Pilot multi-reader study demonstrating potential for dose reduction in dual energy hepatic CT using non-linear blending of mixed kV image datasets

Anja Apel · Joel G. Fletcher · Jeff L. Fidler · David M. Hough · Lifeng Yu ·
Luis S. Guimaraes · Matthias E. Bellemann · Cynthia H. McCollough ·
David R. Holmes III · Christian D. Eusemann

Received: 24 February 2010 / Revised: 8 June 2010 / Accepted: 16 July 2010 / Published online: 29 September 2010
© European Society of Radiology 2010

Abstract

Objective To determine the potential for radiation dose reduction using sigmoidally-blended mixed-kV images from dual energy (DE) hepatic CT.

Methods Multiple contrast-enhanced, DE (80 kV/140 kV) datasets were reconstructed from 34 patients undergoing clinically-indicated examinations using routine CTDI_{vol}. Noise was inserted in projection-space to simulate six dose levels reflecting 25–100% of the original dose. Three radiologists, blinded to dose, evaluated image preference, image quality, and diagnostic confidence (scale 1 to 5) using sigmoidally-blended, mixed-kV images, identifying the lowest acceptable dose (both image quality and confidence scores ≥ 4). At this lowest acceptable dose, the sigmoidal, 0.5 and 0.3 linear blended images were ranked in order of preference.

Results Radiation dose level correlated with image preference (correlation coefficients = 0.94, 0.81, 0.94). However, 82% (28/34) and 97% (33/34) of examinations corresponding to dose reductions of 45% and 30%, respectively, yielded acceptable image quality and confidence for all three radiologists. These frequencies were similar whether or not a lesion was present. Each radiologist had specific preferences between mixed-kV image display techniques ($p \leq 0.006$), with two most often preferring sigmoidally-blended images.

Conclusions There is potential for further dose reduction utilizing DE hepatic CT. Radiologist visual preference for mixed-kV images is idiosyncratic.

Keywords Tomography · X-ray computed · Radiation dosage · Radiography · Dual energy scanned projection · Subtraction technique · Liver

A. Apel · C. D. Eusemann
Siemens Healthcare USA,
Malvern, PA, USA

A. Apel · M. E. Bellemann
University of Applied Sciences Jena,
Jena, Germany

J. G. Fletcher (✉) · J. L. Fidler · D. M. Hough · L. Yu ·
L. S. Guimaraes · C. H. McCollough
Department of Radiology, Mayo Clinic,
200 First Street SW,
Rochester, MN 55905, USA
e-mail: fletcher.joel@mayo.edu

D. R. Holmes III
Department of Biomedical Engineering, Mayo Clinic,
200 First Street SW,
Rochester, MN 55905, USA

Introduction

Dual energy CT was originally designed for classification of the elementary chemical composition of tissues, which is of clinical value in applications such as automated bone removal [1], kidney stone characterization [2], gout detection [3], pulmonary embolism detection [4], endoleak detection [5] and iodine removal (or “virtual non-contrast”) [1]. Relatively little attention has been paid to the potential for enhanced detection of hypervascular pathologies at dual energy CT [6], despite the predicted advantage of low kV imaging in phantom models [7, 8]. Marin et al. recently demonstrated, however, that low kV imaging results in increased conspicuity of hypervascular liver lesions [9]. Material characterization at hepatic CT may be beneficial if increased lesion detection or ability to quantitate fat or iron

are realized. The two different energies have a great impact on the contrast resolution and noise characteristics of the resulting datasets. Moreover, reviewing and displaying dual energy images presents challenges. To alleviate this problem, dual-energy image data is routinely blended to yield a fused volume consisting of both the low and high kV information. Low kV acquisitions (e.g. 80 kV) provide great iodine contrast resolution, but usually present high noise levels. High kV acquisitions (e.g. 140 kV) show low noise levels, but inferior iodine contrast signal.

This unique information contained in high and low kV images has the potential to enhance the diagnostic capabilities of CT, but the question arises as to how to best convey this information to the radiologist in a single, blended image. The standard approach uses a simple linear blending algorithm that utilizes a specific percentage of the low and of the high kV image. A blended image with a blending ratio of 70% 140 kV image and 30% 80 kV has image characteristics similar to a 120 kV single energy image for particular patient sizes. By increasing the blending ratio more towards the 80 kV image, the iodine contrast information will improve, but the noise in the image will increase, since both characteristics are linked. This means that by shifting the blending ratio in the opposite direction, the blended image will contain less noise, but also less iodine contrast information. A second and more sophisticated approach to blend the two images is to combine the images in a non-linear fashion. This can be accomplished via a variety of different algorithms. Eusemann et al. [10, 11] compared six non-linear blending algorithms and showed that a blending ratio based on a sigmoidal blending (i.e. modified sigmoidal blending) provided the best image characteristics of any of the blended images. This sigmoidal blending approach (implemented as Siemens syngo(R) DE Optimum Contrast, Siemens Healthcare, Erlangen, Germany) utilizes low kV image data for image regions that show higher CT number (such as iodine contrast) and high kV image information for imaging regions with lower CT number. Independent of the blending algorithm, dual energy CT requires careful patient selection. Our clinical practice currently employs a patient size threshold of ≤ 36 cm lateral width for selection of dual-energy examinations employing 80 kV as the lower energy x-ray tube, and ≤ 41 cm lateral width for selection of dual-energy examinations employing 100 kV as the lower energy x-ray tube [12]. The use of low kV data will require higher tube currents, which will be limited by CT systems, and will result in non-diagnostic images in large patients. Because images obtained at both energies must be in the quantum limited region, there is the potential for unnecessary overexposure. In our practice we usually match the radiation dose of dual energy CT examinations with corresponding single energy CT examinations. The

goal of this study was to estimate the potential for radiation dose reduction in hepatic dual-energy CT using non-linear blending of high and low kV images.

Methods

Case selection

Institutional Review Board approval was received for this retrospective examination to utilize raw CT data for research purposes. Every participant in this study consented to the retrospective use of clinically-acquired data for research purposes, and this authorization was documented and verified by the investigators. For this study, only abdominal dual energy liver cases acquired using dual-source CT (Siemens Definition, Siemens Medical Solutions, Germany) and utilizing intravenous contrast medium were included. Cases could be either routine portal-phase abdominal dual energy examinations or late arterial phase dual energy acquisitions as part of a multi-phase hepatic CT. Further criteria included 14×1.2 mm detector configuration and archived raw data acquired using Syngo CT2008G software (required for noise insertion). Other inclusion criteria included the ability to insert noise and to generate sigmoidal- and linear-blended mixed kV images.

Figure 1 outlines the inclusion criteria used to select the 34 patients in this study. A total of 107 patients underwent dual energy hepatic CT using dual source CT. The images for 46 of these patients were acquired using a detector configuration other than 14×1.2 mm and the raw data for seven patients was no longer available. In one case, the images could not be blended. Eleven cases could not be reconstructed as a result of an older software version of the CT system. In eight cases, only delayed phase or unenhanced images were obtained.

All clinical examinations were performed on a Siemens Definition dual source CT system (Siemens Healthcare, Erlangen, Germany) utilizing automatic exposure control, a 14×1.2 mm detector configuration, and the following parameters for acquisition and reconstruction for Tube-A (quality reference mAs = 100–200; 140 kVp; slice thickness = 3 mm; reconstruction interval = 2 mm; reconstruction kernel = B30f) and Tube-B (quality reference mAs = 425–808; 80 kVp; slice thickness = 3 mm; reconstruction interval = 2 mm; reconstruction kernel = B30f). This yielded examinations with CTDI_{vol} of either ~ 18 or ~ 25 mGy (representing the sum of CTDI_{vol} values for both tubes) for the standard size patient, corresponding to our routine abdominal (including enterography) or liver CT dose levels. The routine dose levels for our single source routine abdominal and hepatic CT are also 18 or 25 mGy, respectively. These dose levels are within the European Guidelines on Quality

Fig. 1 Outline of the inclusion criteria used to select the study population

Criteria for Computed Tomography of 35 mGy [13]. The CTDIvol of each CT examination was recorded, in addition to the patient lateral width at the level of the liver dome, and the reason for the examination as given by the referring clinician. Eighteen of the cases were used in a prior retrospective study [12], but the reconstructed CT datasets (including all datasets with noise insertion, and non-linear and linear mixed kV blended images) and all observations and comparisons are entirely unique to the present study.

CT noise insertion

Noise in CT images originates from data noise in the projection measurement, which has two principal sources: quantum noise and electronic noise. The electronic noise is the result of electronic fluctuation in the detector photodiode and other electronic components. The quantum noise is due to the limited number of photons collected by the detector. Although a current CT detector is not a photon-counting element but rather an energy integrator that

generates a signal proportional to the total energy deposited in the detector, a photon-counting model is still a good approximation and is widely used for characterizing noise properties of the CT data [14–16]. The actual residual error introduced by a photon counting model is only a few percent for typical photon flux level in clinical CT protocols [17]. In our noise insertion algorithm, we utilized a photon counting model that considered the effect of beam-shaping filter and tube current modulation.

Using this validated noise insertion tool, we simulated noise mimicking the following dose levels into high and low kV data: 25%, 50%, 60% and 70%. Six sets of mixed kV images were created from the 80 and 140 kV datasets using the following combinations:

- No insertion of noise (0% dose reduction)
- $0.7 * 140 \text{ kV} / 0.7 * 80 \text{ kV}$ (70% dose reduction)
- $0.6 * 140 \text{ kV} / 0.6 * 80 \text{ kV}$ (60% dose reduction)
- $0.6 * 140 \text{ kV} / 0.5 * 80 \text{ kV}$ (55% dose reduction)
- $0.5 * 140 \text{ kV} / 0.5 * 80 \text{ kV}$ (50% dose reduction)
- $0.25 * 140 \text{ kV} / 0.25 * 80 \text{ kV}$ (25% dose reduction)

Noise insertion, therefore, resulted in 204 unique CT datasets (34 patients $\times$ 6 dose levels).

Creation of mixed kV images

For non-linear blending we utilized Optimum Contrast (Siemens Syngo CT2008G; Siemens Healthcare, Forchheim, Germany). This commercial non-linear blending software differs from the previously described [11] by slightly denoising the low kV image. To adjust the resulting images to the characteristics required by the physicians, the blending software allows for adjustment of the blending centre, which is the point of inflection, and blending width, which defines the rate of change in transition from a low blending ratio to a high blending ratio. Figure 2 illustrates the non-linear blending function underlying the commercial software.

The preferred image characteristics for the liver (i.e., noise level and contrast) were based on previously conducted studies described by Holmes et al. [11]. Holmes et al. showed that the modified sigmoidal blending function can increase CNR for DE images compared to linear blending approaches, and that blending parameters are dependent on the organ of interest. The blending focus of this study was to maximize the CNR while at the same time maintaining low noise in the liver. For consistency, noise was always determined in liver regions without visible arteries or pathologies and the contrast was always determined as enhancement of the intravenous contrast in the portal vein. The low noise requirement could be best matched by a high fraction of the 140-kV image (low noise and contrast), while a high contrast resolution could be best

Fig. 2 Graphical illustration of non-linear blending algorithms (used with permission from [11])

achieved by a high fraction of the 80 kV image (high contrast and noise). The first actual blending step was to set the blending centre, the point of inflection for the sigmoidal curve. The higher the blending centre value, the more 140 kV information is included in the resulting images and vice versa. The second blending step was to set the blending width, which can smooth the transition between the blend ratios for less noise. The choices for blending centre and width were modified based on visual surveillance for each patient and dose level. Figure 3 illustrates a full dose image and two moidally blended images at 25% and 60% of the original dose.

In addition to non-linear sigmoidal blending we blended images using a 0.3 and 0.5 linear blending method, which means that 30% or 50% of the CT number of each pixel, respectively, is weighed by the 80 kV tube, and the remaining 70% and 50% of the CT number is weighted by the higher energy tube (140 kV).

Assessment of image quality, diagnostic confidence and preference

Three radiologists independently evaluated the six sigmoidally-blended, mixed kV CT datasets for each patient in a blinded, randomized side-by-side comparison using a commercial workstation (GE Advantage Windows, GE Healthcare). Each reader is a board-certified radiologist by the American Board

of Radiology with a fellowship in GI radiology ($n=1$) or abdominal imaging ($n=2$), with 16, 16, and 12 years of experience as a consulting staff radiologist. For comparison of the different dose levels for each patient, the original dose was chosen to be the non-blended baseline for the ranking.

The initial assignment for the radiologists was to rank sigmoidally-blended image series by image preference. The image series were ranked in comparison to the baseline image. The order of the ranked image series was recorded for each patient, with radiologists recording their preference on a scale from 1 to 6, where one was the best place and six the worst.

The radiologists were subsequently asked to determine the lowest acceptable dose based on image quality (IQ) and diagnostic confidence (DC) for each patient. The scoring mechanism for this assignment is outlined in Table 1. The lowest acceptable dose was defined as the lowest dose at which both the image quality and the diagnostic confidence were scored ≥ 4 (i.e., minor insignificant noise or artifacts, most likely will detect all abnormalities).

The final assignment was to compare the sigmoidally-blended mixed kV images against the linear blended mixed kV images. For this assignment, each of the three radiologists compared the image series that were chosen as the lowest acceptable dose level. For purposes of comparison, we used the dose level chosen by Radiologist A as the standard dose level at which all three radiologists would make this comparison. Each radiologist then compared the sigmoidally-

Fig. 3 51-year-old female with cholangiocarcinoma (arrow) and hepatic venous occlusion at dual-energy CT. **a** Sigmoidally-blended mixed kV image at 25% full dose level. **b** Sigmoidally-blended mixed

kV image at 60% dose level. **c** Sigmoidally-blended mixed kV image at full clinical dose

Table 1 Explanation for scoring of image quality and diagnostic confidence for lesion detection or characterization

	Image quality	Diagnostic confidence
1	Non-diagnostic	Non-diagnostic, cannot identify or rule out lesion
2	Compromised by artifact	Will potentially miss lesions
3	Moderate artifacts	Will probably not miss or mischaracterize lesions
4	Minor insignificant noise or artifacts	Most likely will identify all abnormalities
5	Excellent image quality	Can detect lesions without diagnostic compromise

blended mixed kV, 0.3 linear blend (LB) mixed kV and 0.5 LB mixed kV images in a blinded fashion (Fig. 4). The 0.3 LB mixed kV images simulates the CT numbers of 120 kV for patients of certain sizes, and the 0.5 LB mixed kV images provide greater image contrast to assist in the visualization of hypervascular nodules or vessels. The readers evaluated these image series for image quality and diagnostic confidence as well (using the previously defined scale) and also ranked their preference for each mixed kV blending method.

Statistical analysis

Friedman's test was used to assess for significant differences in sigmoidal width and centre between different pairs of noise levels. The capacity for image preference to assign the images in perfect order of the noise introduction was statistically calculated by using Lin's Concordance Correlation Coefficient (CCC), with values ranging from +1 (perfect agreement) to -1 (perfect disagreement) with zero

meaning no agreement. A CCC=1 means that there is perfect concordance between the ranking and the percent noise. The calculation of the CCC compared the radiologist's ranking (1 to 6) with the degree of noise in the image (100%=1 to 25%=6) for each of the 34 patients. Descriptive statistics for the CCC estimates, assuming that the 100% dose level should be the best, followed by the 70% dose level, and so forth until 25% is the worst. The evaluation was then analyzed using Friedman's test.

Results

Eighteen dual-energy routine abdominal or enterography CT examinations had a mean CTDI_{vol} of 18±4 mGy, corresponding to patient sizes of mean lateral width of 34±4 cm. Sixteen dual-energy dedicated hepatic CT examinations had a mean CTDI_{vol} of 28±5 mGy, corresponding to mean patient sizes of 36±4 cm in terms of patient lateral width. These full-dose mixed kV examinations serve as the

Fig. 4 66-year-old female with a new neuroendocrine metastasis to right posterior liver (arrows). A prior RFA defect is seen in the right anterior liver. **a** Sigmoidal-blended image at 25% of standard dose, with unacceptable image quality for two of three radiologists. **b** 0.3 Linear blend image at 50% standard dose, with acceptable image quality for all three radiologists. **c** Sigmoidal-blended image at 50% standard dose,

with acceptable image quality for all three radiologists. **d** 0.5 Linear blend image at 50% standard dose, with acceptable image quality for all three radiologists. **e** 0.3 Linear blend image at full dose, simulating the CT attenuation properties of a 120 kV image. **f** Sigmoidal-blended image at full dose. Note higher CT number of enhancing structures due to higher contribution of 80 kV image data

full-dose comparison examinations for this study. Reasons for referral to CT examination follow-up or detection of hepatocellular carcinoma ($n=13$), detection or staging of Crohn's disease ($n=6$), diarrhea ($n=5$), abdominal pain ($n=3$), and other reasons ($n=6$).

Moidal blending settings—centre and width

The descriptive statistics for the moidal blending Width and Centre values are shown in Table 2. The sigmoidal blending width was similar for all noise level comparisons, with the median width varying between 127 and 132 ($p>0.09$), except for the width levels at 55% and 50% of full dose ($p=0.04$). In contrast, sigmoidal blending centre values were not consistent across noise levels, and slightly increased as dose level decreased.

Image preference and dose level

Preference rank for each radiologist correlated highly with dose level. For radiologist A, the estimates of Lin's correlation coefficient had a mean (SD) of 0.92 (0.07) and a median (min, max) of 0.94 (0.77, 1.0); for radiologist B the estimates of the CCC mean (SD) were 0.81 (0.19) with a median (min, max) of 0.83 (0.26, 1.0); and for radiologist C the estimates of the CCC mean (SD) were 0.93 (0.07) with a median (min, max) of 0.94 (0.77, 1.0). The radiologists A and C showed a somewhat greater concordance between their ranking and the dose level of the reconstructed CT dataset.

Lowest acceptable dose

Table 3 illustrates the lowest acceptable dose levels for each patient as assessed by the radiologists.

Since presence and/or absences of lesions and their detection is of greatest importance, Table 3 list the lowest acceptable dose for patient cases with and without lesions and Table 4 solely the 14 cases with lesions. Both tables illustrate that a majority of the cases (e.g. 65% with lesion vs. 62% with or without lesion) demonstrated acceptable image quality and diagnostic confidence at 50% radiation dose, and that all 14 patients with lesions had acceptable image quality and diagnostic confidence at 30% dose reduction for all three radiologists. 82% (28/34) and 97% (33/34) of examinations corresponding to dose reductions of 45 and 30%, respectively, yielded acceptable image quality confidence for all three radiologists. Figures 4 and 5 show focal hepatic lesions at unacceptable levels of dose reduction and acceptable levels of dose reduction using sigmoidally-blended mixed kV images.

Blending method preference

For the evaluation of the different blending methods, the three blending methods were ranked using the lowest dose level that was deemed acceptable to radiologist A. Two out of three radiologists preferred the sigmoidal blending technique (22/34 for reader 1, $P<0.001$ and 18/34 for reader 3, $p=0.002$), whereas the third radiologist preferred the 0.3 linear blend mixed kV image (20/34 cases, $p=0.006$).

Discussion

Dual energy CT can provide information not available in single energy CT due to the unique potential for material classification. Apart from material classification, however, DE also provides opportunities for disease diagnosis using

Table 2 Descriptive statistics for moidal blending width and centre values

	Mean	Standard Deviation	Median	Minimum	Maximum	25th%	75th%
Moidal Blending Width							
Dose Level 100%	132.941	59.6215	120	70	350	100	140
Dose Level 70%	135.294	70.8492	120	70	410	100	140
Dose Level 60%	135.294	67.9677	115	60	370	100	140
Dose Level 55%	130.282	72.4667	110	80	410	100	130
Dose Level 50%	140.882	74.6453	120	80	430	100	140
Dose Level 25%	127.941	74.1734	110	60	500	100	120
Moidal Blending Centre							
Dose Level 100%	240.588	71.8560	260	90	440	190	280
Dose Level 70%	243.824	65.9667	260	110	430	200	280
Dose Level 60%	249.118	70.6836	245	110	440	200	280
Dose Level 55%	256.471	65.3143	250	130	430	210	300
Dose Level 50%	256.176	71.8169	255	120	450	210	300
Dose Level 25%	266.765	65.8612	275	120	440	220	300

Table 3 Lowest acceptable dose level for all 34 patient cases (i.e. image quality and diagnostic confidence scores of four or higher)

Radiologists	Dose level ^a	Frequency this level is lowest dose	Percentage	Cumulative% of examinations acceptable at given dose level
A	25	0	0.00	0
	50	24	70.59	71
	55	6	17.65	88
	60	2	5.88	94
	70	2	5.88	100
	100	0	0.00	100
B	25	8	23.53	24
	50	22	64.71	88
	55	2	5.88	94
	60	0	0.00	94
	70	1	2.94	97
	100	1	2.94	100
C	25	22	64.71	65
	50	12	35.29	100
	55	0	0.00	100
	60	0	0.00	100
	70	0	0.00	100
	100	0	0.00	100
Overall Radiologists	25	0	0.00	0
	50	21	61.76	62
	55	7	20.59	82
	60	2	5.88	88
	70	3	8.82	97
	100	1	2.94	100

^a“Dose level” is given in percent, relative to the clinically-employed mixed kV dual-energy exam, which was used as a baseline for comparative purposes

mixed kV images. In particular, the incorporation of low kV image information is likely to be diagnostically useful if image noise is not excessive [9]. Despite these potential advantages, however, others have reported that increased radiation dose is required for dual energy CT for various indications on several hardware platforms [18, 19]. The non-linear sigmoidal blending approach utilizes the diagnostic strength of both tube voltages, and permits an avenue for dose reduction using anatomic mixed kV images.

Our study showed that while radiologist preference is closely linked with radiation dose level and different radiologists perceive noise levels differently, this preference does not adversely impact image acceptability in terms of image quality or diagnostic confidence. The lowest acceptable dose level analysis showed that over 80% of the patients could have been examined at 55% of the acquired dose, independent of the presence of lesions/abnormalities detected. To ensure a margin of safety and diagnostic image

quality, however, we believe a 30% reduction in dose will result in acceptable DE mixed kV images, as this dose reduction translated into 100% image acceptability for two readers and 97% for the third. While dual energy is sometimes criticized for increasing radiation dose, our study showed that dose reduction is achievable using dual-energy in combination with sigmoidal blending, even when compared to single energy, single tube CT acquisitions in our practice. This may be particularly beneficial when dual energy is simultaneously being used for material classification.

Using phantoms of varying sizes, Yu et al. recently compared the image quality of single energy CT images to optimized, linearly-blended mixed kV images at the same radiation dose [20]. They found that optimized, linearly-blended dual energy images could produce similar noise and similar or improved iodine contrast to noise ratios

Table 4 Lowest acceptable dose level for the 14 patient cases with hepatic masses

Radiologists	Dose level ^a	Frequency this level is lowest dose	Percentage	Cumulative% of examinations acceptable at given dose level
A	25	0	0.00	0
	50	9	64.92	65
	55	2	14.29	79
	60	2	14.29	93
	70	1	7.14	100
	100	0	0.00	100
B	25	5	35.71	36
	50	9	64.29	100
	55	0	0.00	100
	60	0	0.00	100
	70	0	0.00	100
	100	0	0.00	100
C	25	11	78.57	79
	50	3	21.43	100
	55	0	0.00	100
	60	0	0.00	100
	70	0	0.00	100
	100	0	0.00	100
Overall Radiologists	25	0	0.00	0
	50	9	64.92	65
	55	2	14.29	79
	60	2	14.29	93
	70	1	7.14	97
	100	0	0.00	100

^a“Dose level” is given in percent, relative to the clinically-employed mixed kV dual-energy exam, which was used as a baseline for comparative purposes

Fig. 5 65 year-old male with stable multifocal HCC on chemotherapy with new enhancing focus along superolateral edge of previously treated tumor (arrows). Subsequent imaging confirmed growth of this marginal recurrence. **a** Sigmoidal-blended image at 25% of standard dose, with unacceptable image quality for two of three radiologists. **b** 0.3 Linear blend image at 50% standard dose, with acceptable image quality for all three radiologists. **c** Sigmoidal-blended image at 50%

standard dose, with acceptable image quality for all three radiologists. **d** 0.5 Linear blend image at 50% standard dose, with acceptable image quality for all three radiologists. **e** 0.3 Linear blend image at full dose, simulating the CT attenuation properties of a 120 kV image. **f** Sigmoidal-blended image at full dose. Note higher CT number of enhancing structures due to higher contribution of 80 kV image data

compared to 120 kV single energy CT. Our work represents the logical extension of this principle into patients, but employed sigmoidally-blended mixed kV images to reduce noise and optimize CNR.

The study also illustrated that the blending method preference is user dependent. Two out of three radiologist preferred sigmoidal blending over the linear blended images. The third radiologist preferred the 0.3 linear blending approach. The review of the sigmoidal blending centre settings were affected by dose/noise level but that effects on sigmoidal widths were negligible. Given the small influence on both parameters, we found a general appropriate blending width of 130 and a blending centre of 250 can be used as appropriate default settings for liver display.

Our study has several weaknesses, principally owing to its retrospective nature, non-consecutive patient selection, heterogeneous reasons for referral to CT examination, and lack of reference standard histopathology or imaging follow-up. Some examinations were dedicated hepatic CT examinations whilst others were performed for survey or general indications other than hepatic pathology. We selected hepatic visualization as an imaging endpoint for our study based on the anatomic complexity and heterogeneity of the organ, as reflected by the higher dose levels used in our practice for dedicated hepatic imaging. While the combination of routine abdominal and hepatic CT

indications is justified given the goals of this pilot multi-reader study to validate the ability of non-linear blending to reduce dose at dual-energy CT, further studies comparing the performance characteristics and clinical utility of dose-optimized dual-energy and single-energy CT are needed. Our study also examined a relatively small number of patients and did not selectively evaluate low contrast hepatic lesions. Theoretically the use of lower kV at dual-energy CT combined with the use of a non-linear blending method to emphasize iodine signal differences should improve the CT number differences of subtle lesions; however, we did not test this hypothesis, and only 14 patients had hepatic lesions. In this pilot study, however, six datasets were reconstructed for each patient to determine the lowest acceptable dose. Future studies will be needed to examine the impact of lower dose combined with dual-energy CT and non-linear blending of mixed kV images to assess the impact of these methods on low contrast hepatic lesions compared to single source CT. Finally, we did not substratify our results by patient size. Of course, some patients should not undergo dual-energy CT due to their size (e.g., ≥ 36 cm and ≥ 41 cm based on patient lateral width from the CT scout for dual energy examinations utilizing 80 kV and 100 kV for the lower energy, respectively), which would result in undesirable artifacts that would degrade diagnostic image quality.

Conclusions

Sigmoidal blending of dual energy data can provide an avenue for dose reduction at dual energy, contrast-enhanced abdominal CT. The results of this study suggest lower dose dual energy acquisitions could be utilized in clinical routine, but the effect of such dose reductions on material classification remains unknown. Further investigation on organ-specific default settings is warranted, but the relatively consistent blending parameters for the liver cases suggests that sigmoidal blending center and width can be determined for specific organs and are relatively independent of dose level.

Conflicts of interest Christian Eusemann, PhD - employee of Siemens
J. G. Fletcher, MD, Cynthia McCollough, PhD - grant support, Siemens

J. G. Fletcher, MD, Christian Eusemann, PhD, David Holmes, PhD - co-inventors of non-linear blending method tested

None of the co-inventors or Siemens employees had control of study data or performed the data analysis

References

- Johnson TR, Krauss B, Sedlmair M, Grasruck M, Bruder H, Morhard D, Fink C, Weckbach S, Lenhard M, Schmidt B, Flohr T, Reiser MF, Becker CR (2007) Material differentiation by dual energy CT: initial experience. *Eur Radiol* 17:1510–1517
- Primak AN, Fletcher JG, Vrtiska TJ, Dzyubak OP, Lieske JC, Jackson ME, Williams JC Jr, McCollough CH (2007) Noninvasive differentiation of uric acid versus non-uric acid kidney stones using dual-energy CT. *Acad Radiol* 14:1441–1447
- Johnson TR, Weckbach S, Kellner H, Reiser MF, Becker CR (2007) Clinical image: dual-energy computed tomographic molecular imaging of gout. *Arthritis Rheum* 56:2809
- Cann CE, Gamsu G, Birnberg FA, Webb WR (1982) Quantification of calcium in solitary pulmonary nodules using single- and dual-energy CT. *Radiology* 145:493–496
- Stolzmann P, Frauenfelder T, Pfammatter T, Peter N, Scheffel H, Lachat M, Schmidt B, Marincek B, Alkadhi H, Schertler T (2008) Endoleaks after endovascular abdominal aortic aneurysm repair: detection with dual-energy dual-source CT. *Radiology* 249:682–691
- Apel A, Fletcher J, Siddiki H, Huprich J, Holmes D, McCollough C, Braun N, Bellemann M, Eusemann C (2008) Linear vs Non-Linear Dual Energy Blending—What blending technique leads to a better Contrast-to-Noise Ratio? 94th Scientific Assembly and Annual Meeting, Radiological Society of North America
- Schindera ST, Nelson RC, Mukundan S Jr, Paulson EK, Jaffe TA, Miller CM, DeLong DM, Kawaji K, Yoshizumi TT, Samei E (2008) Hypervascular liver tumors: low tube voltage, high tube current multi-detector row CT for enhanced detection—phantom study. *Radiology* 246:125–132
- Nakayama Y, Awai K, Funama Y, Hatemura M, Imuta M, Nakaura T, Ryu D, Morishita S, Sultana S, Sato N, Yamashita Y (2005) Abdominal CT with low tube voltage: preliminary observations about radiation dose, contrast enhancement, image quality, and noise. *Radiology* 237:945–951
- Marin D, Nelson RC, Samei E, Paulson EK, Ho LM, Boll DT, DeLong DM, Yoshizumi TT, Schindera ST (2009) Hypervascular liver tumors: low tube voltage, high tube current multidetector CT during late hepatic arterial phase for detection—initial clinical experience. *Radiology* 251:771–779
- Eusemann C, Holmes DI, Schmidt B, Flohr T, Robb R, McCollough C, Hough D, Huprich J, Wittmer M, Siddiki H, Fletcher J (2008) Dual Energy CT—How to best blend both energies in one fused image (no. 6918-2). *SPIE Symposium on Medical Imaging*
- Holmes DR 3rd, Fletcher JG, Apel A, Huprich JE, Siddiki H, Hough DM, Schmidt B, Flohr TG, Robb R, McCollough C, Wittmer M, Eusemann C (2008) Evaluation of non-linear blending in dual-energy computed tomography. *Eur J Radiol* 68:409–413
- Guimarães LS, Fletcher JG, Yu L, Huprich JE, Fidler JL, Manduca A, Ramirez-Giraldo JC, Holmes DR Jr, McCollough CH (2010) Feasibility of dose reduction using novel denoising techniques for low kV (80 kV) CT enterography: optimization and validation. *Acad Radiol* 17(10):1203–1210
- European Commission on Radiation Protection (2000) European guidelines on quality criteria for computed tomography (EUR 16262 EN). Luxembourg
- Barrett H, Swindell W (1981) *Radiological Imaging: The Theory of Image Formation, Detection, and Processing*. Academic, San Diego
- Thibault JB, Sauer KD, Bouman CA, Hsieh J (2007) A three-dimensional statistical approach to improved image quality for multislice helical CT. *Med Phys* 34:4526–4544
- Wunderlich A, Noo F (2008) Image covariance and lesion detectability in direct fan-beam x-ray computed tomography. *Phys Med Biol* 53:2471–2493
- Whiting BR, Massoumzadeh P, Earl OA, O'Sullivan JA, Snyder DL, Williamson JF (2006) Properties of preprocessed sinogram data in x-ray computed tomography. *Med Phys* 33:3290–3303
- Ho LM, Yoshizumi TT, Hurwitz LM, Nelson RC, Marin D, Toncheva G, Schindera ST (2009) Dual energy versus single energy MDCT: measurement of radiation dose using adult abdominal imaging protocols. *Acad Radiol* 16:1400–1407
- Lohan DG, Motamedi K, Chow K, Habibi R, Panknin C, Ruehm SG, Seeger LL (2008) Does dual-energy CT of lower-extremity tendons incur penalties in patient radiation exposure or reduced multiplanar reconstruction image quality? *AJR Am J Roentgenol* 191:1386–1390
- Yu L, Primak AN, Liu X, McCollough CH (2009) Image quality optimization and evaluation of linearly mixed images in dual-source, dual-energy CT. *Med Phys* 36:1019–1024