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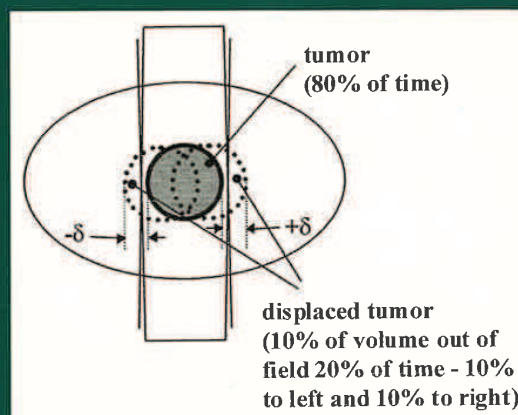
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High-Precision Radiation Therapy of Moving Targets

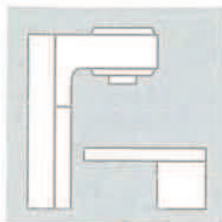
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Seminars in RADIATION ONCOLOGY

Joel E. Tepper, MD, Editor

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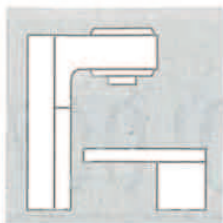
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High-Precision Radiation Therapy of Moving Targets **Thomas Bortfeld, PhD, and George T.Y. Chen, PhD, Guest Editors**

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Cover Illustration: Schematic view of a moving tumor. See Goitein, page 2.

Artifacts in Computed Tomography Scanning of Moving Objects

George T.Y. Chen, Jong H. Kung, and Kevin P. Beaudette

Target volumes in the thorax and abdomen are commonly computed tomography (CT) scanned during light respiration. In this article, we analyze the distortions introduced in helical scanning of moving objects. Objects of known geometry are placed on a moving sled and scanned in a multirow helical CT scanner. The motion of the sled approximates the magnitude and velocity of organ movement in patients during light respiration (amplitude 1 cm, period 4 seconds). Scans of the phantom are obtained in high speed mode at incremental phases of respiration, and the resulting images are compared with scans obtained when the phantom is static. Computer simulations of the scan process are also performed to interpret the results and extend the

analysis to a greater range of parameters in scanning, motion, and object size. Resulting scans show that spherical test objects can be shortened by as much as 2 cm or twice the periodic motion amplitude. Object shape was significantly distorted, and the geometric center of the object was displaced by as much as ± 0.8 cm. Computer simulation results qualitatively agree with the experimentally observed phantom images. These simulations predict that the effect is clearly observable even if the amplitude is decreased to 0.5 cm. Implications of scanning moving objects on treatment planning are discussed.

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Helical computed tomography (CT) simulators are standard technology in modern radiation oncology. Helical scanning¹ has many advantages over conventional CT, including volumetric data acquisition in a single breath hold, reduction of motion artifacts, high speed, improved z-axis resolution,² and excellent image quality. Within the past several years, multidetector-row helical CT has become available and provides 2- to 3-fold improvement in volume coverage speed with comparable diagnostic image quality.³ Unlike diagnostic scans, treatment planning scans in the thorax and abdomen are commonly performed with the patient breathing lightly. The rationale for this scan technique has been to scan the patient the same way he/she is treated (eg, during light breathing). From these images, gross target volumes are outlined and subsequently expanded to account for microscopic extension, setup uncertainty, and organ motion.^{4,5}

Organs in the thorax and abdomen move periodically along the craniocaudal and anterior-posterior axes. Magnitudes of the motions are on the order of several centimeters, with a period of

~4 seconds.^{6,7} When imaging moving organs during CT scanning, distortions can occur.^{8,9} Commonly observed distortions include the discontinuities seen in the coronal multiplanar view of the diaphragm as well as the possibility of artifacts in CT sampling caused by breathing. "Shrink wrapping" the observed target volume and applying a margin are generally considered to be adequate to cover the clinical target volume excursions during respiration.

The accurate 3-dimensional geometric modeling of an organ is a prerequisite for precision treatment planning and directly affects both aperture shape and target center in beam's eye view based planning. The potential for distortions and deformations of organ shape should be understood in detail and strategies for mitigation developed. The objective of this study is to visualize and quantify the distortions of geometric objects scanned under clinical conditions and identify those parameters of organ motion and scan technique that most significantly affect the magnitude of such effects. Ideally, 4-dimensional trajectory information on organ motion in vivo can be of importance in understanding the true dose delivered by advanced beam delivery methods such as intensity modulated radiotherapy. We discuss new 4-dimensional scanning techniques later.

Scanning and Simulating Scanning of Moving Objects

Experimental and analytic data presented include the (1) scanning of a moving phantom, in

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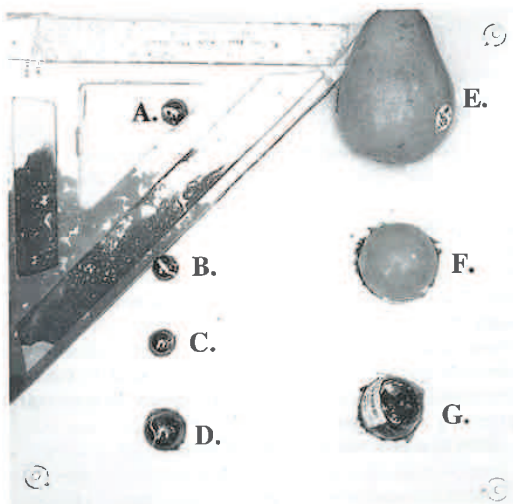


Figure 1. Photograph of the phantom of scanned geometric objects. Objects A through D are glass marbles. Object E is a pear (fruit) and objects F and G are rubber balls. A drafting triangle is shown in the upper left hand region of the Styrofoam block in which the objects are embedded. A rod attached to wheel drives the sled in an oscillatory motion with amplitude 1 cm and period of 4 seconds. The phase of the periodic motion is advanced by $\pi/4$ for each volumetric scan.

which are embedded well-defined geometrically shaped objects, and the visualization of resulting distortions and (2) the modeling and simulation of CT scanning to better understand the dependence of the resulting object shape as a function of motion and scan parameters.

Scanning Moving Phantoms

Spherical objects (diameters ranging from 1.6 cm to 6.3 cm) are embedded in a Styrofoam block as shown in Figure 1. The left column of spherical objects are glass marbles; objects F and G are rubber balls. The true diameters of the spheres are included in Table 1. In some studies, a non-

Table 1. Dimensions of Phantom Spheres

Label	Object	Diameter (cm)
A	Small marble	1.6
B	Small marble	1.6
C	Small marble	1.6
D	Large marble	2.4
E	Pear (length)	9.4
F	Medium ball	4.8
G	Small ball	3.6

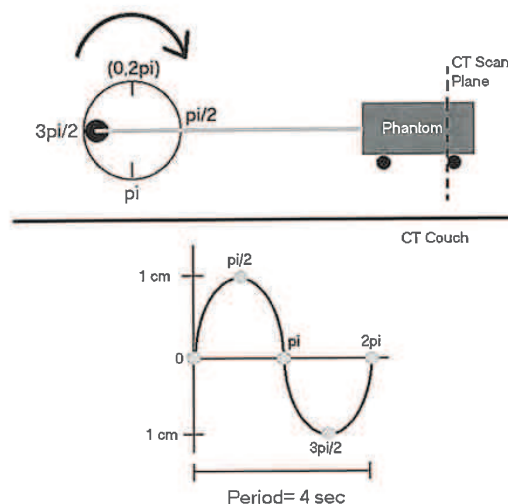


Figure 2. Schematic representation of moving phantom apparatus. (Above) The sled is driven by an electric motor that drives a wheel / arm assembly. (Below) Motion phase is a key parameter at the instant of object scanning.

spherical irregular object (object E, a pear) and a drafting triangle are also placed in the scan field. The phantom assembly is placed on a moving sled, which in turn is placed on the CT scanner couch. A 1-dimensional sinusoidal motion of the sled is vectored along the scanner table axis, simulating craniocaudal motion consistent with respiration. The motorized sled drive is adjusted to provide periodic motion of amplitude 1.0 cm with a period of ~ 4.0 seconds. The experimental setup is shown in Figure 2. The phase of the scan for a specific object is that angle in sinusoidal motion at which the CT scan plane first intersects the test object. This phase angle is correlated with the velocity of the object at the time of object sampling for the sinusoidal model.

The phantom assembly is scanned on a GE (Waukesha, WI) Lightspeed QX/i scanner in helical mode. Scanning is performed in both the high-speed (HS) mode, with slice thickness of 3.75 mm, an effective speed of 0.2 s/slice, and pitch of 1.5 cm, and the high-quality (HQ) mode, with identical slice thickness but an effective speed of 0.4 s/slice and a pitch of 0.75 cm. An initial scan is performed with the sled in static mode (no periodic motion). This is followed by a series of 8 CT scans taken with the sled moving periodically, with different initial phases ranging

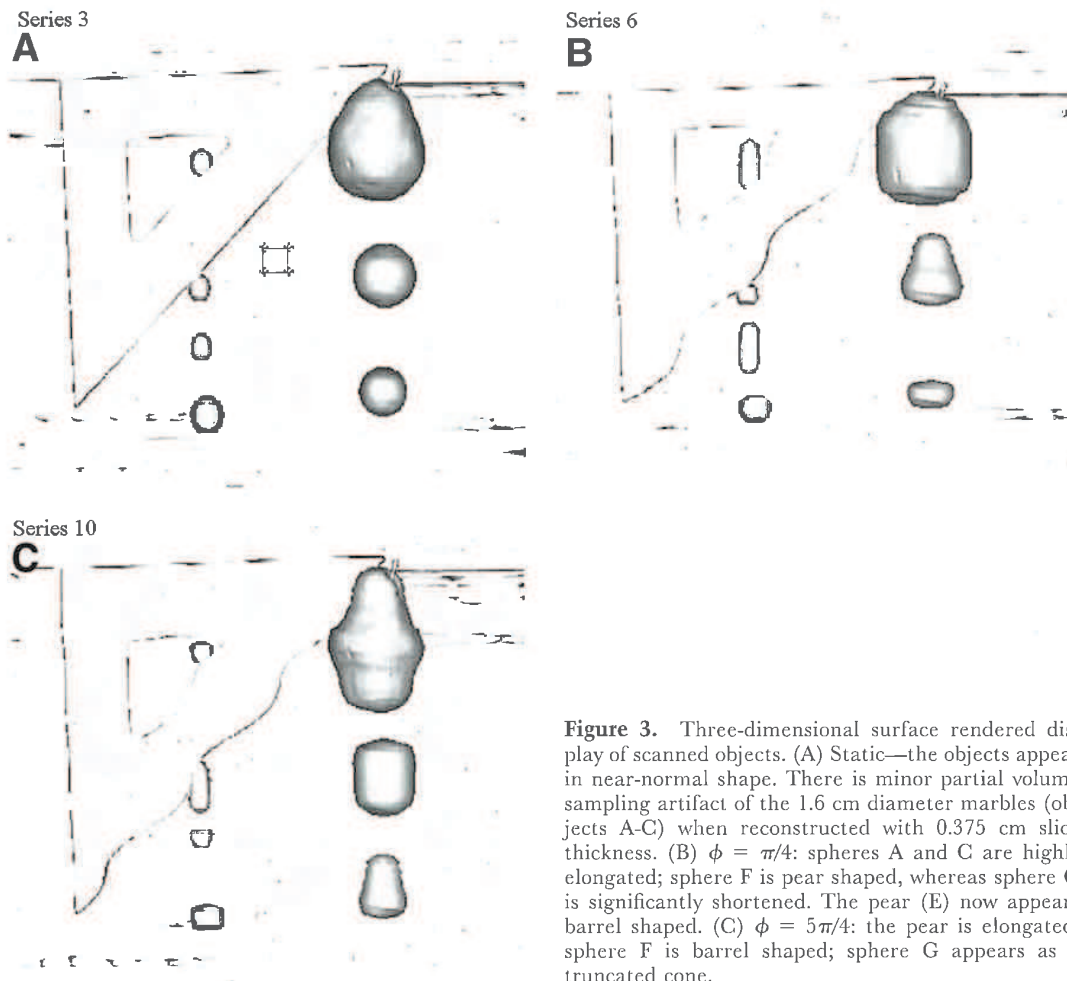


Figure 3. Three-dimensional surface rendered display of scanned objects. (A) Static—the objects appear in near-normal shape. There is minor partial volume sampling artifact of the 1.6 cm diameter marbles (objects A-C) when reconstructed with 0.375 cm slice thickness. (B) $\phi = \pi/4$: spheres A and C are highly elongated; sphere F is pear shaped, whereas sphere G is significantly shortened. The pear (E) now appears barrel shaped. (C) $\phi = 5\pi/4$: the pear is elongated; sphere F is barrel shaped; sphere G appears as a truncated cone.

from relative phase of $\phi = 0$ to 2π in increments of $\pi/4$. An 8-second prescan delay is selected before scanning begins; at the initiation of the prescan delay, the sled motion is activated to bring the assembly to periodic motion, and scans are begun approximately 4 slices above the phantom.

The resulting scans are viewed with GE Advantage Sim software, Version V4.1.09. The objects are displayed in a 3-dimensional surface rendered mode, and object lengths are measured with ruler/measurement software within the simulation package.

Phantom Images

Figure 3A to C show representative volumetric images of the scanned objects in the static sled

and oscillating sled modes. When the sled is static, the spherical objects appear with symmetric longitudinal and transverse diameters within partial volume sampling uncertainties. There is a small distortion of the 1.6-cm diameter marbles (left column) because the slice thickness of 0.375 cm is relatively coarse for small spheres. When the sled oscillates during scanning, the resulting spheres can appear either foreshortened or lengthened along the axis of motion, depending on object scan phase. The object shape is significantly distorted (refer to Fig 1 for object labels). The pear (object E) in Figure 3B appears much more barrel shaped. Sphere F appears “pear shaped” in Figure 3B and barrel like in Figure 3C. Sphere G is transformed to a lengthwise compressed ellipsoid in Figure 3B and barrel

Table 2. Measured Lengths of Phantom Spheres

Series	Length (cm)	Diff from Static (cm)	% Diff
Object A			
3	2.16	0	0
6	3.98	1.82	+84.26%
10	1.74	-0.42	-19.45%
Object B			
3	2.11	0	0
6	1.69	-0.42	-19.90%
10	4.03	1.92	+91%
Object C			
3	2.06	0	0
6	4.08	2.02	+98.1%
10	1.36	-0.7	-34%
Object D			
3	2.95	0	0
6	2.25	-0.7	-23.73%
10	2.11	-0.84	-28.47%
Object E (Pear)			
3	9.43	0	0
6	8.82	-0.61	-6.47%
10	11.25	1.82	+19.3%
Object F			
3	4.79	0	0
6	5.57	0.78	+16.3%
10	6	1.21	+25.3%
Object G			
3	3.8	0	0
6	2.2	-1.6	-42.11%
10	5.01	1.21	+31.84%

shaped in Figure 3C. There is obvious lengthening and distortion of the marbles (objects A-D) in Figures 3B and C relative to 3A. The hypotenuse of the triangle appears to have waves when motion is present. Differences in object lengths for these spheres are shown in Table 2. Objects can be lengthened by as much as 2 cm. In reconstructed scans at other phases, objects are shortened by ~ 2 cm.

Computer Simulations

The process of scanning a moving geometric object and assessing resulting distortions of the reconstructed volume can be approximated through a simplified computer simulation. These simulations were performed by using MATLAB v6.1 by Mathworks (Natick, MA). The program models the geometric process of transaxial imaging by intersecting a scan plane with a sphere, whether static or moving sinusoidally along the scanner couch axis. In this simplified model, the transaxial plane is intersected with the sphere to yield a circular contour, whose radius can be

simply calculated as illustrated in Figure 4. There is no attempt to truly simulate the complex process of collecting ray projections through an object and performing a CT reconstruction. Simulation input variables include slice thickness, effective time for each slice acquisition (Δt), (high quality: $\Delta t = 0.4$ seconds; HS $\Delta t = 0.2$ seconds) sphere radius in cm (R), and motion parameters of the sinusoidal movement, including amplitude (A_0), frequency (f), and initial phase (ϕ).

Consider the simulation geometry in Figure 4, as viewed from above the CT scanner couch. From this viewpoint, the scan plane advances from top to bottom, generating a transaxial section of the sphere at each specified table index. The image on the left represents a scan, when the sphere is static. At the moment of acquisition, for a given table index and time, the scan plane intersects the sphere, resulting in an image of a circle of diameter R , which can be easily calculated from the radius of the object and the table index. On the right side of Figure 4, the oscillating sphere has moved upward toward the head of the couch, and its center appears at a higher z coordinate, resulting in a different intersection of the imaging plane and sphere. The value k' can be expressed as: $k' = k + A_0 \sin(\omega t + \phi)$, where k is table index value for the static condition, $t =$ time, $A_0 =$ amplitude, $\omega = 2\pi f$, and ϕ the phase angle. t (time) is indexed to table position.

As each slice is imaged, the resulting object shape is assembled, and in a classical BEV display

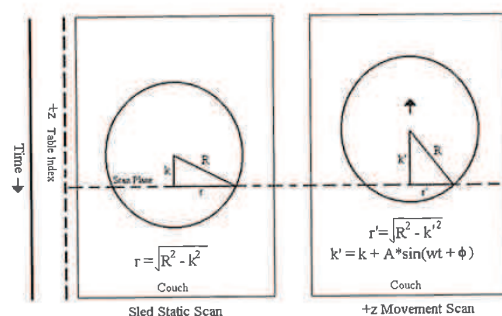


Figure 4. Schematic representation of scanning simulation; (left) a sphere of radius R is scanned at a distance k (determined from table index) from the sphere center. The imaged section is a circle of radius r . (Right) The sphere has oscillated upward and the distance from the sphere center to the scan plane is now k' , resulting in imaging of a circle of radius r' .

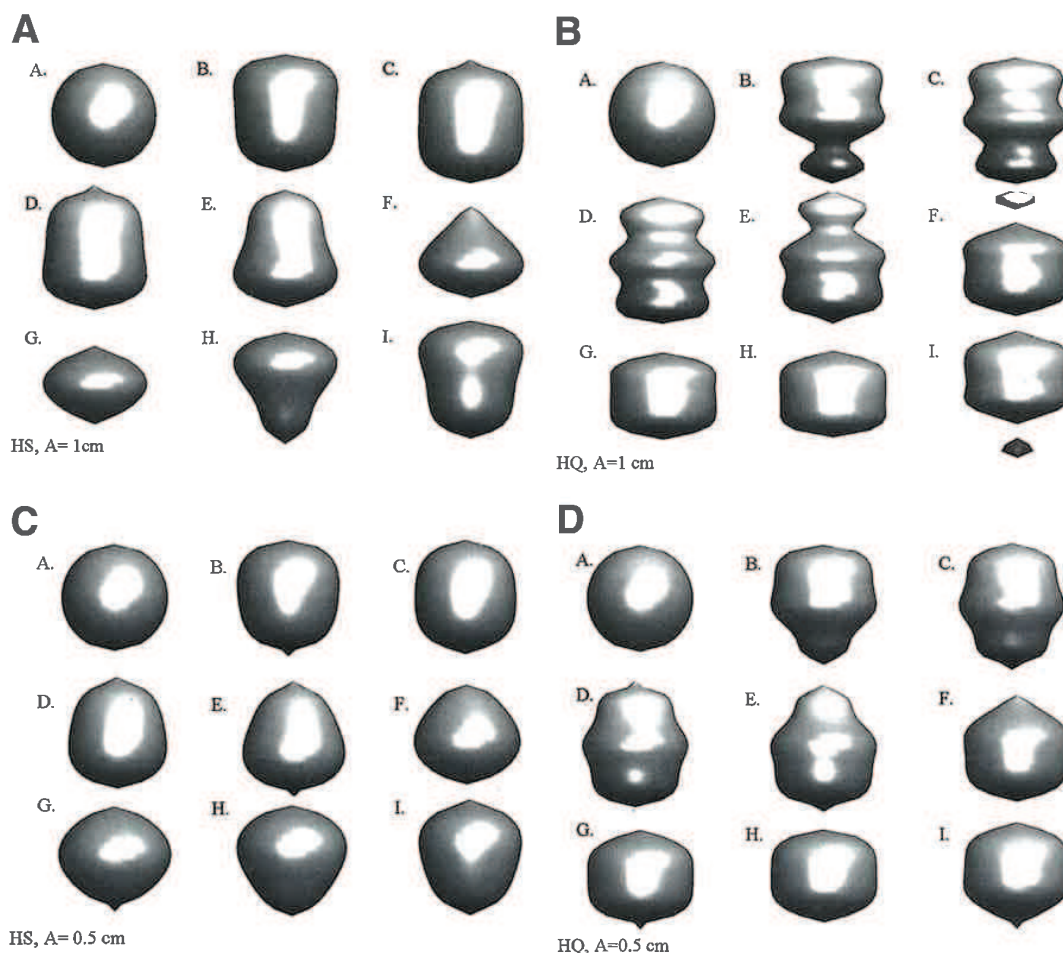


Figure 5. Simulations of scanning a 3-cm radius sphere under different conditions. Object A in each series is a volumetric representation of the object scanned in static mode. Objects B-I represent the sphere as reconstructed beginning with $\phi = 0$, and incremented by $\pi/4$ for each subsequent object. (A) HS mode, amplitude 1 cm; (B) high-quality mode, amplitude 1 cm; (C) HS mode, amplitude 0.5 cm; and (D) high-quality mode, amplitude 0.5 cm.

is seen as a series of circular contours. The multiple transaxial contours can also be rendered as a surface, as shown later.

Simulation Images

Results of the simulation of CT scans of spherical objects are shown in Figures 5A to D. In simulations A and B, the amplitude of periodic motion is 1.0 cm, whereas simulations C and D are performed with amplitude 0.5 cm. Simulated CT scans A and C are performed in high speed HS mode, in which the effective speed of the scanner is 0.2 s/slice. Simulations B and D model data acquisition in HQ mode, in which scan speed is

0.4 s/slice. Each subfigure A through D contains nine volume rendered objects of the sphere, labeled A-I. Figure A displays the sphere as visualized in the static mode. Each subsequent rendered image represents the sphere as imaged with a different initial phase offset. Image B shows the reconstructed object at $\phi = 0$, with subsequent images having an increment of phase angle by $\pi/4$ up to $7\pi/4$.

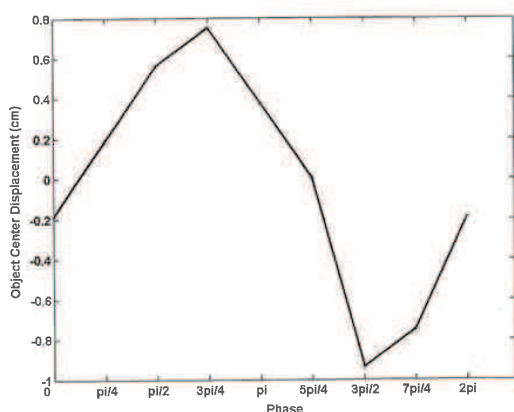
In comparing the objects scanned with HS versus HQ mode, 2 differences are apparent: first, with amplitude constant (comparing 5A to 5B and 5C to 5D), there is the appearance of undulations in the lateral edge of the object in

Table 3. Volume and Length Variations in Simulations

Object	Phase	Volume cm ³	Vol Diff %	Length (cm)	Length Diff %	COM Shift
A	0	112.7	0%	6.0	0	0
B	0	146.2	29%	7.1	19	-0.2
C	$\pi/4$	152.0	35%	7.1	19	0.2
D	$\pi/2$	139.4	24%	7.1	19	0.6
E	$3\pi/4$	110.8	-2%	6.8	13	0.8
F	π	75.1	-32%	5.3	-13	0.4
G	$5\pi/4$	69.2	-38%	4.5	-25	0.0
H	$3\pi/2$	88.2	-22%	6.4	6	-0.9
I	$7\pi/4$	123.7	10%	6.8	13	-0.8

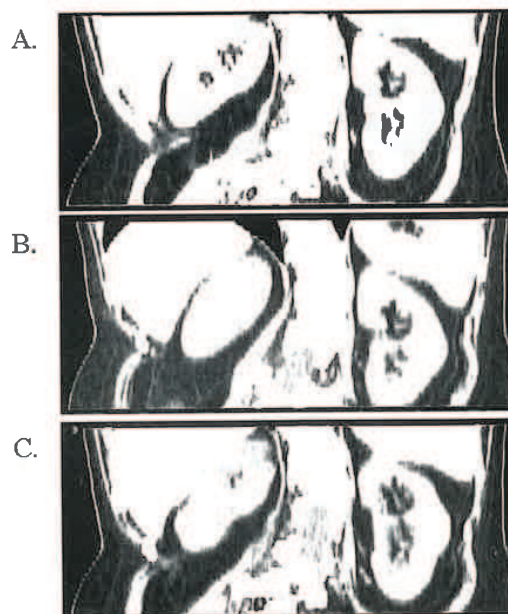
HQ mode that is not present in HS mode, and secondly, there are significant alteration of object shape in comparison to a sphere. The objects are lengthened, shortened, and drastically change shape, depending on phase angle. A table of the resulting lengths and volumes of these distorted objects is shown in Table 3. The effects of reducing the magnitude of sinusoidal motion from 1.0 cm to 0.5 cm can be assessed by comparing Figure 5A versus C for HS mode and 5B versus D for HQ mode. Generally, the degree of distortion is less; however, the effects of distortion because of object movement are still readily apparent visually.

Figure 6 shows the displacement of the object center relative to a static sphere, plotted as a function of phase angle. Depending on initial phase, the centroid of a sphere of 3 cm in radius can wander by $\pm \sim 8$ mm.

**Figure 6.** Center of object displacement for volumes in Figure 5A. Displacements of up to 0.8 cm occur.

Evidence of Artifacts in Planning Scans

Do these perturbations in shape appear in clinical scans in patients? Figure 7 shows data from a series of patient CT scans. In the course of treatment planning for carcinoma of the pancreas, scans were obtained in both breath hold conditions (Figs 7A and B) and continuous light breathing (Fig 7C). Coronal multiplanar reconstructions through the principal plane of the left kidney show undulations in the left kidney con-

**Figure 7.** Coronal sections through kidneys in a patient scanned with breath held at (A) exhale, (B) inhale, and (C) free breathing. The contours of the right and left kidneys are smooth in A and B. Contours are irregular in C, similar to those seen in HQ scanning of spheres (Fig 5B).

tour during respiration that are not seen in breath hold scans.

Discussion

Distortions

Precise treatment of a target begins with an accurate 3-dimensional model of the gross target volume. Modern radiation oncology relies on CT imaging to provide such data. The experiments and simulations in this analysis show that significant distortions of the true shape of an object can be introduced when imaging is performed in the presence of object motion. Distortions along the axis of motion could result in either a lengthening or shortening of the object. This could result in the underirradiation of the target or an unnecessary irradiation of normal structures if adequate margins are not applied for motion. The maximum shortening or lengthening observed in phantom measurements is approximately twice the amplitude of motion for the dynamic range of parameters of the scanning experiment. In addition to shape distortion, the center of the imaged moving object can be displaced by as much as the amplitude of the motion. The phase of object motion at the instant the scan plane reaches the object is the primary determinant of the resulting type of distortion.

An intuitive explanation of the motion-induced distortions can be given as follows. Consider the scan process from the reference frame of the couch. The imaging plane advances along the couch length; in high speed mode, this plane advances at a rate of 0.2 s/slice or 5 slices/s. Given a typical slice thickness of 0.375 cm, this corresponds to the imaging plane advancing at a velocity of ~ 2 cm/s. The moving object being imaged in our experiments oscillate sinusoidally with an amplitude of 1 cm and a period of ~ 4 seconds. The velocity of the object (whether it moves toward or away from the imaging plane) is on the order of 1 cm/s. When the object moves toward the imaging plane, it is shortened. When the object moves away from the advancing imaging plane, it is lengthened. The maximum excursion of a slice through the volume must be within ± 1 cm of the static slice because that is the maximum oscillatory motion.

Consider the distorted image of the pear in Figure 3C (upper right hand corner). Inspection shows that the object is overall elongated. On

further analysis, the superior section of the pear is lengthened; then, as the object reverses direction of motion and moves toward the imaging plane, it is shortened in its midsection. A reversal of direction occurs again, resulting in the final inferior tip of the pear to appear lengthened.

The amplitude of 1.0 cm used in data acquisition or simulation are well within range of those reported by others.^{6,7,10} Peak to trough motion of the liver and diaphragm during normal breathing ranges from 5 to 40 mm and 7 to 38 mm, respectively. Under deep breathing, these motions can be as large as 6 to 8 cm.⁶ Tumors of the pancreas have been reported to move periodically with a peak to trough amplitude of ~ 2 cm.¹⁰

The subject of appropriate margins for radiation apertures has been addressed in the literature.^{6-8,11} This study analyzes the consequences of scanning moving organs, with particular focus on the effects of phase during scanning. Balter et al⁸ described the source of sampling errors and schematically described the errors that might occur. Henkelman and Mah¹² raised issues of the scanning of moving targets well before the development of spiral CT scanners. Movement of tumors during respiration has been studied in patients with ultrafast scanners. Ross et al¹³ studied organ motion principally in the axial plane and documented lateral and anterior posterior motion of individual tumors over 2 to 3 respiratory cycles. The authors cite significant craniocaudal motion of several lower lobe lesions but do not quantify it because of technical limitations of the analysis.

Significant distortions have been shown in phantom studies. Are these distortions also observed in patient scans? Figure 7 suggests that organ movement does distort the 3-dimensional shape of the left kidney. The breath held scans of Figures 7A and B show smooth contours of the kidneys in the coronal plane. When the patient was scanned in the HQ mode with a time interval of ~ 0.4 s/slice, undulations appear. The perturbations are similar to those observed in the HQ simulations of Figure 5B (objects C or D).

Conclusions

Helical scanning of moving organs can lead to distortions of the imaged object shape and center. These distortions can result in a systematic error in design of apertures. Knowledge of the static organ shape and acquisition of CT images

of the object during respiration can be combined to estimate the trajectory of the motion as a function of time.

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