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Original paper

# 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.

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## 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 planning [1]. 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 dose [2,3]. 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 distribution [8] in order to track the dose to targets and OARs and trigger the re-planning, if properly calibrated.

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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 phantom [9] 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 categories [12]. 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 planning [31]. 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 decomposition [35] 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

### 2.1. 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.

### 2.2. Imaging system and parameters

#### 2.2.1. 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 $1024 \times 1024$ arrays of $0.4 \times 0.4$ mm$^2$ pixels (RID1640-A11; PerkinElmer, Waltham, MA, USA). The source-to-axis distance and source-to-detector distance are 1000 mm and 1536 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–Davi s–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

**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 |

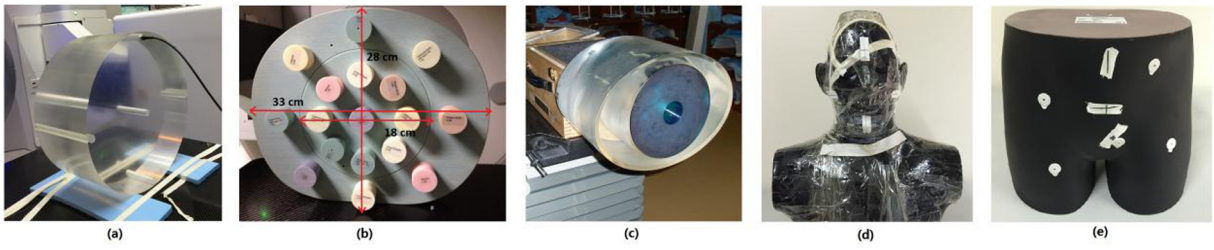


**Fig. 1.** The five phantoms: (a) Dosimetry phantom, (b) CIRS 062, (c) Catphan 503, (d) Alderson Rando Head and Neck, (e) Pelvis.

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