Phantom Investigation of Three-Dimensional Motion Dependent Volume Aliasing During CT Simulation for Radiation Therapy Planning

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INTRODUCTION

Tumor localization and delineation for treatment planning in radiation oncology is commonly performed using computed tomography (CT). Owing to image matrix size, slice thickness, and window and level settings, an overestimation of a static target's physical volume is typically observed due to partial volume sampling effects. [1] Organ motion, most pronouncedly observed in the thorax and the abdomen, further challenges CT-based targeting due to the potential for insufficient temporal sampling of the moving target. Clinically, these uncertainties can result in errors in representation of true tumor location, extent, and associated motion envelop (the 3D-space that is occupied by a target volume over time, due to respiration and other motion inducing positional variations). Thus, it is critical to understand the potential challenges and limitations of CT-based target localization and delineation, as they correlate directly with the ability to accurately deliver a radiation oncology treatment in anatomical sites that are subject to organ motion.

Volume aliasing, or a CT misrepresentation of the true spatial and geometric parameters of well-defined volumes, has been investigated experimentally and/or analytically for targets moving freely in a single dimension (longitudinally or transversally) [2, 3]. The present study investigates the impact of clinically relevant, *three-dimensional* target motion of well-defined geometric targets using a prototype motion phantom (now commercially available from CIRS, Computerized Imaging Reference Systems Inc., Norfolk, VA). The aim of this study was to specifically quantify volume aliasing for known, clinically relevant, three-dimensional tumor motion amplitudes as a function of spiral CT image acquisition time, phase and slice number (single vs. multislice) as well as target size and motion amplitude.

MATERIAL and METHODS

A series of spiral CT scans were acquired using 1) single-slice fast, 3) single-slice slow, and 3) multi-slice fast scanning techniques on dynamic spherical targets (10 and 31.5 mm in diameter), embedded in a dynamic anthropomorphic phantom. Three dimensional target motions typical of clinically observed tumor motion parameters were investigated. Target motion excursions included ±5 mm, ±10 mm, and ±15 mm displacements in the S-I direction synchronized with constant displacements of ±5 mm and ±2 mm in the A-P and lateral directions. For each target, scan technique, and motion excursion, eight different initial motion-to-scan phase relationships were investigated. Target delineation was performed via a user independent, semi-automated process that was confirmed to be reproducible.

RESULTS

True target volume mis-estimation

The key findings are as follows: 1) independent of scan technique and motion amplitude of the target, the large target volume is overestimated by as much as by 1.5 to 2 fold (i.e. 150% to 200% overestimation of target volume), and the small target volume is overestimated by as much as 3 to 11 fold (i.e. 300% to 1100% overestimation of true target volume) ; 2) independent of scan technique and motion amplitude, both small and large target dGTV's are variable as a function of imaging phase ; 3) the small target dGTV volume is highly variable as a function of the phase relationship between target motion and image acquisition i.e. based solely on when the scan was initiated relative to the periodic motion of the target, the 15 mm dGTV ratio can vary between 2.4 and 5.0 (i.e. the overestimation of the true target size can vary from 240% and 500%).

The mean overestimation for single slice fast scan CT technique was as much as 3.38 times (or a 238% increase in) the original small (10 mm diameter) target volume and 1.57 times (or a 57% increase in) the original large (31.5 mm diameter) volume. The mean overestimation for multi-slice fast scan CT technique was as much as 4.65 times (or a 365% increase in) the original small target volume and 2.08 times (or a 108% increase in) the original large volume. Finally, the mean overestimation for single slice slow scan CT technique was as much as 11.1 times (or a ~1000% increase in) the original small target volume and 2.26 times (or a 126% increase in) the original large volume (table 1). For qualitative appreciation of motion-induced volumetric distortion during CT imaging, coronal plane projections of the moving targets are presented in Fig 2., with images of the stationary target depicted in column 1 of each image set.

Phase synchronization related centroid misplacement

Fig. 1 depicts the magnitude of displacement of the imaged centroid from reference centroid location for a moving target. While no predictable relationship between the displacement of the reference centroid and initial motion phase was discerned, key findings include 1) a greater centroid misplacement in the S-I direction for fast scan techniques, 2) a greater centroid misplacement in the transaxial (AP and L-R) directions for the slow scan technique, and 3) for identical motion pattern and phase synchronized, a greater centroid displacement for the smaller target.

Fig. 3 depicts a coronal plane CT representation of: a) the static target at reference position-sGTV (grey shading) b) the calculated true motion envelope representing the known motion envelope of the target (black contour) and c) the dynamically imaged target-dGTV for fast scan example cases of both small and large targets. It is clear from the image that the dGTV (green) differs significantly from the known true potion envelope, or what might logically be viewed as the ideal PTV.

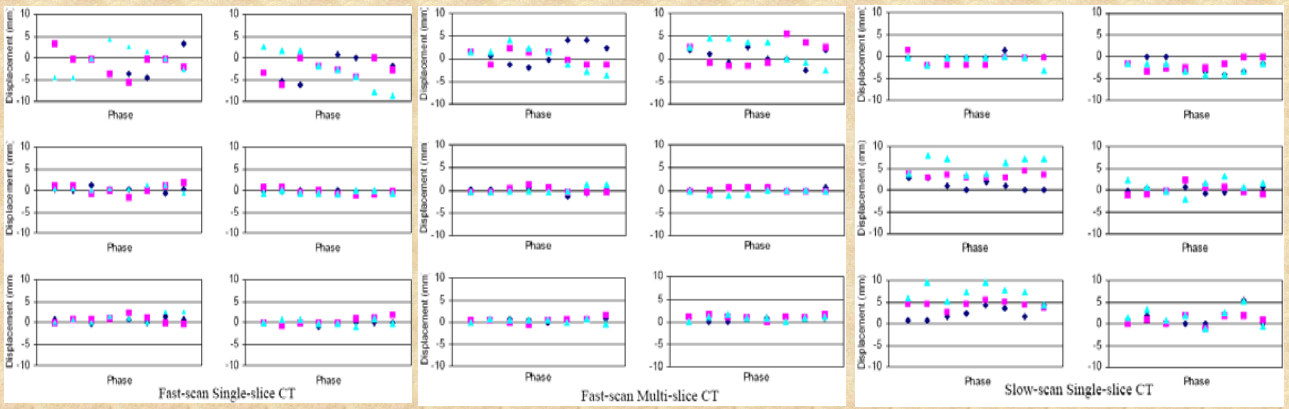

Fig. 1 Reference centroid misplacement as a function of three (3) motion amplitudes and eight (8) initial motion phases for the 10- and 31.5-mm diameter targets. Rows one, two, and three represent misplacement of the reference centroid location in the S-I, A-P, and L-R directions, respectively. The blue, pink and teal colored lines respectively represent motion amplitudes of ±5, ±10, and ±15 mm in the S-I direction. Each S-I motion is synchronized with an A-P and an L-R motion of ±2 and ±5 mm respectively.

Target Diameter (mm)	S-I Motion Amplitude ± (mm)	Single-slice Fast		Multi-slice Fast		Single-slice Slow	
		Mean	1σ	Mean	1σ	Mean	1σ
10	5	2.52	1.02	2.85	0.70	4.57	0.62
	10	2.68	0.21	3.61	0.37	6.84	0.99
	15	3.38	1.07	4.65	0.39	11.1	0.60
31.5	5	1.35	0.08	1.56	0.03	1.60	0.06
	10	1.45	0.03	1.79	0.05	1.89	0.07
	15	1.57	0.05	2.08	0.05	2.26	0.11

Table 1. Range of volume overestimation as a function of motion amplitude. S-I motion was synchronized with a fixed R-L and an A-P displacement of ±2 and ±5mm respectively.

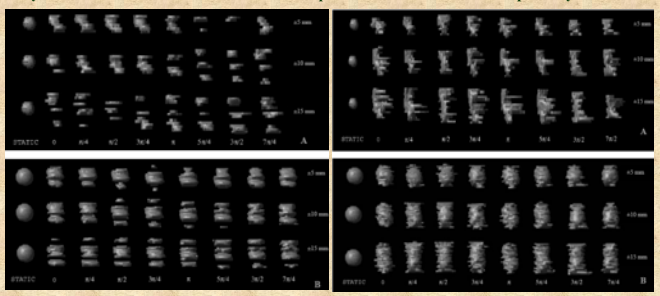

Fig. 2 Volumetric distortion of a 10- (image set A) and 31.5-(image set B) mm diameter targets as a function of motion amplitude and initial motion phase for a fast or 1-sec (left) and a slow or 4-sec scan (right) CT imaging techniques.

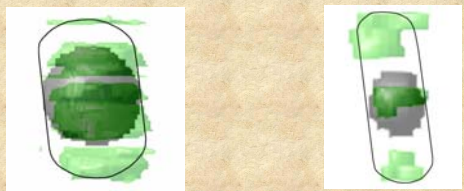

Fig. 3 Coronal plane representations of statically imaged target (sGTV) in grey shading, known true motion envelope TME) for the phantom-based target's motion over time (black contour) and dynamically imaged and delineated target (dGTV) in green shading. Note the significant difference between the TME, or ideal PTV, and the dGTV.

CONCLUSIONS

CT imaging of targets undergoing 3D motion have been quantified to result in significant misrepresentation of target volume, shape and location. These aliasing effects are seen to be difficult to predict due to their phase dependence. For the current study, wherein the true motion envelope (TME) is known for a phantom-based target moving in 3 dimensions, large differences between the TME and dGTV are observed, thus documenting the potential for targeting errors in situations where typical 3D respiratory induced motions are present.

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