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Original paper

Image quality of four-dimensional cone-beam computed tomography obtained at various gantry rotation speeds for liver stereotactic body radiation therapy with fiducial markers

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ABSTRACT

In this study, qualities of 4D cone-beam CT (CBCT) images obtained using various gantry rotation speeds (GRSs) for liver stereotactic body radiation therapy (SBRT) with fiducial markers were quantitatively evaluated. Abdominal phantom containing a fiducial marker was moved along a sinusoidal waveform, and 4D-CBCT images were acquired with GRSs of 50–200° min^{-1}. We obtained the 4D-CBCT projection data from six patients who underwent liver SBRT and generated 4D-CBCT images at GRSs of 67–200° min^{-1}, by varying the number of projection data points. The image quality was evaluated based on the signal-to-noise ratio (SNR), contrast-to-noise ratio (CNR), and structural similarity index (SSIM). The fiducial marker positions with different GRSs were compared with the setup values and a reference position in the phantom and clinical studies, respectively. The root mean square errors (RMSEs) were calculated relative to the reference positions. In the phantom study, the mean SNR, CNR, and SSIM decreased from 37.6 to 10.1, from 39.8 to 10.1, and from 0.9 to 0.7, respectively, as the GRS increased from 50 to 200° min^{-1}. The fiducial marker positions were within 2.0 mm at all GRSs. Similarly, in the clinical study, the mean SNR, CNR, and SSIM decreased from 50.4 to 13.7, from 24.2 to 6.0, and from 0.92 to 0.73, respectively. The mean RMSEs were 2.0, 2.1, and 3.6 mm for the GRSs of 67, 100, and 200° min^{-1}, respectively. We conclude that GRSs of 67 and 85° min^{-1} yield images of acceptable quality for 4D-CBCT in liver SBRT with fiducial markers.

1. Introduction

Stereotactic body radiation therapy (SBRT) has been introduced as an alternative to standard treatment modalities, such as surgical resection and radiofrequency ablation, for liver tumors [1]. Liver SBRT with dynamic treatments in fewer fractions, such as intensity-modulated radiation therapy (IMRT) or volumetric modulated arc therapy (VMAT), needs treatment planning considering the motion blurring effect and the interplay effect on the dose calculation [2,3]. Furthermore, while administering SBRT for liver tumors, accurate target localization using image-guided radiation therapy (IGRT) is important, given the effects of respiration on inter- and intra-fractional tumor motion and the effect of the daily setup error on the patient position [4,5]. Liver tumors can be difficult to visualize using IGRT modalities, due to the lack of soft tissue contrast. Accordingly, fiducial markers are employed as tumor surrogates to enable more accurate target localization than would be achievable using the bone anatomy or diaphragm position [6,7].

Currently, four-dimensional cone-beam computed tomography (4D-CBCT) is used to assess tumor motion during the SBRT of liver tumors [8–10]. In our previous study, we reported that liver tumor motion during the planning simulation using 4D-CBCT could represent liver tumor motion during SBRT and suggested that 4D-CBCT can be introduced for the treatment and planning of liver SBRT as a useful modality for the internal target volume (ITV) definition of tumor motion at institutions that do not have 4D-CT [10]. However, the 4D-CBCT device requires a low gantry rotation speed (GRS; typically, 50° min^{-1}) to generate a sufficient volume of projection data for each phase (of which there are typically 10) during image reconstruction [8–15]. A lower GRS results in a longer acquisition time, which may cause patient

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discomfort and increase intra-fractional tumor and organ-at-risk (OAR) movement. Yoganathan et al. [14] used a dynamic thorax phantom and an image quality phantom to evaluate the effects of the GRS on 4D-CBCT in terms of ITV definition and image quality and demonstrated that a higher GRS might lead to ITV underestimation and reduced image quality. Furthermore, Santoso et al. [15] investigated the effect of the GRS on the image quality and imaging dose using a Varian 4D-CBCT device (Varian Medical System, Palo Alto, CA, USA). They showed that altering the GRS changes the number of projections employed for reconstruction, affecting both the image quality and imaging dose. The imaging dose in a 4D-CBCT scan would linearly increase with a lower GRS.

It should be noted that the above-mentioned previous reports described phantom studies exclusively. Moreover, Ahmad et al. [11] evaluated the effects of the tumor position accuracy at different acquisition times inversely proportional to the GRS by reducing the number of 4D-CBCT projection data for one lung cancer patient receiving radiotherapy and found that the position errors decreased with increasing acquisition time. However, there are no previous reports on evaluating the effects of the GRS on 4D-CBCT-based image guidance for patients who underwent liver SBRT with fiducial markers. Therefore, the objective of the present study was to evaluate the quality of 4D-CBCT images obtained using various GRSs quantitatively as well as the ability of the images to guide liver SBRT appropriately with fiducial markers. Here, we investigated the effects of the GRS on 4D-CBCT images in both phantom and clinical studies.

2. Methods and materials

2.1. 4D-CBCT data acquisition

The 4D-CBCT scans were acquired using an Elekta Symmetry System (Elekta Oncology Systems, Crawley, UK), and the projection data were sorted in 10 respiratory-phase bins (0%–100%) with 10% phase windows [8]. End-inspiration corresponds to the 0% phase, and end-expiration corresponds to 50% phase [9,11]. The projection data were acquired with the exposure parameters of 120 kV, 20 mA, and 16 ms per frame. The frame rate of the kV detector was 5.5 frames/s. The number of projection data points (N) acquired during 4D-CBCT data acquisition can be calculated as follows [14]:

$$\mathrm{N}=\frac{\theta_1-\theta_2}{GRS}\times fps, \tag{1}$$

where θ_1 and θ_2 are stop and start angles of the gantry, respectively. *GRS* and *fps* are the gantry rotation speed (degrees per second) and frame rate (frames per second) of the kV detector, respectively. 4D-CBCT data were acquired in the kV detector "small mode" setup. In the "small mode," the kV detector is placed symmetrically with respect to the center of the field of view (FOV). The small mode was designed to obtain projection data from a 200° gantry angle range with an S20 collimator corresponding to a FOV of 27 cm × 26 cm. Image reconstruction was performed using the Feldkamp filtered-back-projection (FBP) algorithm with a voxel size of 2 mm and a low-resolution reconstruction preset, as used clinically. The number of projection data with GRSs from 50 to 200° min^{-1} ranged from 341 to 1364.

2.2. Phantom study

To assess 4D-CBCT image quality, we used a three-dimensional (3D) abdominal phantom (Model 057A; CIRS Inc., Norfolk, VA, USA), which simulates the abdomen from the thoracic vertebrae (T9/T10) to the lumbar vertebrae (L2/L3) and thereby contains internal structures such as the liver, portal vein, partial kidneys, partial lung, aorta, vena cava, spine, and ribs [16]. A gold fiducial marker (diameter: 2 mm; iGold; Medilit, Tokyo, Japan) was inserted into the liver of this phantom to evaluate the fiducial marker position accuracy. The phantom was then placed over a respiratory motion phantom (QRP Series; Qualita, Nagano, Japan) programmed to move in a sinusoidal waveform in the superior-inferior (SI) direction. We selected a motion amplitude of 10 mm and breathing period of 4 s to simulate plausible liver tumor motion [10,11]. The 4D-CBCT scans were performed with GRSs of 50, 67, 85, 100, and 200° min^{-1}.

Furthermore, based on the 4D-CBCT projection data obtained with a GRS of 50° min^{-1}, we generated 4D-CBCT images at other GRSs by varying the number of projection data points. From Eq. (1), the number of projection data points for GRSs of 67, 100, and 200° min^{-1} was calculated based on the assumption that any change in the GRS does not influence the gantry angle range and the frame rate of the kV detector. By reducing the number of projection data points with a GRS of 50° min^{-1} by factors of 3/4, 1/2, and 1/4 using Elekta XVI software (version 4.5; Elekta Oncology Systems, Crawley, UK), we simulated GRSs of approximately 67, 100, and 200° min^{-1} [11]. The number of projection data points was discretized at the frame sampling angles of 0.2, 0.3, and 0.6°, corresponding to GRSs of 67, 100, and 200° min^{-1} using the "Frame Active/Inactive" function in the Elekta XVI software. Moreover, the 4D-CBCT images generated by a simulation were compared to those obtained by the 4D-CBCT scans using image quality metrics and the fiducial marker position accuracy, as described later.

2.3. Clinical study

This retrospective study was approved by the institutional review board of our hospital. All patients provided informed consent for treatment and the use of 4D-CBCT and its images prior to this study. Six patients who underwent 4D-CBCT image-guided liver SBRT between May 2014 and May 2016 were involved in the study. Abdominal compression was applied to all patients to achieve reproducible tumor motion. One or two gold fiducial markers were implanted into the liver of each patient via the percutaneous transhepatic approach [17]. We then obtained 4D-CBCT projection data for each patient at a GRS of 50° min^{-1}. Moreover, by reducing the number of original projection data points, we simulated 4D-CBCT images at other GRSs in the same manner as the phantom study and obtained 4D-CBCT images at GRSs of approximately 67, 100, and 200° min^{-1} [11].

2.4. Image quality analysis

4D-CBCT image quality was evaluated using the signal-to-noise ratio (SNR), contrast-to-noise ratio (CNR), and structural similarity (SSIM) index [18,19]. The means and standard deviations (SDs) of the measured values over all 10 phases were compared among the GRSs. The SNR was defined over a smooth liver region, R, with a volume of 1 cm^3 (Fig. 1). After measuring the intensities in R using Elekta XVI software, the SNR was calculated using the following formula:

$$\mathrm{SNR}=\frac{\mathrm{Mean(R)}}{\mathrm{SD(R)}}. \tag{2}$$

The CNR was defined based on the intensities of the fiducial marker and surrounding liver (R), as shown in Fig. 1, and calculated using the following formula:

$$\mathrm{CNR}=\frac{I_{marker}-\mathrm{Mean(R)}}{\mathrm{SD(R)}}. \tag{3}$$

The SSIM index was employed to quantify the visibility of errors between two images by using three image characteristics: luminance, contrast, and structure [19]. The SSIM index is defined as

$$\mathrm{SSIM}(x,y)=\frac{(2\mu_x\mu_y+C_1)(2\sigma_{xy}+C_2)}{(\mu_x^2+\mu_y^2+C_1)(\sigma_x^2+\sigma_y^2+C_2)}, \tag{4}$$

where μ_x and μ_y are the local means, σ_x and σ_y are the standard deviations, σ_{xy} is the cross-covariance for images x and y, and C_1 and C_2 are