Development of an Iterative Reconstruction Method for Low Dose CBCT in Proton Therapy Patient Positioning

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***Abstract*—Cone Beam Computed Tomography (CBCT) is used to determine patient position before each irradiation, which results in increased radiation exposure for the patient. To reduce the patient exposure, reduction of either X-ray imaging time or X-ray tube current is necessary. However, this leads to increased noise and reduced contrast in images. To meet this requirement, We have developed an image reconstruction method employing an iterative algorithm that is robust to noise and that can achieve high image contrast. In the iterative image reconstruction process, a system matrix corresponding to the measured geometry is required. In gantry-mounted CBCT, the X-ray tube and the flat panel detector (FPD) deviate some from their ideal position, resulting in a different geometry for each measurement angle. However, full calculation of system matrices requires a large data capacity or significant computation time. To overcome these issues, our method calculates the system matrix for an ideal geometry in advance and performs positional deviation correction on the projection data, allowing us to obtain a reconstructed image with positional deviations taken into account. We applied both the Feldkamp method and the proposed method to digital phantom data with positional deviations and Poisson noise, and by comparing the Detectability Index results, we were able to confirm improved noise reduction and image contrast.**

Keywords*—Cone Beam Computed Tomography; CBCT; Iterative reconstruction; Low dose CT; gantry-mounted CBCT

I. Introduction

CBCT imaging before each irradiation in proton therapy would increase the radiation exposure for a patient. Further, gantry mounted-CBCT [Fig. 1] leads to a long distance between the X-ray tube and flat panel detector (FPD) as compared to a diagnostic CT apparatus. The distance reduces the amount of X-rays incident on the detector, resulting in noisy reconstructed images. Therefore, we have developed an iterative reconstruction method that is robust to noise [1-6]. In this process, a corresponding system matrix is required for a given measurement geometry. In a gantry-mounted CBCT, the X-ray tube and the detector positions deviate some from their ideal position, resulting in a different geometry at each measurement angle, but full calculation of system matrices requires a large data capacity or significant computation time. To overcome these issues, our method calculates the system matrix for an ideal geometry in advance and performs positional deviation correction on the projection data [7].

II. Method

A. Reconstruction Algorithm

We adopted an ordered subset (OS) convex method as an iterative reconstruction algorithm [8]. The equation is shown, as in

$$\mu_j^{\,n,s+1} = \mu_j^{\,n,s} + \mu_j^{\,n,s} \frac{\sum_{i=1}^{D_s} l_{ij} \cdot \left[d_i \cdot \exp\left(-\sum_{k=1}^{B} l_{ik}\mu_k^{\,n,s} \right) - y_i \right]}{\sum_{i=1}^{D_s} \left[l_{ij} \cdot d_i \cdot \exp\left(-\sum_{k=1}^{B} l_{ik}\mu_k^{\,n,s} \right) \cdot \sum_{k=1}^{B} l_{ik}\mu_k^{\,n,s} \right]} \quad (1)$$

Here, $\mu_j^{\,n,s}$ is the linear attenuation coefficient for the j-th voxel, n and s denote iteration and subset number respectively, and d_i is the i-th detector output for the case which there are not objects between the X-ray tube and FPD. Conversely, y_i is i-th detector output for objects between the X-ray tube and FPD, and l_{ij} is the transmission length for the j-th voxel for which X-rays are incident on the k-th detector, whose length is referred to as the system matrix.

B. System matrix

If the transmission length l_{ij} is different from the actual value, artifacts in the image occur. More accurate length produces a reconstructed image closer to the true distribution. Therefore, we divided the source and the detector elements into 400 parts, and calculated the transmission length for all combinations of lines that connected a sub-source and a sub-detector. The calculated length passing through a voxel was averaged for each detector. The average value was adopted as the exact transmission length [Fig. 2]. The equation is shown, as in

$$l_{ij} = \frac{\sum_{s=1}^{N_s}\sum_{d=1}^{N_d} l_{ij}^{sd}}{N_s \cdot N_d} \quad (2)$$

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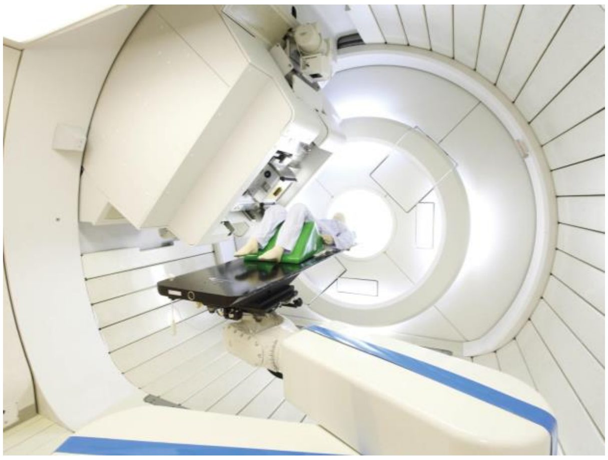
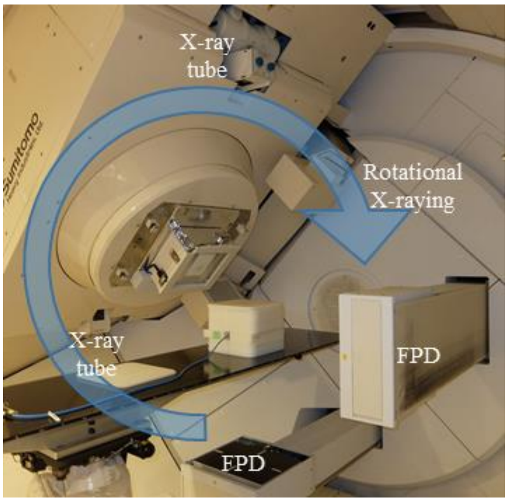
Here, N_s is the number of sub-sources, N_d is the number of detectors.

C. Position Deviation Correction

In the case of a proton therapy system, the X-ray tube and the FPD positions deviate from their ideal positions. Therefore, correcting for position deviation is required. First, the positional deviation correction of the detector performs a re-projection into the ideal position relative to the X-ray tube and linear interpolation using the re-projection data. Second, the position deviation correction of the X-ray tube performs linear interpolation of the system matrix data obtained in advance [Fig. 3].



(a)



(b)

Fig. 1. Overview of a gantry mounted CBCT system: (a) Proton treatment room. FPDs are stored in the gantry under normal conditions: (b) Overview of the system. FPDs are deployed from the gantry during patient positioning and CBCT imaging. While X-ray tubes and FPDs rotate with the gantry, projection data is collected at regular interval.

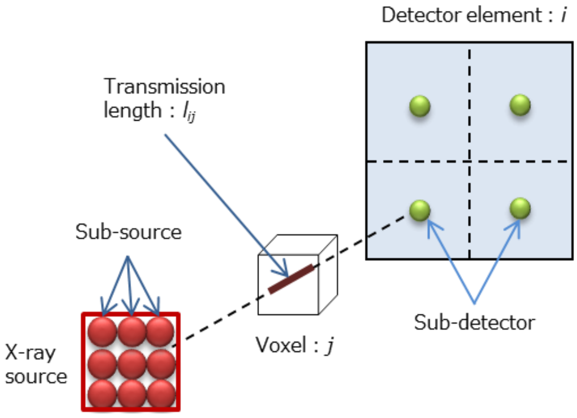


Fig. 2. System matrix calculation: The source and the detector were divided into sub-sources and sub-detectors, and a line was drawn connecting a sub-source and a sub-detector. The transmission length was calculated as the length of the line across the voxel. The calculated length for all combinations of lines was averaged for each detector, and the average value was treated as the transmission length for the voxel and the detecor.

III. Simulation

For verification of the reconstruction algorithm employing positional deviation correction, we simulated 180 projection images of a digital phantom [Fig. 4]. For each projection angle, the positional deviation was set as shown in Fig. 5, and the phantom and the detector were placed accordingly. In addition, by using ray tracing, the transmission length of each substance was calculated. Poisson noise was added to the calculated projection data y_i according to the actual number of photons. The simulated CBCT imaging conditions are shown in Table 1. The maximum standard deviation of the projection data is 2.5% [Fig. 6].

IV. Result & Discussion

Fig. 7 shows the reconstructed images from the Feldkamp method and OS Convex method respectively. Both algorithms have approximately the same spatial resolution. On the other hand, the contrast resolution (5%mm) by OS Convex is better than that of Feldkamp (>15%mm).

TABLE I. CBCT IMAGING CONDITIONS

	Specification
Focus size	1mm
Current	80mA
Voltage	120kV
Pulse width	5ms
FPD size	28.2x40.6cm
Pixel pitch	254μm
Pixels	1152x1600
Frame rate	3fps
SID	2.16m
Projection angle	180°

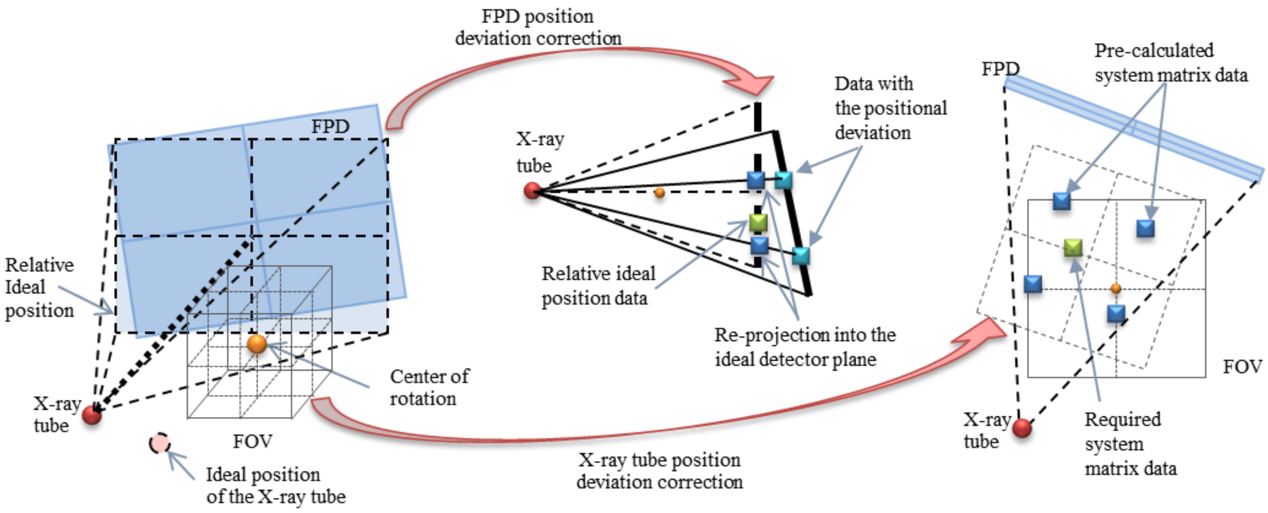


Fig. 3. Positional deviation correction: The positional deviation correction of gantry mounted CBCT systems consists of two kinds of corrections, FPD positional deviation correction and X-ray tube positional deviation correction. The former is performed by calculating the ideal position of the FPD corresponding to the actual position of the X-ray tube, and by re-projecting the projection data onto the FPD of the ideal position. The latter is performed by transforming the coordinates of the system matrix data at the ideal position calculated in advance, and by calculating the data at the coordinate of the actual position of the X-ray tube.

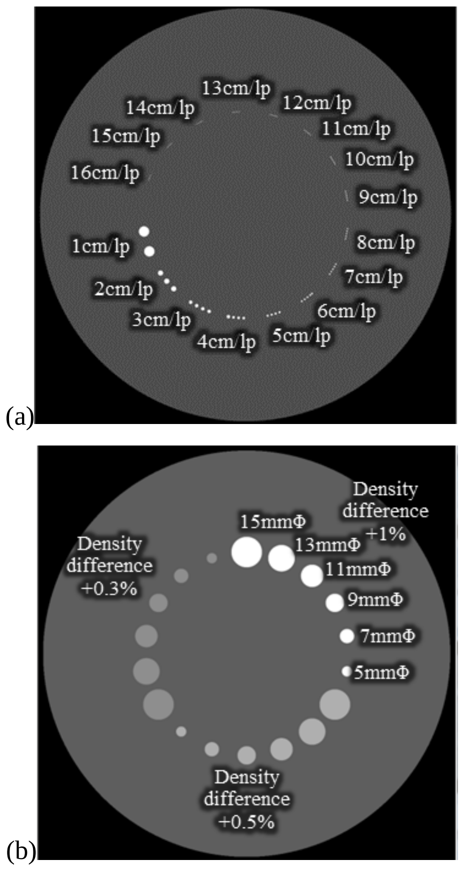


Fig. 4. Digital phantom image: (a) Module for spatial resolution evaluation: (b) Module for contrast resolution evaluation.

In order to evaluate the contrast resolution quantitatively, ROI of 10mm diameter was set to the 15mm diameter region of a 1% concentration difference, and the region located thereunder. Detectability Index, expressed as d' [9], was calculated for each region. The equation for d' is shown below

$$d' = \frac{\bar{\mu}_s - \bar{\mu}_n}{\sqrt{\frac{1}{2}\left(\sigma_s^2 - \sigma_n^2\right)}} \tag{2}$$

Here, $\bar{\mu}_s$ and σ_s represent the average value and the standard deviation of the 15mm diameter region of a 1% concentration difference, respectively, and $\bar{\mu}_n$ and σ_n represent the average value and the standard deviation of the region located thereunder. As a result, d' of the Feldkamp method is 0.52, and that of OS Convex method is 1.0. OS Convex improves the Detectability Index about two fold.

V. Conclusion

We developed an iterative image reconstruction method for proton beam gantry-mounted CBCT, and were able to obtain noise reduction and high contrast resolution images. However, the iterative method requires significant computation time (7 sec / projection / iteration on Core i7-5930K with OpenMP). We intend to reduce computation time by using a GPU.

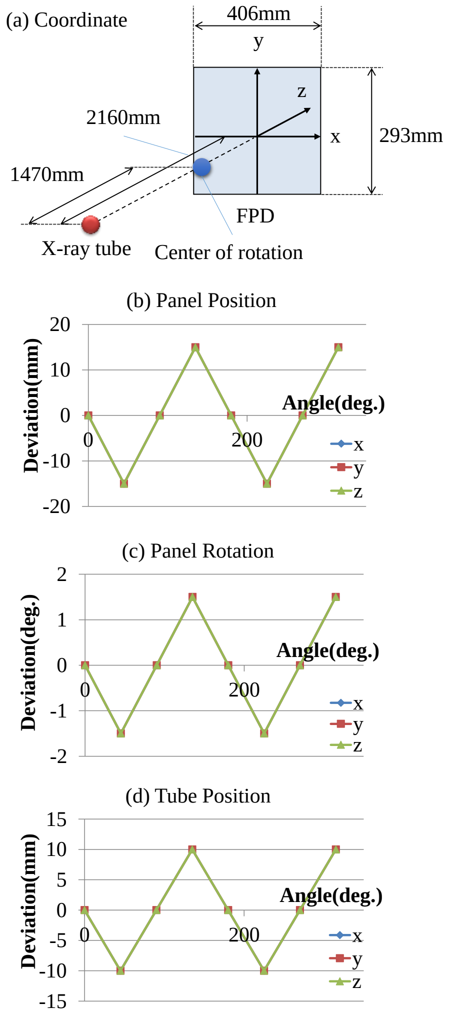


Fig. 5. Positional deviation for each projection angle: (a) Coordinates of the positional deviation: (b) Pannel (FPD) positional deviation: (c) Pannel (FPD) rotational deviation: (c) X-ray tube positional deviation.

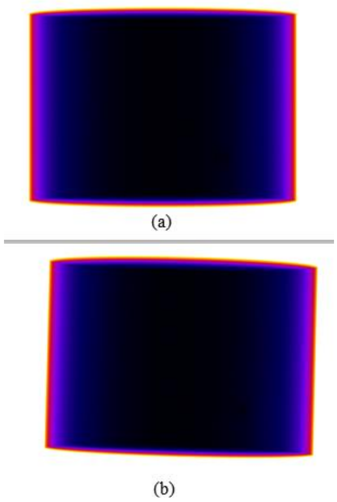


Fig. 6. Projection data with position deviation: (a) 0 degree: (b) 45 degree.

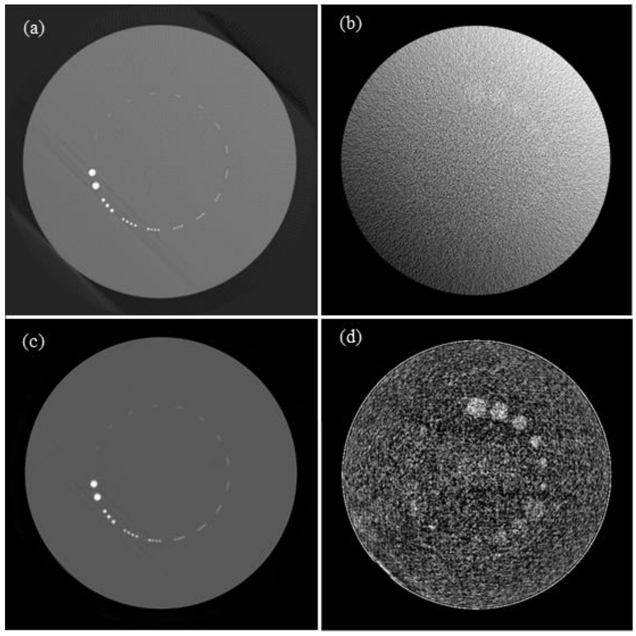


Fig. 7. Reconstruction images: The upper row are the images using the Fledkamp method, and the lower row are that of the OS Convex method: (a),(c) Modules for spatial resolution evaluation: (b),(d) Modules for contrast resolution evaluation.

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