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PHYSICS CONTRIBUTION

DEVELOPMENT AND CLINICAL EVALUATION OF A THREE-DIMENSIONAL CONE-BEAM COMPUTED TOMOGRAPHY ESTIMATION METHOD USING A DEFORMATION FIELD MAP

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Purpose: To develop a three-dimensional (3D) cone-beam computed tomography (CBCT) estimation method using a deformation field map, and to evaluate and optimize the efficiency and accuracy of the method for use in the clinical setting.

Methods and Materials: We propose a method to estimate patient CBCT images using prior information and a deformation model. Patients' previous CBCT data are used as the prior information, and the new CBCT volume to be estimated is considered as a deformation of the prior image volume. The deformation field map is solved by minimizing deformation energy and maintaining new projection data fidelity using a nonlinear conjugate gradient method. This method was implemented in 3D form using hardware acceleration and multi-resolution scheme, and it was evaluated for different scan angles, projection numbers, and scan directions using liver, lung, and prostate cancer patient data. The accuracy of the estimation was evaluated by comparing the organ volume difference and the similarity between estimated CBCT and the CBCT reconstructed from fully sampled projections.

Results: Results showed that scan direction and number of projections do not have significant effects on the CBCT estimation accuracy. The total scan angle is the dominant factor affecting the accuracy of the CBCT estimation algorithm. Larger scan angles yield better estimation accuracy than smaller scan angles. Lung cancer patient data showed that the estimation error of the 3D lung tumor volume was reduced from 13.3% to 4.3% when the scan angle was increased from 60° to 360° using 57 projections.

Conclusions: The proposed estimation method is applicable for 3D DTS, 3D CBCT, four-dimensional CBCT, and four-dimensional DTS image estimation. This method has the potential for significantly reducing the imaging dose and improving the image quality by removing the organ distortion artifacts and streak artifacts shown in images reconstructed by the conventional Feldkamp–Davis–Kress (FDK) algorithm. © 2012 Elsevier Inc.

Imaging dose reduction, Image reconstruction, Deformable registration, Image-guided radiation therapy, Cone-beam CT (CBCT).

INTRODUCTION

On-board cone-beam computed tomography (CBCT) is now becoming a powerful tool for image-guided radiation therapy (1), but its clinical utility may be limited because of its high imaging dose to a large volume of the patient (2). Specifically, in some adaptive radiation therapy contexts where CBCT is taken daily over a 30- to 40-fraction treatment, the accumulated imaging dose may have clinical significance.

Reducing the number of projections acquired in a CBCT scan is a rational strategy for reducing the imaging dose. Two different approaches have been taken to reduce the number of CBCT projections. The first approach, digital tomosynthesis (DTS), acquires projections only within a limited scan angle (20–60°) (3). Our previous studies showed that DTS images reconstructed by the Feldkamp–Davis–Kress (FDK) (4) method can provide accurate rigid body alignment of the patient's bony structures (5). However, DTS images do not provide full volumetric information for target localization due to the limited angle of the DTS acquisition. The second approach is to reduce the number of projections acquired without limiting the scan

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angle. The CBCT images reconstructed by the FDK method in this approach usually have severe streak artifacts because of the limited amount of projection information acquired.

Novel image reconstruction methods have been developed recently to improve the image quality for reconstruction using undersampled projection data. One category of methods is based on the compressed sensing (CS) theory (6), and it has been implemented for image reconstruction from limited-views and limited-angle projection data (7–9). However, these methods generally assume the images studied to be sparse which may not always be true for medical images. Besides, these methods can not recover the full volumetric information in DTS images or completely remove the streak artifacts in limited projection CBCT reconstruction when the projection number is small. Furthermore, the image resolution is often degraded in these methods. Another category of methods uses deformable image registration for image reconstruction (10–16) These methods were mainly developed for four-dimensional (4D) CT and 4D CBCT image reconstruction, and no such method has been comprehensively evaluated for limited-angle DTS reconstruction. Previously we developed a novel CBCT estimation method that uses a deformation field map to estimate CBCT images from limited-angle projections (17). However, this method was developed and tested only in two-dimensional (2D) cases, and it was not clear whether it is feasible for estimation of 3D CBCT images using limited projections.

In the current study, we implemented the method in 3D form using hardware acceleration and multi-resolution scheme, and evaluated it both for limited-angle and for limited-projection image reconstruction using liver, lung, and prostate cancer patient data. Comprehensive evaluation of the effects of projection number, scan angle, and scan direction on the CBCT estimation accuracy was performed, and different estimation parameters of the proposed method were optimized for clinical use.

METHODS AND MATERIALS

CBCT estimation algorithm

Either patient planning CT images or the previous day's CBCT images can be used as the prior images, which are denoted by I_{prior}. The new 3D CBCT images to be estimated are denoted by $CBCT_{new}$. The size of these images is defined to be n×n×n. In our method, the $CBCT_{new}$ image is considered as a deformation of the prior image I_{prior}. The deformation field is represented by $D_m(k,j,i)$, $m = 1,2,3$, $k,j,i = 1\ldots n$, where m represent the three directional components of the deformation field along x, y, and z axes, respectively, and k, j, and i represent the 3D index of the deformation field at each image voxel. Then $CBCT_{new}$ can be expressed as a function of D and I_{prior} as follows:

$$CBCT_{new} = CBCT_{new}\left(D,\ I_{prior}\right) \tag{1}$$

Specifically, each voxel value in $CBCT_{new}$ is interpolated from I_{prior} according to the deformation field D using trilinear interpolation. The equation for calculating the voxel value at (k, j, i) in the new CBCT image is as follows:

$$CBCT_{new}\ (k,j,i) = I_{prior}\ (k + D_1(k,j,i),\ j + D_2(k,j,i),\ i + D_3(k,j,i)) \tag{2}$$

In image reconstruction, the data fidelity constraint has to be met, which can be expressed by the following equation:

$$P^{*}\ CBCT_{new}\left(D,\ I_{prior}\right) = Y \tag{3}$$

where P is the cone-beam projection matrix to describe the x-ray projection measurements, and Y is the new cone-beam projection data acquired. Another constraint that we used to solve the deformation field is the energy constraint, which requires the deformation field D to have the minimum deformation energy. We used the free-form energy defined by Lu *et al.* (18) as the deformation energy. The formula of the free-form energy is as follows:

$$E(D) = \sum_{k=1}^{n}\sum_{j=1}^{n}\sum_{i=1}^{n}\sum_{m=1}^{3}\left(\left(\frac{\partial D_m(k,j,i)}{\partial x}\right)^2 + \left(\frac{\partial D_m(k,j,i)}{\partial y}\right)^2 + \left(\frac{\partial D_m(k,j,i)}{\partial z}\right)^2\right) \tag{4}$$

Based on these two constraints, the CBCT estimation problem is converted into the following unconstrained optimization problem:

$$\tilde{D} = \arg\min_{\forall D} f(D) = \arg\min_{\forall D}\left(\mu * E(D) + \left\|P * CBCT_{new}\left(D, I_{prior}\right) - Y\right\|_2^2\right) \tag{5}$$

where f(D) is the objective function to be minimized, and μ is the relative weight of the deformation energy. A nonlinear conjugate gradient (CG) method is used as the optimizer to solve the optimization problem in Eq. (5) (19). After the deformation field D is solved, the new on-board CBCT image $CBCT_{new}$ is obtained by deforming the prior image I_{prior} based on Eq. (2).

Hardware acceleration

This 3D CBCT estimation algorithm requires forward projection and back projection of the 3D volume in each iteration loop. The forward projection process that generates digital reconstructed radiographs (DRR) is the most time-consuming process. We have implemented this process in the computer graphics card to accelerate its speed (20). By using the three color channels in the graphics card, we are able to extend its data storage range to 0 to 2047 (11-bit). For CBCT estimation using simulated projections of the new CBCT images, both I_{prior} and the new CBCT images are linearly scaled to a data range of 0 to 2,047 before generating DRRs from deformed I_{prior} and simulated new projections from new CBCT images. For CBCT estimation using real cone-beam projections, the DRRs need to be generated from deformed I_{prior} with its original gray value range to match the gray value range of real projection images to meet the data fidelity constraint. To achieve that, we first linearly scale the gray values of the I_{prior} into 11-bit range by using the following formula:

$$I_{prior}{}' = \left(I_{prior} - I_{prior}{}^{min}\right) \Big/ \left(I_{prior}{}^{max} - I_{prior}{}^{min}\right) * 2047 \tag{6}$$

The scaled image volume I'_{prior} is loaded into the three color channels in the computer graphics card. The image volume is