Medical Dosimetry 38 (2013) 153–159

Medical Dosimetry

journal homepage: www.meddos.org

Experimental evaluation of actual delivered dose using mega-voltage cone-beam CT and direct point dose measurement

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ARTICLE INFO

Article history:
Received 23 February 2012
Accepted 25 October 2012

Keywords:
MVCBCT
Point dose measurement
IGRT
Dose calculation
Experimental evaluation

ABSTRACT

Radiation therapy in patients is planned by using computed tomography (CT) images acquired before start of the treatment course. Here, tumor shrinkage or weight loss or both, which are common during the treatment course for patients with head-and-neck (H&N) cancer, causes unexpected differences from the plan, as well as dose uncertainty with the daily positional error of patients. For accurate clinical evaluation, it is essential to identify these anatomical changes and daily positional errors, as well as consequent dosimetric changes. To evaluate the actual delivered dose, the authors proposed direct dose measurement and dose calculation with mega-voltage cone-beam CT (MVCBCT). The purpose of the present study was to experimentally evaluate dose calculation by MVCBCT. Furthermore, actual delivered dose was evaluated directly with accurate phantom setup. Because MVCBCT has CT-number variation, even when the analyzed object has a uniform density, a specific and simple CT-number correction method was developed and applied for the H&N site of a RANDO phantom. Dose distributions were calculated with the corrected MVCBCT images of a cylindrical polymethyl methacrylate phantom. Treatment processes from planning to beam delivery were performed for the H&N site of the RANDO phantom. The image-guided radiation therapy procedure was utilized for the phantom setup to improve measurement reliability. The calculated dose in the RANDO phantom was compared to the measured dose obtained by metal-oxide-semiconductor field-effect transistor detectors. In the polymethyl methacrylate phantom, the calculated and measured doses agreed within about +3%. In the RANDO phantom, the dose difference was less than +5%. The calculated dose based on simulation-CT agreed with the measured dose within ±3%, even in the region with a high dose gradient. The actual delivered dose was successfully determined by dose calculation with MVCBCT, and the point dose measurement with the image-guided radiation therapy procedure.

Introduction

In radiation therapy, treatment planning for patients is performed by using computed tomography (CT) images acquired before start of the treatment course. Here, many patients with head-and-neck (H&N) cancer experience tumor shrinkage or weight loss or both during their treatment course. Thus, there are some deviations between the planning and actual beam delivery.

In Barker *et al.*,[1] gross tumor volume and parotid gland volume decreased by 1.8% and 0.6% per treatment day, respectively. Tumor regression was asymmetric, and the center of the tumor changed position with time.[1] Parotid glands shifted medially toward the high-dose region,[1-5] and this shift was highly correlated with weight loss.[1]

Additionally, daily positional error of patients also causes unexpected deviations from the CT images used for planning. Hong *et al.*[6] analyzed the daily positional error of patients with H&N cancer with conventional shrinking field design. They applied the daily positional errors to H&N intensity modulated radiation therapy plans and simulated dosimetric impact, which revealed gross tumor volume underdosing and normal tissue overdosing.

Therefore, for accurate clinical evaluation, it is essential to identify these anatomical changes and daily positional errors, and consequent dosimetric changes in actual beam delivery. To evaluate the actual delivered dose, direct dose measurement and dose calculation using cone-beam CT (CBCT) are proposed.

For direct evaluation of the actual delivered dose, Marcie *et al.*[7] placed metal-oxide-semiconductor field-effect transistor (MOSFET) detectors on the oral cavity with a molded mouth plate and

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http://dx.doi.org/10.1016/j.meddos.2012.10.005

performed the measurement >300 times for 48 treatment plans. The measured and calculated doses agreed within ±5% in 70% of all measurements, and 5% of the measurements agreed beyond ±10%. For the large discrepancy, the authors doubt the reproducibility of the detector and patient positioning. Improvement of the detector reproducibility and patient positioning is clearly needed to evaluate the actual delivered dose correctly and to determine the dosimetric consequences.

To verify patient positioning, the CBCT mounted on a linac was used as a localization system. For imaging, a kilo-voltage (kV) or mega-voltage (MV) beam with a large cone angle was combined with an electronic portal imaging device (EPID). Dose calculations have been reported with the kVCBCT system. Yang *et al.*[8] evaluated dosimetric performance computed with kVCBCT images, including intrascan organ motion. In a static phantom, doses computed based on the planning CT and the kVCBCT agreed within 1%. Notable differences (maximum 3% in the high-dose region) were found in a motion phantom. Because each patient has different scatter components that affect the Hounsfield Unit (HU) of the kVCBCT, Hu *et al.*[9] developed a region-of-interest (ROI) mapping method to generate a calibration curve of the kVCBCT number *vs* relative electron density. However, metal artifacts are associated with use of a kV beam. Thus, it is difficult to visualize complex anatomical structures of patients with H&N cancer with teeth crowns or dental implants and to obtain reliable CT numbers for dose calculation.

CT images by MV beam have fewer metal artifacts than CT images by kV beam.[10] Therefore, the authors attempted to perform the dose calculation using MVCBCT. Because MVCBCT has CT-number nonuniformity, even in a homogeneous phantom,[11] correction methods have been developed. Petit *et al.*[12] calculated scatter distribution from imaging objects and subtracted it from transmission images. This approach is somewhat difficult to use for routine implementation. Morin *et al.*[11] defined an ellipsoid-shaped geometrical model to characterize the CT-number nonuniformity in a head-sized water phantom. With their uniformity correction, the dose calculated with MVCBCT agreed well (3 mm, 3% in gamma-method) with that obtained by simulation-CT. However, those authors have not yet reported dosimetric evaluation.

In this study, the authors provide essential dosimetric evidence to prove the usefulness of dose calculation using MVCBCT. Dose evaluation was performed with a homogeneous cylindrical polymethyl methacrylate (PMMA) phantom. To improve the CT-number uniformity, a simple empirical correction method was developed, which involved analysis of the CT number of a cylindrical water phantom similar to an H&N site. Treatment processes from planning to beam delivery were performed for an H&N site of an anthropomorphic phantom. The image-guided radiation therapy (IGRT) procedure was utilized for phantom setup to improve measurement reliability. Additionally, the direct point dose measurement with MOSFET detectors under the IGRT was evaluated simultaneously.

Methods and Materials

MVCBCT

MVCBCT mounted on an ONCOR linear accelerator (Siemens Medical Solutions, Concord, CA) was used. A beam with a large cone angle combined with an EPID was employed for MVCBCT imaging. The EPID has an amorphous-silicon flat panel of OptiVue 1000ST (Siemens Medical Solutions, Concord, CA) with a 41 cm × 41 cm active detection area and 1024 × 1024 spatial resolution. The voxel size was 1.07 mm × 1.07 mm × 1 mm and field of view was 27.4 cm × 27.4 cm. The CT images were reconstructed by using 200 projections during 200° gantry rotation. The image reconstruction process and the relationship between patient dose and image quality are described elsewhere.[10,13-17]

Because MVCBCT has a smaller field of view than the trunk width, dose evaluation was performed only for the H&N site. In MVCBCT imaging, the field size was 27.4 cm × 27.4 cm at a source-to-axis distance of 100 cm, and 15 monitor units were exposed.

CT-number uniformity correction

The CT-number uniformity of MVCBCT is <30% in a head-size water phantom.[11] Therefore, when using MVCBCT for dose calculation, the CT-number uniformity should be evaluated and corrected as needed. A water phantom (20 cm in diameter, 20 cm in length) was used as a substitute for the H&N site.

Coordinate axes used in the CT-number analysis were defined as shown in Figure 1A. Figure 1B and C show the flatness of the CT-number profiles along a radial direction, *r*, without and with the uniformity correction, respectively. Without the uniformity correction, the CT number ranged from about ± 50 HU at $z = 0.5$ mm from the center of the longitudinal axis, and from + 50 HU to + 150 HU at $z = 60.5$ mm. The flatness of the CT-number profiles was about 5% at $z = 0.5$ mm and 10% at $z = 60.5$ mm. Without the uniformity correction, the CT-number uniformity was different between $z = 0.5$ mm and $z = 60.5$ mm. The average CT number inside a transverse plane increased with *z*. Furthermore, the CT number increased with *r* at $z = 0.5$ mm, but decreased with *r* at $z = 60.5$ mm.

To improve the CT-number uniformity, correction factors were obtained by using the CT numbers of the water phantom. The CT-number variation of the water phantom, $CT(r, z)$, was expressed by

$$CT(r,z) = a(z)r + b(z) \tag{1}$$

where *a* and *b* are the slope and intercept, respectively. In a machine specification, the CT-number uniformity is described as being within ± 40 HU in a water-equivalent phantom. For this reason, a simple linear approximation (rather than a polynomial or higher approximation) was employed for the fitting. The fitting was performed on each transverse plane.

Correction factors, $CF(r, z)$, were calculated so as to be the CT number at the center of $z = 0$ mm, CT_0, which indicated calibrated CT number for water obtained from the relationship between HU and electron density.

$$CF(r,z) = \frac{CT_0}{CT(r,z)} \tag{2}$$

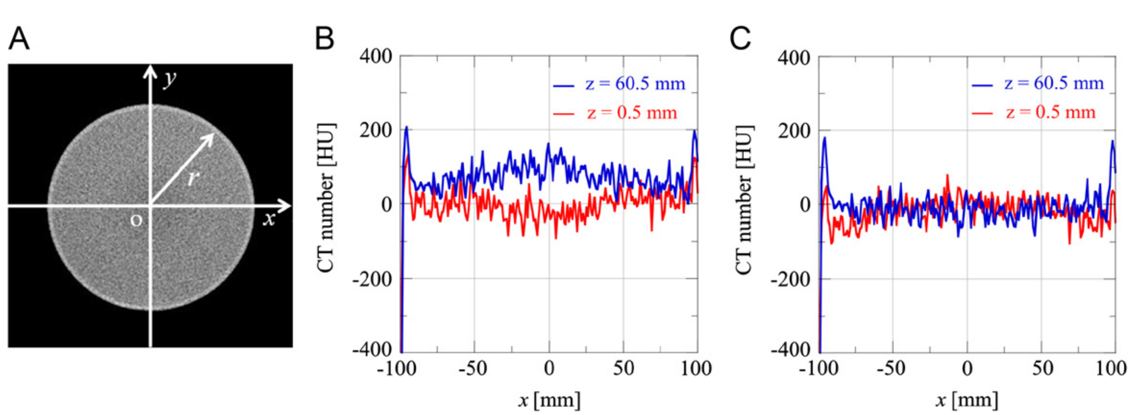

Fig. 1. CT-number uniformity in the water phantom. Profiles of the CT number are shown in the *x* direction at $z = 0.5$ mm and $z = 60.5$ mm from the center of the longitudinal axis. (A) Definition of coordinate axes used in the CT number analysis. Profiles without (B) and with (C) uniformity correction.