The potential for undertaking slow CT using a modern CT scanner

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ABSTRACT. Respiratory motion complicates target volume definition for patients with lung cancer. The use of slow CT to aid in the definition of moving target volumes was investigated. Standard and slow scans of an oscillating phantom were acquired using gantry rotation periods of 0.8 and 1.5 s and pitches of 0.95 and 0.50, respectively. The resultant images of three spheres within the phantom, labelled as A, B and C with diameters of 3.7, 2.8 and 2.2 cm, were analysed. The central co-ordinates of each volume were determined, and the ratio of the target volume outlined on CT (TV_{CT}) to the true target volume (TV) was calculated. For 1.5 cm peak-to-trough (PTT) motion, standard CT mean ratios of 0.8, 0.8 and 0.7 were obtained for spheres A, B and C, respectively, whereas slow CT resulted in a mean ratio of 0.9 for all three spheres. For 2.5 cm motion, standard CT mean ratios of 0.8, 0.7 and 0.7 were obtained, whereas the slow CT mean ratios were 0.9, 0.9 and 0.8. The deviation of the central co-ordinate for the slow CT volumes was within 0.1 cm whereas deviations of up to 0.7 cm were seen using standard CT. This study indicates the potential benefit of using slow CT, even on modern scanners capable of rotation periods only down to 1.5 s, to define moving target volumes more accurately and reproducibly moving target volumes, and aid in the management of respiratory motion for patients with lung cancer.

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Survival statistics for patients with lung cancer are very poor owing to the fact that many present with advanced stage disease. For patients with potentially curable disease, the best outcomes are provided by surgery; however, radical radiotherapy will be used for a significant proportion of patients for whom surgery is not possible. Local recurrence rates following radiotherapy are approximately 50%, and this high rate may be contributed to by the inaccuracy of determining the extent and location of the primary tumour and the difficulty in identifying treatment margins because of tumour motion.

The peak-to-trough (PTT) motion of the diaphragm during normal breathing ranges from 0.7 to 3.8 cm and under deep breathing these motions can be as large as 6.0–8.0 cm [1]. Liu et al [2] assessed respiration-induced tumour motion in 152 patients with lung cancer and found that the proportions of tumours that moved more than 0.5 cm along the superior–inferior (SI), lateral and anterior–posterior (AP) axes during normal breathing were 39.2%, 1.8% and 5.4%, respectively. The tumour motion was found to be associated with diaphragm motion, the SI tumour location in the lung, the gross tumour volume (GTV) and disease stage. To account for this movement, large volume expansions are generally used for thoracic tumours, generating large target volumes. Dose escalation with the aim of improving poor survival rates is limited by the increased risk of normal tissue toxicity associated with irradiating large volumes of normal tissue. This toxicity includes

oesophageal and heart toxicity, although the most important factor is lung toxicity and the risk of pneumonitis [3].

The International Commission on Radiation Units and Measurements (ICRU) Reports 50 and 62 [4, 5] proposed specific definitions for target volumes that are applied in external beam radiation treatment. Report 50 defines the GTV, clinical target volume (CTV) and planning target volume (PTV). Report 62 introduced the concept of the internal target volume (ITV) to incorporate tumour movement as a separate entity into target definition. As a result of tumour motion, single, non-gated rapid CT scans may image tumours in misrepresentative positions within the respiratory cycle and produce subsequent systematic errors in treatment planning and delivery. In addition to this, asynchronous interplay between the advancing imaging plane and internal organ motion can result in severe geometric distortions of the imaged object.

A study by Chen et al [1] analysed the distortions introduced by the helical CT scanning of moving objects and found that the spherical test objects could be shortened by up to twice the periodic motion amplitude. The object shape was also significantly distorted and the geometric centre of the object displaced. The study also indicated that the phase of the object motion at the instant the scan plane reaches the object is the primary determinant of the resulting type of distortion. Lewis and Jiang [6] presented a theoretical model for respiratory motion artefacts in free-breathing scans and explained the relationship between the artefacts and the motion parameters of the scanner, table and tumour. They showed that slow scanning speeds are useful for obtaining accurate ITVs whereas fast scanning speeds are useful for generating accurate images of tumour or organ shape.

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Respiratory motion may be dealt with in a number of different ways. The American Association of Physicists in Medicine (AAPM) produced a report on the management of respiratory motion that describes various technologies including motion-encompassing methods, respiratory gated techniques, breath-hold techniques and forced shallow-breathing methods [7]. The use of slow CT scanning to image moving targets was chosen as the focus for this investigation. Slow CT involves scanning using a slow gantry rotation speed to capture tumour motion during each slice acquisition. An advantage of this method is that patients can continue to be imaged and treated under free-breathing conditions. Slow CT is also an easy technique to implement on existing CT scanners and does not require investment in additional equipment. The drawback of this technique is that the exact tumour size and shape are unknown and the surrounding anatomy will also be blurred.

A study by Wurstbauer et al [8] found that slow CT scans showed larger, but highly constant, depictions of lung tumours in comparison with conventional CT scanning, yielding an integral delineation of almost all positions of the moving tumours. The study concluded that the use of slow planning CT scans enables the drawing of tighter margins in external beam treatment planning of lung cancer. van Sörnsen de Koste et al [9] also found that slow CT scans generated larger and more reproducible target volumes than rapid planning scans. The study concluded that PTVs derived using slow CT produced superior target coverage to that using conventional scans. Data from a phantom study presented by Tanyi et al [10] supported the conclusion that slow scanning image acquisition appears to be the most practical method of acquiring data that may more reliably characterize the time-average position and shape of a moving target. However, standard scanning more reliably defines the boundary between target and normal tissue.

A study by Mori et al [11] found that three-dimensional (3D) planning based solely on slow CT under free breathing may result in underdosing of the target volume and increased toxicity to surrounding normal tissues owing to the excessively large contouring on slow CT images with unacceptable motion artefacts. Seki et al [12] investigated the possibility that more precise definitions of target volumes could be achieved by combining thin-slice CT obtained under breath-hold conditions that are capable of visualizing fine tumour shape, and slow CT capable of visualizing the entire volume of the tumour trajectory.

This study evaluates the potential benefit of using slow CT to image moving targets using a modern CT scanner dedicated to radiotherapy imaging. Most of the published investigations of slow CT use scanners capable of gantry rotation speeds down to 4.0 s per rotation [8, 9, 13, 14]. The aim of this study is to determine whether slow CT can be implemented on a modern CT scanner capable of rotation speeds only down to 1.5 s per rotation, and the magnitude of the benefits that this could provide for target volume delineation and target location identification.

Methods and materials

Equipment

CT scanner and scanning parameters

A Siemens Somatom Emotion 6 slice CT scanner (Siemens AG, Erlangen, Germany) was used for the acquisition of all CT images in this study. This scanner is dedicated to radiotherapy and is used for imaging and virtual simulation. The scanner has a range of scanning protocols with options for gantry rotation times for spiral data acquisition of 0.8, 1.0 or 1.5 s and pitchfactors of between 0.416 and 1.8 depending on the type of scan selected. Scans were carried out using the radical lung therapy imaging protocol available on the scanner. Standard scans were carried out using a gantry rotation time of 0.8 s, 130 kV, 2.5 mm slice thickness and a pitch of 0.95. Slow scans were carried out using the same imaging protocol but changing the gantry rotation time to the longest available, 1.5 s, and reducing the pitch to 0.5. Using a gantry rotation time of 1.5 s and a pitch of 0.5 allows images to be reconstructed using data acquired over 3.0 s and should theoretically capture all motion with a period of less than 3.0 s. Altering the two parameters of gantry rotation time and pitch for the helical scan effectively alters only one variable, which is the time over which data are acquired for each image slice, increasing it from 0.84 s for the standard scan to 3.0 s for the slow scan.

Phantom

A National Electrical Manufacturers Association (NEMA) IEC/2001 phantom filled with distilled water was used with its internal glass spheres left air-filled. The phantom contains six glass spheres of varying size, but only the three largest spheres were considered for this study. The diameter and volume of each sphere used is listed in Table 1. These spheres were chosen as the most relevant considering the typical size of lung tumours treated. Wolthaus et al [15] reported GTV volumes between 2 and 200 cm³ corresponding to tumour diameters between 1.5 and 7.2 cm in their study of 45 patients with non-small-cell lung cancer.

Moving platform

To simulate a range of respiratory cycles, the phantom was moved with a periodic motion along the long axis of

Table 1. Diameters and static volumes of spheres A, B and C, along with the target volume (TV) of each sphere when oscillating with peak-to-trough motion of 1.5 or 2.5 cm

Parameter	Sphere		
	A	B	C
Diameter (cm)	3.7	2.8	2.2
Volume (cm ³)	26.5	11.5	5.4
TV (cm ³)			
1.5 cm motion	42.7	20.7	11.3
2.5 cm motion	53.4	26.9	15.1

the scanner by placing it on a platform driven by a variable speed motor as shown in Figure 1. Only one-dimensional motion was considered to simplify the experimental set-up and allow for clearer interpretation of the acquired data. Motion in the longitudinal direction was chosen for study since the largest component of lung tumour motion is reported to be in this direction [16]. The platform was constructed in-house based on the design of Dietrich et al [17]. The control discs for the motor-driven platform were machined based on the mathematical model, developed by Lujan et al [18], that describes the movement of the diaphragm in one direction. This model is described in Equation 1, where s_0 is the position of the tumour at exhalation, A is the amplitude of the harmonic movement, τ is the breathing cycle period and ϕ is the starting phase.

$$s(t) = s_0 + A \cos^{2n} \left(\frac{\pi t}{\tau} + \phi \right) \quad (1)$$

The value of n alters the shape of the curve, with more time spent in the exhale position than in the inhale position when n is greater than 1; for this study, the value of n was set to 2. This model was verified by Seppenwoolde et al [16] making real-time measurements of 3D tumour motion.

Experimental design

Measurement of motion and "X-ray on" time

The BioPac MP100 system (BioPac Systems Inc., Goleta, CA) was used to record the motion of the phantom. A belt fitted with a strain gauge transducer and connected between two pegs on the platform provided an input signal indicating the phantom motion. A light sensor built in-house was used to detect the output from the "X-ray on" warning light in the CT scanner control room. The resultant signal, in the form of

Figure 1. Phantom positioned on motor-driven platform.

a voltage pulse, was input into the BioPac system. To ascertain whether there was a significant delay between the actual beam on time and the output signal from the warning light, the output from a CT pencil ionization chamber positioned in the centre of the scanner was compared with the output from the light detector by inputting both signals into an oscilloscope. The delay between the signal on times was found to be 40 ms and small enough to be neglected for this investigation. Combining the information on phantom position and "X-ray on" time allows the phantom phase to be accurately identified.

Scan acquisition

Images of the moving phantom were acquired using both the standard and slow CT scanning methods. Images were acquired with the phase of the phantom varied by monitoring the respiration trace and starting the scan at different points in the motion cycle. Four different starting phases, distributed as evenly as possible throughout the motion cycle, were acquired for each scanning set-up.

The period of the phantom motion could be adjusted by varying the speed of the motor driving the platform and the motion amplitude could be varied by fitting different control discs to the spindle of the motor. The period of phantom motion was chosen to be 4 s and PTT motions of 1.5 and 2.5 cm were investigated. These values were chosen based on the clinical values quoted in the reviewed literature [1, 2, 16].

Experimental analysis

Outlining of CT images

All acquired images were imported into Oncentra MasterPlan (OMP) (Nucletron, Veenendaal, the Netherlands) to be outlined. With the aim of minimizing user bias in the outlining process, the automatic outlining tool available in the target delineation module of OMP was used. A threshold of -70 HU was chosen to optimally outline the volumes when viewed using the soft-tissue window (40/350) for all slices of the 3D image. If the moving sphere was imaged as a number of separate volumes, then, following automatic contouring, the union function available in OMP was used to combine the separate volumes with any missing contours interpolated between slices to produce a single outline for each imaged moving sphere. Once the outlines were complete the volume of each outline and its 3D centre were recorded. OMP calculates the 3D centre as the average of the midpoints of the enclosing rectangles of all the contours in the region of interest.

Determination of phase

The phase of the phantom at a particular time was calculated using a saw-tooth waveform as shown in Equation 2, where t_p is the time registered by the BioPac minus the time of the first trough in the respiration

Figure 2. Graphical representation of the relationship between respiration signal and phase.

voltage signal, τ is the period of the respiration signal and $\lfloor x \rfloor$ is the floor function that returns the highest integer less than or equal to x . The respiration signal peak therefore corresponds to 0 or 100% and the signal trough corresponds to a phase of 50% (Figure 2).

$$\% \text{ Phase} = 100 \left(\frac{t_p}{\tau} - \left\lfloor \frac{t_p}{\tau} + \frac{1}{2} \right\rfloor \right) + 50 \quad (2)$$

Calculation of expected volumes

The volumes were analysed by calculating the ratio of the target volume outlined on CT (TV_{CT}) to the true target volume (TV). The TVs were calculated by summing the volume of the sphere and the volume of a cylinder with a diameter equal to that of the sphere and of height equal to the magnitude of the motion. A schematic of the motion envelope generated by the oscillating sphere is shown in Figure 3, and Table 1 lists the TVs for the three sizes of sphere, labelled A, B and C.

Results and discussion

Volumes

Figure 4a shows the 3D volume outlines produced with the phantom moving with a PTT motion of 1.5 cm. Four images, each acquired with the phantom at a different phase at the start of the scan, are shown for each imaging technique. The images demonstrate the variability in the imaged volume depending on the initial phase of the phantom motion when imaging using standard CT. The four slow CT images show greater consistency in the resultant outlined volumes. Figure 4b shows the 3D outlines produced using a PTT motion of the phantom of 2.5 cm. These images again show the inconsistency in the outlined standard CT volumes depending on the initial phase of the phantom, particularly for the smallest volume, C. The green plane shown on each image indicates the closest slice to the true centre of the moving

Figure 3. Schematic diagram of the target volume (TV) generated by the motion envelope of an oscillating sphere.

target. It is clear from the images that some of the standard CT volumes would indicate an incorrect central slice, whereas the slow CT volumes again appear more consistent in terms of size and position.

Figure 5 shows the ratio TV_{CT}/TV for spheres A, B and C for both 1.5 cm and 2.5 cm motion. Each graph shows the relationship between TV_{CT}/TV and the initial phase of the phantom for volumes outlined using either standard or slow CT images. The slow CT technique consistently resulted in larger volumes being imaged, which better represented the true motion envelope volumes and therefore resulted in larger values of the ratio TV_{CT}/TV being calculated than the standard CT volumes. For the largest sphere, A, slow CT scans resulted in ratios in the range of 0.92–0.94 being calculated for both 1.5 and 2.5 cm PTT motion. The ratios calculated for the standard scans were in the range of 0.74–0.88 for both motion amplitudes. Not only did the standard CT images result in smaller volumes being imaged but the volumes were much more variable with the phase of the phantom at the time the scan was started. Volume B showed a similar pattern, with the slow scan ratio varying between 0.88 and 0.93 for both PTT motions whereas the standard scan ratios varied between 0.63 and 0.88. For volume C, the 1.5 cm motion ratios for the slow CT volumes were consistently 0.88 and the 2.5 cm motion ratios ranged between 0.83 and 0.85. For 1.5 cm motion, standard CT volume ratios ranged between 0.62 and 0.80 whereas the 2.5 cm ratios varied between 0.56 and 0.82, showing similar levels of variation with the phase of the phantom. The mean values of these ratios and their standard deviations (SD) are listed in Table 2.

Central location

Figure 6 illustrates the deviation of the calculated central co-ordinate of each imaged volume from the true central co-ordinate. For 1.5 cm motion, the largest deviation of 0.65 cm from the true central position was calculated using the standard CT for volume C. The range of deviation for standard scan volumes A, B and C

Figure 4. (a) Images of three-dimensional (3D) outlines produced using standard CT, top four images, and slow CT, lower four images, with spheres A, B and C shown from left to right in yellow, green and blue, respectively. Outlines depict the target volumes outlined on CT (TV_{CT}) for peak-to-trough motion of 1.5 cm. The green line identifies the true central slice of each moving sphere. Each image was acquired with the phantom at a different phase at the start of the scan. (b) As for Figure 4 (a) but for a peak-to-trough motion of 2.5 cm.

Figure 5. Ratio of the target volume outlined on CT (TV_{CT}) to the true target volume (TV) for image sets with the scan started at different phases in the motion cycle. Graphs for spheres A, B and C are shown. The standard CT scan volumes are compared with the slow CT scan volumes for both 1.5 cm and 2.5 cm peak-to-trough motion.

Table 2. Mean ratios of the target volume outlined on CT images (TV_{CT}) to the true target volume (TV) for spheres A, B and C, imaged using both standard and slow CT scanning while moving with peak-to-trough (PTT) motions of either 1.5 or 2.5 cm

PTT motion and scan type	Mean $\frac{TV_{CT}}{TV}$ ($\pm$ SD)		
	A	B	C
1.5 cm motion			
Standard CT	0.80 ($\pm$ 0.07)	0.76 ($\pm$ 0.08)	0.71 ($\pm$ 0.13)
Slow CT	0.94 ($\pm$ 0.00)	0.92 ($\pm$ 0.01)	0.88 ($\pm$ 0.01)
2.5 cm motion			
Standard CT	0.83 ($\pm$ 0.04)	0.72 ($\pm$ 0.12)	0.67 ($\pm$ 0.09)
Slow CT	0.93 ($\pm$ 0.01)	0.89 ($\pm$ 0.01)	0.84 ($\pm$ 0.01)

was 0.3, 0.4 and 0.9 cm, respectively, while the slow scan volumes all identified the true central co-ordinate (deviation of 0.03 cm). For the spheres moving with a 2.5 cm PTT motion, the widest range of deviations in the central co-ordinate was again seen for the standard scan volumes. Standard scan volumes A, B and C had deviation ranges of 0.9, 0.6 and 0.8 cm, respectively, with the largest deviation of -0.6 cm obtained for a standard scan of volume B. For this larger amplitude of motion some of the slow scans did show a small deviation of 0.1 cm from the true central co-ordinate.

For all the standard CT images, adding a 2.0 cm margin to the TV_{CT} , as is usually applied clinically, ensured that the PTV covered the sphere in all of its positions; however, in many cases it also extended the PTV into normal tissue by up to 2.0 cm. The centre of the volumes outlined using standard scans shows a variation of up to ± 0.7 cm. This inconsistency in positioning the centre of the volumes using the standard CT data sets displaces the PTVs with respect to the true centre in some cases and shows that the 2.0 cm margin is essential to ensure the target volume fully encompasses the true volume. Slow CT scanning was more consistent in imaging the correct size and position of the moving spheres over a range of different phantom phases at the time the scan was

started. This has the potential to allow for a reduction in margin size with the additional information gained from the slow CT scans.

Conclusion

Slow CT scanning can be carried out effectively using a modern CT scanner with a gantry rotation time of 1.5 s and a pitch of 0.5. These imaging parameters resulted in larger and more consistent volumes of moving targets being outlined than with the currently used standard scanning protocol. The volumes outlined using the slow CT images more accurately identified the true central co-ordinate in the direction of motion for the moving targets. It is feasible that slow CT scans acquired using this scanner could be used to more accurately define the volume of lung tumours and provide a more accurate measure of the centre of moving targets.

Slow scans produce blurred images of moving objects and therefore standard scans should still be used to provide an image of the thorax that clearly shows the boundary of the tumour with normal tissue; an additional slow scan would then provide information on the extent of motion and help inform the target volume margins.

Figure 6. (a) Central co-ordinate deviations for spheres moving with peak-to-trough motion of 1.5 cm. and (b) 2.5 cm.

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