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SHORT COMMUNICATION

Radiation dose efficiency of dual-energy CT benchmarked against single-source, kilovoltage-optimized scans

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Objective: This study evaluated the radiation dose and image quality implications of dual-energy CT (DECT) use, compared with kilovoltage-optimized single-source/single-energy CT (SECT) on a dual-source Siemens Somatom® Definition Flash CT scanner (Siemens Healthcare, Forchheim, Germany).

Methods: With equalized radiation dose (volumetric CT dose index), image noise (standard deviation of CT number) and signal-difference-to-noise ratio (SDNR) were measured and compared across three techniques: 100, 120 and 100/140 kVp (dual energy). Noise in a 30-cm-diameter water phantom and SDNR within unenhanced soft-tissue regions of a small adult (50 kg/165 cm) anthropomorphic phantom were utilized for the assessment.

Results: Water phantom image noise decreased with DECT compared with the lower noise SECT setting of

120 kVp ($p = 0.046$). A decrease in SDNR within the anthropomorphic phantom was demonstrated at 120 kVp compared with the SECT kilovoltage-optimized setting of 100 kVp ($p = 0.001$). A further decrease in SDNR was observed for the DECT technique when compared with 120 kVp ($p = 0.01$).

Conclusion: On the Siemens Somatom Definition Flash system (Siemens Healthcare), and for equalized radiation dose conditions, image quality expressed as SDNR of unenhanced soft tissue may be compromised for DECT when compared with kilovoltage-optimized SECT, particularly for smaller patients.

Advances in knowledge: DECT on a dual-source CT scanner may require a radiation dose increase to maintain unenhanced soft-tissue contrast detectability, particularly for smaller patients.

INTRODUCTION

The application of dual-energy (DE) techniques in CT is rapidly expanding.^{1,2} Depending primarily on the CT system manufacturer, dual-energy CT (DECT) is typically facilitated by technology which provides for either fast kilovoltage switching, dual-layer detector technology, or two X-ray sources operating at distinctly different kilovoltages.^{2,3} The latter approach [*i.e.* a dual-source CT system (DSCT)] is employed by the Siemens Somatom® Definition Flash model (Siemens Healthcare, Forchheim, Germany) installed at our facility.

Applications for DECT commonly employed at our facility include renal stone characterization, lung perfusion imaging in pulmonary angiography, iodine mapping in oncology scanning, bone removal in vascular studies and metal artefact reduction. Whilst additional information such as the atomic number characteristics of renal calculi are made available using DECT, traditional CT images consisting of a single linear attenuation coefficient matrix remain available through a combination of data from the two

X-ray sources and corresponding detector banks. Such traditional CT data sets may also be created by scanning with only a single tube. Preferably, there should be no image quality characteristic that is compromised when applying a DE technique at the same radiation dose as the single-source/single-energy CT (SECT) acquisition method.

There is substantial published literature on the radiation dose implications of DECT utilization, some of which is in the context of maintenance or improvement in image quality.⁴⁻⁷ A review of the literature by Henzler et al⁸ concluded that “there is strong evidence that DECT imaging with DSCT is not associated with increased radiation dose levels”. Previous investigations have focused on image noise measurement only, or on contrast-to-noise ratio (CNR) of iodinated contrast-enhanced images, and typically the image protocol standard or reference against which DE benefits are measured has been 120 kVp scans.⁸ Given the evidence for and recent trend towards lower kilovoltage scanning, and the potential for DECT to be used without iodinated contrast, we believed that further

investigation of fundamental DECT radiation dose efficiency may be warranted. Noting that DECT frequently leverages off the atomic number change associated with iodinated contrast, the more challenging condition of unenhanced soft-tissue imaging has been used in this study, along with benchmarking against a technique with the SECT kilovoltage optimized for body size.

METHODS AND MATERIALS

Scans were performed on both a 30-cm-diameter cylindrical water phantom and an anthropomorphic phantom (PBU-60; Kyoto Kagaku Co. Ltd, Kyoto, Japan) (Figure 1). The water phantom size was chosen for its similarity to typical adult patient torso dimensions, whilst the anthropomorphic phantom provided for better representation of soft tissue and variable density anatomy within a small (50 kg/165 cm) adult. For the scanning of each phantom, three sets of scan conditions were employed: traditional SECT at 100 kVp, SECT at 120 kVp and DECT at 100/140 kVp. DE images were processed with a kilovoltage data set mixing ratio of 0.5, which is the Siemens default setting for non-contrast imaging. A fourth image data set was retrospectively produced from the anthropomorphic phantom DECT scan data by applying the Optimum Contrast DECT application software (Siemens Healthcare). This software is designed to improve the contrast of DE scans by blending low- and high-energy data sets on a pixel-by-pixel basis in consideration of actual pixel value.⁹ Assessment of an image set produced by this software was thought to be useful since it has been previously shown to improve CNR for DE phantom studies, albeit scans involving iodinated contrast enhancement.⁴

Figure 1. Small adult anthropomorphic phantom used for study measurements.

Both the average (over scan range) and individual slice volumetric CT dose indices ($CTDI_{vol}$) were equalized as far as practicable for each scan protocol applied to a phantom type. This was achieved by adjusting the milliampere-second (mAs) values for the 120 and 100/140 kVp scans to match the average $CTDI_{vol}$ recommended by the automatic exposure control system for 100 kVp scanning. All other test scanning and reconstruction conditions (e.g. pitch, collimation, rotation time, scan length and kernel) were maintained across protocols and were consistent with those applied during routine thoracic scanning (Table 1). Note that for both water and anthropomorphic phantom scanning, automated kilovoltage selection software (CARE kV; Siemens Healthcare) was used in the first instance to determine the preferred kilovoltage for non-contrast-enhanced, single-source scanning (*i.e.* 100 or 120 kVp).

The accuracy of dose metrics displayed by the scanner, including in DE mode, had been previously confirmed by a physicist during annual system performance testing. In terms of choice of comparative dose metric, whilst effective dose may be considered superior to $CTDI_{vol}$ for risk comparison, application of Monte Carlo-based modelling software from ImpACT (CT Patient Dosimetry Calculator v. 1.0.4, Nicholas Keat for ImpACT, London, UK) confirms no change in effective dose for equalized $CTDI_{vol}$ at 100 vs 120 kVp on the Siemens Definition AS platform (single-source predecessor of Definition Flash model; Siemens Healthcare) and changes of <10% on a range of other platforms. Therefore, our approach of ensuring consistent $CTDI_{vol}$ across protocols was deemed to be an acceptable surrogate for more complex assessment and normalization involving methods such as thermoluminescent dosimetry or Monte Carlo modelling.

For the water phantom images, the image noise, measured as the standard deviation of Hounsfield unit values in a region of interest (ROI) at a central position within five consecutive slices, was used as the key metric to compare image quality between acquisition methods (Figure 2). Comparing noise at equalized radiation dose provided for a simple assessment of dose efficiency of each technique. That is, a higher noise level for images acquired with any technique would imply that radiation dose needs to be increased for that protocol to maintain equivalent image noise levels.

The previously described assessment was extended for the anthropomorphic phantom images to provide for measurement of the signal-difference-to-noise ratio (SDNR) in matching positions on five consecutive slices from each (contiguous) image data set. SDNR was calculated as follows:

$$SDNR = \frac{\text{Signal(liver)} - \text{Signal(fat)}}{\sqrt{(\text{Noise(liver)})^2 + (\text{Noise(fat)})^2} / 2}$$

The liver and adjacent region of intra-abdominal fat were chosen as locations for the ROIs owing to their relative homogeneity compared with most other body structures (Figure 3). Choice of these regions of soft tissue facilitated a clinically significant measure of unenhanced soft-tissue contrast detectability via the SDNR.

Table 1. Key parameters and dose metrics for water and anthropomorphic scans

Parameter	100 kVp scan	120 kVp scan	100/140 kVp scan
Rotation time (s)	0.33	0.33	0.33
Reconstruction kernel (SAFIRE strength)	I30f (3)	I30f (3)	I30f (3)
Reconstructed slice thickness/interval (mm)	3/3	3/3	3/3
Reconstructed field of view (mm)	400	400	400
Collimation (slice count $\times$ mm)	128 $\times$ 0.6	128 $\times$ 0.6	128 $\times$ 0.6
Pitch	0.55	0.55	0.55
CARE kV setting	3 (non-contrast)	NA	NA
Effective mAs—water phantom	143	89	83/71
CTDI _{vol} —water phantom (mGy)	6.0	6.0	6.0
(Average) effective mAs—anthropomorphic phantom	75	46	42/37
(Average) effective mAs (at axial slice measurement location)—anthropomorphic phantom	71	46	42/38
(Average) CTDI _{vol} —anthropomorphic phantom (mGy)	3.2	3.1	3.1

CTDI_{vol}, volumetric CT dose index.

CARE kV and SAFIRE; Siemens Healthcare, Forchheim, Germany.

The statistical significance or otherwise of differences in the mean of the measurement sets for each scan/reconstruction condition were evaluated with Microsoft® Excel® 2010 v. 14.0.6129.5000 (Microsoft, Redmond, WA) using two-tailed, unpaired Student's *t*-tests. In each case, the variances of the data sets were evaluated and verified for compatibility using the F Distribution.¹⁰

Since the noise characteristics of a CT image may be altered by applying reconstruction algorithms with differing spatial resolution characteristics, we needed to verify that any noise change associated with DECT does not result from even subtle differences in DECT vs SECT reconstruction algorithms or kernels. Whilst not specifically a part of this study, previous routine system testing using both a Siemens automated QA routine and manual assessment using a Catphan® 600 phantom (The Phantom Laboratory, Greenwich, NY) confirmed no change in image spatial resolution when applying the DE technique. As such, the measurement of noise and SDNR remain valid comparators, which are not confounded by differences in other aspects of image quality such as spatial resolution.

RESULTS

Initial setup with the default single-source thorax scanning protocol, and using automated kilovoltage selection, yielded a recommended kilovoltage of 100 kVp for both phantoms.

Figures 2 and 3 show results from single sets of slices and demonstrate the methods for the noise and SDNR measurements including the locations for ROI placement within adjacent 3-mm image slices. Adjacent slices within the water phantom contained no appreciable difference in characteristics of the imaged volume, whereas only slight slice-to-slice

anatomical differences were evident within the assessed region of anthropomorphic phantom.

Table 2 summarizes the image noise for water phantom scanning and SDNR within the anthropomorphic phantom. There is a decrease in noise for equal-dose scanning of the water phantom when the kilovoltage is increased from 100 to 120 kVp ($p = 0.026$) and a further decrease in noise for DECT when compared with 120 kVp ($p = 0.046$). Therefore, for a typical body size (30-cm water), and neglecting tissue contrast issues, the application of DECT is superior to both 100 and 120 kVp single-source scanning in terms of radiation dose efficiency.

SDNR measured in the anthropomorphic phantom shows a statistically significant decrease when the kilovoltage is increased from 100 to 120 kVp ($p = 0.001$). SDNR resulting from the DECT technique is decreased when compared with 120 kVp ($p = 0.01$). The SDNR of DECT images is marginally improved by application of the Optimum Contrast image-blending software; however, this increase is arguably not significant ($p = 0.14$). Whilst the SDNR for DECT with Optimum Contrast may be inferior to 120 kVp, this decrease is not significant at the 95% confidence level ($p = 0.10$).

DISCUSSION

DECT has many and varied benefits in clinical practice. On a DSCT system such as the Definition Flash scanner at our facility, system design and available scan modes provide for use of DECT whilst maintaining almost all aspects of scanning performance and features that are available in traditional single-source scanning. This system therefore lends itself to routine DE application including circumstances where there is no prior expectation of specific benefits inherent to DE. Although there may be no prior expectation of benefit, routine DE application

Figure 2. Noise measurement in the 30-cm-diameter water phantom from scans at 100 kVp (top left), 120 kVp (top right) and dual energy 100/140 kVp (bottom left).

may on occasion provide unexpected scan quality improvements such as metal artefact reduction or additional diagnostic information such as the composition of renal stones, averting a requirement for further investigation or repeat scanning. Furthermore, previous evaluations of DECT/DSCT performance have concluded that this technology can be implemented without a radiation dose-related penalty, providing confidence in its application to a wide variety of scan protocols.

The majority of CT examinations performed at our institution involve the intravenous administration of iodinated contrast.

Whilst the literature supports the ability of DSCT systems to maintain CNR without an increase in radiation dose in this circumstance, we believed that the question of whether or not contrast detectability may be maintained in unenhanced images when benchmarked against the emerging standard of single-source scanning at 100 kVp (for non-obese patients) required further exploration. In this study, we have measured against a single-source scanning beam quality set to its most dose-efficient condition, as confirmed by the systems' automated kilovoltage selection. It should, however, be noted that the choice of 100 kVp is strongly influenced by phantom size and so

Figure 3. Signal-difference-to-noise ratio measurement in the anthropomorphic phantom from scans at 120 kVp (top left), 100 kVp (bottom left), dual energy (DE) 100/140 kVp (top right) and Optimum Contrast-blended DE scan (bottom right). Optimum Contrast; Siemens Healthcare, Forchheim, Germany.

the results presented here may not be directly applicable to all patients—particularly those who are significantly larger than our relatively small (50 kg/165 cm) anthropomorphic phantom.

It had been anticipated that these scan conditions would be more challenging than those previously applied in terms of demonstrating the radiation dose efficiency of DECT on a DSCT system. Despite image noise reductions when applying DECT, this proved to be true with small, but statistically significant decreases in SDNR at 120 kVp when compared with 100 kVp

scanning (somewhat validating the automated 100 kVp selection), and similarly with the default DECT technique when compared with 120 kVp single-source scanning. Furthermore, our results confirm that DECT with Optimum Contrast blending software does not provide the substantial benefits in soft-tissue contrast that have been demonstrated with the presence of iodinated contrast.⁴

The SDNR-inferiority of DECT when benchmarked against 100 kV imaging on a relatively small anthropomorphic

Table 2. Image noise and signal-difference-to-noise ratio (SDNR) measurement results for each scan/reconstruction technique

Scan condition	Mean noise (SD of noise) in water phantom (HU)	Mean SDNR (SD of SDNR) in anthropomorphic phantom
100 kV	11.56 (0.28)	9.56 (0.30)
120 kV	11.08 (0.21)	8.72 (0.25)
100/140 kV	10.68 (0.26)	8.23 (0.23)
100/140 kV Optimum Contrast	Not measured	8.46 (0.21)

HU, Hounsfield unit; SD, standard deviation.

Optimum Contrast; Siemens Healthcare, Forchheim, Germany.

phantom is perhaps unsurprising given that the ideal SECT kVp for this body size is 100 kVp whereas there is a large component of higher photon energies (but not lower) when adding the tin-filtered 140 kVp spectrum used in this DECT system. Extending this argument, it might also be expected that the radiation dose efficiency for scanning much smaller bodies including children and infants, where scanning at 70 or 80 kVp may be ideal, could be further compromised with this DECT technique. This assumption is supported by preliminary investigations with a 5-year-old child-sized body model on a Siemens Somatom Definition Force system (Siemens Healthcare) (M. Irvine, 2015, personal communication). It is also worth noting that Siemens have now developed optional software for the Force system that provides for more flexibility in kilovoltage pair selection including the use of lower value pairs which may be beneficial for smaller patients (M. Edmonds, Siemens Healthcare, 2015, personal communication).

Assuming quantum-noise-dominated images, the increase in radiation dose required to restore the SDNR for DECT to that measured for 100 kVp SECT in this study is approximately 35%. This result applies specifically to unenhanced soft tissue contrast in a relatively small adult patient where 100 kVp would be preferred for SECT. As an example, these contrast conditions could apply to the soft-tissue background in a kidney-ureter-bladder scan for the evaluation of renal stones, or where DECT is specifically applied for metal artefact reduction. In practice, we have no intention of applying such dose increases to our DECT protocols since unenhanced soft-tissue contrast is rarely a critical image quality determinant, and because we do not scan children, our measurements are worst-case in terms of patient size. For contrast-enhanced studies, no such dose increase and potentially even a dose decrease may be justified, particularly where Optimum Contrast is used to obtain a blended image with superior iodine contrast enhancement as demonstrated by Schenzle et al.⁴

Consequently, in our adult tertiary referral facility, the results of this study have not discouraged the use of DECT for any protocol that provides some promise of additional DE-specific information or image quality improvements. We believe that with the current restrictions on DE kilovoltage selection, caution should be exercised in applying DE protocols to very small patients or children for the limited circumstances where unenhanced soft-tissue contrast may be a key aspect of image quality.

This is particularly the case for protocols where there is limited anticipated benefit from DE-specific features.

In addition to the main previously noted limitation related to anthropomorphic phantom size, we have only assessed DECT images created with the Siemens default mixing ratio for non-contrast studies. It is possible that an improved SDNR may be obtained for the patient size represented by our phantom with an alternative kilovoltage data set mixing ratio. In particular, it is speculated that a mixing ratio biased towards the 100 kVp data set may provide improved SDNR for smaller patients, whilst a mixing ratio biased towards the 140 kVp data set may be superior for larger patients. We intend to undertake this analysis at our facility in the near future.

The image quality performance analysis in this study required the equalization of patient dose across the compared scan modes. Since it is not possible to ensure the exact same dose to each irradiated organ or body tissue under different beam quality conditions, the best surrogate metric is probably the effective dose. Whilst we have verified that equalization of CTDI_{vol} is a reasonable alternative on this scanner platform in particular, it is possible that this process could be improved. Since our anthropomorphic phantom does not have the facility for insertion of passive dosimeters (e.g. thermoluminescent dosimeters), the only feasible method for a more accurate analysis would be to develop a Monte Carlo model specific to our anthropomorphic phantom and scanner beam spectra. In consideration of the magnitude of dose change required to equalize SDNR between our DECT technique and 100 kVp (approximately 35%), we feel that these efforts to improve radiation dose assessment accuracy are unnecessary.

CONCLUSION

The visibility of subtle, unenhanced soft-tissue density differences may be slightly compromised for DECT using our DSCT system when compared with equal-dose, kilovoltage-optimized SECT of a small adult torso. This aspect of image quality should have little impact on the development and expansion of DECT applications where iodinated contrast is used or other DE-specific benefits are clearly demonstrated.

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