Dual-energy imaging method to improve the image quality and the accuracy of dose calculation for cone-beam computed tomography

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ABSTRACT

Purpose: To improve the image quality and accuracy of dose calculation for cone-beam computed tomography (CT) images through implementation of a dual-energy cone-beam computed tomography method (DE-CBCT), and evaluate the improvement quantitatively.

Methods: Two sets of CBCT projections were acquired using the X-ray volumetric imaging (XVI) system on a Synergy (Elekta, Stockholm, Sweden) system with 120 kV (high) and 70 kV (low) X-rays, respectively. Then, the electron density relative to water (relative electron density (RED)) of each voxel was calculated using a projection-based dual-energy decomposition method. As a comparison, single-energy cone-beam computed tomography (SE-CBCT) was used to calculate RED with the Hounsfield unit–RED calibration curve generated by a CIRS phantom scan with identical imaging parameters. The imaging dose was measured with a dosimetry phantom. The image quality was evaluated quantitatively using a Catphan 503 phantom with the evaluation indices of the reproducibility of the RED values,

high-contrast resolution (MTF_{50%}), uniformity, and signal-to-noise ratio (SNR). Dose calculation of two simulated volumetric-modulated arc therapy plans using an Eclipse treatment-planning system (Varian Medical Systems, Palo Alto, CA, USA) was performed on an Alderson Rando Head and Neck (H&N) phantom and a Pelvis phantom. Fan-beam planning CT images for the H&N and Pelvis phantom were set as the reference. A global three-dimensional gamma analysis was used to compare dose distributions with the reference. The average gamma values for targets and OAR were analyzed with paired *t*-tests between DE-CBCT and SE-CBCT.

Results: In two scans (H&N scan and body scan), the imaging dose of DE-CBCT increased by 1.0% and decreased by 1.3%. It had a better reproducibility of the RED values (mean bias: 0.03 and 0.07) compared with SE-CBCT (mean bias: 0.13 and 0.16). It also improved the image uniformity (57.5% and 30.1%) and SNR (9.7% and 2.3%), but did not affect the MTF_{50%}. Gamma analyses of the 3D dose distribution with criteria of 1%/1 mm showed a pass rate of 99.0–100% and 85.3–97.6% for DE-CBCT and 73.5–99.1% and 80.4–92.7% for SE-CBCT. The average gamma values were reduced significantly by DE-CBCT ($p < 0.05$). Gamma index maps showed that matching of the dose distribution between CBCT-based and reference was improved by DE-CBCT.

Conclusions: DE-CBCT can achieve both better image quality and higher accuracy of dose calculation, and could be applied to adaptive radiotherapy.

Keywords: dual-energy imaging, CBCT, image quality, dose calculation, adaptive radiotherapy

1. INTRODUCTION

During radiotherapy, fan-beam computed tomography (FBCT) is used to generate digital three-dimensional (3D) reconstruction of the body, which is the basis for patient positioning and treatment planning¹. Patients receive radiotherapy *via* a certain number of fractions. Anatomic changes that may be observed during such fractionated therapy introduce large deviations between the planning dose and delivered dose²⁻³. As a result, the tumor target may not receive a sufficient dose, whereas organs at risk (OAR) may receive too much radiation.

Cone-beam computed tomography (CBCT) mounted on a linear accelerator can generate volumetric images with the patient directly at the treatment position. CBCT has been used mainly for image-guided radiotherapy^{4,5} and adaptive radiotherapy (ART)^{6,7}. 3D-CBCT images acquired immediately before each treatment fraction provide exact anatomic information and can be used to recalculate dose distribution⁸ in order to track the dose to targets and OARs and trigger the re-planning, if properly calibrated.

For accurate calculation of dose based on CBCT images, the relationship between CT numbers (in Hounsfield units (HU)) and electron density relative to water (relative electron density (RED)) is required. This strategy enables the treatment-planning system (TPS) to account for tissue heterogeneities in radiotherapy planning. Usually, conversion from HU to RED involves a RED calibration phantom⁹ that has different

tissue-equivalent substitutes with known REDs to create a HU–RED calibration curve. However, compared with planning FBCT images, CBCT images exhibit inferior image contrast and have more artifacts^{10,11} introduced by scatter and beam hardening. Moreover, the HU variability of CBCT has higher sensitivity to motion artifacts with respect to FBCT. Several studies^{12–24} have developed different correction strategies and investigated the reliability of CBCT for dose calculation depending on the properties of the CBCT system. In general, the correction methods are divided into three categories¹². The first category calibrates the voxel values of the CBCT image using a special look-up-table based on CBCT acquisitions of phantoms with inserts of different densities^{13–14}. Rong et al.¹³ proposed a site-specific calibration method to achieve higher accuracy in CBCT image-based dose calculations. Richter et al.¹⁴ generated population-specific conversion curves for head, lungs, and abdomen/pelvis treatment sites. These methods are prone to dosimetric errors resulting from CBCT artifacts. The second category measures or simulates directly the scatter of CBCT and then removes the scatter from the projections to improve image quality^{15–17}. The accuracy of dose calculation on such images has yet to be validated, and implementation of such approaches is complex and difficult to introduce into a clinical workflow. The third category maps HU from the planning FBCT images to the CBCT images based on a rigid^{18, 19} or deformable^{20–22} image registration. A reference CT is needed to compare and correct the CT values for all methods using registration procedures. However, the information from the planning FBCT is no longer correct for the daily CBCT because it is affected by the geometric uncertainties

related to the image-registration errors on the deformable anatomy.

Conversely, the HU is dependent on not only the RED but also on the effective atomic numbers, and the composition of the calibration phantom is different from that of human organs. The error in dose calculation caused by an inaccurate RED can be up to 4%²³. Dual-energy CT can be used to calculate the RED and effective atomic numbers in the human body directly^{24–28}. Some researchers have tried to use dual-energy CT to improve the accuracy of dose calculation^{29–31}. Saito et al.³⁰ presented a simple conversion from the energy-subtracted CT number by means of DECT to RED *via* a single linear relationship. They also used this conversion method for a TPS to provide an accurate and reliable inhomogeneity correction in treatment planning³¹. However, most studies have focused on FBCT and used dedicated dual-energy imaging devices. Some studies have used dual-energy cone-beam computed tomography (DE-CBCT)^{32–34}. Li et al.^{32, 32} implemented a dual-energy imaging method for virtual monochromatic and linearly mixed CBCT to reduce metal artifacts and enhance the contrast in image-guided radiation therapy. Ding et al.³⁴ investigated the feasibility of a three-material compositional measurement of water, lipid, and protein content of breast tissue with DE-CBCT for diagnostic purposes. However, reports focusing on dose calculation based on DE-CBCT are lacking.

Previously, we proposed a projection-based (not image-based) dual-energy decomposition³⁵ to calculate the RED of a head-like CIRS phantom using high- and

low-kV CBCT projections. Here, we describe the implementation of a dual-energy imaging and processing method to improve the image quality and accuracy of the dose calculation based on CBCT images in radiotherapy planning based on a Synergy™ (Elekta, Stockholm, Sweden) system. Performance of the proposed method was verified using three phantoms (Catphan 503; Alderson Rando Head and Neck (H&N); Pelvis).

2. MATERIALS AND METHODS

2.A. Phantoms

Five commercially available phantoms were used in the present study (Fig 1). The usage of these phantoms in this study is listed in Table 1. Please refer to the supplemental material for more technical specifications.

2.B. Imaging system and parameters

2.B.1. CBCT

CBCT was done using the X-ray volumetric imaging (XVI) of a Synergy™ machine (Elekta, Stockholm, Sweden). The X-ray source uses a rotating anode X-ray tube (Dunlee D604; Aurora, IL, USA) with peak tube potential of 150 kV and maximum current of 500 mA. The detector is an indirect detection flat-panel imager with a spatial resolution of $1,024 \times 1,024$ arrays of $0.4 \times 0.4 \text{ mm}^2$ pixels (RID1640-A11; PerkinElmer, Waltham, MA, USA). The source-to-axis distance and

source-to-detector distance are 1,000 mm and 1,536 mm, respectively. Projections can be acquired with small, medium and large field of view (FOV). The number of projections for a full 360° rotation is ≈ 660 . XVI software uses a cone-beam reconstruction process based on the Feldkamp–Davis–Kress algorithm. A S20 collimator and a M20 collimator were applied for the H&N-size phantoms scan (named hereafter as “H&N scan”) and body-size phantoms (named hereafter as “body scan”), respectively. Both S20 and M20 collimators impose a field size of $20 \times 20 \text{ cm}^2$ at the isocenter, but the M20 collimator shifts the imager laterally by 10 cm and creates an asymmetric beam to scan a medium FOV. In the dual-energy method, dual-energy imaging was performed with 70 kV & 16 mAs and 120 kV & 8 mAs for H&N and body scans, whereas the settings were 70 kV & 16 mAs and 120 kV & 10 mAs for body scans, respectively. For conventional SE-CBCT, the parameter settings were 100 kV & 16 mAs for H&N and 120 kV & 16 mAs for body scan. Projections were processed at high resolution to yield projections of dimension $1,024 \times 1,024$ with pixel size of $0.4 \times 0.4 \text{ mm}^2$. CBCT images were reconstructed with a voxel size of $1.0 \times 1.0 \times 1.0 \text{ mm}^3$.

2.B.2. FBCT

FBCT images were acquired on a helical CT system (SOMATOM Definition AS 40; Siemens Healthcare, Forchheim, Germany). Settings were 120 kV and 270 mA for the H&N phantom, and 120 kV and 280 mA for the Pelvis phantom, with a slice thickness of 1.0 mm, and pixel size of $1.0 \times 1.0 \text{ mm}^2$ for both phantoms.

2.C. Measurement of imaging dose

For estimation of the radiation exposure of CBCT imaging, the volume CT dose index ($CTDI_{vol}$) is measured. It equals the weighted CT dose index ($CTDI_w$) for CBCT. The center of the dosimetry phantom was placed at the isocenter position of the machine. An ionization chamber of type DCT10-RS/Lemo (IBA, Schwarzenbruck, Germany) was used for dose measurements. The dose was measured at five different positions of the phantom, starting from the phantom center to the four peripheral holes. Each measurement was repeated three times and the average value was taken. $CTDI_w$ is calculated from the five measurements according to the following formula:

$$CTDI_{vol} = CTDI_w = \frac{CTDI_{center} + 2 \times CTDI_{peripheral}}{3} \quad (1)$$

where $CTDI_{center}$ is the measured value in the center hole and $CTDI_{peripheral}$ is the average of all four peripheral measurements.

2.D. RED calculation methods

The implementation procedure of DE-CBCT imaging for the RED calculation and treatment planning is shown in Fig 2. The detailed steps to calculate the RED could be found in our previous work³⁵. The corresponding high and low X-ray projections acquired under the same imaging angle were registered using the Insight Segmentation and Registration Toolkit (ITK).

SE-CBCT and FBCT voxel values were calibrated with RED by acquisitions of CIRS 062 phantom with inserts of different RED. An aluminum rod (RED = 2.34) and a

titanium rod (RED = 3.79) were added to represent the high-density materials. Solid water (thickness, 15 cm) was attached to the CIRS phantom to simulate the additional scatter. Regions of interest (ROI) of size $5 \times 5 \text{ mm}^2$ within each insert were used to measure the mean value of HU. Two identical inserts were used to address the positional dependence of CT numbers of CIRS 062 inserts. One was arranged on the periphery and the other in or near the center. The mean CT value of the two ROI was used for HU–RED calibration. The HU–RED calibration curves were established for head-like and body-like configurations, respectively. Then, the curves were used to transform the corresponding HU to RED for subsequent phantom studies.

2.E. Analysis of Image quality

The Catphan 503 phantom (The Phantom Laboratory, Salem, NY) was employed in the study to mimic the human head and neck region (named hereafter as “Cat-H&N”), whereas insertion of this phantom into the oval body annulus was used to mimic the human body (named hereafter as “Cat-body”). The quantitative comparison indices were the reproducibility of the RED values, high-contrast resolution (HCR), uniformity, and signal-to-noise ratio (SNR).

Reproducibility of the RED values was measured using the CTP 404 module. It has several inserts comprising different materials: air, polymethylpentene (PMP), low-density polyethylene (LDPE), polystyrene, acrylic, polyoxymethylene (Delrin™) and polytetrafluoroethylene (Teflon). ROI of size $5 \times 5 \text{ mm}^2$ within each insert were used to measure the mean RED value. Correlation between the SE- or DE-CBCT-based RED calculation and the reference was assessed using the Pearson

correlation test. A Bland–Altman plot was used to identify the mean difference and 95% limits of agreement.

HCR was measured using the CTP 528 module. It was evaluated with the modulation transfer function (MTF) using a well-known method³⁶. The evaluation value was 50% of the MTF value ($MTF_{50\%}$) from the MTF curves.

Uniformity and SNR measurements were performed using the CTP 486 module. Five ROI of size $10 \times 10 \text{ mm}^2$, comprising one at the center and four at peripheral positions, were selected to measure the mean RED values. Uniformity was calculated as the maximum percentage difference between any two mean values. SNR was defined as the ratio of the mean to the standard deviation of the central ROI.

Image quality was also evaluated qualitatively with the Alderson Rando H&N phantom and the Pelvis phantom by a radiation oncologist.

2.F. Dosimetric evaluation

A nasopharyngeal cancer was simulated in the Alderson Rando H&N phantom. Target prescription was 69.96 Gy in 33 fractions for planned treatment volume for gross tumor volume and 60.06 Gy in 33 fractions for PTV. Brainstem, spinal cord, left parotid gland, right parotid gland, optic nerve, larynx, and temporomandibular joint (TMJ) were defined as the main OAR. The Pelvis phantom was used to simulate a tumor in the prostate gland with a target prescription of 67.5 Gy in 25 fractions for PTV. Main OAR were rectum, bladder, left femur and right femur.

The Analytical Anisotropic Algorithm in Varian's Eclipse TPS (Varian Medical Systems, Palo Alto, CA, USA) was used to create volumetric-modulated arc therapy (VMAT) plans and to calculate the dose distribution. The dose was calculated on grids of dimension $1 \times 1 \times 1$ mm.

A plan based on the reference FBCT ("Ref-CT") was first created for each phantom. This standard dose distribution was named "Ref-dose". To ascertain if the accuracy of dose calculation was improved by the dual-energy method, the VMAT plan was imported into DE-CBCT and SE-CBCT, respectively. CBCT images were registered rigidly to the corresponding Ref-CT before dose calculation. Rigid registration was carried out using the 'Image Registration' module built into the Eclipse TPS. Then, the dose distribution was calculated with the same planning parameters (field size, beam arrangement, control points, planned monitor units).

Calculated dose distributions on the SE-CBCT and DE-CBCT images were compared to the Ref-dose of the Ref-CT plans using global 3D gamma analysis³⁷. In the gamma analysis, a threshold dose value was set to 10% of the prescription dose, and the dose difference criterion was defined as percentage of the prescription dose. The agreement was assessed at the tolerance levels of 1%/1 mm and 2%/2 mm for the $\gamma < 1$ test. The average gamma values for targets and OAR were also calculated and were analyzed with paired *t*-tests between DE-CBCT and SE-CBCT. All analyses were performed with a p-value set to <0.05 .

3. RESULTS

3.A. Measurements of imaging dose

The results for the measured CTDI_{vol} data are listed in Table 2. The latter shows that the imaging dose of the dual-energy method increased by 1.0% and decreased by 1.3% for the H&N scan and body scan, respectively, compared with the SE-CBCT.

3.B Evaluation of image quality

Figure 3 shows the experimental results on the Catphan 503 phantom. The values of the Pearson correlation coefficient (r) obtained by comparing reference RED values, $\text{MTF}_{50\%}$, uniformity, and SNR are listed in Table 3. The Bland–Altman plot illustrates the mean difference between the SE- or DE-CBCT and the reference (Fig 4). The mean bias was reduced from 0.13 to 0.03 for the H&N scan and from 0.16 to 0.07 for the body scan with the dual-energy method. The 95% limit of agreement was -0.28 to 0.55 , -0.10 to 0.17 , -0.32 to 0.64 , and -0.20 to 0.35 for the SE-H&N scan, DE-H&N scan, SE-body scan and DE-body scan, respectively. The dual-energy method had a significant improvement in image uniformity (57.5% and 30.1%) and SNR (9.7% and 2.3%) in both scans, but did not affect the $\text{MTF}_{50\%}$ with respect to SE-CBCT images. It can also be seen in Fig 3 that the “cup-like”-shaped artifacts and image noise were reduced by the DE-CBCT method, although some artifacts were still residual due to uncorrected scatter.

RED images of the H&N phantom and Pelvis phantom acquired with SE-CBCT,

DE-CBCT and FBCT methods, respectively, are shown with the same window level in Fig 5 and Fig 6. The RED profiles illustrate that the percent difference between DE-CBCT and reference ($1.5\% \pm 6.2\%$, $3.1\% \pm 6.4\%$) was smaller than that between SE-CBCT and reference ($4.3\% \pm 9.1\%$, $8.5\% \pm 8.7\%$) for both H&N and Pelvis phantoms. The volume of the Pelvis phantom in the scan FOV was much larger than that of the H&N phantom, and it induced more scattering and beam hardening. Hence, the RED value of the Pelvis phantom calculated by the dual-energy method was slightly different to the reference, but was better than the single-energy method.

3.C. Dose comparison

3.C.1. Alderson Rando H&N

The dose distribution and gamma index maps for nasopharyngeal cancer are shown in Fig 7. The match of the dose between CBCT and reference was improved with DE-CBCT. The average gamma values of all the targets and OAR were reduced by DE-CBCT with both criteria: 1%/1 mm (0.13 ± 0.10 vs. 0.29 ± 0.20 , $p < 0.05$) and 2%/2 mm (0.08 ± 0.02 vs. 0.17 ± 0.02 , $p < 0.05$). The amount of voxels passing the gamma evaluation for 1%/1 mm (2%/2 mm) criteria was 73.5% (97.3%) to 99.1% (100%) when comparing SE-CBCT with the reference, and 99.0% (99.6%) to 100% (100%) when comparing DE-CBCT with the reference.

For SE-CBCT, gamma analyses using criteria of 1%/1 mm for the TMJ, optic nerve, and larynx had lower pass rate (73.5–94.5%) than other OAR. This may have been because the three organs are bone-/air-like, or around bone, and SE-CBCT could not

calculate the RED of these tissues accurately. The pass rate increased to nearly 100% with criteria of 2%/2 mm.

3.C.2. Pelvis

The dose distribution and gamma index maps for the tumor in the prostate gland are shown in Fig 8. The average gamma values were improved by DE-CBCT: 1%/1 mm (0.19 ± 0.07 vs. 0.38 ± 0.10 , $p < 0.05$) and 2%/2 mm (0.15 ± 0.05 vs. 0.32 ± 0.10 , $p < 0.05$). The pass rate using the gamma index method for SE-CBCT was 80.4–92.7% (1%/1 mm) and 96.1–99.1% (2%/2 mm), whereas for DE-CBCT it was 85.3–97.6% (1%/1 mm) and 98.6–99.7% (2%/2 mm). Largest discrepancies for SE-CBCT appeared in regions with beam-hardening artifacts, which were improved in DE-CBCT.

4. DISCUSSION

A dual-energy imaging method is based on two measurements at different energies. The attenuation differences among different materials vary with the X-ray energy. This variation between low- and high-energy imaging should be obvious enough to perform the basis material decomposition. The peak tube potential commonly used in the XVI system ranges from 70 kV to 120 kV. Hence, we used 70 kV and 120 kV for the dual-energy imaging method. As a contrast, the single-energy imaging method used 100 kV for the H&N scan and 120 kV for the body scan, which are the default settings in clinics. To reduce the imaging dose to patients, the exposure was reduced

for the high-energy scan and the measured CTDI_{vol} was within $\pm 1.5\%$ between dual-energy and single-energy scans.

In the case of Catphan, apparent relative errors were observed for Teflon in the RED calculation for SE-CBCT and DE-CBCT. These differences can be attributed to the effective atomic number (Z_{eff}) of materials³⁸. In the kV range of an X-ray spectrum in a CT scanner, photon interactions are due partially to the photoelectric effect, which has a larger dependence on Z_{eff} . However, Teflon ($Z_{\text{eff}}=8.5$) is not a suitable substitute for bone tissue ($Z_{\text{eff}}\approx 12.5$) because of its lower Z_{eff} , and our method has a larger relative error.

One of the major limitations of our dual-energy method is that the scatter has not been corrected in the proposed method. For cone-beam scanning, the scanning FOV is much larger and so it will induce more scatter, which is well known to be the most important factor affecting the quality of CBCT images. Scatter could cause cup-like-shaped artifacts and HU inaccuracies, and have been evaluated by Siewerdsen and Jaffrey³⁹. This also introduces errors in CBCT-based dose calculations. Hatton et al.⁴⁰ assessed the impact of HU-to-RED calibration phantom insert composition and phantom volume on the accuracy of dose calculation for CBCT. They found that the effect of phantom diameter can lead to dose differences for X-rays of 6 MV and 18 MV under bone inhomogeneities of up to 20% in extreme cases. Houser et al.⁴¹ and Yoo et al.⁴² demonstrated that dose calculation from CBCT

without scatter correction for large bodies had greater inaccuracy if the effects of inhomogeneities were included. One study ⁴³ used an anti-scatter grid to correct the scatter, and showed an accuracy of dose calculation $<2\%$ on phantoms. Another research team ⁴⁴ used body CBCT for dose calculation, with corrections to HU numbers for the effects of scatter and beam hardening. The accuracy of dose calculation was within 5% of the dose difference and 2 mm of the distance to agreement. In our dual-energy method, due to the lack of scatter correction, some artifacts were still residual and the accuracy of dose calculation using the dual-energy method was not as good as the reference, but was better than the single-energy method. Scatter correction using the dual-energy method to improve dose calculation of large bodies is beneficial. The accuracy of a combination of our method and scattering-correction methods merits further investigation.

Another drawback of the presented method will limit the effect of clinical application because of its two consecutive acquisitions. The phantoms used in our study were rigid and the projections acquired at high and low energy under the same imaging angle were registered using ITK software. However, if applied in clinics, the content of the corresponding projections can be different between the two projections because of breathing and other patient motions. Any possible misalignment of the two projections introduced by patient movement may cause artifacts in dual-energy imaging. The deformable registration may be useful and the problem can be resolved thoroughly by developments in the CBCT hardware, such as “rapid voltage switching” ⁴⁵ and “layer detector” ⁴⁶. These technologies perform the two acquisition within milliseconds or even simultaneously, and have been available in dual-energy

fan-beam CT to account for the motion between the two scans.

CBCT could provide the volumetric data of a patient in the treatment position. With evolution of radiotherapy evolving fast towards precise ART, establishment of validated methods to use CBCT data for patient positioning as well as calculation and verification of doses is important. However, the accuracy of RED calculated using the HU-RED calibration curve is usually not sufficient for dose calculation. We showed that a dual-energy method can be used to obtain more accurate REDs at the treatment position and enable CBCT data to be employed for routine calculation of doses. The Eclipse TPS was used for this work, but the method described could be replicated in other TPSs. For TPSs using PD to calculate the dose (e.g., Pinnacle TPS), a RED-PD table can be used to convert the RED to PD.

One potential application of our method is ART for patients whose changes in anatomy are larger compared with the planning CT. The original treatment plan can be mapped onto pre-treatment CBCT images to verify target coverage and OAR dose. If the deviations were too large, the treatment plan could be re-planned based on online CBCT. Moreover, the dual-energy method has a merit in proton stopping power computation⁴⁷, which gives it great potential in online proton treatment planning.

5. CONCLUSION

We investigated the image quality and the accuracy of dose calculation using DE-CBCT for head and pelvic sites. Calibration using the DE-CBCT method provided better image quality and dosimetric agreement with the planning FBCT for all phantoms. However, the volume of the pelvic phantom in the scan FOV was much larger than that of the head phantom, and it induced more scattering, which reduced the accuracy of dose calculation. These observations suggest that the DE-CBCT method can achieve higher accuracy of dose calculation, and could be applied to adaptive radiotherapy, but scatter artifact limits its performance for body sites.

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CONFLICTS OF INTEREST

The authors report no conflicts of interest with this study.

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Figure 1. The five phantoms: (a) Dosimetry phantom, (b) CIRS 062, (c) Catphan 503, (d) Alderson Rando Head and Neck, (e) Pelvis

Figure 2. Flowchart of the proposed method

Figure 3. RED images of the CTP-404 module in a Catphan 503 phantom: (a) and (c) SE-CBCT, (b) and (d) DE-CBCT

Figure 4. Pearson correlation and Bland-Altman plot. (a).The scatter plot graph illustrating the correlation between RED values obtained by CBCT and reference. (b)-(e). The Bland-Altman diagram showing the magnitude of the difference between the RED values obtained by CBCT and reference. The solid line represents the mean difference and the dashed lines represent the lower (-1.96 standard deviation [SD]) and upper limits ($+1.96$ SD) of agreement.

Figure 5. RED images of the Alderson Rando H&N phantom: (a) SE-CBCT, (b) DE-CBCT and (c) reference. (d) RED profiles through the region of interest, denoted by the white dotted line in (c).

Figure 6. RED images of the Pelvis phantom: (a) SE-CBCT, (b) DE-CBCT and (c) reference. (d) RED profiles through the region of interest, denoted by the white dotted line in (c).

Figure 7. Dose distribution and gamma index maps (1%/1 mm) for nasopharyngeal cancer: (a) reference CT image; (b) dose distribution; (c) SE-CBCT vs. the reference; (d) DE-CBCT vs. the reference

Figure 8. Dose distribution and gamma index maps (1%/1 mm) for a tumor in the prostate gland: (a) reference CT image; (b) dose distribution; (c) SE-CBCT vs. the reference; (d) DE-CBCT vs. the reference

Table 1. Experimental usage of the phantoms.

Phantom	Experimental usage
Dosimetry phantom	CTDI measurement
CIRS 062 phantom	HU-RED calibration
Catphan 503 phantom	Image quality analysis
Alderson Rando H&N phantom	Nasopharyngeal cancer simulation
Pelvis Phantom	Prostate cancer simulation

Table 2. Results of CTDI_{vol} measurements

	H&N scan		Body scan	
	SE-CBCT	DE-CBCT	SE-CBCT	DE-CBCT
CTDI _{vol} (cGy)	4.01	4.05	3.01	2.97

Table 3. Results of Catphan 503 phantom studies

	H&N scan		Body scan	
	SE-CBCT	DE-CBCT	SE-CBCT	DE-CBCT
r	0.963	0.997	0.954	0.983
MTF _{50%} (lp/cm)	0.520	0.521	0.403	0.402
Uniformity (%)	4.0	1.7	11.3	7.9
SNR	77.5	85.0	51.8	53.0

Table 4. Dose analysis for the Alderson Rando Head and Neck phantom

ROI	Imaging mode	Average Gamma value		Gamma index pass (%)	
		1%/1 mm	2%/2 mm	1%/1 mm	2%/2 mm
PTV	SE-CBCT	0.12	0.11	99.1	99.6
	DE-CBCT	0.06	0.05	99.7	100
PGTV	SE-CBCT	0.17	0.12	97.6	99.9
	DE-CBCT	0.08	0.06	99.8	100
Brainstem	SE-CBCT	0.32	0.12	97.6	99.9
	DE-CBCT	0.15	0.06	99.7	100
Spinal cord	SE-CBCT	0.09	0.08	99.0	100
	DE-CBCT	0.04	0.04	100	100
Left parotid gland	SE-CBCT	0.19	0.13	96.3	99.8
	DE-CBCT	0.09	0.06	99.2	99.9
Right parotid gland	SE-CBCT	0.18	0.13	96.4	99.8
	DE-CBCT	0.08	0.07	99.5	99.9
Optic nerve	SE-CBCT	0.80	0.23	73.5	100
	DE-CBCT	0.40	0.11	99.7	100
Larynx	SE-CBCT	0.27	0.25	94.5	99.9
	DE-CBCT	0.09	0.09	99.0	100
TMJ	SE-CBCT	0.35	0.26	94.3	100
	DE-CBCT	0.17	0.07	99.5	100
Whole body	SE-CBCT	0.29	0.24	94.7	97.3
	DE-CBCT	0.13	0.10	99.0	99.6

Table 5. Dose analysis for the Pelvis phantom

ROI	Imaging mode	Average Gamma value		Gamma index pass (%)	
		1%/1 mm	2%/2 mm	1%/1 mm	2%/2 mm
PTV	SE-CBCT	0.21	0.15	92.7	99.0
	DE-CBCT	0.08	0.07	96.2	99.6
Rectum	SE-CBCT	0.43	0.40	80.4	99.1
	DE-CBCT	0.26	0.19	85.3	99.7
Bladder	SE-CBCT	0.48	0.29	83.5	96.1
	DE-CBCT	0.24	0.14	89.6	99.1
Left femur	SE-CBCT	0.45	0.41	85.2	96.7
	DE-CBCT	0.22	0.20	90.6	98.6
Right femur	SE-CBCT	0.46	0.40	85.6	97.2
	DE-CBCT	0.21	0.20	90.0	98.9
Whole body	SE-CBCT	0.30	0.28	91.6	98.5
	DE-CBCT	0.15	0.12	97.6	99.3