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FULL PAPER

Optimizing geometric accuracy of four-dimensional CT scans acquired using the wall- and couch-mounted Varian® Real-time Position Management™ camera systems

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Objective: The aim of this study was to identify sources of anatomical misrepresentation owing to the location of camera mounting, tumour motion velocity and image processing artefacts in order to optimize the four-dimensional CT (4DCT) scan protocol and improve geometrical-temporal accuracy.

Methods: A phantom with an imaging insert was driven with a sinusoidal superior-inferior motion of varying amplitude and period for 4DCT scanning. The length of a high-density cube within the insert was measured using treatment planning software to determine the accuracy of its spatial representation. Scan parameters were varied, including the tube rotation period and the cine time between reconstructed images. A CT image quality phantom was used to measure various image quality signatures under the scan parameters tested.

Results: No significant difference in spatial accuracy was found for 4DCT scans carried out using the wall- or couch-mounted camera for sinusoidal target motion. Greater spatial accuracy was found for 4DCT scans carried out using a tube rotation speed of 0.5s rather than 1.0s. The reduction in image quality when using a faster rotation speed was not enough to require an increase in patient dose.

Conclusion: The 4DCT accuracy may be increased by optimizing scan parameters, including choosing faster tube rotation speeds. Peak misidentification in the recorded breathing trace may lead to spatial artefacts, and this risk can be reduced by using a couch-mounted infrared camera.

Advances in knowledge: This study explicitly shows that 4DCT scan accuracy is improved by scanning with a faster CT tube rotation speed.

Motion management techniques are increasingly used to facilitate accurate delivery of radiotherapy to a number of tumour sites, including lung, liver, breast, pancreatic and oesophageal malignancies. Improvements in scanning techniques allow for accurate target delineation with a concomitant reduction in dose delivery to adjacent organs at risk.¹ Historically, population-based margins were used to encompass the presumed respiratory motion of a target but may deliver an unnecessary dose to adjacent normal structures (e.g. lung, spinal cord, oesophagus), and could also limit the therapeutic dose deliverable to the target volume. Whilst motion can occur in subdiaphragmatic sites owing to bladder filling and colorectal volume changes, the majority of research on organ motion has focused on the impact of respiratory motion.

In the majority of patients, respiratory motion secondary to diaphragmatic motion is small (<1 cm). For some patients, the motion of tumours, particularly unfixed

tumours close to the diaphragm, can be >2 cm.² A number of physiological factors (e.g. concurrent infection, pleural effusions, coexisting lung disease) and tumour factors (size, fixity to bone/chest wall, position in lung) can influence tumour motion and can impact on the quality and interpretation of diagnostic and treatment planning scanning modalities.²

Accounting for and personalizing radiotherapy planning based on tumour motion is particularly important in the era of stereotactic ablative radiotherapy (SABR), whereby extreme hypofractionation is used to deliver tumoricidal doses to small volume lesions with a requisite high degree of accuracy.³ In such instances, planning target volumes based on generalized population statistics would potentially lead to unacceptable toxicity, e.g. pneumonitis, and therefore motion management techniques may be advantageous in such treatments. Even for standard fractionation schedules,

the use of four-dimensional CT (4DCT) can lead to a reduction in margins and therefore a reduction in dose to normal tissues that can theoretically allow for dose escalation in radical radiotherapy for non-small-cell lung cancer.¹

Retrospective 4DCT image acquisition involves the reconstruction of multiple phase CT scan sets following a single scan, which captures multiple phase data at each axial slice location. Whilst 4DCT acquisition protocols are standardized in many centres, the subsequent use of images to individualize patient treatment can vary. Centre-specific approaches include the use of geometrical position, time-weighted average position, and average and maximum intensity projections (MIPs).⁴ A common method is to use the MIP, combined with individual breathing phase images, to delineate an internal target volume (ITV) in lung tumours that includes identifiable disease as imaged over the full respiratory cycle. Other methods include the use of a mid-ventilation scan for outlining.^{4,5} Accurate spatial representation of anatomy in each breathing phase is essential for the correct delineation of 4D target volumes. Further to this, 4D volumetric arc therapy (VMAT) has been proposed,⁶ which requires accurate information on the position of the target throughout the breathing cycle of the patient. It is important to identify the sources of 4DCT image artefacts, as the inclusion or exclusion of misidentified structures could have a large dosimetric and clinical impact for small targets, given the relative changes in target volume when treatment margins are applied.

In order to retrospectively sort the acquired CT images into breathing phase bins, a respiratory signal is required. There are various methods of monitoring respiratory motion, including implanted marker position, abdominal height, spirometry, abdominal strain gauge, inductive plethysmography and temperature sensors.⁷ The commercially available solution used in this study measures chest wall motion or abdominal height variation using an infrared (IR) camera system and external patient marker [Varian® Real-time Position Management™ (RPM) system; Varian Medical Systems, Palo Alto, CA]. In this case, the recorded signal acts as a surrogate for internal respiratory-induced anatomical motion. The RPM camera system for detecting the external marker is either mounted on the wall or on the CT couch. The advantage of the wall-mounted camera is that the IR camera is permanently mounted on the wall. In addition to this convenience, the camera set-up remains fixed with the result that a recalibration is not required on a regular basis. It is sufficient to run a quick daily calibration check to confirm the camera positioning to be within tolerance. The RPM wall-mounted camera requires a multistep calibration (when out of tolerance) that involves moving the external marker block to nine different locations on the couch. This calibration is only necessary when the daily calibration check reports a deviation of >5 mm between the registered block position and the expected position when the block is placed at isocentre. The couch-mounted camera, on the other hand, needs to be mounted on the couch using a bracket for each 4DCT scanning session, and the system requires a quick single-step calibration for each 4DCT scanning session. As yet, the difference between the two camera-mounting systems in the ability to accurately detect targets has not been reported.

Target motion velocity was shown to impact 4DCT accuracy for a sinusoidal motion pattern with delineated targets overestimating volumes by up to 91% when the breathing period was 2 s and amplitude was 10 mm.⁸ This confirmed that misalignment and motion-blurring artefacts should be taken into consideration when determining target volumes, especially for fast target motion with a short breathing period. Nakamura *et al*⁸ suggest that using the maximum tube rotation speed available may improve temporal resolution and reduce motion artefact-based geometrical uncertainties, although the CT scanner used only allowed a 1-s rotation speed.

Anteroposterior motion artefacts also demonstrated increasing volume error with increasing target motion speed; hence for this reason, using the maximum tube rotation speed is recommended, which may in turn necessitate higher tube current to maintain a sufficient signal-to-noise ratio.⁹ The image noise increases when using a faster tube rotation speed while maintaining the same mA because images are acquired in less time and therefore constructed from a reduced number of X-ray photons. Thus, in our study we aim to determine if a higher tube current is necessary to compensate for the resulting loss in signal-to-noise ratio. Hurkmans *et al*¹⁰ suggest that a faster rotation speed should be used when possible. To our knowledge, no work has investigated the effect of changing the rotation speed on target delineation and correlated this with target velocity. In the present study, a comparison of image quality is conducted to assess the impact of rotation speed changes.

The importance of temporal resolution in 4DCT scan protocols has been investigated and the inappropriate selection of acquisition parameters as one of the main sources of error that could lead to undesirable dosimetric consequences.¹¹ The tube rotation time equals the acquisition time of projection data used for the reconstruction of any individual image under full segment reconstruction mode. The total extent of phase included in any single image is dependent on the tube rotation speed, and therefore, the temporal resolution can be increased by using a faster scan rotation speed. Tyler Watkins *et al*¹² also show theoretically that a reduction in tube rotation period correlates with a nearly linear reduction in uncertainty and they recommend using the fastest possible rotation speed for 4DCT scanning.

The primary purpose of this work was to quantify geometrical differences in moving target delineation following four-dimensional (4D) retrospective image acquisition on a 16-slice CT scanner equipped with either a wall-mounted or a couch-mounted Varian RPM camera system. We investigated the accuracy of 4DCT scanning using various 4D scan parameters, including the time between images and the CT tube rotation speed with the secondary objective to identify sources of motion artefact errors in order to minimize their influence in clinical practice.

METHODS AND MATERIAL

Four-dimensional CT scanning and phantom set-up
The Varian 4DCT RPM camera and software allow for the binning of CT images into phases of the breathing cycle when

a series of CT images are sequentially acquired over a period of time comparable to the time required for each breathing cycle. The GE Optima CT580 W CT scanner (GE Healthcare, Waukesha, WI) can be integrated with the Varian RPM system for the retrospective binning of images acquired over successive axial scan couch positions. The variations in the mounting of the RPM infrared camera used in this study are: (1) wall-mounted camera and (2) couch-mounted camera. The wall-mounted camera is more convenient as the camera is a permanent fixture, whereas the couch-mounted camera requires the attachment of the camera on a bracket for 4DCT scanning sessions as illustrated in Figure 1.

Retrospective binning of the CT data set was accomplished using the GE Advantage 4D (GE Healthcare) software package installed with the GE AdvantageSim v. 8 virtual simulation platform (GE Healthcare). Acquired images were sorted into 0–90% phase images at 10% intervals with 0% corresponding to end inspiration and 50% end expiration. The MIP was also generated using Advantage 4D. The phase sorting of individual images was accomplished via synchronized time stamping between the recorded breathing trace and the digital imaging and communications in medicine image files. The phase is allocated based on the automatic identification of the peak inspiration points using the RPM software with manual adjustment if required.

The cube within the imaging insert ($3 \times 3 \times 3$ cm) of the QUASARTM respiratory motion phantom (Modus Medical Devices Inc., London, ON) was imaged using a series of breathing periods (2, 4 and 6 s) and amplitudes (1.5, 2.0 and 2.5 cm from peak inhale to peak exhale). These breathing periods and amplitudes were selected to represent the clinically relevant range anticipated in radiotherapy delivery that would most benefit from 4DCT.⁴ The QUASAR programmable respiratory motion v. 3.0 software was used to drive the phantom with a sinusoidal motion pattern. For both the wall- and couch-mounted camera systems, 4DCT scans were acquired of the phantom using the scan parameters shown in Table 1. The cine duration is the length of time for which scanning takes place at each couch position. As each cine duration should capture one full breathing cycle, 1 s was added to the breathing period to minimize any possible data loss. The cine time between images determines how many images will be retrospectively constructed per breathing phase from the collected scan data. In order to bin the data into 10 phases, this time was initially set as the breathing period divided by 10. The time between cine images was

Table 1. Table of scan parameters used for four-dimensional CT scanning

Rotation time	Cine duration	Time between images
1 s	$P + 1$ s	$P/10$
0.5 s (0.6 s for $P = 6$ s)	$P + 1$ s	$P/10$
0.5 s (0.6 s for $P = 6$ s)	$P + 1$ s	$P/20$

P is the breathing period.

changed to equal the breathing period divided by 20 during acquisition but reconstructed into 10 phases using the AdvantageSim workstation. The clinical protocol for 4DCT within our centre used a 1-s CT tube rotation time. The QUASAR phantom was scanned either using a speed of 1 s for all breathing periods or 0.5 s for 2- and 4-s breathing periods and 0.6 s for a 6-s breathing period. All CT scans were acquired at 2.5-mm CT slices, and the temporal scanning parameters that were varied are listed in Table 1. For all scan sequences, the tube potential was set to 120 kV and the tube current was set to 250 mA.

In-phase and maximum intensity projection analysis Following sorting of the images into the 10 phases and generating the MIP, images were transferred to the EclipseTM v.11 treatment planning system (Varian Medical Systems), where the 10 phase images and the MIP data set were contoured. The cube embedded within the QUASAR phantom imaging insert was contoured using the CT ranger tool to include voxels with CT numbers $>20\%$ above the background. The 20% level was deemed appropriate for isolation of the cube against the background and resulted in the correct volume of the static phantom of 27.0 cm^3 . For the individual phase images, the 20% level corresponded with 175 HU. 190 HU was found for the MIP images owing to the higher mean CT number. The superior–inferior length of the cube (along the centre of the cube) was measured on screen using the ruler function to measure the distance between the superior and inferior contours. This distance was used as the surrogate for contouring accuracy to allow estimation of margins using MIPs and also mid-phase target volume delineation.

For the phase images, the length was plotted against the maximum velocity of the cube within each phase bin. The method for determining the maximum velocity was the same as a previously used method for sinusoidal motion patterns where the velocity for any given phase $V(p)$ is given by:

$$V(p) = -\frac{2\pi A}{P} \sin\left(\frac{2\pi}{100}p\right)$$

where A is the motion amplitude and P is the period of motion.⁸ Comparisons were made between the maximum cube velocity and delineated cube length using Pearson correlations for both the wall- and couch-mounted camera systems for all phases with peak-to-peak amplitudes of 1.5, 2.0 and 2.5 cm and breathing periods 2, 4 and 6 s using the scan parameters shown in Table 1. To better evaluate where the changes in length occurred during

Figure 1. (a) Couch-mounted Real-time Position ManagementTM (RPM) camera (Varian® RPM system, Varian Medical Systems, Palo Alto, CA), (b) wall-mounted RPM camera.

phases, the superior and inferior positions of the block outlined for each phase were recorded. This was analysed for the amplitude of 2.5 cm for each breathing phase (2, 4 and 6 s). Plotting these positions against the expected sinusoidal wave pattern positions was used to demonstrate any difference in elongation of the cube with each breathing period and scanner setting tested.

Differences between cube length using a wall- or couch-mounted camera system were tested for statistical significance using the Mann–Whitney *U*-test with a statistical significance set at 0.05. The same test was also used to compare a 1.0- and 0.5-s rotation time and the difference between using a time between cine images of the breathing period divided by 20 rather than 10.

Image quality

The Catphan® 500 (The Phantom Laboratory Inc., Salem, NY) was scanned in a static position using both a 1.0-s and a 0.5-s tube rotation speed. The QUASAR phantom was positioned inferior to the Catphan on the CT couch with the marker block visible in the IR camera. The QUASAR phantom was used to generate a 6.0-s per breath signal for 4DCT acquisition using the clinical scan parameters (2.5-mm CT slice thickness; 120 kV; 250 mA, 0.6 s between images, 7.0-s cine time). Both 4DCT scans were phase binned and the 0% phase, average intensity projection and MIP data sets were analysed to assess the relative change in image quality associated with the increased tube rotation speed while maintaining equal dose exposure levels. The Catphan phantom was also scanned using a clinical helical lung scan protocol, setting the tube current manually to 250 mA. The image quality was assessed using the PIPSPRO™ software package (Standard Imaging Inc., Middleton, WI) to determine the modulation transfer function (MTF), subjective low-contrast resolution and measured low-contrast resolution. The image noise level was also measured using the uniformity module.

Acoustic noise

The noise output during a scan was measured using a Testo 815 sound level meter (Testo AG, Lenzkirch, Germany) in order to quantify any relative difference in the frequency-weighted sound output. The detector was positioned at the patient's head location, above the couch, for a phantom 4DCT scan using 1.0 and 0.5 s per rotation.

Peak misidentification

In order to sort the individual CT images into phase bins, each image must be tagged with a phase stamp. The accuracy of the phase stamp is dependent on the correct identification of the 0% full inspiration peak on the patient breathing trace using the RPM software. A regular breathing trace displaying clear peaks facilitates easy identification. However, this process is more difficult for irregular breathing traces. Breathing traces recorded using the wall-mounted camera are characterized by superimposed (non-breathing-related) peaks corresponding to couch movements. These unwanted peaks can result in 0% phase misidentification. The couch-mounted camera maintains a fixed distance from the marker block and is not affected by this problem. The effect of peak misidentification was tested by performing a 4DCT scan of the

Figure 2. Graphs showing the measured imaging cube length as a function of the maximum cube velocity within each phase bin. Data for both wall- and couch-mounted cameras are shown. In (a), the CT tube rotation time (RT) was set to 1.0 s with the time between images set to breathing period (P)/10. (b, c) Show data for RT = 0.5 s with the time between images set to $P/10$ and $P/20$, respectively. Max., maximum.

Table 2. Observed phase ranges and differences between maximum intensity projection (MIP) and actual length (including motion) for all amplitudes and breathing periods studied

Mount	Rotation time (s)	Time between image (s)	Phase range (%)		MIP difference (cm)		Linear fit parameters			
			Median	Range	Median	Range	Slope	Intercept	Pearson correlation	<i>p</i> -value
Couch	1	$P/10$	9	6–18	−0.01	−0.31 to 0.00	0.038	3.08	0.80	<0.001
Wall	1	$P/10$	8	5–14	−0.02	−0.35 to 0.00	0.031	3.18	0.72	<0.001
Couch	0.5	$P/10$	8	4–10	−0.02	−0.14 to 0.10	0.025	2.99	0.71	<0.001
Wall	0.5	$P/10$	8	5–13	−0.02	−0.19 to 0.20	0.017	3.06	0.56	<0.001
Couch	0.5	$P/20$	4	1–5	−0.04	−0.20 to 0.17	0.015	3.07	0.63	<0.001
Wall	0.5	$P/20$	4	3–5	−0.01	−0.12 to 0.02	0.015	3.08	0.56	<0.001

P is the breathing period.

The range is defined as minimum and maximum. Linear fit parameters of plots shown in Figure 2.

QUASAR phantom using the wall-mounted camera and allowing peak misidentification. The resulting phase scans were compared against the same 4DCT scan with correctly identified peaks.

RESULTS

Phase range and maximum intensity projection analysis

No differences in reported phase ranges were observed when using wall- or couch-mounted cameras. The reported phase error increases when images are filtered out by the RPM normal breathing predictive filter or when peaks are misidentified, leading to a reduction in the number of images available in a breathing period at a scan location.¹³ For the tests carried out in this study, a sinusoidal motion was used to drive the insert, which facilitated easy peak identification. Increasing the number of images per breathing cycle to 20 (while still binning the data into 10 phases) by reducing the cine time between images had the effect of reducing the reported phase error. The median differences between expected and delineated length measured using the MIP were <0.5 mm. Again, no differences were observed between camera mounts. The MIP-delineated length underestimated the actual length, including motion by up to 3.5 mm for 1-s rotation periods and up to 2.0 mm for a 0.5-s rotation period. The larger errors that were observed were 2-s breathing periods and amplitudes of 2.5 cm. All errors using the faster rotation speeds were less than the CT slice thickness of 2.5 mm.

Interphase analysis

Figure 2a shows that there was a significant correlation between maximum velocity of the target and the length of the cube for both the couch-mounted camera (slope = 0.038, intercept = 3.08 cm, $R = 0.80$, $p < 0.001$) and the wall-mounted camera (slope = 0.031, intercept = 3.18 cm, $R = 0.72$, $p < 0.001$) when the rotation time was set to 1 s. Table 2 and Figure 2b,c show that the slope reduced when the rotation time was set to 0.5 s. No significant difference was observed between the wall- and couch-mounted scanners for any of the scan parameters shown in Table 1. Measured cube lengths were closer to the actual 3-cm length when the scan rotation speed was increased to 0.5 s ($p < 0.001$). No statistical

difference was observed in lengths when the time between images was set as the breathing period divided by 20, rather than 10, irrespective of the camera mounting used.

Figure 3. The superior (S) and inferior (I) contour positions of the imaging block plotted against phase for a motion amplitude of 2.5 cm. Plots for breathing periods of 6, 4 and 2 s are shown for tube rotation times of 1.0 and 0.5 s and time between images of breathing period (P)/10 and $P/20$. Rot, rotation.

Figure 4. Catphan® 500 (The Phantom Laboratory, Salem, NY) contrast and line pair pattern images for four-dimensional CT and helical scans. To facilitate visual comparison, the windowing levels for the contrast pattern images are set to window = 180; level = 130. The windowing levels for the line pair pattern images is set to window = 2500; level = 1250.

Figure 3 shows deviations of the superior and inferior outlines from the expected position with amplitude of 2.5 cm under the different conditions outlined in Table 1. For a 6-s breathing period, it is clear that the differences are minimal when the scan is acquired with any of the parameters although differences are largest when the 1-s rotation speed is used. When the breathing period is reduced to 4 s, there is an increase in difference between

the observed and expected block position. This gives a clear indication as to the areas where margins should be increased and scan parameters should be optimized if phase images are outlined and used for target definition in circumstances such as gating or 4D-VMAT. The largest differences are observed for fast breathing periods (2 s), where again, a 1-s rotation time is not adequate for determining the target in phase images.

Table 3. The Catphan® 500 image quality results derived using PIPSpro software™ package

Rotation speed (s)	Phase	Smallest detail visible (mm)	Measured contrast (HU)	50% modulation transfer function (cy cm^{-1})	Image noise (%)
1.0	Helical	5.0	10.4	3.59	8.2
0.5		6.0	9.4	4.6	12.3 (+50%)
1.0	0%	3.0	8.8	3.06	5.8
0.5		4.0	10.2	3.05	8.4 (+45%)
1.0	Average	3.0	8.7	3.08	2.5
0.5		3.0	9.2	3.05	2.7 (+8%)
1.0	Maximum intensity projection	2.0	9.0	3.08	3.6
0.5		3.0	9.4	3.06	4.6 (+28%)

Catphan® 500 was obtained from The Phantom Laboratory Inc., Salem, NY; and PIPSpro software™ package from Standard Imaging Inc., Middleton, WI.

Table 4. Scanner noise values for four-dimensional CT scans using 0.5- and 1.0-s rotation times

Tube rotation speed	Sound level during run-up (dB)	Sound level during scan (dB)
1.0 s per rotation	73.9	74.4
0.5 s per rotation	76.2	75.9

Image quality

The contrast pattern and line pair pattern images are shown in Figure 4, and the extracted image quality values, using the PIPspro software, are given in Table 3. The image noise (defined as the standard deviation of the CT numbers within a region of interest at the centre of the Catphan uniformity insert divided by the mean Hounsfield unit number) is seen to increase when the scan rotation time is decreased to 0.5 s. This is owing to the reduced number of X-ray photons collected per image when the current is fixed at 250 mA. The loss of image quality is not considered clinically significant when comparing the low- and high-contrast resolution values. The 4DCT images show better low-contrast (perceived detail) resolution than do the helical scans with no difference seen in the measured contrast. No difference can be seen in the measured 50% MTF values for 4DCT scans with 1.0- and 0.5-s tube rotations. The improved 50% MTF values for the helical scans can be attributed to the higher image resolution of 0.5 mm per pixel for the helical scan than 1.0 mm per pixel for the 4DCT scans.

Acoustic noise levels

The measured scanner acoustic noise values are shown in Table 4 for the sound level detected during the run-up stage and during a 4DCT scan. For each scan, the detector was set to record the maximum sound level with a sampling period of 125 ms. The measurements indicate that the sound output levels are similar for scan rotation times of 0.5 and 1.0 s.

Peak misidentification

It may be difficult to distinguish between peaks owing to irregular breathing and peaks owing to couch movements seen

when using the wall-mounted camera system. However, it is possible to identify the impact of misidentified peaks when a sinusoidal pattern is used. Significant deviations from the sinusoidal pattern were observed in breathing traces using the wall-mounted camera (Figure 5) but not with the couch-based camera. Peak misidentification leads to geometric artefacts in the phase images as illustrated in Figure 6.

DISCUSSION

The difference between the RPM wall- and couch-mounted camera systems for the acquisition of breathing signals for 4DCT imaging has been investigated using different amplitudes, breathing periods and scan parameters. Differences measured in MIP or phase image sets were found to be independent of the camera mounting. Differences were found to be highly dependent on scanner rotation speed with smaller differences observed between the outlined target and known target size for 0.5-s rotation speeds. Artificial peaks in the breathing trace were observed for the wall-mounted camera system, which may have been misidentified during free breathing, leading to the incorrect phase binning of images.

The correlation between maximum speed and delineated length follows a similar trend to that observed by Nakamura et al⁸ for 1-s rotation speed. A number of authors suggested that a faster rotation speed should result in reduced artefact,^{8–10} although this had not been investigated using maximum velocity with the same detail as presented by Nakamura et al⁸ for a 1-s rotation time. The presented work further shows that using a faster tube rotation speed results in less delineation errors for outlining mid-phase targets with the least error shown at maximum inhalation and maximum exhalation, consistent with previous observations.⁹ This is particularly true for high amplitude movement combined with fast-breathing periods.

It has also been shown that no difference in length was observed between acquiring CT scans with the cine time between images set to 10 or 20 intervals across the breathing period. Generating 20 images per breathing period improves the reported phase error in AdvantageSim but does not improve 4DCT scan accuracy for a regular breathing pattern. The reported phase error is the deviation in phase between the nominal phase and the phase value derived from the time stamp of each individual CT image. Increasing the number of images within any phase will therefore reduce the reported phase error but will not reduce the total phase

Figure 5. Example of recorded breathing trace using wall-mounted Real-time Position Management™ (Varian® Medical Systems, Palo Alto, CA) camera with correctly (grey lines) and incorrectly (dotted lines) identified 0% phase peaks.

Figure 6. (a) 20% phase coronal image of imaging insert cube generated using the correctly identified 0% peaks shown in (Figure 5). (b) Coronal projection showing a motion-induced artefact in the same phase image caused by the misidentification of 0% peaks.

spanned by any image. Reconstructing the data into 20 images increases the oversampling of projection data as a full tube rotation is still required for any individual image. The change in reported phase error may be more important for patients who have an irregular breathing pattern. One limitation of the presented study is that a regular breathing pattern was used although this allowed for very precise knowledge of the expected cube position.

Previous studies have concentrated on target volume changes.^{8–10,12,14} This study presented length changes to allow better estimations of the margins expected if all phases are used for target definition in building an ITV or for treatment using 4D-VMAT.⁶ Indeed, it has recently been shown that the MIP may not be adequate in defining the target for the oesophagus,¹⁵ and therefore an increasing amount of information is required for target definition across all phases. Detailed analysis of where the delineated positions were compared with expected positions again pointed towards the requirement for as fast a rotation speed as possible. Even with this, margins up to 0.9 mm for the target defined at each phase may be required to account for differences owing to speed and intraslice artefacts. MIP differences in length from what was expected were slightly larger for the slower rotation speed. The expected MIP lengths were slightly less than expected lengths, and this should be accounted for in any margin. This was typically less than the CT slice width of 2.5 mm.

The presented data show for the first time that there are no differences between RPM wall- or couch-mounted camera systems for the definition of targets using the MIP or any phase of the breathing cycle for a regular breathing pattern. It has been shown that using the wall-mounted camera system may result in an increased risk of peak misidentification resulting from incorrectly identified 0% phase peaks on the breathing trace owing to couch motion between subsequent axial scan positions. An extreme example of an imaging step artefact owing to peak misidentification has been presented, although for patients with less regular breathing, it may be possible that a peak is more likely to be misidentified. Any reduction in these artefacts on the breathing trace, as with the couch-mounted camera system, can

improve the 4DCT overall image quality. The difference between the two camera systems is even less significant in the context of the advantages gained by optimizing the rotation speed to 0.5 s. The issue of peak misidentification is especially of concern when free-breathing 4DCT technique is used. Breath control methods will result in a more regular breathing pattern that will facilitate the isolation and identification of true peaks from couch motion peaks. Further gains could be achieved by even faster rotation times than the limit of 0.5 s available on the CT scanner utilized in this work.

An increase in the resulting acoustic noise level was investigated as a possible issue when scanning with a 0.5-s rotation time. This was tested using a sound meter that showed only a slight increase in the sound level when moving to 0.5 s. The image quality was analysed for individual 4DCT images acquired using 0.5- and 1.0-s rotation settings, and a deterioration was found for scans acquired using 0.5 s. However, the image quality was found to be better than a standard helical scan using the same tube potential and tube current settings. Acoustic noise levels were <80 dB(A), which is the threshold above which it is suggested that hearing protection should be used for MRI.¹⁶

The present study highlights that by optimizing scanning methods, improvements in image quality and reduction in artefacts can occur. The clinical relevance of these improvements is an area of potential further research. Motion management techniques and 4DCT are not suitable for all patients and must be selected on an individual basis. The potential benefits of using such techniques have been recommended for patients with significant tumour motion (>1.5 cm), and indeed, optimizing their scanning protocol may have additional gains in the therapeutic index, with particular application to SABR.³ Reducing the risk of peak misinterpretation through use of a couch-mounted camera may be part of this optimization.

CONCLUSIONS

No significant difference was found in geometric accuracy between 4DCT scans acquired using wall- and couch-mounted RPM cameras for a phantom driven by a regular sinusoidal motion pattern. However, the scan accuracy is dependent on the identification of peaks within the recorded breathing trace. The breathing trace recorded using the wall-mounted system displays errors that may lead to the misidentification of peaks resulting in image artefacts. 4DCT scans acquired with the maximum GE tube rotation speed of 0.5 s per rotation are more geometrically accurate than scans acquired with 1.0 s per rotation. Binning the data into more images per breathing cycle reduces the reported phase errors but has no effect on the final geometrical accuracy. Methods to improve optimization of radiotherapy planning by using couch-mounted cameras and increasing tube rotation speed can lead to significant improvements in image quality and a reduction in artefacts. This may have significant importance in the era of increased utilization of motion management techniques in conjunction with SABR for the delivery of radical radiotherapy.

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REFERENCES

1. Cole AJ, O'Hare JM, McMahon SJ, McGarry CK, Butterworth KT, McAleese J, et al. Investigating the potential impact of four-dimensional computed tomography (4DCT) on toxicity, outcomes and dose escalation for radical lung cancer radiotherapy. *Clin Oncol (R Coll Radiol)* 2014; **26**: 142–50. doi: [10.1016/j.clon.2013.11.024](https://doi.org/10.1016/j.clon.2013.11.024)
2. Seppenwoolde Y, Shirato H, Kitamura K, Shimizu S, van Herk M, Lebesque JV, et al. Precise and real-time measurement of 3D tumor motion in lung due to breathing and heartbeat, measured during radiotherapy. *Int J Radiat Oncol Biol Phys* 2002; **53**: 822–34.
3. Distefano G, Baker A, Scott AJ, Webster GJ; UK SABR Consortium Quality Assurance Group. Survey of stereotactic ablative body radiotherapy in the UK by the QA group on behalf of the UK SABR Consortium. *Br J Radiol* 2014; **87**: 20130681. doi: [10.1259/bjr.20130681](https://doi.org/10.1259/bjr.20130681)
4. Cole AJ, Hanna GG, Jain S, O'Sullivan JM. Motion management for radical radiotherapy in non-small cell lung cancer. *Clin Oncol (R Coll Radiol)* 2014; **26**: 67–80. doi: [10.1016/j.clon.2013.11.001](https://doi.org/10.1016/j.clon.2013.11.001)
5. Wolthaus JW, Sonke JJ, van Herk M, Damen EM. Reconstruction of a time-averaged midposition CT scan for radiotherapy planning of lung cancer patients using deformable registration. *Med Phys* 2008; **35**: 3998–4011.
6. Chin E, Loewen SK, Nichol A, Otto K. 4D VMAT, gated VMAT, and 3D VMAT for stereotactic body radiation therapy in lung. *Phys Med Biol* 2013; **58**: 749–70. doi: [10.1088/0031-9155/58/4/749](https://doi.org/10.1088/0031-9155/58/4/749)
7. Lu W, Low DA, Parikh PJ, Nystrom MM, El Naqa IM, Wahab SH, et al. Comparison of spirometry and abdominal height as four-dimensional computed tomography metrics in lung. *Med Phys* 2005; **32**: 2351–7.
8. Nakamura M, Narita Y, Sawada A, Matsugi K, Nakata M, Matsuo Y, et al. Impact of motion velocity on four-dimensional target volumes: a phantom study. *Med Phys* 2009; **36**: 1610–17.
9. Simon L, Giraud P, Servois V, Rosenwald JC. Initial evaluation of a four-dimensional computed tomography system using a programmable motor. *J Appl Clin Med Phys* 2006; **7**: 50–65.
10. Hurkmans CW, van Lieshout M, Schuring D, van Heumen MJ, Cuijpers JP, Lagerwaard FJ, et al. Quality assurance of 4D-CT scan techniques in multicenter phase III trial of surgery versus stereotactic radiotherapy (radiosurgery or surgery for operable early stage (stage 1A) non-small-cell lung cancer [ROSEL] study). *Int J Radiat Oncol Biol Phys* 2011; **80**: 918–27. doi: [10.1016/j.ijrobp.2010.08.017](https://doi.org/10.1016/j.ijrobp.2010.08.017)
11. Mutaf YD, Brinkmann DH. An investigation of temporal resolution parameters in cine-mode four-dimensional computed tomography acquisition. *J Appl Clin Med Phys* 2008; **9**: 2819.
12. Tyler Watkins W, Li R, Lewis J, Park JC, Sandhu A, Jiang SB, et al. Patient-specific motion artefacts in 4DCT. *Med Phys* 2010; **37**: 2855–61.
13. Malone C, Rock L, Skourou C. Efficacy evaluation of retrospectively applying the Varian normal breathing predictive filter for volume definition and artefact reduction in 4D CT lung patients. *J Appl Clin Med Phys* 2014; **15**: 4315. doi: [10.1120/jacmp.v15i3.4315](https://doi.org/10.1120/jacmp.v15i3.4315)
14. Rodriguez-Romero R, Castro-Tejero P. The influence of respiratory motion on CT image volume definition. *Med Phys* 2014; **41**: 041701. doi: [10.1118/1.4866889](https://doi.org/10.1118/1.4866889)
15. Chen X, Lu H, Tai A, Johnstone C, Gore E, Li XA. Determination of internal target volume for radiation treatment planning of esophageal cancer by using 4-dimensional computed tomography (4DCT). *Int J Radiat Oncol Biol Phys* 2014; **90**: 102–9. doi: [10.1016/j.ijrobp.2014.05.015](https://doi.org/10.1016/j.ijrobp.2014.05.015)
16. MHRA. *Safety guidelines for magnetic resonance imaging equipment in clinical use*, DB 2007(03).