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Obtaining breathing patterns from any sequential thoracic x-ray image set

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Abstract
A technique is presented to allow a breathing pattern to be obtained from any multi-slice CT, cone-beam or other series of sequential chest x-ray image sets. The technique requires no extra signals to be recorded and does not need specific external or internal oscillating structures to be visible in the field of view. The breathing pattern is instead acquired from analysing the variation in pixel values between projection images. For cone-beam image sets, slowly varying changes, due to an angular attenuation dependence, must be corrected before the breathing trace analysis can begin. All the results of the new technique were checked visually and were in good agreement. If the studied image set could be analysed using the existing ‘Amsterdam shroud’ technique, then the results it provided were also used for comparison. In cases that allowed comparison by both techniques, the results were in agreement. The new technique was also shown to provide a usable signal when applied to cardiac motion.

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1. Introduction
The advent of cone-beam CT imaging (CBCT) on linear accelerators has highlighted the anatomical motion that occurs during a radiotherapy treatment. Traditionally, this intrafraction motion is accounted for by treating a larger area around the tumour. The area is chosen to encompass the envisaged location of the tumour at all breathing phases (van Herk 2004). If the standard of cone-beam reconstructed images could be improved to give more detailed information on the anatomical motion, this extra information may allow verification or reduction of the treatment margins. The reduction of margins may allow dose escalation and improved tumour control. If the cone-beam reconstructions are further improved, to provide 4D reconstructions, the reconstructions can be used to provide measurement of the
tumour motion trajectory at treatment time. The treatment-time verification allows new treatments such as frameless, stereotactic body radiotherapy (SBRT) to be attempted (Sonke et al. 2008). This motion measurement would also be very useful for motion compensating therapy schemes, such as tracking (Keall et al. 2001, McQuaid and Webb 2006), gating (Kubo and Hill 1996) or any treatment attempting to account for tumour motion without reducing it. The variation in motion trajectories, viewed over a course of radiotherapy treatment, would also be very valuable for accurately assessing the benefit of motion compensation schemes.

A CBCT scanner acquires data over the time scale of 1 or 2 min. Over this time scale a scanned patient will have taken several breaths. If the thoracic region is scanned, the respiratory action will cause movement of many of the imaged structures. The images recorded will no longer be projections of an identical object at different angles. Instead, they are projections of a time-varying object. Thus when standard cone-beam reconstructed techniques are used the reconstructed object is a blurred average of the time-varying object, resulting in a poor quality image. It is possible to use motion models and compensating reconstruction algorithms for the time-varying changes (Rit et al. 2008) though this does require a previous 4DCT scan.

One may virtually gate the reconstruction by reconstructing using only images taken within a certain breathing phase. As the images are still acquired over all breathing phases, any phase can be chosen to reconstruct. More informatively, all phases can be reconstructed separately and saved. These reconstructions can then be viewed sequentially to give a time-varying or 4D reconstruction. As the image acquisition is continuous, to allow virtual gating, a technique for sorting the images on breathing phase is required. Previously, an image-processing technique known as the Amsterdam shroud technique has been developed to allow assigning the breathing phases to individual images (van Herk et al. 2007, Zijp et al. 2004). This method is demonstrated in figure 1.

The Amsterdam shroud technique is very powerful but it is not suitable for all image sets. It is limited to image sets in which a distinct oscillating structure is visible for all projection angles. If a patient is being treated for a tumour in the upper part of the lung and the tumour does not provide a suitable signal, then the field of view (FOV) needs to be extended. For example, often the field of view is extended to the diaphragm. Hence the necessity of seeing a distinct oscillating structure in each projection means that extra healthy lung tissue may be irradiated. To allow the processing of image sets previously gathered without a oscillating structure visible and to allow small-field-of-view 4D scanning, a new technique has been developed.

2. Method

In order to understand this new technique, consider an Amsterdam shroud (formed using the method shown in figure 1) for an image set that does not include a distinct oscillating structure throughout the scan. An example of such a shroud is plotted in figure 2. Examination of the plot reveals several ridges running parallel with the row-number axis. These ridges are due to the attenuation varying as the scanned patient breaths. The variation is due to the inhaled air displacing higher density tissue from the field of view.

If an individual pixel value at a position \((x, y)\) in a projection is given by \(V(x, y)\) then the line summation for each projection \(P\) is given by

\[
S(P) = l_2 \sum_{y=l_1}^{x_{\text{max}}} \sum_{x=1}^{l_2} V(x, y),
\]

where \(x_{\text{max}}\) is the width of the image in pixels. The limits \(l_1\) and \(l_2\) are chosen to include all the rows in the image covering part of the lung. Calculating \(S(P)\) for the image set used to
Figure 1. Figure demonstrating the Amsterdam shroud technique. The logarithm of a standard x-ray projection (a) is taken, and vertical edge detection is used to show the diaphragm more clearly (b). This altered image is then horizontally averaged into a line image (c). Edge detection can be used to further highlight the diaphragm’s or another distinct structural edge (d). This new image can then be cropped to a smaller region containing the edge (e). Concatenating the resultant line images over multiple frames gives an Amsterdam shroud image (f).

Figure 2. A 3D surface plot of part of an Amsterdam shroud produced from an image set with no distinct oscillating structure. The arrow shows a feature at the start that fades away at later projections as the angle is changed.
Figure 3. Plot of the projection angle versus the projection pixel summation (for the image set partly used to form figure 2). The direction corresponding to a projection angle of $0^\circ/360^\circ$ is indicated by the small icon in the top right; this is drawn from a viewpoint that faces towards the Linac gantry. From this viewpoint the scan was taken in the clockwise direction.

Figure 4. Example of how the field of view and patient anatomy cause anisotropic attenuation. Part (a) shows a geometry in which the imaging beam goes entirely through the table. Part (b) shows a geometry in which the beam only partly passes through the table but does transverse more patient tissue.

form figure 2 gives a result shown in figure 3. This is effectively just a summation of pixel values for each projection. The form of the plotted function arises from several factors which can be divided into slowly varying and rapidly varying effects. Hence the function $S(P)$ can be considered as

$$S(P) = S'(P) + S''(P),$$

(2)

where $S'(P)$ is a function describing the rapidly varying effects, mostly due to respiration, and $S''(P)$ is a function describing all features that vary slowly with $P$. The slowly varying features are due to the x-ray beam traversing varying thickness of materials as the angle of irradiation is changed. These attenuation changes can be due to exterior structures such as the table top and supports the patient lies on passing through the x-ray beam. Alternatively, the changes can be due to patient-borne effects such as differing angles of irradiation penetrating extra tissue or tissue of different density. The patient-borne effect is minimized if the field of view captures the whole width of the patient’s chest, but due to differing scatter at different angles still exists. Figure 4 gives a diagrammatic representation of two images of the right lung with very different attenuation. There are several possible ways to remove the function, $S''(P)$, from $S(P)$ such as applying a high-pass band pass filter to $S(P)$ or simply subtracting a
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Figure 5. The variation in attenuation values plotted against the angle of projections taken after slowly varying effects have been removed. Counting the peaks reveals 36 breaths were taken during the approximately 2 min scan.

Figure 6. The breathing patterns on the right are obtained from a fluoroscopic movie, of which a frame is shown on the left. The thin line is the signal and the thick line is the signal after smoothing, to remove cardiac motion.

The breathing patterns on the right are obtained from a fluoroscopic movie, of which a frame is shown on the left. The thin line is the signal and the thick line is the signal after smoothing, to remove cardiac motion.

smoothed version of \( S(P) \). The remaining function after slowly varying changes are removed from the example shown in figure 3 is given in figure 5. If the peaks are not distinct, the summation limits in equation (1) should be altered. Often a smaller range of limits may witness bigger fractional changes. It is easy to select the optimal range by using a simple dynamic viewing program to display the resultant breathing signal as the summation limits are altered. Variations in the amplitude of the peaks of the final breathing trace will always be present. It is difficult to assess from the trace alone if these are true breathing variations, as the amplitude is also dependent on angle and tissue geometry. However, the amplitude is not required to allow image sorting by breathing phase. Thus the amplitude variation can be ignored, provided the peaks are distinct from the background. Isolated irregular features in the breathing patterns may exist, but these will normally have unusual length and amplitude. As such, normally, they can still be detected without the amplitude information. Extremely irregular breathing patterns might be a problem, but patients with these breathing patterns are unlikely to be suitable for 4D scanning, in the near future.

2.1. Tests of the new algorithm

Previously, it has been shown that varying attenuation values in fluoroscopy produced a clear enough signal to gate radiotherapy (Berbeco et al 2005). The technique described in this paper was similarly tested on existing fluoroscopy movies to attempt to provide breathing
patterns. An example is shown in figure 6. Analysis of the fluoroscopy movies with the new technique resulted in clear breathing patterns for all 21 available movies. It is important to note that, unlike CBCT image sets, there is no inter projection angular variation; thus no slowly varying background correction is needed. Therefore, for breathing patterns derived from fluoroscopy movies the amplitude of the breathing pattern can relate to the position of attenuating structures, such as the diaphragm. It is therefore possible to directly compare the pattern produced by the new technique to that produced by the Amsterdam shroud technique. Thus, where possible, the fluoroscopy scans were also analysed by the Amsterdam shroud technique. The two methods agreed very well, as shown in figure 7. The new technique was further tested by checking the maxima in the trace aligned with the frames visually. Cardiac motion was also visible on some breathing traces. Whilst this motion is easy to remove by smoothing, it is also possible to isolate this signal. This isolation is performed by altering the extraction of the breathing signal (section 2) to apply to cardiac rather than respiratory time scales. An example of the resultant pattern due to cardiac motion is shown in figure 8. It should be noted that this was only possible on fluoroscopy scans due to the fast acquisition rate of 25 frames a second, which is adequate to sample cardiac motion. CBCT scans are generally acquired at a slower rate which is insufficient to sample the rapid changes due to cardiac motion. As such cardiac motion traces obtained from standard CBCT scans are likely to have a very low signal-to-noise ratio and be of very limited clinical use.
2.2. Binning

Once the CBCT respiratory trace is obtained, the local maxima and minima are identified. Noise in the trace can generate erroneous local minimum and maximum, especially in the region of the breathing signal where the amplitude is very small. To guard against noise inducing extra turning points the maximum and minimum lists were examined. Any turning points with a separation smaller than 75% of the average separation were removed. The projections can be placed into bins using the phase or amplitude of the extracted breathing signal. Using either scheme results in a very similar outcomes for 10 or more bins; however, for less bins the choice can have a noticeable effect. Phase-binning produces a similar number of projections in each bin. This results in relatively uniform image quality across the individual reconstructions of the projection bins. However, in phase-binned projection sets often the majority of the organ motion occurs within a small fraction of the bins. Consequently, the reconstruction of these bins will have the majority of the motion blurring. This makes the benefit of 4D conversion, namely visualizing organ motion more difficult to see. Alternatively amplitude-binning spreads the motion approximately equally across all the bins. Subsequently, the bin reconstructions all have a similar level of motion blurring. However, amplitude-binning also results in some bins having significantly fewer projections than the others; this will result in these reconstructions being of poorer quality. For this paper, amplitude-binning was chosen merely so motion was more clearly visible. As the amplitude of the breathing signal obtained using the present method is related to the angle, the absolute amplitude cannot be used. Instead, the amplitude relative to the nearest local minima and nearest local maxima is used. Hence each projection image was sorted according to equation (3):

\[ B_p = \frac{S'(P) - \text{Min}(P)}{\text{Max}(P) - \text{Min}(P)}, \]

where \( B_p \) is the breathing phase, \( \text{Min}(P) \) is a function giving the nearest local minimum in \( S'(P) \), and \( \text{Max}(P) \) is a function giving the nearest local maxima in \( S'(P) \). \( B_p \) is a fraction between 1 and 0. Whilst 1 and 0 do correspond to two points in the breathing cycle, they do not necessary relate to maximum inhale and exhale. The projections can then be sorted into any number of bins and the individual bins reconstructed. As an example, figure 9 shows two slices from reconstructions, one reconstruction in which all projections are included, and one formed with only the projections taken in the first breathing phase, when the projections are sorted into six breathing phases. All reconstructions were formed using Elekta’s XVI program. Each sorted image set only has roughly a sixth of the standard number of projections, and thus the reconstruction is severely degraded, compared to the standard reconstruction. Nevertheless, the breathing phase-sorted reconstruction still has noticeable structural differences to the standard reconstruction. As more breathing-phase bins are used, the number of files in each bin decreases and so the quality of each breathing-phase-sorted reconstruction decreases.

3. Discussions and conclusions

The new technique generates a suitable breathing signal for projection sorting, even for image sets not suited to the existing (Amsterdam shroud) method. The selection of the limits in the summation in this new algorithm is an interesting issue for cone-beam scans. Often if the limits are varied, it is possible to alter the angular dependence of the breathing signal amplitude and the phase at which the local maxima and minima occur in the trace. The angular dependence is the result of the decreasing magnification of structures with distance from the x-ray source. As an example, consider analysis with limits optimized using only the diaphragm of the left
Figure 9. An example of two slices from reconstructions; before the projections are sorted (a) and after sorting projections into six phases (b). The structure (outlined in the left image) above the diaphragm that is clear in (a) is absent in (b) due to respiratory motion. Organ motion can also be seen in the movie supplied along with this paper (available at stacks.iop.org/PMB/54/4879).

Figure 10. Breathing traces obtained using different summation limits. The trace with the larger amplitude in the centre of the figure (red on-line) is formed using limits around the diaphragm of the right lung. The other trace (blue on-line) is obtained using limits around the top of the left lung.

lung, when viewing in the anterior–posterior direction. This analysis will struggle when the beam is going from right to left (120° to 240°). The difficulty will arise due to structures on the right side of the chest, as any changes in these will be magnified and may now extend into the selected limits. Figure 10 demonstrates this effect by showing two sets of breathing trace from the same patient with differing angular dependence. It can also be seen from figure 10 that the trace from the top of the lung clearly is π out of phase with the bottom of the lung. This is not a hysteresis effect but is due to the movement of a structure across the field of view. The effect is demonstrated is figure 11.

The strong dependence on limits is not a problem for this technique as any phase differences stay constant. If the limits were to vary dynamically so would the phase. As
Figure 11. Diagrammatic representation of how if the lung moves as a whole it can cause apparent phase variation. The row summation results are shown to the right of each image in black. The variation of summation of pixel values is $\pi$ out of phase between summations with limits at the top of the lung and those with limits at the bottom.

such, it is not possible to alter the summation limits dynamically to obtain the clearest trace. However, normally this is not required, as constant limits can be found that provide a suitable trace across the range.

Reconstruction of the breathing phase-sorted images yields reconstructions which clearly show the motion. However, artefacts arise due to the reduced and non-uniform angular sampling of the sorted image set. The non-uniform sampling could be accounted for by dropping frames closely spaced together. However, this would result in the majority of frames being unused. Alternatively, one could reduce the gantry rotation rate, have faster rotations of the gantry with multiple rotations or indeed have several scans. However, these solutions are not ideal and they increase scan time, push hardware safety constraints or increase dose to the patient, respectively. As an alternative, recent reconstruction algorithms have been developed specifically to work with reduced non-uniformly sampled image sets (Leng et al 2008a, Leng et al 2008b). These algorithms allow reconstructions to be formed from small non-uniformly sampled image sets with much less streaking than reconstructions using standard reconstruction methods. The new breathing-phase-sorting technique described above could be combined with these more optimized algorithms to allow any cone-beam image set to be reanalysed into a 4D reconstruction. This means that a vast number of patients would now be available to allow assessment of tumour motion and how it changes through out a radiotherapy course. Perhaps more importantly the option of small-field-of-view 4D CBCT might allow the reduction of radiation to some future patients. For example, consider a tumour at the top of a lung. By using a filter giving a field of view with a small lateral extent one can reduce the radiation load to the patient, when compared to a filter giving full lateral coverage of the lung down to the diaphragm.

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