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4D cone-beam computed tomography (CBCT) using a moving blocker for simultaneous radiation dose reduction and scatter correction

Cong Zhao¹, Yuncheng Zhong², Xinhui Duan³, You Zhang², Xiaokun Huang², Jing Wang², and Mingwu Jin^{1,*}

¹Dept. of Physics, University of Texas at Arlington, Arlington, TX 76019

²Dept. of Radiation Oncology, University of Texas Southwestern Medical Center, Dallas, TX 75390

³Dept. of Radiology, University of Texas Southwestern Medical Center, Dallas, TX 75390

Abstract

Four-dimensional (4D) X-ray cone-beam computed tomography (CBCT) is important for a precise radiation therapy for lung cancer. Due to the repeated use and 4D acquisition over a course of radiotherapy, the radiation dose becomes a concern. Meanwhile, the scatter contamination in CBCT deteriorates image quality for treatment tasks. In this work, we propose to use a moving blocker (MB) during the 4D CBCT acquisition (“4D MB”) and to combine motion-compensated reconstruction to address these two issues simultaneously. In 4D MB CBCT, the moving blocker reduces the X-ray flux passing through the patient and collects the scatter information in the blocked region at the same time. The scatter signal is estimated from the blocked region for correction. Even though the number of projection views and projection data in each view are not complete for conventional reconstruction, 4D reconstruction with a total-variation (TV) constraint and a motion-compensated temporal constraint can utilize both spatial gradient sparsity and temporal correlations among different phases to overcome the missing data problem. The feasibility simulation studies using the 4D NCAT phantom showed that 4D MB with motion-compensated reconstruction with 1/3 imaging dose reduction could produce satisfactory images and achieve 37% improvement on structural similarity (SSIM) index and 55% improvement on root mean square error (RMSE), compared to 4D reconstruction at the regular imaging dose without scatter correction. For the same 4D MB data, 4D reconstruction outperformed 3D TV reconstruction by 28% on SSIM and 34% on RMSE. A study of synthetic patient data also demonstrated the potential of 4D MB to reduce the radiation dose by 1/3 without compromising the image quality. This work paves the way for more comprehensive studies to investigate the dose reduction limit offered by this novel 4D MB method using physical phantom experiments and real patient data based on clinical relevant metrics.

*Correspondence to: Mingwu Jin, Ph.D., Univ. of Texas at Arlington, Dept. of Physics, 502 Yates Street, BOX 19059, Arlington, TX 76019, mingwu@uta.edu, Phone: 817-272-6262, Fax: 817-272-3637.

Keywords

4D CBCT; motion-compensated 4D reconstruction; moving blocker; scatter correction; dose reduction

1. Introduction

X-ray Cone-beam computed tomography (CBCT) has been widely used in radiation therapy for image-guided radiation beam delivery with accurate tumor localization and holds the potential for calculation of the actual dose on the treatment day for an adaptive therapy (Xing *et al.*, 2006). Since there are multiple fractions in each treatment course and CBCT scan may be repeatedly conducted at each treatment fraction. Lowering the radiation dose of CBCT, while maintaining or improve image quality, is important to realize the full potential of CBCT as the repetitive ionization radiation becomes a concern for patients (Brenner and Hall, 2007; Kalender, 2014).

New data acquisition strategies combined with hardware and software advances have been vigorously pursued and shown the great potential to substantially lower the CT radiation dose without compromising the image quality (Yu *et al.*, 2009; McCollough *et al.*, 2009; Hsieh, 2003). There are two main strategies for CT dose reduction in data acquisition: 1) the X-ray flux reduction by lowering the operating current, potential and exposure time of an X-ray tube, which have been adapted in some clinical practices; and 2) the sparse sampling including fewer projection views, limited coverage angle, and interior scan, which are still in active research stage. They could also be combined to achieve the maximum dose reduction. In 4D CBCT for radiation therapy, CT numbers for dose calculation and the tumor location and motion for beam delivery are of clinical interests. The performance of these clinical tasks, different from diagnostic purposes, may be well maintained even if the dose reduction, such as sparse sampling, is applied. In this work, in addition to dose reduction using fewer projection views, a moving blocker inserted between the X-ray source and the imaging subject, whose main purpose is to achieve the measurement based scatter correction, can lead to the subsampling in each projection view for further radiation dose reduction. To our best knowledge, this is the first study to investigate dose reduction potential through subsampling of both the view numbers and the projection images, particularly for 4D CBCT.

There are many factors that deteriorate the CBCT image quality. The inferior image quality as compared to the planning multi-detector CT (MDCT) becomes the major hurdle for CBCT to fulfill aforementioned dose calculation tasks. Among them, scatter is one of major problems due to the use of large-area flat panel detector (FPD). Various scatter correction (SC) methods have been proposed to address this problem and achieved notable success. These methods generally fall into three categories: 1) hardware scatter suppression methods, e.g. anti-scatter grids (Chan and Higashida, 1985; Kyriakou and Kalender, 2007; Siewerdsen *et al.*, 2004; Wiegert *et al.*, 2004); 2) computational methods, either analytical modeling or Monte Carlo simulation (Boone and Seibert, 1988; Jarry *et al.*, 2006; Kyriakou *et al.*, 2006; Li *et al.*, 2008; Zhao *et al.*, 2014; Zhu *et al.*, 2009a); and 3) measurement-based methods using various kinds of blockers or modulators (Dong *et al.*, 2013; Jin *et al.*, 2010; Lee *et al.*,

2012; Ning *et al.*, 2004; Ouyang *et al.*, 2013; Siewerdsen *et al.*, 2006; Wang *et al.*, 2010; Yan *et al.*, 2010; Zhu *et al.*, 2006; Zhu *et al.*, 2005; Zhu *et al.*, 2010; Zhu *et al.*, 2009b; Zhao *et al.*, 2017; Ren *et al.*, 2012). Although there is lack of a comprehensive comparison of these SC methods, using moving-blockers between the X-ray source and the image subject has achieved excellent SC performance with one conventional CBCT scan, which can reduce the CT number error less than 100 HU, satisfying less than 2% dose calculation error requirement (Zhao *et al.*, 2017). An added benefit of use of the moving blocker is that the blocked X-ray intensity effectively reduced the radiation exposure to the patient as aforementioned subsampling of each projection image.

Another degrading factor for CBCT is the patient motion since the acquisition speed is not fast enough to yield motion-frozen images with complete data. This is particular troublesome for the radiation therapy of lung cancer as the respiratory motion can be as high as 40 mm along superior-inferior direction and 20 mm along anterior-posterior direction (Keall *et al.*, 2006; Pretorius *et al.*, 2014). If only one 3D CBCT volume image was reconstructed from the whole data, the motion blur would introduce severe artifacts in the reconstructed image. It has been shown that respiratory motion gated 4D CBCT can greatly reduce these artifacts and improve image quality (Bergner *et al.*, 2009; Krinski *et al.*, 2005; Li *et al.*, 2006b; Lu *et al.*, 2007; Sonke *et al.*, 2005; Wang and Gu, 2013). Furthermore, 4D CBCT provides an opportunity to generate a tumor motion-of-the-day model right before each treatment fraction for a more precise treatment than using a universal motion model. Motion-compensated 4D reconstruction methods have been actively pursued to yield a series of images depicting tumor motion (Andersen and Kak, 1984; Blume *et al.*, 2010; Gilland *et al.*, 2002; Gravier *et al.*, 2006; Jacobson and Fessier, 2006; Jin *et al.*, 2009; Li *et al.*, 2010; Li *et al.*, 2007; Li *et al.*, 2006a; Qiao *et al.*, 2006; Rit *et al.*, 2009; Staub *et al.*, 2011; Zhang *et al.*, 2010; Wang and Gu, 2013; Tang *et al.*, 2012; Taguchi and Kudo, 2008; Gao *et al.*, 2012). Among them, a simultaneous motion estimation and image reconstruction (SMEIR) algorithm has been shown to yield high-quality respiratory phase images and a tumor motion model (Wang and Gu, 2013). Nevertheless, the scatter problem has not been addressed in these previous studies, although some computational methods may be adapted (Xu *et al.*, 2015).

In this work, we propose to use moving blockers in a respiratory motion gated 4D CBCT acquisition and to develop an effective reconstruction method. This work is founded on our previous developments on moving blocker SC for 3D CBCT (Zhao *et al.*, 2017; Wang *et al.*, 2010; Ouyang *et al.*, 2013) and SMEIR for 4D CBCT (Wang and Gu, 2013). In this novel imaging method, there are two challenges for the highly incomplete data reconstruction: 1) for each of total K phases of a respiratory cycle, there are only approximately $1/K^{\text{th}}$ projection views are available for reconstruction; and 2) for each projection view, the data is not complete (i.e. partially blocked). By adapting the techniques of compressed sensing reconstruction, motion-compensated reconstruction (particularly SMIER in this study), and moving-blocker based SC, we develop a 4D moving-blocker (4D-MB) CBCT method that can overcome these challenges and serving both dose reduction and scatter correction purposes. In this feasibility study of the proposed 4D-MB CBCT method, we demonstrated that high quality 4D images could be reconstructed at substantially reduced radiation dose (30% lower) with much reduced scatter artifacts using both non-uniform rotational B-spline-

based cardiac-torso (NCAT) (Segars *et al.*, 2003) phantom simulation and pseudo-real patient data.

2. Methods

2.A. Moving-blocker based 4D (4D-MB) CBCT acquisition

The moving-blocker based 4D (4D-MB) CBCT acquisition is illustrated in Fig. 1 (transverse view). A blocker made of parallel lead strips is placed between the X-ray source tube and the image subject and holds in the plane parallel to the FPD plane. During the data acquisition, the blocker is moving along the axial direction (perpendicular to the plane shown in Fig. 1) as the gantry rotates around the subject (more details about moving blocker setup can be found in our previous work (Zhao *et al.*, 2017; Wang *et al.*, 2010; Ouyang *et al.*, 2013)). In the meantime, the projection data is binned with respect to different phases of the respiratory motion of the subject based on the motion amplitude gating. Therefore, at any projection angle, the projection data corresponding to a particular respiratory phase is partially blocked (determined by the blocker position). Given non-synchronized respiratory motion and blocker motion, for each phase, the projection views are sparse around the subject with each projection data randomly sampled as lateral strips locate at different positions. Both compressed sensing techniques, e.g. total variation (TV) constraints exploiting the image gradient space sparsity, and motion-compensated reconstruction techniques, such as SMEIR, utilizing the temporal correlation can be adapted to reconstruct 4D images using these highly undersampled data.

The blocker serves two purposes here: 1) to provide scatter information in the blocked region; and 2) to lower the radiation dose by blocking substantial amount of X-ray flux. As long as the movement of the blocker is not synchronized with the respiratory motion so that the sampling data for each respiratory phase will randomly span the whole projection domain, which can be effectively reconstructed by compressed sensing techniques. In addition, the scatter signal can be estimated from the blocked regions. Thus, the primary signal in the unblocked regions for scatter-corrected reconstruction can be recovered by subtracting the scatter signal from the total signal. As we showed in our previous work (Zhu *et al.*, 2010), TV-based iterative reconstruction can effectively suppress the scatter artifacts and produce high-quality CBCT images using the primary signal only in the unblocked regions. It is important to note that the proposed 4D-MB will have much less projection data for each 3D phase volume compared to moving-blocker based 3D CBCT (3D-MB). Indeed, if the number of respiratory phases is K , the projection views for each phase in 4D-MB are only approximately $1/K^{\text{th}}$ of 3D-MB. The temporal redundancy must be taken into account to maintain high quality reconstruction. As a proof of concept study, we will adapt our previously developed SMEIR method (Wang and Gu, 2013), although other motion-compensated reconstruction methods can be equally applied (Andersen and Kak, 1984; Blume *et al.*, 2010; Gilland *et al.*, 2002; Gravier *et al.*, 2006; Jacobson and Fessier, 2006; Jin *et al.*, 2009; Li *et al.*, 2010; Li *et al.*, 2007; Li *et al.*, 2006a; Qiao *et al.*, 2006; Rit *et al.*, 2009; Staub *et al.*, 2011; Zhang *et al.*, 2010).

2.B. Scatter correction and image reconstruction

From Fig. 1, we have the total signal received at phase k of a respiratory motion gated acquisition (of a total of K phases) as follows,

$$S_k^{Total} = S_k^{Primary} \sum_{i=1}^B \prod_i + S_k^{Scatter}, \quad k = 1, 2, \dots, K, \quad (1)$$

where the element of $S_k^{Primary}(j) = I_0 e^{-\int_{L_j} \mu_k(x, y, z) dl}$ is the primary signal detected in the detector bin j from an incident source intensity I_0 through the ray path L_j . $\prod_i$ is the i^{th} rectangle window function (1 inside and 0 outside) for $i=1, 2, \dots, B$ blockers, $S_k^{Scatter}$ is the scatter signal and S_k^{Total} is the total signal that is the sum of the primary and scatter signals. The scatter signal $S_k^{Scatter}$ can be determined from the blocked regions and then be interpolated and subtracted from the total signal S_k^{Total} to recover the primary signal $S_k^{Primary}$ in the unblocked regions. Afterward, the normalized logarithm can be taken on the estimated primary signal $S_k^{Primary}$ to obtain line integrals p_k to have the following linear forward projection model,

$$p_k = A_k \mu_k, \text{ for } k = 1, 2, \dots, K, \quad (2)$$

where A_k is the projection matrix for phase k , whose element is the intersection length of a particular voxel with the ray reaching a particular detector bin. Note that A_k is different for different phases due to different view angles and blocker positions.

The phase-by-phase 3D reconstruction can be achieved by using constrained TV minimization,

$$\mu_k^* = \arg \min_{\mu_k} \|\mu_k\|_{TV}, \text{ s.t. } \|A_k \mu_k - p_k\|_2^2 \leq \epsilon, \text{ for } k = 1, 2, \dots, K, \quad (3)$$

where $\|\mu_k\|_{TV} = \sum_{x, y, z} \sqrt{(\mu_{x, y, z} - \mu_{x-1, y, z})^2 + (\mu_{x, y, z} - \mu_{x, y-1, z})^2 + (\mu_{x, y, z} - \mu_{x, y, z-1})^2} + \delta$ with a small constant $\delta (10^{-6})$ to avoid zero in the denominator in the gradient of TV and ϵ is an error bound for the quadratic data fidelity constraint. The alternated update methods can be used to solve Eq. (3), i.e. the data fidelity enforced by simultaneous algebraic reconstruction technique (SART) (Andersen and Kak, 1984) and the TV minimization achieved by the steepest descent method (with an initial step size of 0.2 and a relaxation factor 0.8) (Wang and Gu, 2013) are optimized in an interleaved fashion to update μ_k . However, this phase-by-phase 3D reconstruction will suffer from substantially undersampled data and can be improved by 4D reconstruction that utilizing the motion correlations among the phase volumes.

If a motion model is available to describe the transformation from phase to phase, an image at any phase k can be derived from a reference phase, say phase 1 for convenience,

$$\mu_k = D_{1 \rightarrow k} \mu_1, \text{ for } k = 1, 2, \dots, K, \quad (4)$$

where $D_{1 \rightarrow k}$ is the deformation matrix to transform the image at phase 1 to that at phase k . Then Eq. (2) can be re-written as

$$p_k = A_k D_{1 \rightarrow k} \mu_1, \text{ for } k = 1, 2, \dots, K. \quad (5)$$

Letting $\mathbf{p} = [p'_1, p'_2, \dots, p'_K]'$ denote the projections for all phases and $\mathbf{A} = (A_1 D_{1 \rightarrow 1}, A_2 D_{1 \rightarrow 2}, \dots, A_K D_{1 \rightarrow K})$ denote the 4D projection matrix, the 4D-MB model can be expressed as:

$$\mathbf{p} = \mathbf{A} \mu_1 \quad (6)$$

Thus, the projections from all phases can be used to reconstruct the image at the reference phase 1, whereas the image at phase k can be obtained by using Eq. (5) if the deformation matrix $D_{1 \rightarrow k}$ is available.

Since the deformation matrix $D_{1 \rightarrow k}$ is not known beforehand, the motion model has to be estimated and updated with the image reconstruction. Here we adapt the motion estimation method used in SMEIR (Wang and Gu, 2013) by matching the forward projections at each phase. For the convenience of expression of motion estimation, we define a deformation vector field (DVF) $\mathbf{d}_{1 \rightarrow k}$ as the displacement vector from phase 1 to phase k , which leads to the following equivalency,

$$D_{1 \rightarrow k} \mu_1 = \mu_1(\mathbf{d}_1 + \mathbf{d}_{1 \rightarrow k}), \text{ for } k = 1, 2, \dots, K, \quad (7)$$

where $\mathbf{d}_1$ is the coordinate vector. To obtain the inverse consistent forward ($\mathbf{d}_{1 \rightarrow k}$) and backward ($\mathbf{d}_{k \rightarrow 1}$) DVFs required by 4D-MB reconstruction, a symmetric energy function is used as,

$$\mathbf{d}_{1 \rightarrow k}^* = \arg \min_{\mathbf{d}_{1 \rightarrow k}} \|p_k - A_1 \mu_1(\mathbf{d}_1 + \mathbf{d}_{1 \rightarrow k})\|_2^2 + \beta R(\mathbf{d}_{1 \rightarrow k}),$$

$$\mathbf{d}_{k \rightarrow 1}^* = \arg \min_{\mathbf{d}_{k \rightarrow 1}} \|p_1 - A_k \mu_k(\mathbf{d}_k + \mathbf{d}_{k \rightarrow 1})\|_2^2 + \beta R(\mathbf{d}_{k \rightarrow 1}),$$

$$\text{s.t. } \mathbf{d}_{1 \rightarrow k} \circ \mathbf{d}_{k \rightarrow 1} = \mathbf{d}_{k \rightarrow 1} \circ \mathbf{d}_{1 \rightarrow k} = 0, \quad (8)$$

where $R(\bullet)$ is a smoothness regularization function for DVFs, β is a parameter balancing the first data fidelity term and the second smoothness regularization term, and $\circ$ is a composition operation defined as $\mathbf{d}_{1 \rightarrow k} \circ \mathbf{d}_{k \rightarrow 1} = \mathbf{d}_{1 \rightarrow k}(\mathbf{d} + \mathbf{d}_{k \rightarrow 1}) + \mathbf{d}_{k \rightarrow 1}$ that is equal to zero to enforce the inverse consistent constraint of DVFs. An interleaved optimization can be used to solve Eq. (8), where a nonlinear conjugate gradient algorithm with back-tracking line search is used for two regularized minimizations and the inverse-consistency of DVFs is automatically enforced by using the solution of one minimization as the initialization of the other minimization. The β value was set as 0.05 for the phantom data and 0.5 for the patient data through the trial-and-error test. More details of SMEIR can be found in (Wang and Gu, 2013).

2.C. NCAT phantom simulation

A 4D NCAT phantom (Segars and Tsui, 2002; Segars *et al.*, 2003) was used to simulate and to quantitatively evaluate the 4D-MB method. Ten respiratory phases of a male phantom were generated with a respiratory period 4 s. The maximum diaphragm motion of 20 mm and the maximum anterior-posterior motion of 12 mm were used to model the respiratory motion. Three orthogonal views for phase 1 are shown in Fig. 3a. A spherical 3D tumor with a diameter of 10 mm was also included and indicated by a red arrow in Fig. 3a. The dimension of the phantom and the reconstructed images were $256 \times 256 \times 100$ with a voxel size of $2 \times 2 \times 2 \text{ mm}^3$ and the source to detector distance and the source to isocenter distance was 750 mm and 500 mm, respectively, same as our previous work (Wang and Gu, 2013). The Siddon's ray-tracing algorithm (Siddon, 1985) was used to generate projection images with a detector matrix of 384×150 and a detector pixel size $2 \times 2 \text{ mm}^2$. The distance between the blocker and the X-ray source is 75 mm. The moving blocker is made of lead strips with 2 mm width and 4 mm pitch (i.e. 4 mm gap between two strips). The ideal block of X-ray by the lead strips is assumed in the simulation so that the X-ray incident intensity is reduced by 1/3 when the blocker (blocked region : unblocked region = 1 : 2) is used. The noise was added as the sum of Poisson-distributed counting noise (1×10^5 counts/incident ray) and Normal-distributed electronic noise (variance of 10) (Wang *et al.*, 2006; Wang and Gu, 2013). The blocker motion is simulated to have six positions with a step-and-shoot mode. The blocker position related to the view angle is shown in Fig. 2, where the blocker stays in the same position for each respiratory cycle (with 10 phases). Although in this ideal simulation case the position of the blocker has a repetitive pattern, the randomness arises from that these positions occur at the different projection view angles. It is important to note that the blocker positions are evenly distributed for each phase so that there is no missing projection area or overexpressing of a certain area. Both 4D-CBCT with and without the moving blocker were simulated for comparison. We name the acquisition with the moving blocker as "MB" and that without the moving blocker as "Conventional".

In the first experiment, the scatter was not included. Two cases were simulated: one with 300 projection views (i.e. 30 projections for each respiratory phase) and the other with 200

projection views (i.e. 20 projections for each phase), both evenly distributed around the phantom for 360°. This experiment will be used to compare the reconstruction performance using two different dose reduction methods for 4D CBCT: 1) lowering the number of the projection views (sparse view); or 2) reducing the X-ray fluence in each projection view (4D-MB). The moving blocker will be used for the simulation using 300 projection views and be compared with the results using 300 and 200 projection views without the moving blocker. Note that the dose level using 300 projection views with each view blocked by 1/3 is equivalent to that using 200 views without the moving blocker ($=300 \times 2/3$).

In the second experiment that used 300 projection views, we investigate the effect of the scatter correction rendered by 4D-MB. Specifically, the scatter signal within each projection image is simulated by convolving the primary signal with a point spread function derived from an experimental study (Gaussian function, $\sigma = 0.8$ mm, size 141×141) (Li *et al.*, 2008). The total observed signal becomes the sum of the scatter signal and the projection (primary) signal generated in the first experiment (Wang *et al.*, 2010). Two 4D CBCT acquisitions were simulated: one with the moving blocker (4D-MB) and the other without. In 4D-MB, the blocked regions record the scatter signal that will be used for scatter correction (See Section 2.B.).

In summary, we simulated the following three cases for the first experiment without including the scatter effect: 1) total 300 views for 10 phases without the moving blocker (“Conventional 30 views/phase”); 2) total 200 views for 10 phases without the moving blocker (“Conventional 20 views/phase”); and 3) total 300 views for 10 phases with the moving blocker (“MB 30 views/phase”). For the second experiment with the scatter effect, we simulated two cases: 1) total 300 views for 10 phases without the moving blocker (“Conventional 30 views/phase”); and 2) total 300 views for 10 phases with the moving blocker (“MB 30 views/phase”). Two reconstruction methods were used for the simulation data: 1) phase-by-phase 3D iterative reconstruction with a TV constraint (“3D TV”); and 2) 4D reconstruction combining both spatial TV minimization and motion compensated regularization (“4D SMEIR”).

Structural similarity index (SSIM) (Wang *et al.*, 2004) and root mean square error (RMSE) in Hounsfield unit (HU) are used as quantitative evaluation criteria. SSIM measures the similarity between two images on three aspects: intensity, contrast and structure and RMSE measures the overall absolute reconstruction error. The SSIM index and the RMSE are calculated in a volume of interest ($160 \times 90 \times 50$) centered at the NCAT phantom including the tumor of the first phase (Fig. 3a), unless otherwise stated. Specifically, the SSIM index is calculated on the VOIs (denoted as x for the ground truth and y for the estimate) of two images as follows:

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

where μ 's are the average of VOIs, σ 's are variance and covariance, and C_1 and C_2 are two variables to stabilize the division with weak denominator. In this work, C_1 is calculated as

the square of 0.01 times of the dynamic range of the estimated image and C_2 is calculated as the square of 0.03 times of that. The RMSE is defined as:

$$RMSE = \sqrt{[\sum_i^n (x_i - y_i)^2]/n},$$

where i denotes the voxel index and n is the total number of voxels in the VOI.

In addition, the maximum and mean deviations from the tumor motion trajectory of the phantom for different methods are used to measure the motion recovery performance.

2.D. Patient study

To demonstrate the proposed method in a more clinically relevant setting, we generated the moving-blocker data from a set of real patient data. This patient underwent radiotherapy for lung cancer. CBCT was acquired using an Elekta Synergy® XVI system (Elekta AB, Stockholm, Sweden) with a M20 collimator and a F1 bow-tie filter. The acquisition protocol parameters were: 120 kVp and 1.6 mAs per projection, a total of 631 projection (512×512 pixels with a pixel size of $0.8 \times 0.8 \text{ mm}^2$) views evenly distributed in 360° , and acquisition time of 2 min. The source to detector distance was 1536 mm and the source to isocenter distance was 1000 mm. The CBCT projections were grouped into eight phases based on the diaphragm position at each projection and further downsampled by half, which results in around 49 projections for each phase. The reconstructed images are $300 \times 300 \times 150$ with a voxel size of $1.6 \times 1.6 \times 1.6 \text{ mm}^3$. A moving blocker blank acquisition was simulated using the same acquisition protocol of the patient data so that the blocked regions in each projection of the real patient data can be identified. The pseudo-real moving blocker patient data were obtained by setting the intensity in the blocked regions of the real patient projection data as zero. The removed data (i.e. those with intensity set as zero) is about 1/3 of the original data, which leads to about 1/3 dose reduction for the pseudo-real moving blocker patient data. Since the real moving-blocker data is not available at this point, it is not possible to evaluate the scatter correction effect, but will be investigated in future work. In this work, we aim to demonstrate the feasibility of 4D-MB with 1/3 dose reduction on real patient data. It will be compared with reconstruction results at the regular dose with and without motion compensation.

3. Results

3.A Results of 4D NCAT phantom

3.A.1. Results without scatter included—Fig. 3 shows the reconstructed images of phase 1 NCAT phantom in three orthogonal views: transverse (left), coronal (middle) and sagittal (right). In all cases, the tumor is visible, but clearer for 4D reconstruction. Figs. 3b–d illustrate the images reconstructed by 3D TV for conventional 30 views/phase, conventional 20 views/phase, and MB 30 views/phase, respectively. The latter two acquisitions (Fig. 3c and 3d) have 1/3 dose reduction compared to the first one (Fig. 3b). With the number of projection views decreased from 30/phase to 20/phase, the images in Fig. 3c suffers more blurring and blocky artifacts than that in Fig. 3b because the stronger TV regularization has to be applied to address the streak artifacts caused by the missing-data. Some fine structures

in the lungs are missing due to this effect (e.g. yellow and blue arrows in Fig. 3c). It is interesting to see that, at the equivalent dose level, 3D TV with MB 30 views/phase (Fig. 3d) recovered this fine structure to a better extent than that with conventional 20 views/phase, likely attributing to the effect of different undersampling methods. However, some horizontal streak artifacts can be seen in the coronal and sagittal views (e.g. green arrow in Fig. 3d), likely due to use of the blocker. With 4D motion-compensated reconstruction (4D SMEIR), the reconstructed images are less blurred and maintain better fine structures than those with 3D TV. The images of 4D SMEIR with conventional 30 views/phase (Fig. 3e) achieve the closest similarity to the original phantom (Fig. 3a). The reduction of 1/3 of the number of projection views suffers notable blocky artifacts and lowered contrast in the reconstructed images (Fig. 3f). At the equivalent dose level, the proposed method of MB 30 views/phase with 4D SMEIR (Fig. 3g) seems to suffer less blocky artifacts and to maintain more fine structures compared to that in Fig. 3f. The tumor contrast is also notably higher in Fig. 3g than Fig. 3f. In addition, the horizontal streak artifacts in the coronal and sagittal views in MB 30 views/phase with 3D TV (Fig. 3d) are effectively suppressed in MB 30 views/phase with 4D SMEIR (Fig. 3g).

The quantitative measures are listed in Table 1. The rankings using SSIM are generally the same as that using RMSE, i.e. a higher SSIM value leads to a lower RMSE value. It can be seen that the best quantitative results are obtained with conventional 30 views/phase with 4D SMEIR, followed by MB 30 views/phase with 4D SMEIR (2/3 dose of the former). It is worth noting that 4D SMEIR at reduced dose level through either a reduced view number or the moving blocker can achieve higher quantitative accuracy than conventional 30 views/phase with 3D TV reconstruction. This demonstrates that 4D motion-compensated reconstruction is crucial to improve or maintain image quality at lowered imaging dose for 4D CBCT. In summary, both the reconstructed images and the quantitative results in this experiment seem to favor undersampled projection rays (30 views/phase with MB) over undersampled view angles (20 views/phase without MB) for a fixed dose level. Further investigation on these two undersampling strategies is needed to generalize this finding.

3.A.2. Results with scatter included—Results in this section are with 30 projection views/phase and scatter signal included in the CBCT simulation. Fig. 4 illustrates the reconstructed images with conventional and MB acquisitions and different reconstruction methods. Without scatter correction (SC), both 3D TV (Fig. 4a) and 4D SMEIR (Fig. 4b) suffer the shading artifacts and low contrast, although 4D SMEIR images are much better than 3D TV (in the middle region of the field of view). With the MB acquisition, the scatter signal can be effectively removed from the projection data, which results in greatly improved image quality as shown in Fig. 4c using 3D TV and Fig. 4d using 4D SMEIR, in addition to the benefit of 1/3 reduction of the radiation dose. Similar to the findings in the simulations without scatter, images reconstructed by 4D SMEIR show improved tumor contrast and preserve more structural details than that reconstructed by independent phase-by-phase 3D TV reconstruction.

The line profiles (the red dashed line in Fig. 3a) across the tumor in the transverse slice in Fig. 4 are shown in Fig. 5 to compare the different acquisition and reconstruction methods. Both 3D TV and 4D SMEIR with scatter correction recover the attenuation coefficient

profile much better than those without scatter correction, even though the radiation dose with the MB acquisition is just 2/3 of the conventional one. Although 4D SMEIR without scatter correction outperforms 3D TV without scatter correction, it is inferior to 3D TV with scatter correction in general (except at the tumor location indicated by the arrow). The proposed 4D MB with motion-compensated reconstruction (MB-SMEIR) yields the closest profile to the phantom as it takes the advantages of both spatiotemporal 4D reconstruction and moving-blocker based scatter correction. It is important to note that this substantial reconstruction accuracy improvement of MB-SMEIR is accompanied with the dose reduction benefit of 1/3, compared to SMEIR with the conventional acquisition.

The quantitative measures are listed in Table 2. It can be seen that 4D SMEIR outperforms their counterpart using phase-by-phase 3D TV reconstruction under the same condition of data acquisition and scatter treatment. Although SMEIR without scatter correction improved the SSIM metric to the similar level of 3D TV with the moving-blocker based scatter correction, it suffers a much large absolute reconstruction error (131 HU vs. 89 HU). With 1/3 dose reduction, the proposed 4D MB method with SEMIR improves the SSIM measure by 37% and the RMSE measure by 55% compared to the conventional SMEIR without the moving blocker based scatter correction. At the same dose level, SMEIR with the MB acquisition and scatter correction outperforms 3D TV under the same condition by 28% on SSIM and 34% on RMSE.

3.A.3. Tumor motion recovery—The tumor motion trajectories of all respiratory phases are calculated for each method and compared with the tumor motion trajectory of the 4D NCAT phantom (serving as the true tumor motion). Specifically, the deformation vectors in a small volume ($2 \times 2 \times 2$ voxels) at the tumor center are averaged to get the motion from the reference phase (i.e. phase 1) to the target phases (i.e. phase 2~10). The deviations from the true tumor motion for different methods are plotted in Fig. 6. The tumor motion recovered by 4D SMEIR with MB and SC (“MB-SMEIR with SC”, green lines) matches well with the true motion with deviations generally less than 2 mm, except for phases 8 and 9. TV with SC (blue lines) also achieves good motion recovery performance with deviations more than 2 mm only for three phases (3, 4, and 8). Although SMEIR without SC (magenta lines) recovers the tumor motion much better than TV without SC (red lines), their deviations are worse than the methods with SC. The maximum and mean deviations over all phases are listed in Table 3. The proposed MB-SMEIR method outperforms other methods in both criteria, with a maximum deviation of 3.01 mm and a mean deviation of 1.48 mm.

3.A.4. Reconstruction convergence—The objective function to be minimized for $d_{1 \rightarrow k}$ at phase 5 ($k=5$ in the first objective function in Eq. (8)) is used to show the convergence of 4D-MB reconstruction. As shown in Fig. 7, 4D-MB reconstruction indeed converges with continuous decreasing of the objective function values. It can be seen that the first 50 iterations, there is a slight jump of the objective function value because the SART and TV minimization used for image reconstruction update the images after every 10 DVF estimation. This jump diminishes as the iteration goes on beyond 100. It is worth noting that different view angle and projection ray undersampling patterns, which result in different

system matrices, may influence the convergence. This topic is out of the scope of the current work and will be investigated in future work.

3.B. Results of the lung cancer patient

In Fig. 8, we show the images reconstructed from the real patient data and the pseudo-real moving blocker patient data for two respiratory phases. The first two columns are the coronal view and the last two columns are the sagittal view. It can be seen that 4D SMEIR effectively suppresses the noise compared to 3D TV. In both cases of with and without moving blocker, 4D SMEIR achieves less noisy and higher contrast images, thus leading to better-defined structures. Comparing 4D SMEIR at the regular dose and at 2/3 of the regular dose with the moving-blocker, the reconstructed images look similar except that the latter suffers slight more noise due to the reduced imaging dose. This is consistent with the findings in the NCAT simulation study without scatter (Section 3.A.1. and Fig. 3). This preliminary patient study indicates that the substantial dose reduction without compromising the image quality may be achievable. The proposed 4D MB method provides a promising way to achieve the dose reduction goal through random sparse sampling in each projection view and advanced 4D reconstruction.

There is generally no ground truth in real patient images, which is why the phantom study was conducted in the first place. To quantify the performance of different methods in patient study, we use SMEIR without MB as the ground truth and calculate the SSIM and RMSE of the VOIs (shown in red dashed boxes in Fig. 8) for other methods. The results using phase 1 as the reference phase for SMEIR are shown in Table 4. It can be seen that 4D SMEIR can produce better quantitative results than 3D TV. It is interesting to note that 3D TV without MB and with MB have the similar performance, which may implicate the data redundancy. Also, the results for phase 3 seem to deteriorate a bit, particularly for 3D TV methods. It is likely due to the fact that the ground truth of phase 3 was deformed from phase 1 using the estimated DVF in 4D SMEIR, which may not be perfect. To further characterize the motion influence, 4D SMEIR without MB using phase 3 as the reference is used as the ground truth and the corresponding results are shown in Table 5. The phase 3 results using phase 3 as the reference are in the similar range as the phase 1 results using phase 1 as the reference. The improvement of the current SMEIR method will be investigated in future studies as detailed in Discussion.

4 Discussion

In this work, we demonstrated the working principle of a novel 4D moving blocker method, which is the first time, to our best knowledge, to combine a moving blocker with 4D CBCT to achieve simultaneous dose reduction and scatter correction. Although there were a few studies using blockers/modulators for scatter correction in 3D CBCT (Wang *et al.*, 2010; Ouyang *et al.*, 2013; Zhu *et al.*, 2006; Zhu, 2016) or using moving multi-slit collimator (MSC) for dose reduction in MDCT (Muckley *et al.*, 2017; Koesters *et al.*, 2017), all of them are limited to 2D or 3D reconstruction. The current study is the first study that utilizes the random under-sampling in each projection view for imaging and reconstruction of a moving object (lung here), which can be modeled as a 4D signal in nature. With advanced

compressed sensing (i.e. image gradient sparsity enforced by TV constraints) and motion-compensated 4D (SMEIR) reconstruction, the satisfactory reconstruction images with 1/3 dose reduction were obtained for both simulation and pseudo-real patient data. In addition, the proposed 4D MB method can provide a simple observation based scatter correction for CBCT, which is demonstrated by a simulation study with large improvement in reconstruction accuracy (31% on SSIM and 58% on RMSE). Note that this type of scatter correction requires only interpolation and subtraction in the projection domain, and thus is less computational demanding than computational SC methods.

Furthermore, the dose reduction can also be achieved by reducing the projection views. In this regard, we compared the equivalent dose scheme: 1) 200 views of 4D CBCT without MB; and 2) 300 views of 4D MB CBCT. The simulation results showed that 4D MB CBCT achieved the similar reconstruction performance, if not better, to 200-view 4D CBCT without MB. Considering the capability of simple SC rendered by MB and the constant switch on-and-off of the X-ray tube current required by the fewer view acquisition, 4D MB CBCT may holds a great potential for both dose reduction and scatter correction purposes. On the other hand, adding a moving blocker to the CBCT system may pose some challenges. Nevertheless, a motor-controlled moving blocker has been demonstrated to be feasible in our previous study (Ouyang *et al.*, 2013) due to its lightweight for kilovoltage CBCT. Some optimal conditions for SC have been investigated in our previous work (Chen *et al.*, 2017) and most results are satisfactory (with CT number error < 50 HU). Thus, there is no strict requirement on the blocker moving speed or style as long as the blocker position is evenly distributed for each phase. For example, another possible blocker motion for the step-and-shoot mode is shown in Fig. 9, which led the similar satisfactory results to that used in Fig. 2. In reality, due to the irregularity of patient breath, the blocker motion needs to be controlled to make the distribution of the blocker position for each phase roughly even. Nonetheless, the study of the optimal blocker design, e.g. ratio of blockers, gaps, moving speed, and moving style (continuous or step-and-shoot) and the minimal number of projection views will be important future work to discover their impact on 4D CBCT reconstruction and clinically relevant tasks using more realistic physics modeling or real physical phantom/patient experiments. The effect of sparse sampling due to different view angles and blocker positions on reconstruction convergence and parameter selection will be investigated and provide the practical guidance for the adaption of such a system.

In this proof-of-concept study, the CT imaging physics is not fully modeled, such as the omission of penumbra effects due to either blocker penetration or limited size of the X-ray source, and the lag effects (Siewerdsen and Jaffray, 1999). These factors may need further investigation to answer whether they will significantly impact the reconstruction performance. However, based on our studies on MB based scatter correction for 3D CBCT using physical phantoms (Zhao *et al.*, 2017; Wang *et al.*, 2010; Ouyang *et al.*, 2013; Chen *et al.*, 2017), it is reasonable to expect the similar effectiveness for 4D MB CBCT. The current simulation study is sufficient to demonstrate the feasibility of 4D MB CBCT in principle. In addition, the blocker could generate scatter photons that would contribute extra radiation dose to the patient. As a preliminary investigation of this problem, two air scans were performed: one with the moving blocker (blocked region : unblocked region = 1:1) and the other without it using the Varian TrueBeam OBI system (678 projections). The ratio of the

intensity (as a surrogate of irradiation dose) of all the projections between that with the blocker and that without the blocker is about 48%, which is slightly less than the expected value of 50%. As the lead strips are embedded on a 5 mm thick acrylic board, the 48% ratio (<50%) can be attributed to that the attenuation of the acrylic board outweighs the contribution of the scatter photons from the blocker. Furthermore, the radiation dose was measured on an Elekta Versa HD system. The scan was a standard lung protocol with the following scanning parameters: 120 kVp, 2 min nominal rotation time, 264 mAs and the beam collimation of 26.5 cm × 26.5 cm. A 32 cm diameter PMMA CTDI phantom was aligned in the center of the scanning field of view. A 10 cm long pencil chamber was placed in the center or 12 o'clock position. The CTDI was calculated using the CBDI method (Amer *et al.*, 2007),

$$CTDI_w = \frac{1}{3}D_c + \frac{2}{3}D_p,$$

Where D_c and D_p denote the measured dose in the center and peripheral locations. This definition is an approximation of average dose along the 10 cm range in the middle of the phantom. CTDI measurements were performed without a blocker and with a blocker (1:3 pitch matching the simulation setting). Three repeated measurements were performed at each position and the averaged value was used. The calculated $CTDI_w$ was 5.7 mGy without the blocker and 3.5 mGy with the blocker, which is consistent with the pitch of the blocker for 1/3 dose reduction.

On the other hand, the scatter photons generated by the blocker can reach the detector, which could possibly degrade the projection data. However, the scatter photon from the blocker is of lower energy compared to the incident photons. After passing through patients, the probability of the blocker generated scatter reaching the detector would be very small. Although we are not able to separate the blocker generated scatter photons and patient induced photons in our experiments, satisfying scatter correction performance was achieved in our previous studies using physical phantoms (Zhao *et al.*, 2017; Wang *et al.*, 2010; Ouyang *et al.*, 2013; Chen *et al.*, 2017). A detailed Monte Carlo simulation study can potentially separate scatter photons from different sources at the detector, which will be investigated in a future study.

Finally, there are several potential improvements of the current study worth mentioning. First, the TV was used to enforce spatial sparsity of image gradient. It works well for images with piece-wise constant features, like NCAT images, but may not be suitable for the real patient data with subtle textures. In future studies, more advanced spatial priors will be investigated, such as normal-dose image induced TV (Huang *et al.*, 2013), total generalized variation (Niu *et al.*, 2014), TV-stokes (Liu *et al.*, 2014), l_0 gradient (Yu and Zeng, 2015), Haralick texture measures (Zhang *et al.*, 2016). Second, SMEIR seemed to achieve a large improvement on NCAT phantom data over 3D reconstruction, but only a moderate one on the pseudo real patient data (see Fig. 8.). One reason is that the complication of the real clinical data deteriorates the image quality of 3D reconstruction, which results in larger errors in the initial DVF estimation. These errors may cause SMEIR to trap into a local optimal, instead of the global optimal. Either better initial DVF estimation by a better initial

3D reconstruction or constrained DVF estimation with a reliable prior model can mitigate this problem and will be investigated in future. It is also worth looking into the way of estimating motion for two phases far away. In this study, the direct motion estimation for any pair of phases was used. Instead, the motion can be estimated for neighboring phases and that for far-away phases can be derived from the neighboring motion since the motion estimation can deteriorate quickly when the motion is too big. Meanwhile, patient respiration irregularity and hysteresis may cause additional difficulties in 4D MB CBCT and will need to be addressed in future studies. Last, in the future studies of improving the proposed 4D MB CBCT, a task-based performance evaluation, such as tumor motion tracking and organ segmentation for dose calculation, will need to be employed and compared with clinically adapted low-dose CT methods, such as lowering tube current, and other scatter correction methods for its real clinical impact.

5. Conclusions

In this work, we proposed a moving blocker based 4D CBCT acquisition (4D MB) and combined motion-compensated image reconstruction method with the moving blocker based scatter correction. In the NCAT simulation study, 4D MB with motion-compensated reconstruction at 2/3 of the regular dose can achieve much higher reconstruction accuracy reflected by image appearance and quantitative measurements, compared with the conventional SMEIR without scatter correction at the regular dose. A synthetic patient study also shows the potential of the proposed method to reduce the radiation dose by 1/3 without compromising the image quality. These promising results encourage more comprehensive studies to use realistic physical phantoms and patient data to interrogate the dose reduction limitation of the proposed 4D MB method using the clinical relevant metrics, e.g. tumor and organ segmentation and CT number for dose calculation. More sophisticated motion estimation methods and advanced spatial priors other than TV are also worth exploring to further improve the performance of this novel method.

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Fig. 1. Moving-blocker based 4D CBCT with respiratory gating (transverse view). Different colors of blockers and FPDs denote that the projection data at a particular angle (view) is a combination of a particular respiratory phase and a particular blocker position. (RM: respiratory motion; FPD: flat panel detector. Illustration only, not to scale.)

Fig. 2.

The blocker position versus the projection view angle in the simulation. The red dot denotes the blocker position of the phase 1, where each blocker position has five occurrences. The other phases have the same blocker position distribution.

Fig. 3.

NCAT phantom and reconstructed images using different 4D CBCT acquisition methods (Conventional acquisition and MB acquisition) and reconstruction methods (3D TV and 4D SMEIR). No scatter was included in simulation. (Left: transverse; middle: coronal; right: sagittal). The red dashed line in (a) transverse view is the profile position of Fig. 5. The red arrow in (a) coronal view indicates the tumor. The three dot squares in (a) are the volume of interest (VOI) for quantitative measures of SSIM and RMSE. The yellow and blue arrows indicate the fine structures lost in 3D TV with fewer views and the green arrow points the artifact likely due to use of the blocker.

Fig. 4.

Reconstructed NCAT images using different 4D CBCT acquisition methods (conventional acquisition and MB acquisition) and reconstruction methods (3D TV and 4D SMEIR) with scatter included in the simulation (SC: scatter correction). Note that the radiation dose in (c) and (d) is 1/3 lower than that in (a) and (b) because of use of the moving blocker.

Fig. 5. Line profiles across the tumor in a transverse slice (denoted by a red dashed line in Fig. 3a) for different methods shown in Fig. 4. The arrow indicates the tumor location.

Fig. 6.
The deviations of reconstructed tumor motion from the true tumor motion of the NCAT phantom for different methods.

Fig. 7.
Convergence of 4D MB reconstruction surrogated by the motion estimation objective function defined in Eq. (8).

Fig. 8.

Reconstructed images (from left to right: coronal phase 1, coronal phase 3, sagittal phase 1, and sagittal phase 3) from different reconstruction methods for real patient data at the regular dose and pseudo-real moving blocker data at $2/3$ of the regular dose. The red dashed boxes are for quantitative measures in Table 4 and 5.

Fig. 9.

The blocker position versus the projection view angle. The red dot denotes the blocker position of the phase 1. All phases have the same blocker position distribution.

The structure similarity index (SSIM) and the root mean square error (RMSE) in HU for different acquisition and reconstruction methods, both calculated on the volume of interest (VOI) centered at the phantom.

Table 1

		3D TV		4D SMEIR	
	Conventional 30 views/phase	Conventional 20 views/phase	MB 30 views/phase	Conventional 20 views/phase	MB 30 views/phase
SSIM	0.78	0.65	0.67	0.81	0.86
RMSE	70	91	89	62	59

Table 2

The SSIM and RMSE values for different acquisition methods (conventional acquisition and MB acquisition) and reconstruction methods (3D TV and 4D SMEIR) with/without scatter correction (SC). Note that the radiation dose of the MB acquisition is only two thirds of the conventional acquisition.

	Without MB and SC		With MB and SC	
	3D TV	4D SMEIR	3D TV	4D SMEIR
SSIM	0.58	0.63	0.67	0.86
RMSE (HU)	136	131	89	59

Table 3

The maximum and mean deviations from the true tumor motion (from the phantom) for different acquisition methods (conventional acquisition and moving-blocker (MB) acquisition) and reconstruction methods (3D TV and 4D SMEIR) with/without scatter correction (SC).

	Without MB and SC		With MB and SC	
	3D TV	4D SMEIR	3D TV	4D SMEIR
Max (mm)	6.15 mm	3.50 mm	3.19 mm	3.01 mm
Mean (mm)	3.14 mm	1.94 mm	1.58 mm	1.48 mm

SSIM and RMSE for different reconstruction methods using 4D SMEIR without MB (phase 1 as the reference phase for SMIER) as the ground truth for phase 1 and 3.

Table 4

	Without MB		With MB			
	3D TV		3D TV		4D SMEIR	
	Phase 1	Phase 3	Phase 1	Phase 3	Phase 1	Phase 3
SSIM	0.949	0.900	0.940	0.900	0.978	0.967
RMSE (HU)	38	62	42	59	34	44

Table 5

SSIM and RMSE for different reconstruction methods using 4D SMEIR without MB (Phase 3 as the reference phase for SMEIR) as the ground truth for phase 3.

	Without MB	With MB	
	3D TV	3D TV	4D SMEIR
SSIM	0.947	0.938	0.981
RMSE (HU)	34	41	31