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FULL PAPER

Comparative study of hepatic venography using non-linear-blending images, monochromatic images and low-voltage images of dual-energy CT

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Objective: To investigate the use of non-linear-blending and monochromatic dual-energy CT (DECT) images to improve the image quality of hepatic venography.

Methods: 82 patients undergoing abdominal DECT in the portal venous phase were enrolled. For each patient, 31 data sets of monochromatic images and 7 data sets of non-linear-blending images were generated. The data sets of the non-linear-blending and monochromatic images with the best contrast-to-noise ratios (CNRs) for hepatic veins were selected and compared with the images obtained at 80 kVp and a simulated 120 kVp. The subjective image quality of the hepatic veins was evaluated using a four-point scale. The image quality of the hepatic veins was analysed using signal-to-noise ratio (SNR) and CNR values.

Results: The optimal CNR between hepatic veins and the liver was obtained with the non-linear-blending images.

Compared with the other three groups, there were significant differences in the maximum CNR, the SNR, the subjective ratings and the minimum background noise ($p < 0.001$). A comparison of the monochromatic and 80-kVp images revealed that the CNR and subjective ratings were both improved ($p < 0.001$). There was no significant difference in the CNR or subjective ratings between the simulated 120-kVp group and the control group ($p = 0.090$ and 0.053 , respectively).

Conclusion: The non-linear-blending technique for acquiring DECT provided the best image quality for hepatic venography.

Advances in knowledge: DECT can enhance the contrast of hepatic veins and the liver, potentially allowing the wider use of low-dose contrast agents for CT examination of the liver.

CT venography (CTV) is an important non-invasive examination to assess the hepatic veins and plays an important role in the pre-operative evaluation of liver transplants and the diagnosis of hepatic venous diseases.^{1,2} Compared with CT hepatic artery angiography or multiphasic liver CT, CTV often requires a larger dose of the contrast agent to achieve sufficient contrast for filling in the hepatic veins.³ Increasing the contrast agent not only increases the economic burden of the patient but also raises the incidence of side effects and complications related to the contrast agent. One of the goals in the advancement of CT techniques is to continuously improve the image quality and clinical applications while reducing radiation exposure and promoting the reasonable use of contrast agents. Several studies suggest that low tube voltage CTV reduces radiation and improves vascular contrast⁴⁻⁶ because iodinated contrast material is more conspicuous in low-kilovolt peak (kVp) images with an approximately 80% increase in CT attenuation at 80 kVp compared with that at 140 kVp.⁶⁻⁸

Dual-source CT (DSCT) was recently introduced into clinical practice. It can simultaneously acquire low- and high-energy image data using two X-ray tube and detector systems mounted in one gantry.⁹ Dual-energy CT (DECT) could improve the contrast and thereby the image quality of CTV images by virtual monochromatic imaging¹⁰ and non-linear-blending⁸ and linear-blending techniques.^{7,11,12} Studies have shown that a DECT non-linear-blending technique could improve the conspicuity of myocardial delayed enhancement.⁸ The clinical application of DSCT undoubtedly greatly aids the choice of a suitable application from a variety of post-processing techniques that can significantly improve the contrast enhancement of hepatic veins. However, the ability to improve the conspicuity of hepatic veins via a dual-source DECT non-linear-blending technique and the performance of non-linear-blending and monochromatic imaging techniques have not been studied. Therefore, the purpose of our study was to improve the image quality of hepatic venography over single-energy CT

by using DECT virtual monochromatic imaging and a non-linear-blending technique.

METHODS AND MATERIALS

Patients

The instruments and software used in this study were all approved commercial equipment. The study was not supported by any technical or financial sponsor. One coauthor is an employee of Siemens Healthcare, Beijing, China, and all other coauthors are hospital physicians who have no conflicts of interest with Siemens Healthcare or other institutions and claim ownership of all research data and results. This prospective study was approved by the institutional ethics committee, and informed consent forms were obtained from all patients.

From May to October 2013, 90 patients undergoing upper-abdominal contrast-enhanced DECT in the portal venous phase were successively enrolled into this study. The inclusion criteria for all the patients were that the three major branches of the hepatic vein could be evaluated and there was no history of surgical procedures in the liver. Eight patients were excluded for the following conditions: three patients had no contrast agent filling the hepatic veins at the portal vein phase; two patients had a large intrahepatic tumour, making it difficult to assess the intrahepatic veins; two patients had liver cirrhosis, resulting in poor display of the hepatic veins and difficulty in assessing them; and one patient was excluded owing to the failure of the contrast agent injection at the antecubital vein and, consequently, insufficient enhancement. Finally, a total of 82 patients were enrolled in this study. The patients were aged 34–83 years, with an average age of 55.14 ± 13.64 years; the average weight was 65.73 ± 12.07 kg (43–96 kg); and there were 45 males and 37 females. Among the 82 patients, 20 had cirrhosis (including liver cancer), 37 needed an enhanced liver CT scan before and/or after the treatment of a malignant tumour in other parts of the body, 9 had hepatic haemangioma, 6 had fatty liver, 3 had gallstones, 2 had biliary obstruction and 5 had liver cysts. To set up the control group with single-energy liver CT angiography, we used the same inclusion and exclusion criteria used for the experimental group to enrol 45 patients into the control group for routine single-energy CT scans at the portal venous phase. The control group patients were aged 45–85 years, with an average age of 53.65 ± 14.51 years. The average weight was 63.17 ± 15.34 kg (45–91 kg), and there were 28 males and 17 females. The body mass index (BMI) was calculated as follows:

$$\text{BMI (male and females)} = \frac{\text{Weight (kg)}}{[\text{Height (m)}]^2}$$

CT protocol

All CT scans were performed on a 128-slice DECT scanner (Somatom® Definition Flash; Siemens Medical Systems, Forchheim, Germany). The unenhanced liver scan and enhanced scans at the arterial and portal venous phases of the control group were performed using a standard-of-care scan protocol of our hospital. For the DECT group, the unenhanced and arterial phases

were performed using the same standard-of-care protocol as the control group, whereas the portal venous phase was performed using the dual-energy scan protocol. The standard-of-care and dual-energy scan protocols at the portal venous phase are described in Table 1. For both the DECT and control groups, the arterial and portal vein phases were acquired 30 and 70 s after beginning the injection of the contrast material. All the scans used the automatic tube current modulation. For all patients, the slice thickness of the reconstruction was 1 mm with a 1-mm interslice distance. Iopromide (300 mg I ml^{-1} of Ultravist®; Schering Pharmaceutical Co. Ltd, Guangzhou, China) was injected at a rate of 3 ml s^{-1} via a 22-gauge intravenous catheter into an antecubital vein. If the patient weighed <75 kg, 90 ml of contrast material was administered and 110 ml was administered if the patient weighed >75 kg.

Data processing of the dual-energy CT images

The linear-blended images were reconstructed on the CT console using the dual-energy venous-phase data set with the vendor default weighting factor of 0.5. The image quality of this default weighting factor was considered as an equivalent to a standard 120-kVp image. Therefore, the linear-blended image was considered to be a simulated 120-kVp image and served as a reference for inpatient comparison.

All of the dual-energy images were sent to a commercial workstation for the calculation of the monochromatic and non-linear-blending images (MultiModality Workplace; Siemens Healthcare). For each patient, 31-keV levels of the virtual monochromatic spectral (VMS) imaging sequences were calculated from 50 to 80 keV with 1-keV intervals. The non-linear-blended images were created by choosing the blending widths of 0, 10, 20, 30, 40, 50 and 60 HU and the blending centre of the mean CT value of the hepatic parenchyma and hepatic veins. Therefore, there were seven non-linearly blended image sets for each patient. The principle of the non-linear-blending technique was described previously.¹¹ Briefly, the blending width and centre defined a range of CT values. For a pixel with higher CT values than the defined range, the CT value of the low-kVp image was assigned to the pixel. For a pixel with lower values, a larger high-kVp proportion was blended into the pixel.

There were three steps in the data analysis: first, the optimal VMS level of the hepatic veins of each patient was determined; second, the optimal non-linear-blending image of the hepatic veins of each patient was determined; third, the four image sets, including the optimal-kiloelectron volt (keV), optimal non-linear-blending, 80-kVp and simulated 120-kVp images, were compared to determine the optimal vascular imaging protocol.

Image analysis was conducted as follows: the optimal VMS images for hepatic venography were selected from the monochromatic image sets based on the maximum contrast-to-noise ratio (CNR). The optimal non-linear-blending images were also selected based on the highest CNR. The objective evaluation of the image quality included the signal-to-noise ratio (SNR) and CNR, which were calculated using the following equations:

Table 1. The scan protocols for dual-energy and single-energy CT

Parameter	Dual energy	Standard
Scanning parameters		
Number of X-ray sources	2	1
Tube voltage (kVp)	80/Sn140	120
Quality reference tube current (mAs)	335/135	210
Collimation (mm)	2 × 32 × 0.6	128 × 0.6 by z-flying focus spot
Pitch	0.6	0.9
Rotation time (s)	0.5	0.5
Reconstruction parameters		
Section thickness (mm)	1	1
Reconstruction algorithm		
Kernel	B30	B30
Radiation exposure for portal venous phase		
Volume CT dose index (mGy)	11.87 ± 2.63	13.84 ± 3.79
Dose-length product (mGy cm) ^a	265.26 ± 67.71 ^b	360.98 ± 95.27
Effective dose (mSv) ^a	3.99 ± 1.01 ^b	5.41 ± 1.43

^aData are expressed as the mean ± standard deviation.

^bIndicates a statistically significant difference ($t = 6.462$; $p < 0.001$).

$$\text{SNR} = \frac{\text{ROI}_v}{I_N}$$

$$\text{CNR} = \frac{(\text{ROI}_v - \text{ROI}_l)}{I_N}$$

where region of interest (ROI)_l is the mean attenuation value of the liver parenchyma, ROI_v is the mean attenuation value of the hepatic veins and I_N is the image noise. The image noise was defined as the standard deviation of the attenuation value measured in the liver parenchyma. The vascular CT value was measured using the axial images of portal vein phase. In general, the three major branches of the hepatic vein drain into the inferior vena cava. The measured areas were the trunk of the right hepatic vein, the middle hepatic vein and the left hepatic vein at the nearby secondary portal of the liver; the size of the ROI was adjusted according to the vascular diameter. ROI_v was the average of the three values. For ROI_l, the CT attenuation value of liver parenchyma at the scan level of the secondary portal of the liver was measured, three ROIs at the same scan level were selected and a circle encompassing at least 50 pixels was drawn in each ROI (Figure 1). The average of the three values was ROI_l. The ROIs were selected by a radiologist (FX, who had 5 years' experience). The equation for the CT value of the hepatic vein was as follows:

$$\text{enhancement value} = \text{post-contrast attenuation} \\ - \text{pre-contrast attenuation.}$$

The images of the portal vein phase of the control group and optimal-keV images, optimal non-linear-blending images, 80-kVp images and simulated 120-kVp images (reconstruction thickness

of 1 mm) of the experimental group were imported into the workstation three-dimensional (3D) software. Maximum-intensity projection (MIP) and volume rendering (VR) were used to visualize the hepatic venous structure. Images of hepatic venography of the experimental group and the control group were assessed subjectively. Image quality assessment included the hepatic vein branches, the noise of the liver parenchyma and the sharpness of

Figure 1. Axial dual-energy CT image at the level of the nearby secondary portal of the liver with the blending width, 400 HU; centre, 80 HU. The mean CT values and standard deviation (Hounsfield unit) of the hepatic veins and the liver parenchyma were measured. The mean CT value of the right hepatic vein, middle hepatic vein and left hepatic vein was 247.3 HU. The mean CT value of the three regions of interest of the liver was 101.4 HU, and the noise was 15.5 HU.

the vascular edge. Because MIP displayed small intrahepatic vessels better than VR, this study used 3D-MIP images for subjective ratings.¹³ A four-point scale was used to describe hepatic venography, in which branch order was defined as first order, second order, third order, fourth order or higher. A score of one was given to bad images displaying only first-order branches with high noise, a score of two for moderate-quality images showing second-order branches with moderate graininess, a score of three for good images showing third-order branches with sharp edges and no graininess and a score of four for excellent images showing fourth-order branches with sharp edges and no graininess. All parameters were objectively analysed and subjectively scored by consensus by two radiologists (QW and WL, with 17 and 10 years' experience in CTV, respectively). The window level and window width of MIP and VR images were adjusted based on the actual situation of each image, and the appropriate window level and window width were defined as those providing the clearest view of all branches of hepatic veins.

Statistical analysis

All analyses were performed using the statistical software SPSS® II for Windows v. 17.0.1 J (SPSS Inc., Chicago, IL). A $p < 0.05$ was considered to indicate a statistically significant difference. The comparison between standard CT and DECT images was conducted using an unpaired Student's *t*-test for continuous variables and the Wilcoxon rank sum test for categorical variables. Analysis of variance was used to compare the differences in CNR, SNR, CT attenuation value of hepatic veins, background noise, optimal non-linear-blending images, optimal monochromatic spectrum images, 80-kVp images and simulated 120-kVp images among the groups. Bonferroni adjustment was used for pairwise comparisons between groups. A non-parametric test (Friedman two-way assessment rank variance test) was used to compare the vascular ratings among experimental groups. Pearson's correlation coefficient *r* was calculated for the association between the CNR and BMI values among the experimental group.

RESULTS

Image quality of monochromatic images

The contrast and image noise decreased when the keV level increased from 50 to 80 keV. Although the maximum CT attenuation value of the hepatic veins (309.16 ± 51.86 HU) was obtained at 50 keV, the noise was also at the highest level (54.01 ± 15.52 HU). Therefore, the mean CNR of all patients reached a maximum at 66–68 keV and decreased as the keV level increased further (Figure 2). We selected the optimal monochromatic image with the best CNR for each individual patient from the monochromatic images of 50–80 keV. The images with the best CNR fell between 60 and 76 keV (Figure 3), and 76.8% (63/82) of them were between 63 and 69 keV. The average of the best CNR for all patients was 3.65 ± 1.52 .

Image quality of non-linearly blended images

All seven data sets of non-linear-blending images of the patients are shown in Table 2. The average CNR value obtained at a bandwidth (BW) of 0, 10, 20 or 30 was significantly higher than the value obtained at a BW of 40, 50 or 60 ($p < 0.001$) (Figure 4). The average CNR value obtained at a BW of 20 was the highest (9.28 ± 3.27), but there was no significant difference in the pairwise comparisons with the values at BWs of 0, 10 and

Figure 2. Graph showing the average contrast-to-noise ratio (CNR) variation of the hepatic veins of all patients in the experimental group. The grey band area of the graph shows that the maximal CNR was reached in the range of 66–68 keV. VMS, virtual monochromatic spectral imaging.

30, suggesting that as the BW value increased, the CT attenuation value of hepatic veins decreased, while the CT attenuation value of the hepatic parenchyma and background noise increased ($p < 0.001$). We selected the optimal non-linearly blended image with the best CNR for each individual patient from seven blending width settings. The average CNR of the optimal non-linear-blending images was 10.04 ± 3.44 . It was observed that 93.9% (77/82) of the optimal non-linear-blending images used a BW value between 0 and 30. Only in one patient was the maximum CNR value obtained at a BW of 60.

Comparison of image quality on VMS, non-linear-blending, 80-kVp and simulated 120-kVp images Table 3 lists the mean enhancement CT values of the hepatic veins; the difference in CT values between the hepatic veins and the

Figure 3. Distribution of maximum contrast-to-noise ratio (CNR) values of monochromatic images of all patients in the experimental group. The y-axis represents the CNR value, and the maximum CNR on virtual monochromatic spectral (VMS) images was distributed in the range of 63–69 keV.

Table 2. Comparison of the image quality parameters of non-blending images of the studied population

Image quality parameter	Blending width (HU)						
	0	10	20 ^a	30	40	50	60
CNR	9.14 ± 3.05	9.23 ± 3.11	9.28 ± 3.27	8.94 ± 3.38	8.47 ± 3.26	8.22 ± 3.19	7.82 ± 2.99
Attenuation at hepatic vein (HU)	213.61 ± 44.65	212.48 ± 44.83	212.18 ± 44.95	211.62 ± 44.98	210.39 ± 45.64	209.39 ± 46.26	207.91 ± 46.78
Attenuation at liver (HU)	101.07 ± 12.27	101.07 ± 11.93	101.52 ± 11.75	101.97 ± 11.89	102.82 ± 11.80	103.38 ± 12.00	104.17 ± 12.53
Noise	12.79 ± 3.27	12.55 ± 3.38	12.54 ± 3.53	12.92 ± 3.52	13.36 ± 3.79	13.55 ± 3.75	13.76 ± 3.75

BW, bandwidth; CNR, contrast-to-noise ratio; HV, hepatic vein.

n = 82.

^aCNR, HV, liver and noise from BW 0 to BW 60 were compared by analysis of variance (*p* < 0.001).

liver parenchyma; the image noise of the liver parenchyma; and the CNR, SNR and subjective image quality scores of the 80-kVp, simulated 120-kVp images, optimal monochromatic and optimal non-linear-blending images. The CNR, SNR and difference in CT value between the hepatic veins and liver parenchyma on enhanced CT scans of non-linear-blending images were all significantly higher than the same values for the other three groups (*p* < 0.001). The CNR value of the simulated 120-kVp image was the lowest, whereas there was no significant difference between the pure 80 kVp and the optimal VMS. There was no significant difference in the mean enhancement of the hepatic vein between the pure 80-kVp and optimal non-linear-blending images. The mean enhancement of the hepatic vein was the lowest in the simulated 120-kVp image. The noise values for the 80-kVp and optimal VMS images were significantly higher than that of the simulated 120-kVp images, and the noise of non-

linear-blending images was the lowest (*p* < 0.001). The SNR value of the non-linear-blending images was the highest, followed by those of the simulated 120-kVp images. There were significant differences in the subjective image quality score of the hepatic vein images (*p* < 0.001). The subjective score of the non-linear-blending images was higher than that of the other three groups. The subjective image quality score for hepatic veins was scored as “Good” (Score 3) or better for the 80-kVp images in 54 patients, the simulated 120-kVp images in 45 patients, the optimal monochromatic images in 55 patients and the optimal non-linear-blending images in 75 patients (Figures 5 and 6). The average BMI of all the patients in the experimental group was $24.38 \pm 3.27 \text{ kg m}^{-2}$. Evaluating the influence of BMI on the CNRs of the four methods for improving hepatic venography revealed that the optimal non-linear-blending images were influenced more strongly than all the other methods (*r* = −0.715), whereas the simulated 120-kVp images had a weaker correlation (*r* = −0.46) with the BMI, and the 80-kVp (*r* = −0.536) and optimal VMS images (*r* = −0.568) had stronger correlations with the BMI than those obtained with the simulated 120-kVp method.

Comparison of image quality on simulated 120-kVp and standard single-energy CT images

The CT values of the hepatic veins and of the liver parenchyma on simulated 120-kVp images were both significantly higher than the corresponding values of the control group (*p* = 0.008 and *p* < 0.001, respectively) (Table 4). The average background noise of the simulated 120-kVp images was also higher than that of the control group (*p* < 0.001). However, there was no significant difference in the CNR or subjective rating between the simulated 120-kVp images and the control group (*p* = 0.09 and 0.053, respectively). A Good (three points) or better subjective rating was given to 54.88% (45/82) of the simulated 120-kVp images, while 68.89% (31/45) of the control group had this rating. The SNR of the control group was slightly but significantly higher than that of the simulated 120-kVp image group (*p* = 0.041).

DISCUSSION

The depiction of the hepatic vein and branches has become increasingly important for liver surgery,^{3,14,15} elevating standards for hepatic CTV image quality. The challenge of hepatic

Figure 4. Distribution of maximum contrast-to-noise ratio (CNR) value of non-linear-blending images of all patients in the experimental group. The y-axis represents the CNR value, and the maximum CNR on virtual monochromatic spectral (VMS) images was distributed in the bandwidth (BW) range of 0–30. For two patients, the maximum CNR values (4.81 and 14.75, respectively) were reached at BW 40; for two patients, the maximum CNR values (11.05 and 12.2, respectively) were reached at BW 50; for one patient, the maximum CNR value (11.13) was reached at BW 60.

Table 3. Comparison of image quality parameters of different hepatic vein (HV) CT angiography techniques in the studied population

Image quality parameter	80 kVp	Simulated 120 kVp	Optimal virtual monochromatic spectral	Optimal non-linear-blending image ^a
CNR	3.67 ± 1.58	3.19 ± 1.41	3.65 ± 1.52	10.04 ± 3.44
SNR	9.17 ± 3.45	10.68 ± 3.6	9.76 ± 3.63	19.32 ± 5.96
Attenuation of HV (HU)	210.06 ± 44.96	151.73 ± 26.48	184.52 ± 34.89	214.94 ± 44.70
Contrast between HVs and liver parenchyma (HU)	85.88 ± 30.36	46.21 ± 17.85	68.18 ± 23.55	113.87 ± 36.15
Noise	25.37 ± 8.18	15.49 ± 4.55	22.76 ± 17.83	11.89 ± 3.29
Subjective image score ^b	2.83 ± 0.95	2.63 ± 0.92	2.86 ± 0.92	3.51 ± 0.69

CNR, contrast-to-noise ratio; SNR, signal-to-noise ratio.

$n = 82$.

^aCNR, SNR, attenuation in HV, the difference in CT value on an enhanced CT scan between hepatic veins and liver parenchyma, and the noise were compared among the four groups by analysis of variance (ANOVA) ($p < 0.001$).

^bFriedman two-way ANOVA by ranks was used to compare the subjective vascular ratings of the four groups ($p < 0.001$).

venography is that the contrast agent concentration in the hepatic vein depends on the circumfluence from the superior mesenteric and splenic veins and the presence of abnormalities

in the liver parenchyma,⁸ requiring higher dosages of the contrast agent for hepatic venography than that for conventional contrast-enhanced liver CT. A study by Ichikawa *et al*³ suggested

Figure 5. Male, 39 years of age, three-dimensional multidetector-row CT maximum-intensity projection (MIP) angiography of hepatic veins in the same anatomical position. Simulated 120-kVp (a) and pure 80-kVp images (b) were reconstructed. Dual-energy monoenergetic $E = 68$ keV (c), dual-energy blended, $c = 150$, $w = 30$ (d), with appropriate adjustment of the window level and window width of each image. The subjective rating of the dual-energy blending and monoenergetic hepatic venous MIP was four points, and the small hepatic venous branches at the liver edge are clearly displayed. The small hepatic venous branches are faint in the simulated 120-kVp images. In the pure 80-kVp images, the small hepatic venous branches are not clearly displayed owing to significant background noise. The two remaining images were rated at three points.

Figure 6. Female, 50 years of age, three-dimensional multidetector-row CT volume rendering (VR) angiography of hepatic veins in the same anatomical position. Simulated 120-kVp (a) and pure 80-kVp images (b) were reconstructed. Dual-energy monoenergetic $E = 69$ keV (c), dual-energy blended, $c = 150$, $w = 10$ (d), with appropriate adjustment of the window level and window width of each image. The difference in CT value on enhanced CT scan of the portal vein phase between hepatic veins and liver parenchyma was 41HU, manifested as difficulty displaying hepatic venous VR images on simulated 120-kVp images clearly. After non-linear-blending, the difference in CT value on enhanced CT scan between hepatic veins and liver parenchyma reached 157 HU, with clear display of the small branches of hepatic veins.

that $\geq 630 \text{ mg I kg}^{-1}$ should be administered to achieve sufficient hepatic vein enhancement. The mean enhancement of the hepatic vein achieved was 177 HU. Most subsegment (third order) and more peripheral branches were displayed clearly. In our study, the routine contrast agent dose for the enhanced liver CT scan was used for hepatic venography. The average contrast agent dose was $415.38 \text{ mg I kg}^{-1}$. After the DECT non-linear-blending technique was applied, the image quality was rated as Good or better for 91.46% of the patients. The mean enhancement of the hepatic vein achieved was 179.09 ± 45.38 HU. Therefore, DECT improved the image quality of hepatic venography using less contrast agent.

Our results showed that 54.87% of the simulated 120-kVp images received a score of Good or higher, primarily because the opacification of small hepatic venous branches was limited due to the small amount of contrast agent. However, accuracy in the precise reproduction of the morphology of the identified veins is very important, with the possibility of including a tunable safety margin around each segmented vessel.¹⁶ Our study showed that, compared with the optimal monochromatic images and pure 80-kVp images, the CNRs of the optimal non-linear-blending images were 175.07% (3.65:10.04) and 173.56% (3.67:10.04) higher, respectively, and the SNRs were 97.95% (9.76:19.32) and 110.69% (9.17:19.32) higher, respectively. These results showed

Table 4. Comparison of image quality parameters of hepatic veins (HVs) on dual-energy simulated 120-kVp and standard CT images

Image quality parameter	Standard 120 kVp	Simulated 120 kVp	<i>p</i> -value ^a
CNR	3.68 ± 1.66	3.19 ± 1.41	0.090
SNR	12.05 ± 3.17	10.68 ± 3.60	0.041
Attenuation on HV	140.26 ± 18.82	151.73 ± 26.48	0.008
Liver parenchyma	97.63 ± 30.36	105.52 ± 12.55	<0.001
Noise	12.26 ± 2.99	15.49 ± 4.55	<0.001
Subjective image score	2.98 ± 0.87	2.63 ± 0.92	0.053 ^b

CNR, contrast-to-noise ratio; SNR, signal-to-noise ratio.

Simulated 120 kVp referred to experimental group ($n = 82$), standard 120 kVp referred to control group ($n = 45$).

^aCNR, SNR, HV, liver parenchyma and noise between simulated 120-kVp and standard images were compared by the unpaired Student's *t*-test.

^bWilcoxon rank sum test was used for pairwise comparison of subjective vascular ratings.

that the non-linear-blending technique enhanced the hepatic vein contrast while reducing the soft tissue noise. The non-linear-blending technique individually weighed the CT value of each pixel and shifted the low Hounsfield unit values toward the 140-kVp CT values and the high Hounsfield unit values toward the 80-kVp CT values.⁸ In the present study, the average CT value of the liver parenchyma and hepatic veins was used as the blending centre, and a narrow blending width of 0–30 was recommended. The purpose of this blending centre and width set-up was to have the pixels in the hepatic veins receive all the 80-kVp attenuation values, increasing vessel lumen opacification, and have the pixels in the parenchyma receive the 140-kVp attenuation values at the same time, reducing the background noise. Therefore, after the non-linear-blending process, the image showed high attenuation in the hepatic veins and low background noise in the parenchyma. Compared with the simulated 120-kVp images, the CT value of the hepatic veins increased from 151.73 to 214.94 HU, and the image noise was reduced from 15 to 11 HU.

Monochromatic imaging has been considered a potential method to reduce beam-hardening artefacts and improve image quality.^{11,17–19} Our results also demonstrated that lower keV levels provided a higher CT value in the hepatic veins. However, owing to the significantly increased noise, the optimal CNR was often achieved at 66–68 keV. Unlike the non-linear-blending technique, the monochromatic technique did not process each pixel individually and thus was unable to use the high attenuation of the low keV and the low noise of high keV within one image set. Therefore, in this keV range with the optimal CNR, the image quality ratings were not as high as the non-linear-blending image ratings, and there was no significant difference compared with the pure 80-kVp images. A novel technique (MEI+, monoenergetic+) was applied in the low-keV images. This technique can reduce the noise and maintain improved contrast of the low-keV images. In contrast to the study by Schabel et al,²⁰ the image quality with the monoenergetic+ technique was lower than that with the non-linear-blending technique. Recently, the iterative reconstruction technique became available, which resulted in a significant reduction of the image noise at the same radiation dose in an abdominal CT.^{21,22} The improvement of noise

performance would help to further improve the image quality of DECT.²³

In addition, this study showed that pure 80-kVp imaging increased the CT value of the hepatic veins, with no significant difference from the non-linear-blending images. The contrast enhancement of the hepatic veins on pure 80-kVp images reached 174.21 ± 45.87 , but the background noise was also the highest. Clearly, the advantage of the non-linear-blending images was the improved CT values of the 80-kVp images and the background noise of the 140-kVp images in hepatic venography.

Our study had limitations. First, the iterative reconstruction technique was not used for the DECT data. The main purpose of the present study was to compare the abilities of the non-linear-blending technique and monochromatic imaging with that of DECT to improve the image quality. Whether the iterative reconstruction method could improve the non-blended or monochromatic image quality or lead to a reduction in the radiation dose should be a topic for future research. Second, we did not enrol patients with hepatic vein thrombosis or stenosis. Although the non-linear-blending technique improved the conspicuity of the branches of the hepatic veins, whether this technique could clearly depict the shape and size of emboli is a topic of interest. Third, the average body weight of the patients in this study was relatively small. It might be necessary to verify our results by applying them to a population with a large body size. Moreover, the tube voltage of 120 kVp, the standard-of-care protocol in our department, was used in the control group. Compared with the DECT group, this group had a higher radiation dose, which might have led to a better noise performance in the control group. Nevertheless, the CNR for the non-linear-blending images in the DECT group was significantly higher than that of the control group.

In conclusion, this study demonstrated that, with optimal monochromatic imaging and the non-linear-blending technique, DECT venography improved the conspicuity of the hepatic veins with a lower dose of contrast agent. According to our study, the non-linear-blending technique with DECT images provided the best image quality for hepatic venography.

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