Feasibility of correcting for realistic head motion in helical CT

Jung-Ha Kim, Tao Sun, Johan Nuyts, Zdenka Kuncic, and Roger Fulton

***Abstract*—Recently we proposed a method to accurately estimate and compensate for six degree-of-freedom (d.o.f.) rigid motion in helical CT by restoring projection consistency using data from a motion tracking system, and reconstructing with an iterative fully 3D algorithm. The method was demonstrated in simulations and in real CT scans using a 3D Hoffman brain phantom. From these experiments, it was found that certain combinations of motion and helical pitch can lead to data insufficiency particularly at higher pitch values. To examine the practical value of the motion correction method, it is of interest to develop a procedure to assess data sufficiency for different human head motion patterns and different brain imaging protocols. We propose to investigate this with simulations. In this work, we describe the method of creating and validating simulations by comparing the simulation results to the outcome of real CT scans. We also collected various head motion patterns from three volunteers using an optical tracking system and applied them to simulated CT scans. Further, the effect on data completeness for different motion patterns was assessed by computing the local Tuy values. The simulated images both with and without motion correction were highly correlated and almost identical to real CT images. Significant improvement was found in the motion corrected images, with the motion artefacts effectively suppressed. Only one of 12 motion corrected images appeared to have residual artefacts due to the data insufficiency based on the local Tuy value map. This simulation technique provides us with an insight into types of head motion that can be effectively compensated by using the motion correction technique.**

***Index Terms*—Motion estimation, motion compensation, computed tomography, image reconstruction, reconstruction algorithms.**

I. Introduction

PATIENT head motion is one of the major causes of image artifacts particularly in young patients and patients suffering from claustrophobia, a mental or behavioural incapacity or head trauma. Only a few methods have been published to retrospectively compensate for rigid motion artifacts in fan beam [1] or helical CT [2, 3], all which are limited to 2D in-plane motion, and also have only been investigated in simulations.

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We have previously demonstrated the feasibility of tracking six d.o.f. rigid motion during helical CT scans using an optical motion tracking system [4] and applying these data to retrospectively compensate for arbitrary rigid motion in actual CT scans [5]. Our results showed that the motion correction can recover an almost undistorted image in the presence of large and rapid rigid motion during helical CT scanning. However, some residual motion artefacts were observed in some parts of the phantom after motion correction when a helical pitch of ≥ 1 was used. This suggested that certain combinations of motion and helical pitch can lead to data insufficiency.

This study extends our previous work by developing and validating a simulation method, which can accurately represent and reproduce the data acquired from the real CT scanner. The aims of this study are to develop a reliable simulation tool to accurately predict the location and extent of residual data insufficiency-related artefacts following application of the motion correction method, and use it to assess (a) the likelihood of such artefacts given a range of feasible head motion patterns derived from normal volunteers, and (b) the relationship between the incidence of artefacts and CT scan parameters.

II. Methods

A. Motion Correction Principle

Projection inconsistencies are introduced when the imaged object moves during the projection acquisition in helical CT, which may result in motion artefacts if reconstructed without correction. The projection consistency can be restored by by applying the inverse of the object motion to source-and-detector pairs, which creates a virtual scanner geometry. Theoretically, by reconstructing an image from the consistent projections in the motion-corrected virtual scanner trajectory, the motion effects can be eliminated. However, it is important to note that this virtual scanner trajectory may introduce gaps in the projection views and therefore may not be sufficient for exact reconstruction.

The motion correction algorithm requires a method to accurately estimate the time series object motion during helical CT scans, which was obtained using an optical motion tracking system [4]. It also requires a fully 3D iterative reconstruction algorithm, which can reconstruct an image from the projection data in any arbitrary scanner geometry. For that purpose, the maximum-likelihood algorithm for transmission tomography (MLTR) was used [5, 6].

B. Simulation validation

1) CT Scans: The motion correction method was tested in a Siemens Biograph mCT PET/CT scanner (Siemens Medical Solutions, Malvern, PA), which incorporates a standard Somatom Definition AS 64-row helical CT scanner. CT scans were performed with the phantom held in an elevated position on the curved surface of the bed by a wedge. In this position a stationary helical CT scan of the phantom was carried out which served as a reference. Then without moving the phantom, a repeat CT scan was performed, during which the wedge was removed remotely by pulling a string, causing the phantom to undergo an oscillatory rolling motion. A clinical paediatric head imaging protocol was used with the following scan parameters: tube voltage $120\,\mathrm{kV_p}$; tube current $615\,\mathrm{mAs}$; collimation $64 \times 0.6\,\mathrm{mm}$; axial pitch of 1.5.

The scan of the stationary phantom was reconstructed with standard MLTR while the moving phantom scan was reconstructed with MLTR with and without motion correction. The reconstructed images had a dimension of $512 \times 512 \times 276$ with a voxel size of $1 \times 1 \times 1\mathrm{mm}^3$. The reconstruction of the stationary phantom was used as a motion-free reference, and as a voxelized phantom in simulations.

2) Motion tracking: An optical motion tracking system (Polaris Spectra, Northern Digital Inc., Waterloo, Canada) was used to record a time-series of six d.o.f. pose estimates of a rigid-body target that was attached to the front-end of the phantom. An additional target was attached to the front edge of the bed to record bed motion. Collected pose measurements were converted to CT isocenter coordinates using a 4×4 transformation matrix determined from a cross-calibration procedure described in [4]. Components of the phantom motion that were due to bed motion were removed from the phantom pose estimates, and a time-series of motions relative to the initial pose of the phantom was calculated. A 17-point second degree Savitzky-Golay (SG) polynomial [8] was then fitted to the data for each d.o.f. by least squares to reduce measurement jitter in the pose estimates. These smoothed motion data were synchronised with the projection data based on time stamps. The inverse of the motion data was applied to the scanner geometry for motion correction.

3) Simulations: In simulating the CT data acquisition process, the voxelized phantom was forward projected using the projector in a normal CT geometry. The projector was defined to match the acquisition parameters that were used for the real CT scans. To simulate the scan with motion, the phantom was transformed at each projection view according to the motion data acquired during the experiment described in Sec. II-B2 while no motion was applied to the image for stationary phantom data.

Simulated scans with motion were reconstructed using MLTR with and without the motion correction whereas the stationary phantom data were reconstructed without motion correction. For motion correction, the inverses of the motion of the phantom were applied to the scanner geometry, and (back)projection was calculated using this virtual scanner geometry.

Reconstructions of real and simulated CT scans subjected to the same motion were compared visually and by calculating Pearson correlation coefficients over all voxels in the 3D volume.

C. Simulations with Volunteer Head Motion

To assess the efficacy of the motion correction method for more realistic head motion, human volunteer motion data were collected using an Optitrack (Natural Point Inc, Corvallis, OR USA) from three healthy volunteers. Seven retro-reflective spherical markers were attached to the volunteer's head. The volunteer was asked to execute four types of motion: no motion, slight, moderate and severe motion for one minute each. The tracker data were then processed in the same way as described in Sec. II-B2.

A voxelized 3D anthropomorphic head phantom (PBU-60, Kyoto Kagaku Co. Ltd, Kyoto, Japan) with a voxel size of $1.5 \times 1.5 \times 1.5\,\mathrm{mm}^3$ was used as an object for volunteer motion simulation. Simulated scan parameters were determined based on a standard brain imaging protocol with $120\,\mathrm{kVp}$; collimation of 64×0.6; rotation time of $1.0\,\mathrm{s}$; and a helical pitch of 0.8.

For the stationary reference scan, the object image was forward projected in a conventional CT orbit, whereas, in the motion scan, a 3D transformation derived from the acquired volunteer motion was applied to the object image prior to forward projection at each view angle. Simulated sinograms with motion were reconstructed with and without motion correction.

The accuracy of the motion correction was evaluated quantitatively by comparing the motion corrected image to the stationary reference image over all voxels using three metrics: root-mean square error (RMSE), Pearson correlation coefficient (CC) and structural similarity image measure (SSIM).

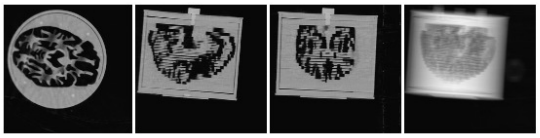
D. Data insufficiency assessment

The completeness of each irregular CT trajectory produced by the volunteer motion was assessed at each voxel of the motion corrected image by computing the degree to which the local Tuy condition [7] was satisfied. Local Tuy values range from 0 to 1. A lower value ($\simeq 0$) indicates sufficient data for exact reconstruction, and as the value becomes higher, more severe violation of the local Tuy condition is indicated. To assess the data insufficiency for the given volunteer motion, this Tuy value map was computed and compared to motion corrected images.

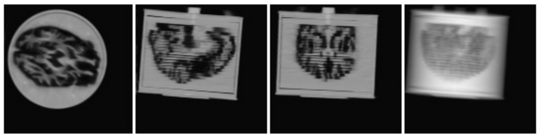
III. Results and Discussion

Fig. 1 shows reconstructions of real and simulated CT scans with and without motion. The simulated images appear more blurred than the real images due to interpolation of voxel values when transforming the images. Nevertheless, the simulated and real images exhibited similar artefacts and had Pearson correlation coefficients of 0.97 and 0.96 for stationary and motion-corrected images, respectively. For comparison, the Pearson correlation coefficient of the stationary vs motion corrected real CT images was 0.80.

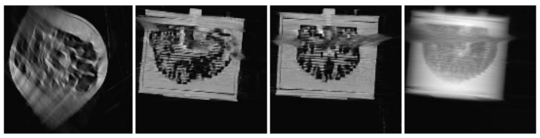
The four different types of head motion for one of the volunteers are plotted in Fig. 2. Rotations and translations in



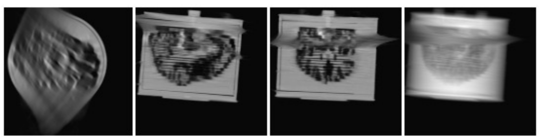
(a) Stationary (Real CT)



(b) Stationary (Simulated CT)



(c) Motion corrected (Real CT)

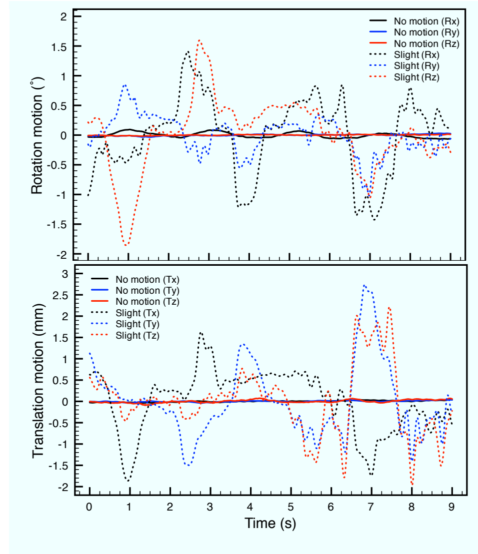


(d) Motion corrected (Simulated CT)

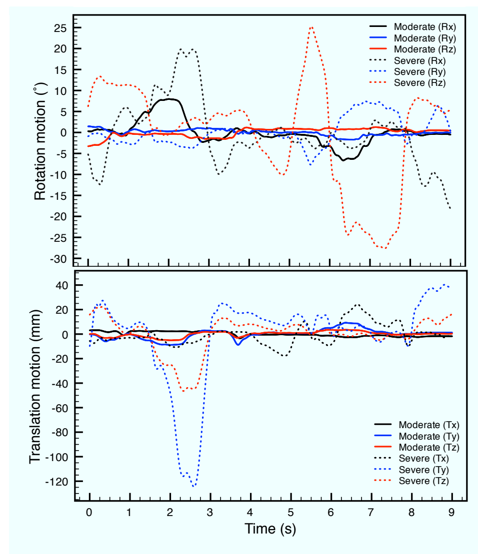
Fig. 1. Reference (stationary) and motion-corrected images for real and simulated CT scans. Each column shows, from left to right, transaxial, coronal, saggital and projection images.

the three axes for this volunteer were up to -30° and -120 mm, respectively. Reconstructed slices from the simulations with and without motion correction are shown in Fig. 3. Motion artefacts in uncorrected images are shown to be effectively suppressed with motion correction in all cases. Substantial improvements in the motion corrected images can also be confirmed with the average metric values in Fig 4.

For one of the volunteers, some residual artefacts were observed for the severe motion. A Tuy value map of this motion is shown in Fig. 5. Regions with insufficient data indicated by high Tuy values are shown to correspond well with the residual artifacts in the motion corrected images. This strongly suggests that these artefacts are due to incomplete sampling.



(a) No & Slight motions



(b) Moderate & Severe motions

Fig. 2. Time series motion of the measured head motion from one volunteer.

IV. Conclusion

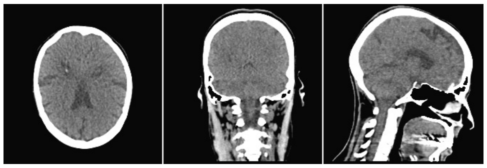
We have successfully developed and validated a simulation tool that appears able to accurately predict the location and severity of residual artefacts in the motion-corrected CT image for any given incomplete source/detector trajectory. We have also begun collecting a variety of head motion data from volunteers in order to assess likelihood of occurrence of residual artefacts due to data insufficiency if the motion correction method is applied clinically. Only for one of the 12 volunteer motion sets, which simulated severe motion, some artefacts were observed after motion correction. Analysis of the corresponding Tuy value map indicated that these artefacts were due to data insufficiency. The next steps are to gather some more volunteer motion data to make a fuller assessment of the likelihood of artefacts with realistic motion, and to optimise the CT scan parameters, prior to a clinical trial.

Acknowledgment

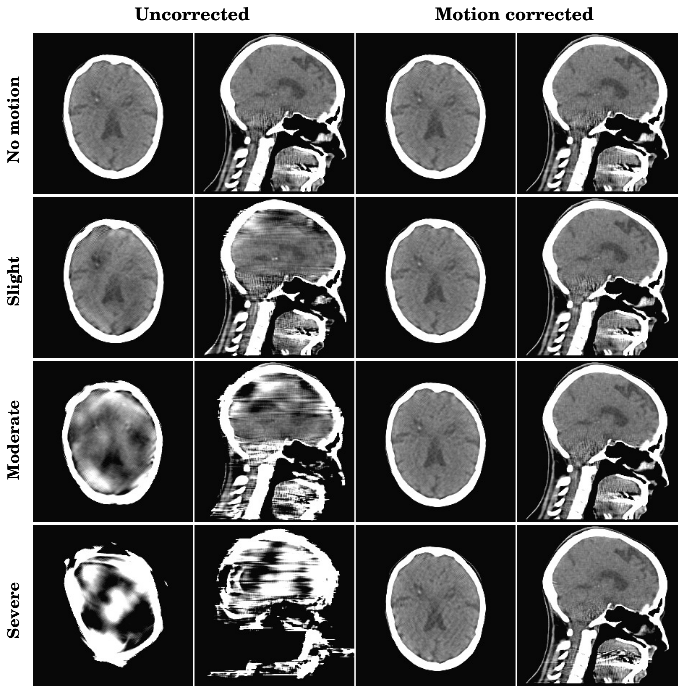
This work is supported by National Health and Medical Research Council Australia Grant 632677, by the KU Leuven IMIR project and by the MIRIAD SBO project of IWT, Flanders. J.-H. Kim is supported by a University Postgraduate Award from the University of Sydney. We gratefully acknowledge Siemens Medical Solutions, USA (Inc.) for assistance with reading the raw CT data.

References

[1] H. Yu and G. Wang, "Data consistency based rigid motion artifact reduction in fan-beam CT," *IEEE Trans. Med. Imag.* **26** 249–260 (2007).

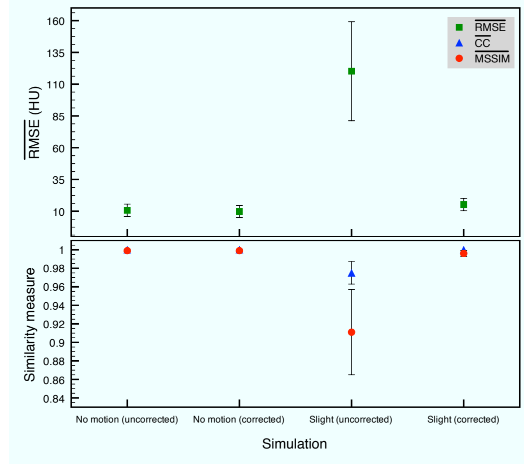


(a) Object image

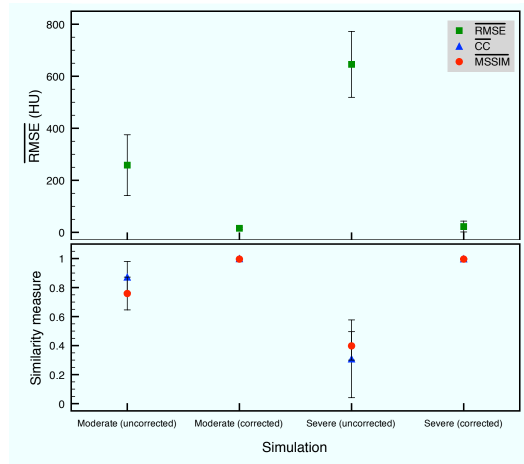


(b) Simulated images

Fig. 3. (a) 3D slices of the object image, and (b) Transaxial and sagittal slices of the simulated images of uncorrected (left two columns) and motion corrected (right two columns) images for different volunteer head motion types of one volunteer [WL=+40 HU, WW=+140 HU].



(a) No & Slight motion



(b) Moderate & Severe motion

Fig. 4. Average RMSE, CC and SSIM calculated on a 3D volume of images simulated with (a) No and slight motion patterns; and (b) Moderate and severe motion patterns of one volunteer.

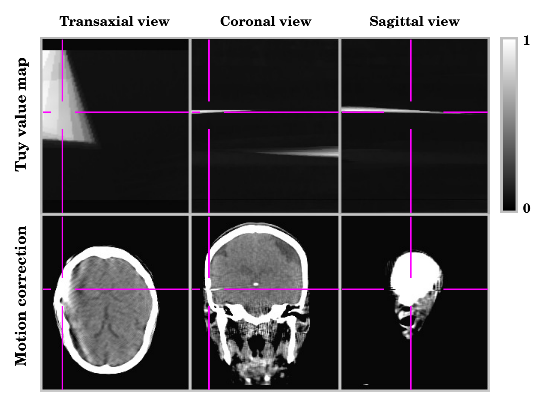


Fig. 5. 3D slices of the Tuy value map (Top) and corresponding motion corrected images (Bottom) for one volunteer severe motion.

[2] G. Wang and M. W. Vannier, "Preliminary study on helical CT algorithms for patient motion estimation and compensation," *IEEE Trans. Med. Imag.* **14**(2) 205–211 (1995).

[3] S. Zafar, "Post scan correction of step, linear and spiral motion effects in CT scans," *Int. J. Comput. Appl.*, **35**(10) 13–19 (2011)

[4] J.-H. Kim, J. Nuyts, Z. Kuncic and R. Fulton, "The feasibility of head motion tracking in helical CT: A step toward motion correction," *Med. Phys.*, **40**(4) 041903-1–4 (2013).

[5] J.-H. Kim, A. Kyme, J. Nuyts, Z. Kuncic and R. Fulton, "A rigid motion correction method for helical computed tomography (CT)," *Phys. Med. Biol.* (in press) (2015).

[6] J. Nuyts, B. De Man, P. Dupont, M. Defrise, P. Suetens, and L. Mortelmans, "Iterative reconstruction for helical CT: A simulation study," *Phys. Med. Biol.*, **43** 729–737 (1998).

[7] T. Sun, R. Clackdoyle, R. Fulton and J. Nuyts, "Quantification of local reconstruction accuracy for helical CT with motion correction," *2014 IEEE Nuclear Science Symposium Record*, (2014).

[8] A. Savitzky and M. J. E. Golay, "Smoothing and differentiation of data by simplified least squares procedures," *Anal. Chem.*, **36**(8)1627–1639 (1964).