Reduction of motion artefacts in on-board cone beam CT by warping of projection images

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Objective: We describe the development and testing of a motion correction method for flat panel imager-based cone beam CT (CBCT) based on warping of projection images.

Methods: Markers within or on the surface of the patient were tracked and their mean three-dimensional (3D) position calculated. The two-dimensional (2D) cone beam projection images were then warped before reconstruction to place each marker at the projection from its mean 3D position. The motion correction method was tested using simulated cone beam projection images of a deforming virtual phantom, real CBCT images of a moving breast phantom and clinical CBCT images of a patient with breast cancer and another with pancreatic cancer undergoing radiotherapy.

Results: In phantom studies, the method was shown to greatly reduce motion artefacts in the locality of the radiotherapy target and allowed the true surface shape to be accurately recovered. The breast phantom motion-compensated surface was within 1 mm of the true surface shape for 90% of surface points and greater than 2 mm from the true surface at only 2% of points. Clinical CBCT images showed improved image quality in the locality of the radiotherapy target after motion correction.

Conclusion: The proposed method is effective in reducing motion artefacts in CBCT images.

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Motion artefacts in cone beam CT (CBCT) occur as a result of movement of the patient during scan acquisition leading to inconsistent data for three-dimensional (3D) reconstruction. This is a particular problem for flat panel imager-based CBCT systems used in image-guided radiotherapy, in which the scanner rotation speed is limited, leading to acquisition times of 1–2 min. Physiological motions such as breathing or internal gas movements lead to anisotropic disturbance, with consequential blurring and streak artefacts in the reconstructed CBCT images [1, 2]. These impede the accurate local delineation of tumours, organs and body surface, which is important for image-guided radiation therapy (IGRT). In IGRT, it is common to acquire a CBCT image immediately prior to treatment delivery for the purpose of verifying that the patient and the relevant internal structures are positioned as intended. If motion artefacts can be removed, making the position of these important objects easier to identify, then the value of the images for geometric verification in radiotherapy may be greatly enhanced.

A number of methods have been used to reduce the severity of motion artefacts in CBCT. Elimination of the motion at source, for example through use of breath-hold during acquisition [3], is effective but not applicable in all situations. Sorting of projections into different breathing phases to produce respiratory-correlated CBCT reconstructions has also been reported [4–6]. This requires acquisition of additional projections (with correspondingly increased

patient dose), and is applicable only to periodic motions such as breathing.

Methods to compensate motion effects during the reconstruction can be applied in 3D, in which the attenuation distribution to be reconstructed is treated as a function of time and the motion path of each voxel is derived from a prior motion model [7–10]. However, a prior motion model is not always available or may be inaccurate.

Alternatively, corrections can be applied to the projection data before back projection. Lu and Mackie [11] described a motion correction for fan beam CT, tracking in-plane motion of nodal points in the sinogram and using this to derive patient motion according to a simple model. The sinogram data were then adjusted to correct for this motion.

A method of shifting CBCT projection images based on the position of markers attached to a moving rigid phantom has been shown to reduce motion artefacts [12]. Perrenot et al [13] and Schäfer et al [14] used two markers attached to a coronary stent to define an affine transformation of each projection image in order to match the marker positions to the forward projection of their position at a chosen reference time. A more complex projection-based motion correction by warping projection images was described by Hansis et al [15]. Projection images from 3D coronary angiography were warped to reduce discrepancies between measured vessel positions and forward projected vessel positions from an initial echocardiogram (ECG)-gated reconstruction.

These two-dimensional (2D) corrections, applied in the projection image domain, are more approximate in nature than the 3D methods. Overlying structures in a projection image, which do not have identical motions,

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Figure 1. Schematic projection image (a) before and (b) after warping to place marker seeds at their projected mean position (dark crosses). Superimposed grid indicates smoothly interpolated warp field.

cannot be corrected by manipulating data in the projection domain. However, the corrections can be valid in a local region or where motion may be considered to be approximately rigid. Correction of motion artefacts in a local region can be particularly useful for CBCT images used for target position verification in radiotherapy. Here, sufficient image quality in the region of the radiotherapy target is necessary to allow assessment of its position in relation to the applied radiation beams.

Corrections in the projection domain can be simpler to apply than full 3D corrections and do not require a prior motion model.

In this paper we describe a motion compensation method for CBCT using a limited number of radio-opaque markers tracked in the projection data. This allows the mean 3D position of each marker to be determined [16]. Projection images are then warped to place each marker at the forward projected mean position for that marker.

Figure 2. (a) Breast phantom and motorised, tilting platform. White arrow indicates the motion direction of the phantom. (b) Projection image from cone beam CT acquisition of the moving breast phantom. The three marker beads placed on the surface are indicated.

Figure 3. Simulated projection images showing virtual phantoms at their mean position. (a, b) Lateral and vertical projections through the first virtual phantom. Note that the phantom surface is not visualised owing to the narrow viewing window optimised to show low-contrast object and markers. (c, d) Lateral and vertical projections through the second virtual phantom. Red asterisks indicate the locations of the surface markers (not visible directly because of the large dynamic range of the image). Ant, anterior; Post, posterior.

The method is demonstrated for both markers placed on the surface of the patient and implanted fiducial markers within the patient, as are widely used in radiotherapy [17, 18].

We apply the proposed motion compensation method to images from a wide-angle CBCT scanner integrated with a radiotherapy linear accelerator. Improvements in image quality are demonstrated for both phantom and clinical images.

Methods and materials

Projection image warping

A set of markers is identified that are visible in all of the CBCT projection images. These may be artificially added markers either on the surface or within the object, or natural features of the object itself. The marker positions are then identified on each projection image,

yielding a set of image co-ordinates as a function of the gantry angle for each marker $(u_{i,j}, v_{i,j})$, where $u_{i,j}$ and $v_{i,j}$ are the u and v positions respectively of the i th marker on the orthogonal major axes of the j th rectangular projection image. Axis v is parallel with the axis of rotation of the CBCT scanner, which coincides with the inferior–superior direction of a patient lying head first into the scanner. Since the gantry angle θ_j of each projection image is known, the marker positions can be fitted to

functions describing the image co-ordinate of a projected static 3D point as a function of the gantry angle [16]. The projection image co-ordinates $u(\theta)$ and $v(\theta)$ of the point (x, y, z) at gantry angle θ are given by

$$u(\theta) = \frac{(x \sin \theta + y \cos \theta) \times \text{SDD}}{\text{SAD} - (x \cos \theta - y \sin \theta)} \quad (1)$$

Figure 4. Virtual phantom reconstructed images. (a–c) Static phantom images. (d–f) Moving phantom without motion compensation. (g–i) Moving phantom with motion correction applied. Left, middle and right columns show axial, coronal and sagittal slices, respectively.

$$v(\theta) = \frac{z \times \text{SDD}}{\text{SAD} - (x \cos \theta - y \sin \theta)}, \quad (2)$$

for a CBCT system with source to detector distance (SDD) and source to axis of rotation distance (SAD).

This fit yields the mean 3D position of each marker, $(\bar{x}_i, \bar{y}_i, \bar{z}_i)$, and the corresponding co-ordinates of the projection of that mean position in each projection image, $(\bar{u}_{ij}, \bar{v}_{ij})$.

Each projection image is then warped using the identified marker positions, (u_{ij}, v_{ij}) , as control (tie) points in the input image and the projected mean positions as the corresponding control points in the warped image. The corners of each projection image were used as additional tie points with assumed zero motion. The images were warped smoothly using thin-plate spline interpolation [19]. Image processing was carried out using IDL (v. 6.3; ITT Visual Information Solutions, Boulder,

Figure 5. Virtual phantom reconstructed images. (a–c) Static phantom images. (d–f) Moving phantom with no motion correction. (g–i) Moving phantom with motion correction applied. Left, middle and right columns show axial, coronal and sagittal slices, respectively. Dashed lines indicate the surface position of the static phantom overlaid onto each image.

CO). After warping, the image of each marker lies at the forward projection of its measured mean position. The warped projection images are then used as input for CBCT reconstruction. Figure 1 shows schematically a projection image with six markers before and after warping. The crosses indicate the projected mean positions of each marker, and the grid is superimposed, for illustration purposes only, to show the smoothly interpolated warp field applied to the image.

Image data

CBCT images were acquired with Elekta Synergy (XVI 3.5, Elekta, Crawley, UK). Projection images were acquired using a 512×512 sampling resolution (pixel size 0.5 mm, scaled to the isocentre). Breast phantom and patient images were acquired using a small field of view (imaging panel centred) protocol with 350 projection images evenly spaced over 200° with technique factors of 120 kV, 16 mA, 16 ms per frame (scan dose 3.5 mGy). Clinical pancreas images were acquired using a medium field of view (panel offset in the u direction) protocol with 650 projection images evenly spaced over 360° and technique factors of 120 kV, 10 mA, 40 ms per frame (scan dose 8.6 mGy). CBCT images were reconstructed by a Feldkamp Davis Kress filtered back projection algorithm [20] using a commercial software package (Cobra v. 5, Exxim, Pleasanton, CA), with 1 mm voxel size in all directions.

Breast phantom

Images of a realistically shaped breast phantom were used to test the proposed motion correction method. The breast phantom was constructed using a rapid prototyping system that can produce a smooth, continuous outer shell supported by an underlying honeycomb structure. The breast phantom was placed onto a motorised, tilting platform to simulate periodic motion of the phantom during image acquisition (shown in Figure 2a). Three

spherical glass marker beads of 4 mm diameter were placed onto the surface of the phantom, which could be tracked in the CBCT projection images. CBCT images were acquired with the static phantom and with the phantom moving with a period of approximately 4 s and a maximum amplitude of 15 mm at the inferior edge of the phantom. The amplitude used here is larger than typically observed patient motion in breast radiotherapy [21]; hence, motion artefacts are expected to be more severe than those observed in typical patient images.

The marker positions were identified in each 2D projection image of the moving phantom and used to warp the projections, as described above, before reconstruction. The moving phantom 3D reconstructed images with and without motion compensation were compared. The motion-compensated moving phantom reconstruction was also compared with that of the static phantom. The two images were co-registered in 3D and the phantom surface shapes compared. This registration step removed differences in the phantom position between the static image and the motion-compensated image, which shows the mean phantom position.

A closest point metric was used to compare the shape of the isosurfaces extracted from the two volumes. Isosurfaces were created by thresholding at an appropriate value (approximately half way between the air value and the peak value in the phantom wall) and smoothed to remove noise using an erode-dilate procedure (element size 0.5 mm). The two thresholded image surfaces were co-registered using the rigid body iterative closest point algorithm [22] prior to using the final closest point distances as the error metric.

Virtual phantom

Further testing of the proposed motion correction method was carried out using virtual phantoms, which allowed known deformations to be applied, as may be encountered in patient images. Two virtual phantoms were used. The first represents a low-contrast ellipsoidal "tumour" moving within a static cylindrical "body" contour (an example of simulated projection images is

Figure 6. Projection image u and v co-ordinates of one marker bead in the moving breast phantom image as a function of gantry angle. Points indicate observed positions, while the thick line indicates the projected position of the mean marker position.

shown in Figure 3a,b and reconstructions are shown in Figure 4a–c). Six fiducial markers are located within the tumour object, and the tumour is both translated along an elliptical path of peak-to-peak amplitude 4, 10 and 20 mm in the LR, AP and SI dimensions, respectively, and deformed by stretching with a factor of 1.02, 1.05 and 1.10 in the LR, AP and SI directions, respectively. The second virtual phantom represents a cylindrical patient outline moving with a stylised breathing motion consisting of a volume-conserving warp with the

posterior surface remaining fixed (indicated by dotted lines in Figure 5a). Five simulated fiducial markers are situated on the anterior surface of the object (locations indicated on the example of simulated projection images shown in Figure 3c,d) and the position of the anterior surface moves by ± 1.6 cm.

Projection images of the moving virtual phantoms were simulated using the Take software (v. 2.1) [23]. Three images of each phantom were reconstructed: (1) static phantom, (2) moving phantom and (3) moving phantom

Figure 7. Cone beam CT reconstructed images of the moving breast phantom with and without motion compensation. (a) Transaxial slice without motion compensation; (b) transaxial slice with motion compensation; (c) sagittal slice without motion compensation; and (d) sagittal slice with motion compensation.

Figure 8. (a) Surface map of the breast phantom with colour shading indicating distances between the static phantom surface and the motion-compensated dynamic phantom surface. White/grey, discrepancy less than 1 mm; green, discrepancy between 1 and 2 mm; red, discrepancy greater than 2 mm. (b) Histogram showing the number of surface points as a function of distance to agreement between the surfaces and the definition of colour shading of surface map.

with motion correction. Motions for both phantoms were applied with a simulated period of 4 s and so that the mean phantom position/shape was the same as the static phantom image. This allowed direct comparison of the motion-corrected and static phantom images.

Patient images

The proposed motion correction method was also tested for effectiveness on clinical data using projections from CBCT images of a patient with pancreatic cancer and a patient with breast cancer. The patient with pancreatic cancer had six gold seeds (1×5 mm) placed into the pancreas at surgery, a common procedure to improve tumour localisation for radiotherapy planning and delivery [24–26]. The patient with breast cancer had radiographic marker wires, as routinely used for radiotherapy treatment-planning scans, placed onto the skin surface at the entry points of the medial and lateral radiotherapy beams. Three positions on each marker wire could be tracked (each end plus the centre). The position of each marker was manually identified in each projection image and used to warp the projection images as described above before reconstruction. For the breast patient image, some of the markers could not be identified in all projection images owing to a low signal-to-noise ratio or overlying structures. This was the case for 11% of all possible marker identifications, although at least three points were identified for all projection images. Markers were omitted from the list of tie points for warping of projection images in which they could not be identified. The reconstructed images with and without motion compensation were compared.

Results

Breast phantom

Figure 2b shows a projection image from the moving breast phantom CBCT acquisition. The positions of the

three surface marker beads are indicated. Figure 6 shows the projection image u and v co-ordinates as a function of the gantry angle of one of the marker beads in the moving breast phantom image. The projected position of the determined mean marker position is also shown.

Figure 7 shows CBCT reconstructed images of the moving breast phantom with and without motion compensation. The quality of the images without motion compensation (Figure 7a,c) is observed to be significantly poorer. Severe streak artefacts are present, which make it very difficult to even define the surface position. The quality of the motion-compensated images (Figure 7b,d) is observed to be much improved. The streak artefacts are much less severe and the surface position can now be defined.

Figure 8 shows the breast phantom isosurface extracted from the motion-compensated CBCT image,

Figure 9. Projection image from a cone beam CT scan of a patient with pancreatic cancer. The longitudinal positions of the two slices shown in Figure 10 are indicated.

with shading representing the distance between the surfaces segmented from the static phantom CBCT image and the dynamic phantom motion-compensated CBCT image. The majority of the surface (90%) has a distance between the surfaces of less than 1 mm (indicated by white/light grey). Regions with a distance between the surfaces of between 1 and 2 mm cover 8% of the area (shown in green). Only a very few points (2% of the surface) have distances greater than 2 mm (shown

in red). The mean distance between the surfaces is 0.6 mm.

Virtual phantom

Figure 4 shows reconstructed images from the virtual phantom with a moving and deforming "tumour" object within a static body outline. Figure 4a–c show slices

Figure 10. Slices from a cone beam CT image of a patient with pancreatic cancer without motion compensation (a,c,e,g) and with motion compensation (b,d,f,h). (a, b) First transaxial slice; (c, d) second transaxial slice; (e, f) coronal slice; and (g, h) sagittal slice.

from an image of the phantom with no motion. Figure 4d–f show the same slices from an image of the moving phantom. The edges of the tumour are severely blurred and the fiducial markers are no longer clearly visible. Figure 4g–i show the same slices with the proposed motion correction applied. The sharp edges of the tumour object have been restored at the correct mean position, and the fiducial markers are seen clearly. Some artefacts have been introduced in other areas of the image. In particular, there are streak artefacts emanating from the high-density objects representing the spine, and the surface of the phantom has been distorted in places.

Figure 5 shows reconstructed images from the second virtual phantom experiment with a cylindrical object

subjected to a simulated breathing motion. Figure 5a–c show slices from an image of the phantom with no motion. The dashed grey lines indicate the surface position of the static phantom. Figure 5d–f show the same slices from an image of the moving phantom. The anterior and lateral surfaces of the phantom are very blurred and are not reconstructed in their correct mean position (as indicated by the dashed line). Figure 5g–i show the same slices with the proposed motion correction applied. The anterior and lateral surfaces of the phantom are now reconstructed much closer to their true mean position. It is noted that the posterior surface of the phantom appears distorted in the motion-corrected image. This is because the tracked markers were placed

Figure 11. Detail from original (a,c,e) and motion-compensated (b,d,f) patient images showing a close-up of the region around seeds and stent. (a, b) First transaxial slice; (c, d) second transaxial slice; and (e, f) coronal slice.

only at anterior and lateral positions on the surface. Also, the quality of the motion correction degrades superiorly and inferiorly (e.g. Figure 5h,i). This reflects the increasing distance from any of the tracked markers.

Patient images

Figure 9 shows a projection image from the CBCT scan of a patient with pancreatic cancer. The implanted gold seeds are visible close to the centre of the image. The patient also has a stent which is visible in the projection image, although the stent position was not used for the motion compensation.

Figure 10 shows slices from the CBCT images with and without motion compensation of the patient with pancreatic cancer. Images without motion compensation are shown in Figure 10a,c,e,g. Two transaxial slices are shown, the first at the level of the most superior seed (indicated as slice 1 in Figure 9), and the second at the level of three other seeds (indicated as slice 2 in Figure 9). Clear streak artefacts emanating from the seeds are visible in both of the transaxial slices without motion compensation. The image of the stent is also significantly blurred in the images shown in Figure 10a,e,g. Streak artefacts from the seeds are much reduced in the motion-compensated images and the stent boundary is much sharper. However, the bony anatomy (e.g. the spine) in the motion-compensated images is observed to be less clear. Detailed images of the region around the seeds and stent are shown in Figure 11.

Figure 12 shows an example projection image from the patient with breast cancer, illustrating the wire markers tracked as part of the motion correction procedure. Figure 13 shows two axial slices from the images from the patient with breast cancer before and after motion correction was applied. Streak artefacts were observed to be greatly reduced in the motion-corrected image, both from the wire markers themselves and from nearby bony structures. Figure 14 shows details from the same images in which the artefact reduction can be seen more clearly. This reduction in streak artefacts made automatic

Figure 12. Projection image from a cone beam CT acquisition of a patient with breast cancer showing marker wires on the surface of the patient.

contouring of the breast surface position more robust. The image quality of internal structures such as the ribs was also observed to be improved in the motion-corrected image.

Discussion

The reported method is intended to provide local motion blur and artefact correction that is effective for a specific region of interest so that anatomical structures can be better delineated for image-guided therapy. Such structures exist at depth throughout the body and as appendages to the body surface. The primary advantage of the motion correction method proposed in this paper is that all motion information is derived from the CBCT projection images themselves, with no requirement for prior imaging or motion models. However, the motion correction is achieved at a cost of possibly increased artefacts in areas far from the region of interest.

The breast phantom images presented above show that the 3D surface of a phantom can be reproduced with accuracy of the order of 1 mm after tracking only a small number of markers placed on the surface. Although a rigid phantom was used, a tilting motion was applied, rather than a simple translation. Hence, a correction based on simple shifting of the projection images, as proposed in [12], does not perform as well (data not shown).

The virtual phantom tests demonstrate that the proposed motion correction method is also effective in the case of non-rigid motion. In the first simulated experiment, the tumour object's mean shape was accurately reproduced despite significant deformation being applied over the motion cycle. This is because the tracked markers acted as a reliable surrogate for the motion of the tumour edges. The virtual phantom tests also demonstrated two different scenarios in which the motion correction method may be useful: reconstruction of an object moving internally within the body, and reconstruction of body surface position.

In the patient images, motion artefacts, such as blurring and streaking emanating from the markers themselves, were greatly reduced by the motion compensation method. In addition, the markers themselves are reconstructed at the mean position of their 3D motion trajectory. The outline of the stent in the images from the patient with pancreatic cancer was also much sharper after application of the motion compensation algorithm. Since the position of the stent itself was not used for the motion compensation, this demonstrates that the motion compensation is effective for other objects in the vicinity of the tracked marker(s). In the CBCT of the patient with breast cancer, the image quality of the ribs and chest wall were observed to be improved after motion compensation, even though only surface markers were tracked. However, motion blurring of the diaphragm and lower lung tissue was not reduced. This is as expected, since the motion of the ribs and chest wall is reasonably similar to that of the surface, whereas the diaphragm moves with larger amplitude.

As expected, motion correction on a local scale introduces artefacts in regions distant from the tracked markers, for example blurring of the spine in Figure 10. This is because the tracked seeds are moving relative to

Original

Motion compensated

(a)

(b)

(c)

(d)

Figure 13. Original (a, c) and motion-corrected (b, d) slices from a cone beam CT reconstruction of a patient with breast cancer.

the spine, which is essentially static. Projection images from directions where the tracked seeds are superimposed onto the spine will be warped to place the seeds at their mean positions. This will introduce an error into the projected spine position, causing blurring in the reconstructed image. This illustrates the essential weakness of a 2D motion compensation technique: it cannot separate different objects projected onto the same pixel of a projection image. A possible solution to this problem may be to apply warping in 3D to the reconstructed volume, based on the fiducial marker positions detected in each projection image. This would require estimation

of the 3D position of each detected marker in each projection image, for example using the method described by Poulsen et al [27]. The resulting motion correction method would be similar to that described by Li et al [10], although using a 3D warp function derived from observed marker positions in the CBCT projections themselves rather than from a prior motion model.

The motion compensation method presented here is somewhat similar to the method of Hansis et al [15], although we use a sparse set of radio-opaque markers as the tracked objects rather than contrast-enhanced coronary arteries. Also, our method does not require a prior

Figure 14. Detail from images shown in Figure 13. (a, c) Original image slices. (b, d) Motion-corrected image slices.

gated reconstruction to define the ideal artery positions; rather, the mean marker positions are determined from the CBCT data themselves.

There are also similarities with the motion correction method of Lu and Mackie [11], although here we apply the correction to wide-angle CBCT rather than fan beam CT. This means that markers are tracked in 2D projection images rather than one-dimensional CT profiles. In this manner, out-of-plane motion can be tracked as well as in-plane motion, which is likely to be significant for respiratory motion. Marker tracking in cone beam projections is also more robust, for example by continuing to track markers which move longitudinally between slices. In addition, the method presented here does not rely on a simplistic patient motion model to allow an approximate global correction. Rather, it is accepted that motion will only be fully corrected in the vicinity of the tracked markers, which are located in regions of specific clinical interest. This pragmatic approach is more likely

to be applicable in real-world situations, and we have demonstrated it with real phantom and clinical images as well as simulated data.

In this study we used manual tracking of implanted markers. Automatic tracking of the markers, using methods such as that of Tang et al [28], would be a feasible extension to speed up the process. A further desirable extension would be to use natural markers within the body to avoid the need for implantation of seeds. However, it is challenging to find natural markers which are sufficiently visible to be identified from an appropriate number of projection angles.

Conclusion

A motion compensation method for CBCT based on warping of projection images has been developed and tested. Radio-opaque markers within or on the surface of the patient are tracked and their mean 3D position is

calculated. The projection images are then warped before reconstruction to place each marker at the forward projection of its mean position. The method has been shown to greatly reduce motion artefacts in CBCT images of a moving breast phantom, allowing the true surface shape to be accurately recovered. Simulated images of a virtual phantom demonstrated good performance of the motion correction method for non-rigid motion. Clinical CBCT images of a patient with pancreatic cancer and another with breast cancer undergoing radiotherapy showed improved image quality in the locality of the radiotherapy target.

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