Feasibility of proton tomosynthesis in proton therapy

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ABSTRACT

Currently, the patient dose of proton therapy and treatment planning are based on the x-ray photon CT (computed tomography) data. However, material interactions between protons and photons are different. Proton tomograms are useful to make with accuracy the dose calculations for planning and positioning of patients. For applying CT techniques, many projection images during at least half rotation are need. It makes extremely high patient dose. We apply the proton tomosynthesis, which is the limited angle tomography using proton-beams. The proton tomosynthesis may provide more accuracy of dose calculation and verifications with less patient dose. We describe the concept of the proton tomosynthesis and demonstrate its performance with GEANT4 simulation and experiments. We confirmed the feasibility of proton digital tomosynthesis through comparative analysis with the x-ray photon methods. This study is useful for proton therapy planning and verification.

Keywords: Proton therapy, Proton tomosynthesis

1. INTRODUCTION

Proton therapy is a rapidly expanding modality and the number of new facilities being established continues to show substantial growth. The advantage of proton relative to photon beams for radiation therapy lies in their superior dose distributions due to the Bragg peak [1-2]. In the proton-beam radiation therapy, a prescribed radiation dose needs to be delivered to a tumor region while minimizing the dose to surrounding normal tissues and healthy organs at risk. Currently, the patient dose calculation of proton therapy and treatment planning are based on the x-ray photon CT (computed tomography) data. However, material interactions between protons and photons are different. Proton tomograms are useful to make with accuracy the dose calculations for planning and positioning of patients [3-4]. For applying CT techniques, many projection images during at least half rotation are need. It makes extremely high patient dose. Besides the pass length of beam is limited. We apply the proton tomosynthesis, which is the limited angle tomography using proton-beams. Due to tomosynthesis allows only acquire few projection, this method is more...
accessible and make a simple. Although the theoretical framework describing tomosynthesis had been initialized in 1930s, its renewed interest has been in the spotlight since large-area flat-panel detectors had been commercially available in the late 1990s [5].

We apply the proton tomosynthesis, which is the limited angle tomography using proton-beams. The proton tomosynthesis may provide more accuracy of dose calculation and verifications with less patient dose. The purpose of this study is to investigate the feasibility of proton tomosynthesis technologies in the gantry treatment room of a proton therapy system.

2. MATERIALS AND METHODS

2.1 System description

A proton-beam therapy facility at the National Cancer Center (NCC) in Korea utilizes hydrogen ions, which can be accelerated up to 230 MeV by the cyclotron with a magnetic field intensity of 2.3 Tesla. The maximum beam current is 300 nA. Specially designed the range modulators and the beam current modulation were used for protons to attenuate as the mega-voltage x-rays. In this study, we employed the extended dose range film (EDR, Eastman Kodak Co., Rochester, NY) for measuring the attenuated proton-beams. For comparing the quality of proton tomosynthesis data, we acquired the photon tomosynthesis image using on-board x-ray imaging system under the same geometric condition. In the gantry treatment room for proton therapy, two pairs of x-ray imaging systems (x-ray tubes and large-area flat-panel detectors) are installed in orthogonal directions, as shown in Figure 1. For acquiring x-ray tomosynthesis data, a set of x-ray tube and detector located in parallel to the proton-beam direction was used. In this set, the x-ray tube (A2777, Varian Medical Systems, Inc., USA) is embedded inside the proton-beam nozzle. The detector (PaxScan 4030E, Varian Medical Systems, Inc., USA) is based on arrays of hydrogenated amorphous silicon (a-Si:H) thin-film transistors (TFTs) in combination with a-Si:H photodiodes, and the overlying phosphor screen. The field-of-view (FOV) of the detector is ~30 × 40 cm². However, considering the shadow due to the nozzle of the proton-beam head, the actual FOV is reduced to ~25 × 36 cm².

Figure 2. A sketch describing image reconstruction in cone-beam geometry. For the reconstruction of voxel value at \( r = (x, y, z) \), the contribution of projection value at \( (\xi, \eta) \) in the planar detector obtained at the projection angle \( \beta \) is illustrated. The projection signal is back-projected along the line, which is contained in the tilted fan beam.
2.2 Digital tomosynthesis

The basic reconstruction algorithm for tomosynthesis is the shift and add (SAA) method that shifts each of the projection images by a given amount and then adds them together to make objects in a given plane to sharpen while objects from other planes to be blurred [6]. This SAA method is basically equivalent to the simple backprojection (BP) method [7]. It is, however, insufficient to use SAA alone for high-quality tomosynthesis because of blur from outside of the plane of interest. Therefore, the SAA or BP method requires deblurring algorithms to compensate for the out-of-plane blur.

For the cone-beam digital tomosynthesis (CBDT), we adopt CBCT with insufficient projection data obtained from a limited angle scan. The strategy is based on the conventional filtered backprojection (FBP) method with additional filters to control the incomplete frequency responses due to sampling in limited angular range in cone-beam geometry.

Cone-beam geometry with planar detectors is very attractive because of the simple geometry and quick scan for large volumetric objects, and thus it is popular in many technical applications. However, since a single circular x-ray source trajectory in cone-beam geometry does not provide complete Radon data, the exact 3D image reconstruction is not available following the Tuy-Smith sufficiency condition [8]. Instead, an approximate version of the exact reconstruction developed by Feldkamp, Davis, and Kress, the so-called FDK method [9], is typically used. The FDK method is a simple extended version of the conventional FBP method in longitudinal direction by considering the cone angle.

Figure 2 illustrates a reconstruction scheme of the object \( f(r = (x, y, z)) \) with projection \( p(\xi, \eta) \) obtained at angle \( \beta \).

Following the notations described in Figure 2, the FDK reconstruction formula for the range of projection angle from \( \beta_{\text{min}} \) to \( \beta_{\text{max}} \) can be given by

\[
f(r) = \int_{\beta_{\text{min}}}^{\beta_{\text{max}}} \frac{L^2}{(L - r \cdot (n_\xi \times n_\eta))^2} \left[ \tilde{p}_\beta(\xi, \eta) * h(\xi, \eta) \right] d\beta,
\]

where

\[
\tilde{p}_\beta(\xi, \eta) = \frac{L_D}{\sqrt{L_D^2 + \xi^2 + \eta^2}} p_\beta(\xi, \eta).
\]

The symbol "*" implies one-dimensional (1D) convolution operation. If we express more explicitly, Eq. 1 would be modified to

![Figure 3](image-url) Modeled nozzle at National Cancer Center (Korea) designed using GEANT4 toolkit. A total of eight components were modeled and placed in the beam line. And phantom and film were located on center of gantry.
\[ f(x, y, z) = \int_{\beta_{\text{max}}}^{\beta_{\text{min}}} d\beta \left[ \frac{L^2}{\Lambda} \times \int^\infty_{-\infty} d\xi \left( \frac{L_D}{\sqrt{L_D^2 + \xi^2 + \eta^2}} \right) p_\beta(\xi, \eta) \times h \left( \frac{L_D x}{\Lambda} - \xi, \frac{L_D y}{\Lambda} \right) \right]. \]  

The weighting term on the projection signal is to correct the cone beam into the parallel beam, and thus it can be simply expressed as \( \cos(\phi) \). The other weighting term \((L/\Lambda)^2\) is to correct the contribution of projection signal toward an element or voxel constituting the object by considering distance or magnification. The high-pass filter \( h \) is used to compensate the distance-weighted blur arising in the backprojection operation and only depends upon \( \xi \) direction in projection signals. Therefore, the FDK method or Eq. 3 describes that the 1D filtered pre-weighted projection signals are added in each virtual voxel, which eventually constitutes the object, considering magnification between the voxel and the corresponding projected pixel over the scanned angular range.

If we designate \( u, v, \) and \( w \) as the Fourier conjugates corresponding to \( x, y, \) and \( z \), respectively, in the space Cartesian coordinates, the high-pass or ramp filter in Eq. 3 can be described in the Fourier domain as

\[ H_{k_{\text{SA}}} (u, v) = 2 \beta_{\text{scan}} \sqrt{u^2 + v^2}, \]  

where \( \beta_{\text{scan}} \) denotes half the total scan angle. Since this ramp filter is vulnerable to high-frequency noise, we combined it with a spectral apodizing filter [10],

\[ H_{k_{\text{SA}}} (u) = \frac{1}{2} \left[ 1 + \cos \left( \frac{\pi u}{k_{\text{SA}}} \right) \right], \]  

which limits the bandwidth of the projection images hence reduces high-frequency noise and aliasing. The parameter \( k_{\text{SA}} \) determines the bandwidth and is typically multiples of the Nyquist frequency in \( u \) direction, \( u_{\text{N}} \).

Incomplete sampling over the limited angular range results in an out-of-plane artifact in the reconstructed images. We incorporated an additional slice thickness filter to limit the frequency response in the depth or \( z \) direction such that [10]

![Figure 4.](image)

**Figure 4.** Electron density cylinder block with diameter of 28 mm and height of 70 mm (solid water, lung and cortical bone). Making a difference with depth, we added block with several height 30, 50 and 70 mm. And in order to demonstrate spatial resolution, resolution hole pattern between 2.5 and 6.25 lp/cm were used.
\[ H_{ST}(w) = \frac{1}{2} \left[ 1 + \cos \left( \frac{\pi w}{k_{ST}} \right) \right]. \] (6)

The slice thickness filter plays the same role as the spectral apodizing filter but the application direction is different. Similarly, the parameter \( k_{ST} \) determines the bandwidth and is typically multiples of \( u_x \times \tan(\beta_{\text{scan}}) \).

It should be noted that we realized two apodizing filters with the Hann window functions [11].

2.3 Simulation

Figure 3 shows the modeled National Cancer Center (Korea) nozzle, which is designed using GEANT4 toolkit. A total of eight components were modeled and placed in the beam line [12]. This toolkit allowed us to successfully model the beam delivery system installed at the National Cancer Center of Korea. And phantom and film were located on center of gantry. The phantom for simulation and experiment is shown in Figure 4. We used electron density cylinder block with diameter of 28 mm and height of 70 mm (solid water, lung and cortical bone). Making a difference with depth, we added block with several height 30, 50 and 70 mm. And in order to demonstrate spatial resolution, we used resolution hole pattern between 2.5 and 6.25 lp/cm.

2.4 Experimental

Specially designed beam current modulation of the proton beam provides the beam eye’s view of proton. We can use the special beam condition to verify the CT calibration of the radiation treatment planning system and to examine the proton stopping power of various materials [3-4]. The depth dose distribution of the proton beam used in irradiation of EDR film. Since normal SOBP (Spread-Out Bragg Peak) beams can not make optical density differences in depth, this photon-like proton beam is needed. For acquiring tomosynthesis data, the phantom was scanned with a rotational angle step of 4° from -20° to 20°. Therefore the total number of projections was 11 views. The region was scanned at 85 kVp and 40 mAs for the photon digital tomosynthesis. The reconstructed three-dimensional (3D) image format was \( 512 \times 512 \times 186 \) voxels with a voxel size of 0.4 mm and slice thickness of 1 mm. The elapsed time for the backprojection operation was approximately 6 seconds with a PC (Intel Quad Core 3.0 GHz CPU, 4 G RAM).

![Figure 5. The normalized dose profile is the dose in water along the beam axis measured by the ion-chamber in water bath. Region of entrance dose within 20 mm is non-linear. Measured proton beams have linearity from 20 mm to 160 mm. This linear region is fitted by the first order polynomial regression. Regression factor is 0.9971.](https://example.com/image.png)
2.5 Geometric calibration

Protons accelerated in cyclotron were delivered like as the point beam through the beam delivery system with several magnets. The delivered point beam was scattered by scatterer. Finally, scattered one was like parallel-beam after scatterer. Thus we assumed that proton geometry is parallel and film is located in iso-center of gantry. To determine geometric parameters for photon, such as the source-to-detector distance and tilted angles of the detector, we employed the analytic algorithm and corresponding calibration phantom, which were successfully demonstrated in the megavoltage cone-beam computed tomography system [13]. The rotationally tilted angles of the detector are defined around three axes; yaw around x-ray beam path, roll around the axis-of-rotation of gantry, and pitch around the sagging axis of the detector [13]. The phantom consists of two sets of 12 steel ball bearings (BBs) spaced evenly at 30° circularly and each set is separated by 160 mm in height along the acrylic cylinder. Each BB has a diameter of 4.7 mm. The BBs trajectories are calculated from the projection data for 360° gantry rotation. The position of BBs is measured by the circle voting method of the Hough transformation about the edge strength map of each projection. Circular patterns of top- and bottom-BBs layers looks like two ellipse in projection acquired at each angle. The geometric parameters are estimated from the length of short-axis, centroid, slanted angle and parallel line of the top-to-bottom BBs [13].

3. RESULTS

Figure 5 shows that normalized dose profile is the dose in water along the beam axis measured by the ion-chamber in water bath. Region of entrance dose within 20 mm is non-linear. For considering this region, we added acrylic stage with thickness of 20 mm in front of film. Measured proton beams have linearity from 20 mm to 160 mm. This linear region is fitted by the fist order polynomial regression. Regression factor is 0.9971. The attenuated radiation in photon is calibrated by minus logarithm. However, in proton case, fitted curve is used for calibration [3-4].

Based on the measurements of the calibration phantom and analytic algorithm to calculate phantom trajectories, the tilted angles of the detector were analyzed during gantry rotation. The trajectory of the source and detector was evaluated and the excursion was well within 0.5 mm. The distances from the x-ray source to the gantry iso-center and from the detector to the iso-center were estimated to 1517.50 mm and 591.12 mm with a variation of 12 mm or less, respectively.

![Figure 6. Acquired radiograms using photon and proton. (a) and (b) are images of x-ray radiography, and (c) and (d) are ones of proton after scanning. (a) and (c) are acquired at vertically position with zero degree of gantry angle. (b) and (d) are got with 20 degree of gantry angle.](image-url)
Acquired radiograms using photon and proton are shown in Figure 6. (a) and (b) are images of x-ray radiography, and (c) and (d) are ones of proton after scanning. (a) and (c) are acquired at vertically position with zero degree of gantry angle. (b) and (d) are got with 20 degree of gantry angle. Big circle in figure is shadow from snout of nozzle. Four small balls are steal ball for alignment of each image. Due to films can’t be rotated with gantry and located on couch, shadow of nozzle is like an oval in Figure 6(d).

The reconstructed proton and photon tomosynthesis images with 11 projection views for the scan angle of 40° are demonstrated in Figure 7. The quality is very promising. Figure 7 compares the cross-sectional images at $z = 0$ mm (or central slices [Figure 7(a,d)]), $z = 20$ and 40 mm reconstructed with photon and proton-beam: (a-c) photon and (d-f) proton. We can distinguish up to 6.5 lp/cm in photon, and 3.5 lp/cm in proton. The reconstructed three-dimensional (3D) image format was $512 \times 512 \times 186$ voxels with a voxel size of 0.4 mm and slice thickness of 1 mm.

4. DISCUSSION AND CONCLUSIONS

The maximum penetrated length is shorter than 10 cm in this simulated and experimented phantom. Thus, the effects of scattering in material are little [14]. If the length of penetration is up to 20 cm, it is hard to reconstruct tomogram using this simple method. Further, we have to consider the method with scatter, experiment the phantom in the water bath [15]. Moreover, we must optimize the number of projections and scan angle. Geant4 simulation is still running on Grid.

Acquisition of the projection data for proton digital tomosynthesis in this study has been carried out manually at every step angle of gantry rotation because of the pulsed x-ray beam output and the snapshot operation of the flat-panel detector. Accordingly the acquisition time was about 10 minutes for 11 projection views. This inconvenience would be easily remedied if we replace the digital proton imaging system with the proton source and the digital proton detector.

Proton digital tomosynthesis for image-guided radiation therapy has been developed and currently practically utilized. However, there has been very little attention paid to proton therapy. In this study, we have investigated the feasibility of proton tomosynthesis for image-guided proton therapy. From the reconstructed phantom images, the proton digital tomosynthesis system in the gantry treatment room will be very useful as a primary patient alignment system for image-
guided proton therapy. The proton tomosynthesis may provide more accuracy of dose calculation and verifications with less patient dose.

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