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FULL PAPER

Contrast enhancement in abdominal computed tomography: influence of photon energy of different scanners

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Objective: Different CT scanners have different X-ray spectra and photon energies indicating that contrast enhancement vary among scanners. However, this issue has not been fully validated; therefore, we performed phantom and clinical studies to assess this difference.

Methods: Two scanners were used: scanner-A and scanner-B. In the phantom study, we compared the contrast enhancement between the scanners at tube voltage peaks of 80, 100 and 120 kVp. Then, we calculated the effective energies of the two CT scanners. In the clinical study, 40 patients underwent abdominal scanning with scanner-A and another 40 patients with scanner-B, with each group using the same scanning protocol. The contrast enhancement of abdominal organs was assessed quantitatively (based on the absolute difference between the attenuation of unenhanced scans and contrast-enhanced scans) and qualitatively. A two-tailed independent Student's *t*-test and or the Mann-Whitney *U* test were used to compare the discrepancies.

Results: In the phantom study, contrast enhancement for scanner-B was 36.9, 32.6 and 30.8% higher than that for scanner-A at 80, 100 and 120 kVp, respectively. The effective energies were higher for scanner-A than for scanner-B. In the quantitative analysis for the clinical study, scanner-B yielded significantly better contrast enhancement of the hepatic parenchyma, pancreas, kidney, portal vein and inferior vena cava compared with that of scanner-A. The mean visual scores for contrast enhancement were also significantly higher on images obtained by scanner-B than those by scanner-A.

Conclusion: There were significant differences in contrast enhancement of the abdominal organs between the compared CT scanners from two different vendors even at the same scanning and contrast parameters.

Advances in knowledge: Awareness of the impact of different X-ray energies on the resultant attenuation of contrast material is important when interpreting clinical CT images.

INTRODUCTION

Contrast enhancement in CT is affected by numerous interacting factors¹ that may be divided into three main categories: patient factors (e.g. body size, age, gender, cardiac output), contrast-material factors (e.g. contrast concentration, contrast volume, injection duration, flow rate) and CT-imaging factors (e.g. scan timing, tube voltage, reconstruction methods). In addition, the X-ray photon energy used in CT scanners also affects contrast enhancement.

Knowledge of the X-ray spectrum is important for diagnostic CT and its applications, including contrast-enhanced

scanning, low tube voltage scanning and dual-energy analysis. Iodinated contrast material (CM) attenuation increases as the tube voltage decreases because the photon energy in the X-ray beam gets closer to the iodine *k*-edge (33 keV).² This increases photoelectric effect and decreases Compton scattering that translates into a higher iodine attenuation value.

Different clinical CT scanners have different X-ray spectra and photon energies, even at the same peak tube voltage (kVp) settings. The differences in photon energies between CT scanners can influence CM enhancement, with Cheng

et al³ finding that contrast enhancement was different in different CT scanners in a phantom study. In clinical practice, CM enhancement would preferably be equivalent for the different available CT scanners, which would allow meaningful comparisons. Recently, a phantom study conducted by Taguchi et al showed that a 320-row scanner yielded the strongest contrast enhancement among CT scanners from different vendors.⁴ However, the differences in photon energy and contrast enhancement among different CT scanners have not been fully validated in clinical settings.

In this study, we aimed to compare the contrast enhancement achieved with different CT scanners from different vendors during phantom and clinical studies of the abdomen.

METHODS AND MATERIALS

Phantom study

Initially, two phantom studies were performed using either scanner-A (Brilliance-64; Philips Healthcare, Cleveland, OH) or scanner-B (Aquilion ONE ViSION, Toshiba Medical Systems, Otawara, Japan). These studies aimed to assess the iodine contrast enhancement and the effective energies between the two scanners.

Assessment of contrast enhancement on different CT scanners

Six plastic test tubes (tube material, polypropylene; 12 mm diameter, 100 mm long, 1 mm wall thickness) were filled with varying concentrations of iodinated solution (3, 6, 10, 12, 15 and 30 mgI ml⁻¹) and one tube was filled with purified water only. The solutions were prepared with iterative dilutions using a formulated concentrate of CM (Iopamiron 300; Bayer Healthcare, Bayer Healthcare, Osaka, Japan). Each test tube was inserted separately into the cavity in the centre of a 32-cm-diameter cylindrical plastic phantom (model 20CT14; RadCal, Monrovia, CA) located in the centre of the gantry and scanned at 80, 100 and 120 kVp. The phantom was examined at these three voltage peak settings with a constant radiation dose (CT dose index volume [CTDI_{vol}] = 12.8 mGy), and projection data were reconstructed by filtered back projection. Using a cursor with a 1 cm² circular region of interest (ROI), the CT attenuation of each tube was measured from images obtained at the three peak-voltage settings with the two scanners. The measurements were made on five slice images, and the mean of the five mean values for five slice images was taken as the CT attenuation at the given setting.

Measurement of the effective energies on different CT scanners

The effective energies of the two CT scanners were measured using the aluminum half-value layer method with the lead shielding technique, which do not require suspending the X-ray tube rotation.^{5,6} The pencil ionization chamber, covered by a 4-mm thick lead case (the aperture located at the upper surface), was positioned at the isocentre. Aluminium filters were placed 15 cm above the aperture of the lead-covered case (Figure 1).⁵ Exposure doses were then measured by using the following exposure conditions: 80, 100 or 120 kVp tube voltage; 300 mA tube current, 1.0 s gantry rotation time, a detector configuration of 64 × 0.625 mm for scanner-A and 80 × 0.5 mm for scanner-B, a large beam-shaping filter and a large focal spot size (Table 1).

Figure 1. Aluminium half-value layer method with lead-shielding technique, which does not require suspending rotation of the X-ray tube. The pencil ionization chamber, covered by a 4-mm thick lead case (the aperture located at the upper surface), is positioned at the isocentre. Aluminium filters are placed 15 cm above the aperture of the lead-covered case.

Effective energies were estimated from the half-value layer and the linear-attenuation coefficient of aluminium at 80, 100 and 120 kVp by using the basic data of Seltzer and Hubbell from the National Institute of Standards and Technology.⁷

Clinical study

This retrospective study was approved by the institutional review board; patient informed consent was waived.

Patients

Between July and October 2015, 80 patients (18 females, 62 males; mean age, 69.1 ± 10.9 years; range, 20–89 years) who underwent hepatic multiphasic CT were included in our study. The exclusion criteria for contrast-enhanced CT were history of an allergic reaction to iodinated CM, severe renal insufficiency

Table 1. Scanning parameters of phantom study for assessing effective energies

	Scanner-A	Scanner-B
Detector collimation (mm)	64 × 0.625	80 × 0.5
Section/interval thickness (mm)	5.0/5.0	5.0/5.0
Gantry rotation time (ms)	500	500
Tube voltage (kVp)	80, 100 and 120	80, 100 and 120
Tube current (mA)	300	300
Beam pitch	0.769	0.813
Image reconstruction	FBP	FBP
Reconstruction kernel	B	FC03
Bowtie filter	Large	Large
FOV (cm)	30	30

FBP, filtered back projection; FOV, field of view.

Table 2. Patient demographics

	Scanner-A (n = 40)	Scanner-B (n = 40)	p-value
Sex (male/female)	31/9	31/9	1.00
Age (years)	69.6 ± 10.4	68.7 ± 11.5	0.72
Body height (cm)	161.9 ± 9.8	160.8 ± 7.8	0.58
Body weight (kg)	61.3 ± 10.5	58.6 ± 8.7	0.21
BMI (kg m ⁻²)	23.3 ± 2.9	22.6 ± 2.9	0.30
eGFR (ml per min per 1.73 m ²)	75.3 ± 11.9	71.6 ± 12.5	0.18

BMI, body mass index; eGFR, estimated glomerular filtration rate.

Note—data are the mean ± SD.

(estimated glomerular filtration rate, <30 mL/min/1.73 m²), proved or suspected pregnancy and an unstable clinical condition. We then collected the patients who underwent scans with one of the two scanners in a 1:1 ratio; 40 patients each were examined with a scanner-A and scanner-B. Each individual patient underwent only a single CT scan by a single scanner. In the group scanned by scanner-A, the mean weight, height and body mass index were 61.3 ± 10.5 kg (range, 40.9–87.3 kg), 161.9 ± 9.8 cm (range, 143–180 cm) and 23.3 ± 2.9 kg m⁻² (range, 18.3–30.9 kg m⁻²), respectively; the corresponding values in the group scanned by scanner-B were 58.6 ± 8.7 kg (range, 40–74.2 kg), 160.8 ± 7.8 cm (range, 143–177 cm) and 22.6 ± 2.9 kg m⁻² (range, 17.6–27.9 kg m⁻²). There were no significant differences in the sex distribution ($p = 1.00$), age ($p = 0.72$), height ($p = 0.58$), body weight ($p = 0.21$) or body mass index ($p = 0.30$) between the groups (Table 2). The final diagnosis was hepatocellular carcinoma ($n = 62$; post-treatment, $n = 58$), intrahepatic cholangiocellular carcinoma ($n = 3$; post-treatment, $n = 2$), liver cirrhosis ($n = 4$), metastatic liver tumour ($n = 2$; post-treatment, $n = 2$) and liver transplant donor ($n = 2$), together with hepatic haemangioma, liver cysts, liver abscess, primary biliary cirrhosis, focal nodular hyperplasia, biliary cystadenocarcinoma and autoimmune hepatitis ($n = 1$ each).

CT scanning and CM infusion protocols

The scan parameters for scanner-A were as follows: detector configuration, 64 × 0.625 mm (detector collimation); gantry rotation time, 0.5 s; helical pitch (beam pitch), 0.769; and tube voltage, 120 kVp. The tube potential and tube current were determined by automatic exposure control (Dose Right; Philips Healthcare, Cleveland, OH) on the basis of X-ray attenuation on anteroposterior and lateral scout images, with a reference tube current time product of 285 mAs (effective). The parameters for scanner-B were: detector configuration, 80 × 0.5 mm; gantry rotation time, 0.5 s; and helical pitch, 0.813; and tube voltage, 120 kVp. An automatic tube current modulation program (SURE Exposure 3D, Toshiba Medical Systems, Otawara, Japan) was used, and the noise index (one SD of the regional CT radiodensity value) was set at 10 Hounsfield units (HUs) for images at 5.0 mm slice thicknesses. For both scanners, the standard CM dose tailored to patient's total body weight (600 mgI kg⁻¹) was delivered over 30 s followed by flushing with 30 ml of a saline solution

at the same rate as the CM. In all examinations, we delivered iohexol (Omnipaque 300; Daiichi-Sankyo, Tokyo, Japan) or iopamidol (Iopamiron 300 or 370; Bayer Healthcare, Osaka, Japan) via a 20 G catheter inserted into an antecubital vein with a dual-head power injector (Dual Shot GX V; Nemoto Kyorindo, Tokyo, Japan). To minimize the influence of the difference in the CM concentration, an injection protocol that applied a fixed fractional dose set at 20 mgI kg⁻¹ s⁻¹ was used.

All helical scans were started at the top of the liver and proceeded in a cephalocaudal direction to acquire unenhanced and three-phase contrast-enhanced scans of the entire liver. During scanning, the patients were instructed to hold their breath at the end of inspiration, and three-phase contrast-enhanced CT images of the liver during hepatic arterial-, portal-venous- and equilibrium phases (AP, PVP and EP, respectively) were acquired. An automatic bolus-tracking method was used to trigger the begin of scanning for each phase. Monitoring ROI cursor (1.0–2.0 cm²) was placed in the abdominal aorta at the L1 vertebral body level. Real-time serial monitoring began 10 s after starting of CM injection. The trigger threshold was set at 150 HU, and the AP, PVP and EP scans were started at 18, 55 and 160 s later, respectively.

Image reconstruction was in a 30–35 cm field-of-view display, which was dependent on the patient's physique. The reconstruction section thickness and the section interval were 5.0 mm. CT images were reconstructed with hybrid-type iterative reconstruction algorithms in both scanners, as follows: iDose⁴ (Philips Healthcare, Cleveland, OH) in “level 3” mode with kernel C for scanner-A; and AIDR-3D (Toshiba Medical Systems, Otawara, Japan) in “weak” mode with kernel FC03 for the scanner-B. The image acquisition parameters for each scanner is summarized in Table 3.

Table 3. Imaging- and CM parameters of the two scanners

	Scanner-A	Scanner-B
Detector collimation (mm)	64 × 0.625	80 × 0.5
Section/interval thickness (mm)	5.0/5.0	5.0/5.0
Gantry rotation time (ms)	500	500
Tube voltage (kVp)	120	120
Tube current (mA)	AEC ^a	AEC ^b
Beam pitch	0.769	0.813
Image reconstruction	iDose ⁴ (level 3)	AIDR-3D (weak)
Reconstruction kernel	B	FC03
CM dose (mgI kg ⁻¹)	600	600
Injection duration (s)	30	30
Bolus tracking threshold (HU)	150	150
Scan delay (s)	18, 55, 160	18, 55, 160

AEC, automatic exposure control; CM, contrast material; HU, Hounsfield unit.

^aReference tube current time product was set at 285 mAs (effective).

^bThe noise index was set at 10 HU for a section thickness of 5 mm.

Figure 2. Graph showing the CT attenuation of the seven test tubes imaged at 80 kV (a), 100 kV (b) and 120-kV (c) on scanners-A and -B. There is a significant linear correlation between the CT attenuation and the increase in the iodine concentration at all tube voltage settings for both CT scanners ($R^2 = 0.99$, $p < 0.01$). The increase became greater as the tube voltage reduced in both scanners.

Quantitative image analysis

A board-certified radiologist (SO) with 11 years experience reading abdominal CT scans, who was blinded to the scan information, performed the quantitative image analysis of the unenhanced, AP, PVP and EP images. The CT attenuation values of the abdominal aorta, hepatic parenchyma, pancreas, kidney, spleen, portal vein, inferior vena cava (IVC), bilateral psoas muscles and the subcutaneous fatty tissue were measured on the unenhanced CT. Then, the enhancement values of the abdominal aorta were measured in the AP images together with the values for the hepatic parenchyma, pancreas, kidney, spleen and portal vein in the PVP images. Enhancement of the hepatic parenchyma and IVC were also determined in the EP images. The values obtained in the unenhanced, AP, PVP and EP images were measured by placing a manually defined circular ROI. Contrast enhancements of the abdominal aorta, hepatic parenchyma, pancreas, kidney, spleen, portal vein and IVC were calculated for each enhancement phase as the absolute difference between the attenuation from unenhanced scans and the attenuation during each enhancement phase. The attenuation of the hepatic parenchyma was measured with a circular ROI placed at two liver sites in the left and right lobes on images obtained at the level of the main portal vein. To minimize bias from single measurements, the average of the measurements for each ROI was calculated. Also, visible blood vessels, bile ducts and artefacts were carefully

avoided in the ROI measurements of the hepatic parenchyma. The CT attenuation of the pancreas was measured with a circular ROI placed at pancreatic head and body, and the average was calculated. The CT attenuation of the kidney was measured as the average of circular ROIs on images of both kidneys obtained at the level of the renal hilum. In addition, the attenuation (and SD) was measured to evaluate the image noise at the bilateral psoas muscle, then the contrast-to-noise ratios of the hepatic parenchyma during PVP and EP were calculated. We compared these values between the CT scanners.

Qualitative image analysis

Qualitative image analysis was performed with standard abdominal window settings (level and width of 50 and 350 HU, respectively). Two board-certified radiologists (NT and TN with 6 and 19 years experience interpreting abdominal CT scans, respectively) were asked to grade the images independently for contrast enhancement. The following were graded and compared: the abdominal aorta in AP images; the hepatic parenchyma in PVP and EP images; the pancreas, kidney, spleen and portal vein in PVP images; and the IVC in EP images. The CT data sets were randomized and the reader was blinded to the scan settings. A 4-point subjective scale was used to grade contrast enhancement of each site, as follows: 1 = unacceptable enhancement for interpretation; 2 = poor enhancement, limited diagnostic value; 3 =

Table 4. Quantitative assessment of image quality

	Scanner-A	Scanner-B	<i>p</i> -value
CT attenuation_unenhanced images			
Abdominal aorta (HU)	41.6 ± 3.9	42.8 ± 4.3	0.21
Hepatic parenchyma (HU)	60.8 ± 5.2	59.7 ± 6.0	0.35
Pancreas (HU)	44.0 ± 7.3	44.3 ± 5.0	0.81
Kidney (HU)	35.4 ± 1.7	34.6 ± 2.2	0.05
Spleen (HU)	50.4 ± 3.2	49.4 ± 3.8	0.21
Portal vein (HU)	40.8 ± 3.2	41.9 ± 4.5	0.24
IVC (HU)	40.5 ± 4.1	40.1 ± 4.4	0.68
Muscle (HU)	53.0 ± 3.9	51.3 ± 5.8	0.12
Subcutaneous fatty tissue (HU)	-98.1 ± 9.6	-103.4 ± 6.8	<0.01
Contrast enhancement_AP images			
Abdominal aorta (HU)	316.9 ± 47.3	335.0 ± 50.2	0.10
Contrast enhancement_PVP images			
Hepatic parenchyma (HU)	55.9 ± 9.6	62.9 ± 9.1	<0.01
Pancreas (HU)	58.4 ± 12.1	63.7 ± 11.8	0.04
Kidney (HU)	158.7 ± 23.7	169.5 ± 19.7	0.03
Spleen (HU)	81.2 ± 11.5	86.0 ± 11.9	0.07
Portal vein (HU)	137.8 ± 19.2	147.7 ± 17.8	0.02
Contrast enhancement_EP images			
Hepatic parenchyma (HU)	36.3 ± 5.4	42.2 ± 6.9	<0.01
IVC (HU)	68.1 ± 8.5	76.5 ± 10.1	<0.01
Contrast-to-noise ratio			
Hepatic parenchyma (PVP)	6.2 ± 1.6	6.4 ± 1.4	0.52
Hepatic parenchyma (EP)	4.0 ± 0.8	4.3 ± 0.9	0.15

AP, arterial phase; EP, equilibrium phase; HU, Hounsfield unit; PVP, portal venous phase.

Note—data are the mean ± SD.

acceptable enhancement, not interfering with the interpretation and 4 = excellent enhancement.

Estimation of the radiation dose

Based on each patient's examination information, the volume CTDI_{vol} (in mGy) for each scan phase was recorded.

Statistical analysis

Quantitative data are expressed as the mean ± SD. All qualitative and quantitative image parameters obtained with scanners-A and B were compared, and the CTDI_{vol} values were also assessed for each scan phase. Differences in the mean values between the two CT scanners were determined by two-tailed independent Student's *t*-test and or the Mann–Whitney *U* test by whether the data were normally or non-normally distributed, respectively. The degree of agreement in the visual evaluation between the two observers was measured with the κ statistic, where a value of 0 indicated no agreement, 0 to 0.20 indicated poor agreement, 0.21 to 0.40 indicated fair agreement, 0.41 to 0.60 indicated moderate agreement, 0.61 to 0.80 indicated substantial agreement and 0.81 to 0.99 indicated almost perfect agreement. A *p*-value < 0.05

was considered to indicate a statistically significant difference. We used JMP v. 9.0.2 (SAS Institute, Cary, NC) software for the statistical analyses.

RESULTS

Phantom study

Contrast enhancement on different CT scanners

There was a significant linear correlation between the mean CT attenuation and the increase in the iodine concentration at all tube voltage settings for both CT scanners ($R^2 = 0.99$, $p < 0.01$) (Figure 2), and this increase was greater at lower tube voltages. At 80, 100 and 120 kVp, the contrast attenuations were 36.9, 32.6 and 30.8% higher, respectively, for scanner-B than for scanner-A.

Effective energies on different CT scanners

For scanner-A, the effective energies were 49.8, 60.8 and 62.4 keV at 80, 100 and 120 kVp, respectively. The corresponding energies for scanner-B were 43.1, 48.6 and 54.4 keV, respectively. The effective energies were, therefore, higher for scanner-A than for scanner-B at all tube voltage settings.

Figure 3. Box-and-whisker plots for the quantitative CT attenuation results on unenhanced images obtained with the 120-kVp protocol. The medians, interquartile ranges and extremes are shown. There are no significant differences in the CT attenuation at any measurement site, except for the subcutaneous fatty tissue wherein attenuation is significantly higher for scanner-A than for scanner-B ($p < 0.01$). HU, Hounsfield unit; IVC, inferior vena cava.

Clinical study

Quantitative image quality analysis

The results of our quantitative assessment of image quality are shown in Table 4. In the unenhanced images, there were no significant differences in CT attenuation at any measurement site, except for the subcutaneous fatty tissue, where the CT attenuation was significantly higher on scanner-A ($p < 0.01$) (Figure 3). The difference in contrast enhancement of the abdominal aorta on AP images between the two CT scanners was not significant ($p = 0.10$) (Figure 4). Contrast enhancement of the hepatic parenchyma, pancreas, kidney and portal vein was significantly higher on the PVP images for scanner-B than on those for scanner-A,

but the difference was not significant for the spleen (Figure 5). Contrast enhancement of the hepatic parenchyma and the IVC on EP images was significantly higher with scanner-B than with scanner-A (Figure 6).

Qualitative image analysis

The results of our qualitative assessment of image quality are shown in Table 5. The mean visual scores for contrast enhancement of the kidney and the spleen on PVP images, and for contrast enhancement of the hepatic parenchyma on EP images, were significantly higher on scanner-B ($p < 0.01$). There were

Figure 4. Box-and-whisker plots for the quantitative contrast enhancement results on AP images obtained with the 120-kVp protocol. The medians, interquartile ranges and extremes are shown. There are no significant differences in the contrast enhancement of the abdominal aorta between the two CT scanners ($p = 0.10$). AP, arterial phase; HU, Hounsfield unit.

Figure 5. Box-and-whisker plots for the quantitative contrast enhancement results on PVP images obtained with the 120-kVp protocol. The medians, interquartile ranges and extremes are shown. Contrast enhancement of the hepatic parenchyma, pancreas, kidney and portal vein is significantly higher for scanner-B than for scanner-A. HU, Hounsfield unit; PVP, portal-venous phase.

Figure 6. Box-and-whisker plots for the quantitative contrast enhancement results on EP images obtained with the 120-kVp protocol. The medians, interquartile ranges and extremes are shown. Contrast enhancement of the hepatic parenchyma and the IVC is significantly higher for scanner-B than for scanner-A ($p < 0.01$). EP, equilibrium phase; HU, Hounsfield unit; IVC, inferior vena cava.

no significant differences in the visual scores for the remaining organs. Interobserver agreement for visual grading at each evaluated site was substantial (κ -value = 0.66–0.79). Representative cases are shown in Figure 7.

Radiation dose

Table 6 shows the CTDI_{vol} data. The mean CTDI_{vol} without contrast was significantly higher for scanner-A than for scanner-B ($p < 0.01$). But, there were no significant differences in the mean CTDI_{vol} in the AP, PVP or EP images.

DISCUSSION

The results of the phantom study show that contrast enhancement for scanner-B was more than 30% higher than that for scanner-A, and that the effective energies were higher for scanner-A than for scanner-B. In the clinical study, scanner-B yielded

slightly higher contrast enhancement than scanner-A in the quantitative analysis, with significant differences in the hepatic parenchyma, the pancreas, the kidney, the portal vein and the IVC. This was supported by the mean visual scores for contrast enhancement, which were higher on images obtained by scanner-B than for those obtained by scanner-A. Our results suggest that the contrast enhancement of abdominal organs was different among the CT scanners with different photon energies, even when the same scan and contrast parameters were used.

The typical X-ray tube voltage used in clinical CT examinations is 120 kVp (*i.e.* peak tube voltage of the X-ray spectrum). However, different CT scanners have different shapes of X-ray spectrum, even when the same 120 kVp is used; this results in different contrast enhancements among the CT scanners. To our knowledge, this is the first work to validate the difference in contrast enhancement of different CT scanners with different X-ray photon energies in clinical settings.

Variations among clinical CT scanners in attenuation for the same material, location and orientation were first reported more than 30 years ago.^{8,9} In recent years, there have been notable advances in CT technology that have had, or are expected to have, a significant clinical impact; these include extreme multidetector, high-resolution and iterative reconstruction techniques. The question arises, however, as to how these modern CT techniques might reduce the inconsistencies in CT attenuation.^{8,9} Using tissue characterization phantoms, Cheng et al³ compared the variation in CT attenuation among 18 clinical CT scanners from 5 different vendors. They reported that the minimum attenuation tended to be 7–10% lower than the maximum attenuation for all substitute tissue materials and all CT scanners. Indeed, even for CT scanners from the same vendor, it is possible that the CT attenuation could vary. Many factors may contribute to the variation seen in CT attenuation, including the X-ray spectrum of the CT scanners, variation in the material composition of the subject, beam filtration (*e.g.* bowtie filter), scatter conditions, anti-scatter grids and data processing (*e.g.* beam-hardening corrections and reconstruction algorithms). Taguchi et al performed a phantom study using four modern CT scanners (Aquilion ONE ViSION Edition, Toshiba Medical Systems, Otawara, Japan; Light Speed VCT, GE Healthcare, Milwaukee, WI; Brilliance 64, Philips Healthcare, Cleveland, OH; and SOMATOM Definition AS+, Siemens Healthcare, Forchheim, Germany) and showed significant variations in CT attenuation.⁴ These phantom findings are concordant with our clinical results.

Generally, for hepatic multiphasic CT, an iodinated CM dose of about 600 mgI kg⁻¹ is required to obtain adequate hepatic contrast enhancement.^{10,11} A recent study by Awai et al¹² showed that lean body weight had the strongest correlation with aortic and hepatic enhancement, and indicated that it was the best index of body size for determining the appropriate CM dose for hepatic multiphasic CT. They reported that to achieve an aortic enhancement of 280 HU¹³ and a hepatic enhancement of 50 HU,¹⁴ the required iodine doses were 687 and 701 mgI kg⁻¹ lean body weight, respectively, for hepatic multiphasic CT; the corresponding values were 679 mgI kg⁻¹ for

Table 5. Qualitative assessment of image quality (visual scores for contrast enhancement)

	Scanner-A	Scanner-B	<i>p</i> -value
Abdominal aorta (AP)	4.0 ± 0.0	4.0 ± 0.0	1.00
Hepatic parenchyma (PVP)	3.6 ± 0.5	3.8 ± 0.4	0.12
Pancreas (PVP)	3.1 ± 0.4	3.2 ± 0.6	0.31
Kidney (PVP)	3.3 ± 0.5	3.6 ± 0.5	<0.01
Spleen (PVP)	3.3 ± 0.5	3.6 ± 0.6	0.04
Portal vein (PVP)	3.3 ± 0.5	3.5 ± 0.5	0.11
Hepatic parenchyma (EP)	3.5 ± 0.5	3.8 ± 0.4	0.02
IVC (EP)	3.2 ± 0.5	3.3 ± 0.6	0.34

AP, arterial phase; EP, equilibrium phase; IVC, inferior vena cava; PVP, portal venous phase.

Note—data are the mean ± SD.

Figure 7. Representative cases of patients scanned with scanners-A and -B obtained with the 120-kVp protocol. The images in a–d are for a 55-year-old male weighing 62.7 kg (BMI, 21.7 kg m⁻²) who was scanned with scanner-A, whereas those in e–h are for a 60-year-old male weighing 60.2 kg (BMI 21.4 kg m⁻²) who was scanned with scanner-B. Representative transverse CT images in the unenhanced phase (a, e), AP (b, f), PVP (c, g) and EP (d, h) are shown (window level 50 HU and width 350 HU). Scanner-B yielded slightly higher contrast enhancement of abdominal organs than scanner-A. AP, arterial phase; BMI, body mass index; EP, equilibrium phases; HU, Hounsfield unit; PVP, portal-venous phase.

males and 761 mgI kg⁻¹ for females. In terms of uniformity and reproducibility of contrast-enhanced CT examination, it may be necessary to optimize the iodine doses by the photon energy of each CT scanner.

Recent innovations in CT hardware and software have allowed implementation of imaging with low tube voltages (*e.g.* 80 or 100 kVp) in clinical CT protocols.¹⁵ Such low tube voltages yield higher contrast enhancements than standard 120 kVp techniques because the X-ray output energy is closer to the *k*-edge of iodine (*i.e.* 33 keV).¹⁶ Indeed, compared with 120 kVp tube voltages, scanning with low tube voltage of 100 and 80 kVp may increase contrast enhancement by 25 and 65%, respectively.¹⁷ As the difference in iodine attenuation between the two scanners increased as the tube voltage decreased in our phantom study, we should recognize, especially when interpreting the low tube voltage images, that there could be significant differences in contrast enhancement of the organs among different CT scanners, even when using the same scan parameters. Adjustment of the CM dose may be needed for each scanner when scanning with low tube voltages.

Table 6. Radiation exposure

	Scanner-A	Scanner-B	<i>p</i> -value
Mean CTDI _{vol} (mGy)			
Unenhanced scan	15.4 ± 2.9	13.4 ± 3.3	<0.01
Arterial phase	13.7 ± 2.4	13.5 ± 3.3	0.70
Portal venous phase	13.7 ± 1.7	13.7 ± 3.2	0.96
Equilibrium phase	12.1 ± 2.1	12.7 ± 2.8	0.34

CTDI_{vol}, volume CT dose index.

Note—data are the mean ± SD.

Dual-energy CT imaging has recently become available on clinical scanners. With this method, the image data from two energy levels—high and low polychromatic X-ray energy—are merged using appropriate algorithms that exploit the energy-dependent differences in the materials found in the data sets for the two tube voltage scans. With dual-energy scans, virtual monochromatic spectral images can be reconstructed from a pair of material density images and mass attenuation coefficients can be produced.^{18,19} These images depict an object as if it were imaged with a monochromatic beam at the intended X-ray energy.^{18,19} Using the same virtual monochromatic X-ray energy images in dual-energy CT may mitigate the variations in CT attenuation among different CT scanners. This concern about virtual monochromatic imaging requires further study.

This study has some limitations. First, we do not evaluate how the results affect diagnostic performance. Instead, we focus on comparing the actual difference in contrast enhancement between the two CT scanners. Second, we evaluated only contrast enhancement, and did not assess other image quality parameters (*e.g.* image noise and contrast-noise ratio). Although the CT scan protocols were almost the same for the two scanners, there were differences in the automatic tube current modulation program, the reconstruction algorithms and the kernels between the two scanners, and these could not be controlled because they were produced by different vendors. Therefore, comparison of image quality was difficult, as was the comparison of X-ray exposure dose, although we evaluated the CTDI_{vol} data. Generally, the differences in photon energy of different CT scanners are considered to affect the radiation dose given to patients (*i.e.* effective dose per unit dose-length product conversion factors). However, it has been reported that intervendor differences in the effective dose per unit dose-length product for standard scans are only very small.²⁰ Third, a slight difference probably exists in scan timing between the

scanners because the scanners used an automatic bolus-tracking program, which adopts the same trigger threshold (in HU), despite the fact that the scanners used different X-ray photon energies. This difference constitutes a potential confounding factor. Fourth, because we compared the CT data from different patient groups, patient-related factors such as cardiac output may influence the results. Finally, this study compares only two CT scanners, an Aquilion ONE VISION Edition (Toshiba Medical Systems, Otawara, Japan) and a Brilliance64 (Philips Healthcare, Cleveland, OH), so the applicability of our findings

to other CT scanners is uncertain. Further study is needed to elaborate on this issue.

CONCLUSION

There were slight but significant differences in the contrast enhancement of abdominal organs between the two CT scanners compared from different vendors, even when the same scan and contrast parameters were used. Awareness of this issue, which is caused by the difference in photon energy of the CT scanner, is important when interpreting clinical CT images.

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