± 2.60	2.86 (-0.23; 4.16)	.010

## B: Subjective Image Quality

Image Quality	Standard-Dose T1 CT Images Acquired at 100 kVp and Reconstructed with FBP (n = 27)	Reduced-Dose T2 CT Images Acquired at 80 kVp and Reconstructed with the Raw Data-based Iterative Technique (n = 27)	P Value
<b>Subjective image noise</b>			
Mediastinal images	...	...	.046
Minimal (score 1)	5 (18.5)	1 (3.7)	...
Moderate (score 2)	22 (81.5)	26 (96.3)	...
Marked (score 3)	0	0	...
Lung images	...	...	.046
Minimal (score 1)	5 (18.5)	1 (3.7)	...
Moderate (score 2)	22 (81.5)	26 (96.3)	...
Marked (score 3)	0	0	...
Overall image quality	...	...	.046
Excellent (score 1)	5 (18.5)	1 (3.7)	...
Good (score 2)	22 (81.5)	26 (96.3)	...
Nondiagnostic (score 3)	0	0	...

Note.—In part A, unless otherwise indicated, data are means ± standard deviations, and statistical analysis was performed with the two-sided Student *t* test. In part B, unless otherwise indicated, data are numbers of patients, data in parentheses are percentages, and statistical analysis was performed with the McNemar test.

\* Data are means, and data in parentheses are 95% confidence intervals.

kg/m<sup>2</sup> ± 4.8 (median BMI, 25.26); our study comprised six underweight patients (BMI <18.5 kg/m<sup>2</sup>), 30 normal-weight patients (BMI of 18.5–24.9 kg/m<sup>2</sup>), 33 overweight patients (BMI of 25.0–29.9 kg/m<sup>2</sup>), and 11 obese patients (BMI ≥30 kg/m<sup>2</sup>). There was no significant difference in the mean BMI of patients between the two scans (*P* = .952). The median interval between CT examinations was 138.5 days, with a range of 7 to 345 days (mean interval, 145.2 days).

### Scanning Conditions at Initial and Follow-up CT

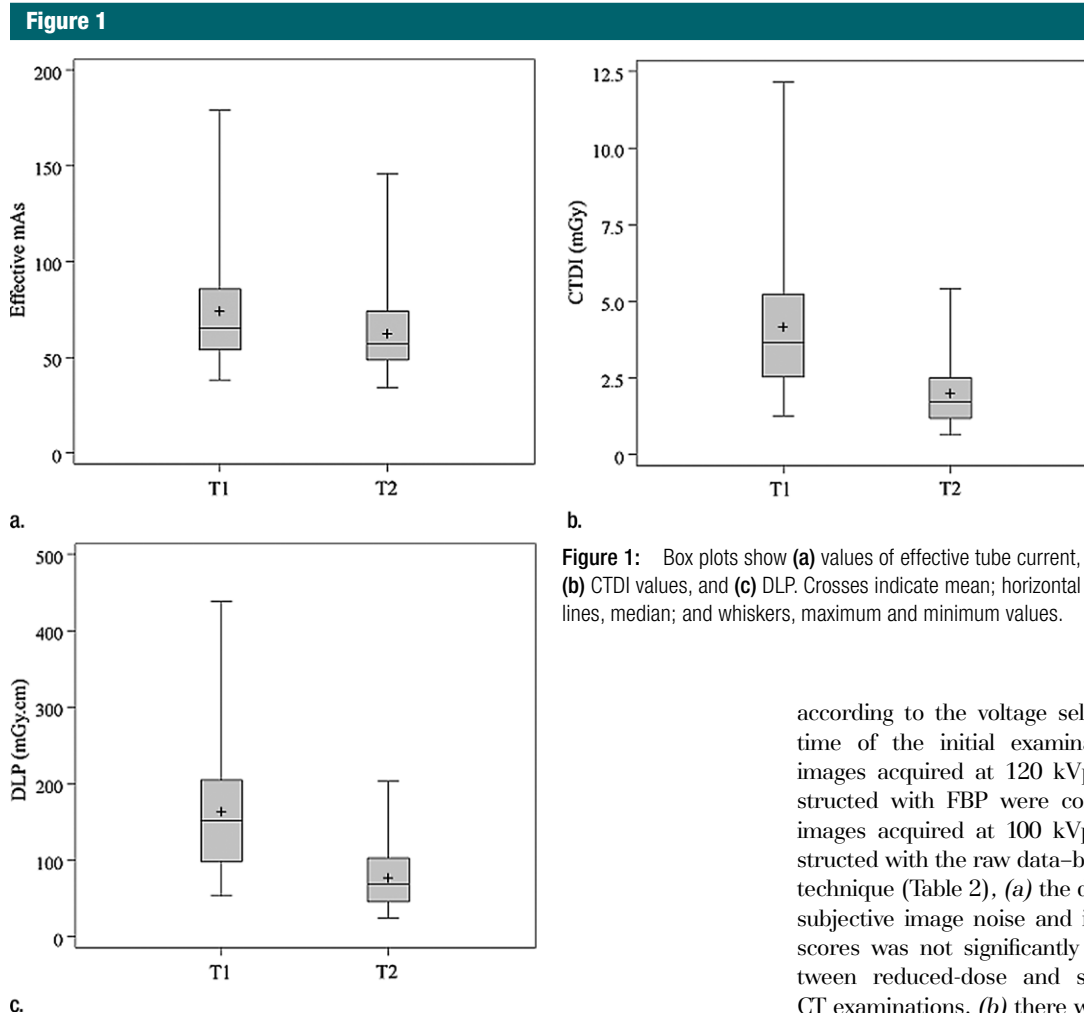
Follow-up CT angiograms were obtained at 100 kVp (instead of at 120 kVp as in the initial examination) in 53 patients (66.2%) and at 80 kVp (instead of at

100 kVp as in the initial examination) in 27 patients (33.8%).

At T1, mean effective tube current was 73.9 mAs (median, 65 mAs; range, 38–179 mAs); mean DLP, 163.6 mGy-cm (median, 152 mGy-cm; range, 53–438 mGy-cm); mean CTDI, 2.78 mGy (median, 2.58 mGy; range, 0.90–7.45 mGy); and mean effective dose, 2.78 mSv (median, 2.58 mSv; range, 0.99–7.45 mSv).

At T2, mean effective tube current was 62.5 mAs (median, 57 mAs; range, 34–146 mAs); mean DLP, 77.3 mGy-cm (median, 69.50 mGy-cm; range, 25–203 mGy-cm); mean CTDI, 1.18 mGy (median, 1.18 mGy; range, 0.43–3.45 mGy); and mean effective dose, 1.31 mSv (median, 1.18 mSv; range, 0.43–3.45 mSv).

These results are summarized in Figure 1. Specific attention was directed toward the dose values observed in the subgroup of 25 patients with a body weight of 70–75 kg: At T1, mean volume CTDI was 3.3 mGy (median, 2.7 mGy; range, 1.97–7.19 mGy); mean DLP, 128 mGy-cm (median, 106 mGy-cm; range, 71–281 mGy-cm); and mean effective dose, 2.18 mSv (median, 1.80 mSv; range, 1.21–4.78 mSv). At T2, mean volumetric CTDI was 1.58 mGy (median, 1.31 mGy; range, 0.87–3.27 mGy) (*P* < .001); mean DLP, 60.88 mGy-cm (median, 52 mGy-cm; range, 32–124 mGy-cm) (*P* < .001); and mean effective dose, 1.03 mSv (median, 0.88 mSv; range, 0.54–2.11 mSv) (*P* < .001).



**Figure 1:** Box plots show (a) values of effective tube current, (b) CTDI values, and (c) DLP. Crosses indicate mean; horizontal lines, median; and whiskers, maximum and minimum values.

### Comparisons of Image Quality between Reduced-Dose and Standard-Dose CT Angiographic Examinations

Table 1 provides results on the comparative analysis of subjective and objective image quality between the two groups. Interobserver agreement was 0.78 for T1 examinations (95% confidence interval: 0.68, 0.88) and 0.83 for T2 examinations (95% confidence interval: 0.72, 0.93). When considering the overall study group, the difference between mean noise measured at the level of the trachea on mediastinal T2 images (raw data-based iterative reconstruction) and that measured on mediastinal T1 images (FBP) was  $-3.15$  (95% confidence interval:  $-4.29$ ,  $-2.01$ ) (Table 1). The upper

limit of the confidence interval was less than 1.7, which led us to conclude that raw data-based iterative reconstruction was not inferior to standard-dose FBP.

As shown in Table 1, there was a significant reduction in the objective noise measured at the level of the trachea on lung and mediastinal images and significantly higher SNR and CNR in T2 examinations than in T1 examinations. No significant difference between T1 and T2 examinations was found in objective noise at the level of the aorta on mediastinal images, subjective image noise on lung and mediastinal images, or distribution of overall image quality scores.

Tables 2 and 3 compare image quality of the initial and follow-up examinations

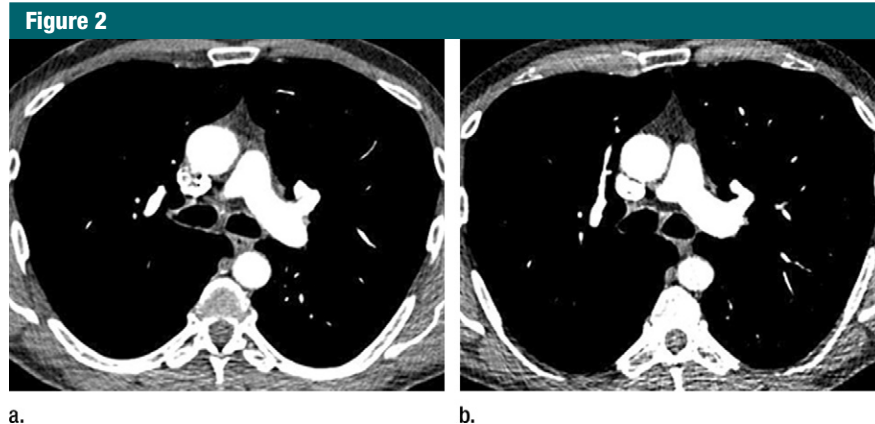
according to the voltage selection at the time of the initial examination. When images acquired at 120 kVp and reconstructed with FBP were compared with images acquired at 100 kVp and reconstructed with the raw data-based iterative technique (Table 2), (a) the distribution of subjective image noise and image quality scores was not significantly different between reduced-dose and standard-dose CT examinations, (b) there was significant reduction in objective noise at the level of the trachea on both lung and mediastinal images but no significant difference at the level of the aorta, and (c) CNR and SNR were significantly higher.

When images acquired at 100 kVp and reconstructed with FBP were compared with images acquired at 80 kVp and reconstructed with the raw data-based iterative technique (Table 3), significant differences were found in the distribution of subjective image noise and overall image quality scores. On 80-kVp images, (a) subjective image noise increased, and the proportion of examinations rated as having good image quality (score 2) increased at the expense of examinations rated as having excellent image quality (score 1); (b) there was a significant reduction in objective noise at the level of the trachea on both lung and mediastinal images and

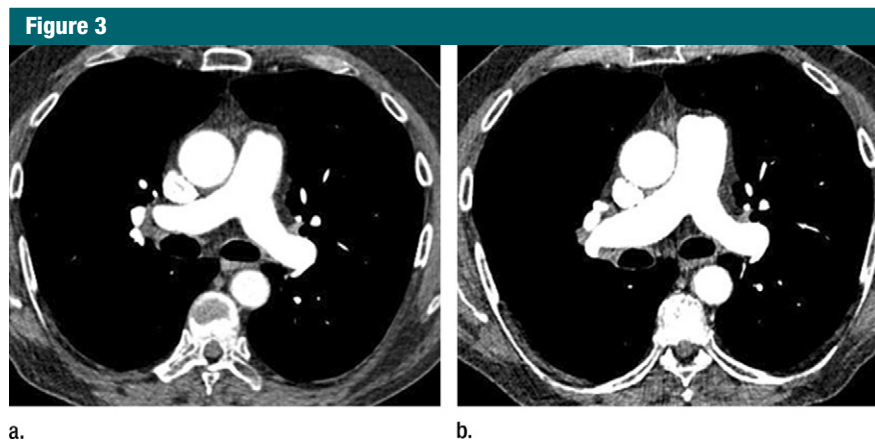
no significant difference at the level of the aorta; and (c) CNR and SNR were significantly higher. Figures 2–4 show the image quality that can be achieved with raw data–based iterative reconstruction according to the patient’s BMI. These images were obtained in one normal-weight patient (Fig 2), one overweight patient (Fig 3), and one obese patient (Fig 4).

### Discussion

In the present investigation, we have shown that iterative reconstructions performed with the newly developed SAFIRE algorithm can yield image quality similar to that achieved with FBP at a 50% lower dose when combined with the reduced tube voltage of CT examinations. Despite moderate image noise due to the systematic weight-adapted selection of voltage in our routine, diagnostic image quality was maintained in all CT examinations and was rated good or excellent. These results show that iterative reconstructions are a suitable tool that can be used to overcome the main limiting factor of low-voltage scanning, namely the decrease in subjective image quality of reduced-dose CT images when reconstructed with FBP (4–6). In regard to objective image noise, we observed a significant reduction in image noise at the level of the trachea on both lung and mediastinal images (the usual site for assessment of objective image noise). Because of the presence of highly attenuating structures in the lower part of the chest, we also included measurements of objective noise at the level of the ventricular cavities to determine if reduced-dose images would be hindered by marked graininess and its subsequent risk of altering the overall diagnostic value of images. In a previous study, Kim et al (6) reported altered image quality of the aorta on 80-kVp images of the chest because of the increased noise and streak artifacts associated with contrast media. Although we did not specifically investigate streak artifacts, we observed that objective image noise did not differ between standard- and reduced-dose images, providing indirect evidence that reliable image quality was maintained in the lower part of the chest. By using the



**Figure 2:** Comparison of T1 and T2 images obtained 2 months apart in a 49-year-old man (normal BMI, 22 kg/m<sup>2</sup>). (a) FBP reconstructed mediastinal image obtained at initial standard-dose CT (100 kVp; 48 mAs [effective]; DLP, 83 mGy·cm) at the level of the carina shows reference image quality (objective noise at the level of the trachea, 29.4 HU; subjective image noise rated minimal). (b) Follow-up reduced-dose mediastinal CT image reconstructed with the raw data–based iterative technique (80 kVp; 44 mAs [effective]; DLP, 38 mGy·cm) at the same level as a. Note the reduction in objective noise measured at the level of the trachea (21.6 HU) and the improved CNR (13.45 vs 10.60) and SNR (14.19 vs 11.90). Subjective image noise was rated similarly when compared with T1 images.



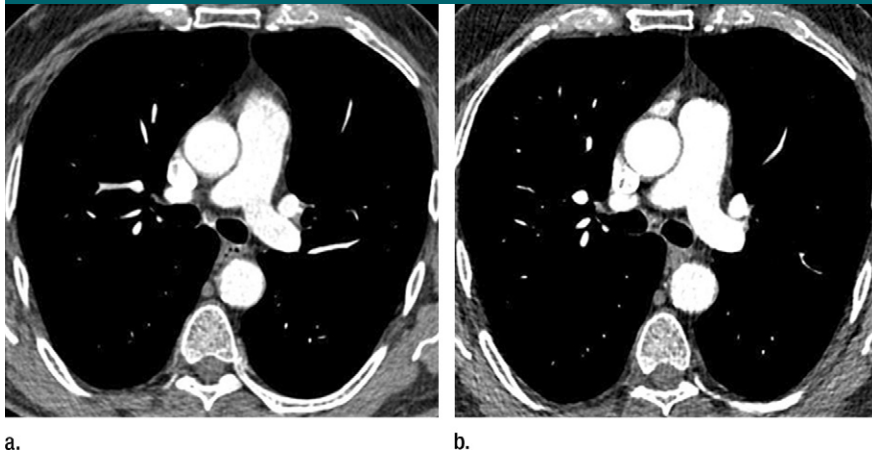
**Figure 3:** Comparison of T1 and T2 images obtained 6 months apart in a 61-year-old overweight man (BMI, 25.2 kg/m<sup>2</sup>). (a) FBP reconstructed mediastinal image obtained at initial standard-dose CT (120 kVp; 64 mAs [effective]; DLP, 168 mGy·cm) at the level of the carina shows reference image quality (objective noise at the level of the trachea, 28.8 HU; subjective image noise rated minimal). (b) Follow-up reduced-dose mediastinal CT image reconstructed with the raw data–based iterative technique (100 kVp; 40 mAs [effective]; DLP, 82 mGy·cm) obtained at the same level as a. Note the reduction in objective image noise measured at the level of the trachea (21.2 HU) and the improved CNR (15.61 vs 10.64) and SNR (15.22 vs 11.62). Subjective image noise was similarly rated when compared with T1 images.

raw data–based iterative reconstruction algorithm, standard image quality was attained on all reduced-dose CT angiograms, with a mean DLP of 77.3 mGy·cm and a mean effective dose of 1.31 mSv. These results compare favorably with the average effective dose of 7–8 mSv reported with diagnostic chest

CT (17,18). Because of differences in the average body weight of Western and Eastern populations, we analyzed more specifically the subgroup of patients with a body weight of 70–75 kg, the latter of which corresponds to the upper limit of observed body weights in Asian patients. In this subgroup of patients, it is



Figure 4



**Figure 4:** Comparison of T1 and T2 images obtained 6 months apart in a 61-year-old obese man (BMI, 34.5 kg/m<sup>2</sup>). **(a)** FBP reconstructed mediastinal image obtained at initial standard-dose CT (120 kVp; 114 mAs [effective]; DLP, 325 mGy·cm) at the level of the pulmonary trunk shows reference image quality (objective noise at the level of the trachea, 16.2 HU; subjective image noise rated minimal). **(b)** Follow-up reduced-dose mediastinal CT image reconstructed with the raw data-based iterative technique (100 kVp; 84 mAs [effective]; DLP, 148 mGy·cm) obtained at the same level as **a**. Note the similar objective image noise measured at the level of the trachea (17.2 HU) and the improved CNR (9.90 vs 7.89) and SNR (11.45 vs 9.31). Subjective image noise was rated as moderate.

noteworthy that mean DLP was 60.88 mGy·cm, with a mean effective dose of 1.03 mSv.

When we stratified our study group according to the voltage of the initial examination, results similar to those reported for the overall population were observed in the subgroup of patients initially scanned at 120 kVp and subsequently scanned at 100 kVp. In the second subgroup of patients, in whom tube voltage decreased from 100 kVp to 80 kVp, we also observed significant reduction in objective image noise. However, a significantly higher proportion of reduced-dose studies was rated as having good image quality at the expense of studies initially rated as having excellent image quality. Because patient weight varied between 50 and 80 kg in this subgroup, it is likely that 80-kVp tube voltage was suboptimal to scan the largest patients. However, it is noteworthy that all examinations remained of diagnostic quality, suggesting that iterative reconstruction will enable broader use of low-tube-voltage scanning in routine clinical practice. The benefits of 80-kVp scanning have been particularly emphasized in the literature. A chest phantom study suggested that both small- and medium-sized

patients may benefit from the use of an 80-kVp tube current in pulmonary CT angiography; however, it also predicted a significant increase in image noise in large patients (19). In a clinical study, Szucs-Farkas et al reported that 80-kVp pulmonary CT angiography permits diagnostic image quality in patients weighing up to 100 kg (8).

On low-voltage images reconstructed with the raw data-based iterative technique, we observed significant improvement in SNR and CNR, taking advantage of the reduced level of objective image noise due to the reconstruction algorithm and the increased vascular attenuation at low voltage. Despite methodologic and anatomic differences, our results are comparable with those of Marin et al (20). When Marin et al investigated a different iterative reconstruction algorithm (adaptive statistical iterative reconstruction), they found mean CNR of the abdominal aorta to be significantly higher with a low-voltage protocol and iterative reconstruction. These findings further reinforce the superiority of iterative reconstructions over FBP reconstruction of reduced-dose images. In the latter situation, an identical SNR and CNR are reported owing to the coincidence of increasing background

noise and increasing pulmonary vessel enhancement (4,5,8). Applied to clinical situations in which adequate vessel enhancement is a crucial point, such as for the depiction of acute pulmonary embolism, one would expect a major clinical usefulness of reduced-dose scanning combined with iterative reconstruction. Several studies have already shown that readers delineated pulmonary arteries to a more distal ramification at the lower tube voltage (4,8,10). An additional advantage of low-voltage scanning should also be considered, namely the possibility to reduce the iodine load. By using 80 kVp in pulmonary CT angiography, Szucs-Farkas et al (8) reduced contrast medium volume by 25% compared with that at 100 kVp, without deterioration of image quality. In regard to the scanning protocol used in the present study, one should emphasize that with a pitch factor of 3 and the corresponding high table speed, tube current modulation cannot react as quickly to changing anatomy as it can with lower pitch values. As a consequence, transitions from high-dose to reduced-dose regions (from the shoulder to the lungs) and from reduced-dose to high-dose regions (from the lung to the liver in case of craniocaudal scan direction) may not be represented exactly in the dose profile and may be smoothed. For the entire scanning range, however, this effect is expected to average out to a certain extent. Thus, we assume that similar scanning parameters but one CT source would yield similar averaged CTDI and DLP. Even if the full level of potential dose reduction might not have been reached with the high-pitch spiral mode, our results will not be affected because we used the same scan mode for all our comparisons.

There were some limitations to our study. First, we did not have strict similarities in the patients' status (body weight, chest abnormalities, or both) at the time of initial or follow-up examinations, which may have influenced image noise. However, we paid attention to these potential biases when we selected our population, and we considered this approach more appropriate than comparing distinct groups of patients. Second, we did not attempt to evaluate the influence of iterative reconstructions on

lesion conspicuity between the initial and follow-up CT examination. This was imposed by our study design, in which any change in the morphologic aspect of individual small-sized lesions could have also been attributable to changes over time. Because iterative reconstructions can yield diagnostic image quality, it is mandatory to investigate the influence of dose settings on the visualization of thoracic changes to provide potential users with a complete evaluation of these reconstruction algorithms. Third, we considered 120-kVp scanning the standard protocol, as it represents the most widely used protocol in our daily routine, and we did not test the 140-kVp protocol. Because the latter is often used in other countries, this might be a limitation. Fourth, qualitative analysis of image quality was obtained by consensus between two readers; thus, we did not include inter- and intraobserver variability. Fifth, although the readers independently analyzed the two series of lung and mediastinal images, they were not blinded to acquisition and reconstruction parameters. This was the consequence of the different visual appearance of iterative reconstructions, which made them easily recognizable from FBP reconstructions. It is noticeable that we did not find any significant difference in the  $\kappa$  values of T1 and T2. The clinical implementation of iterative reconstructions in routine clinical practice requires significant hardware efforts to avoid excessive image reconstruction times. Since the time of this study, a commercially available product has been installed in our CT system, enabling reconstruction of images in real time.

In conclusion, the raw data-based iterative reconstruction technique yielded low-voltage CT angiograms of equivalent image quality when compared with standard-dose FBP CT images, with a 50% dose reduction and an average DLP of less than 80 mGy·cm in the studied population.

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