Respiratory Displacement Modelling in Cone Beam Computed Tomography

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Abstract

Any organ motion during the projection acquisition causes blurring artefacts in the Computed Tomography (CT) reconstructions. It is a drawback in applications of Cone Beam CT (CBCT) for which, the complete scanning cannot be performed during a single breath hold.

Our study proposes a new approach for estimation of the organ's respiratory displacement, from the same projection data, required for the CBCT with sampling over a circular trajectory. The processing includes semiautomatic outlining of the organ's boundaries on the X-ray projections, creating approximate shadow images, and constructing, from them, of a convex 3D organ-like structure, with the property to perform the same respiratory displacement as the one of the organ of interest. The motion is derived as a function of the respiration and can be used for respiratory motion correction in the following CBCT algorithm.

Simulation results reveal significant improvements of the image quality in a cardiac tomogram reconstruction.

1. Introduction

One of the problems encountered in the dynamic volumetric computed tomography (CT) based imaging is the motion due to respiration. If neglected, it causes artifacts in the tomographic reconstruction.

Isocentric imaging equipment, such as in-theatre C-arm units and radiotherapy simulators, equipped with high quality flat panel detectors, are potential platforms for cone beam CT (CBCT) imaging applications. However, due to limitations concerning the C-arm rotating speed and the maximum detector readout rate, the complete scanning cannot be accomplished within a single breath hold and application of respiratory motion compensation is required.

Several algorithms for respiratory motion correction have been proposed in the literature. Crawford et al. [1, 2] proposed a model-based method where the respiration, in an object cross-section, is modelled by time varying magnification and displacement. Ritchie et al. [3] refined their method, assuming that the motion model is correct only in a small region around each object, and developed a pixel-specific filtered backprojection algorithm. Wang et al. [4] proposed a knowledge-based helical multi-slice/CBCT approach for dynamic volumetric imaging. Their approach uses the time varying anatomical information of the heart and is suitable for CBCT. McLeish et al. [5] derived a patient-specific model of the respiratory movement and the deformation of the heart's surface for the MRI imaging. It is based on relatively low-resolution pre-scans with breath holds in different positions of the respiratory cycle, and monitoring landmarks on the reconstructed surface of the heart.

Our study concerns CBCT with scanning over circular trajectory and concurrent respiratory monitoring. The objectives of this study are to develop a model for the displacement of the organ of interest during the respiratory cycle, and to apply it in the tomographic reconstruction algorithm, in order to account for organ movement due to respiration.

Considering heart imaging, using simulations, we were able to demonstrate that its respiratory displacement can be estimated from the same projection data, required for a CBCT reconstruction. Our work includes semi-automatic segmentation of the heart on the projection images, reconstruction of a 3D-heart object that performs the same motion as the heart during respiration and applying this information in the CBCT reconstruction algorithm. The results reveal significantly improved quality of the computed cardiac tomograms.

Although the feasibility of this approach has been demonstrated for heart imaging, it is generic and applies to imaging of any other organ, with a convex hull, that is subject to respiratory displacement during scanning.

2. Methods

The work includes simulation of the acquisition of projection images, deriving a model of the heart motion due to respiration, modifying a CBCT reconstruction with exploitation of the derived motion information.

All computations and simulations were programmed and executed in MATLAB and C/C++.

2.1. Projection acquisition

The study is based on simulations using the 4D MCAT phantom [6], which mimics the human torso, including respiratory and cardiac movements. 3D CT-like volume

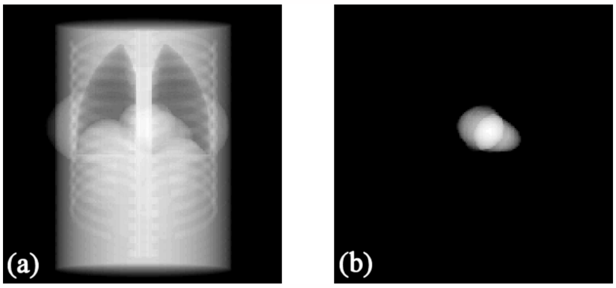

Fig. 1. MCAT-based projections of (a) the thoracic region and (b) the heart

matrices, containing the values for the attenuation coefficients for a given energy, can be obtained over any respiratory and/or cardiac cycle phase.

The acquisition of a complete set of projection images over 360 degrees with a retrospective ECG-gating was simulated as based on the ECG signal, the time instances of the detector frames, closest to the selected ECG phase, were determined and the MCAT phantom was run to create the corresponding volume matrices. Next, by using the Symphan simulator [7, 8], the phantom volumes were irradiated, assuming X-ray cone geometry, and simulated thoracic images (see Fig. 1a), corresponding to different gantry angles and respiratory positions, were obtained. Similarly, projections of the isolated phantom heart were created (Fig. 1b) for reference purposes.

The diaphragm position, taken from the phantom, was used as signal for the chest extension.

2.2. Motion estimation and modeling

The estimation of the heart motion involves the assumption that the heart may be approximated by a 3D-oval surface. Likewise, its projection images may be considered as having approximately oval contours. A second assumption is that, although the exact contour is a complex curve, an approximate 2D projection model may be sufficient for the purposes of estimation of the heart displacement due to respiration. Such model may be mathematically described and identified by semi-automatic segmentation techniques.

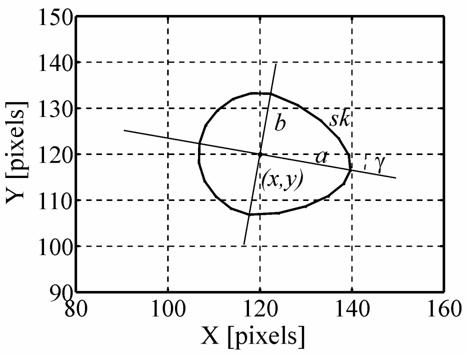

Fig. 2. The contour model used for heart segmentation

Deformable templates [9] are well suited for this type of segmentation. A parametric model with egg-like shape to match the heart contour was used (see Fig. 2). This template is described by 6 parameters: the coordinates of its center – x, y; a major and minor semi axes – a and b; the tilt angle - γ; and a skewness parameter sk, that distorts the oval contour.

The initial contour is user defined as close to the heart shape as possible on the projection image, closest to 0 degree gantry angle. By selecting its center, tip and corpulence, the user draws a contour template on the image and corrects its shape and position until best matching to the heart contours is achieved. The approximate center of the contour is further marked on the projections taken at gantry angles closest to 90, 180, and 270 degrees. The heart segmentation is then performed automatically. The optimization of the model parameters to achieve best matching with the heart contour is performed on the original projection images, after specific edge detection and edge enhancement processing. Optimisation of the model parameters is performed in MATLAB, using functions for constrained non-linear minimization from the Optimisation Toolbox.

The whole range of respiratory position values is divided in 10 intervals and the images from the initial set are grouped into 10 smaller subsets, corresponding to 10 fixed respiratory positions. For each position, the contour parameters, from the corresponding projection images, are interpolated, and a total of 36 heart contour models uniformly distributed, at each 10 degrees, are computed and converted into binary shadow images of the heart.

The shadow images from each respiratory position interval are used for a reconstruction of a 3D heart object, as a common intersection of the shadow cones. The obtained object is an acceptable approximate of the heart for the purposes of computing its motion due to respiration.

The position of the heart inside the human body may be denoted by the position of its center of mass. Although this value, as computed on the simulated object, may not be an accurate measure of the location of the heart, it is adequate for the purposes of defining the origin of a system of coordinates, associated to the heart. The directions of its axes are computed using a 3D Hotteling transform over the heart object volume. In this system of coordinates the heart is static.

The position and the orientation of the associated to the heart objects systems of coordinates for the ten fixed respiratory positions, are used to infer interpolation functions, which analytically describe heart displacement with respect to the respiratory monitoring signal. Linear regression, used for fitting the coordinates of the mass center, proved to be suitable in terms of tomogram quality. The x_c, y_c, and z_c coordinates of the mass center, therefore, are calculated according to (1):

$$\begin{bmatrix} x_c \\ y_c \\ z_c \end{bmatrix} = \begin{bmatrix} \alpha_{x_c} \\ \alpha_{y_c} \\ \alpha_{z_c} \end{bmatrix} + \begin{bmatrix} \beta_{x_c} \\ \beta_{y_c} \\ \beta_{z_c} \end{bmatrix} .RP \qquad (1)$$

where *RP* is the respiratory position value, and *b* and *a* are the coefficients given by the regression formula.

In order to describe the change of the orientation vectors as function of the respiration, the Euler angles (rotation at φ around z, at θ around x, and at ψ around z), which transform the unity vectors of the global system of coordinates to the computed orientation vectors, are calculated. Linear regression is used again to obtain functions that give the Euler angles with respect to the respiratory position value.

2.3. CBCT with motion correction

The derived motion information is then involved in the CBCT reconstruction process. An FDK algorithm [10] for reconstruction of tomographic images of the heart was modified to use it. The heart anatomical planes of interest are defined in the heart's system of coordinates. Having identified the position and the orientation of this system as a function of the respiration, and having the recorded respiration-monitoring signal, the location and orientation of any plane of interest during acquisition may be calculated. The correction applied in the reconstruction algorithm consists of performing the backprojection of each projection profile to the plane of interest as it was at the moment of recording of that profile.

2.4. Initial evaluation of the method

The motion estimation method was tested for translational linear respiratory displacement of the heart.

The respiratory motion of the center of mass is described by the change of its coordinates, i.e. their derivatives with respect to the respiratory position. In the case of linear interpolation, the derivative is equivalent to the slope β of the line that fits the data. The accuracy of slope determination primarily determines the efficacy of the motion identification. In a similar way, the rotation of the heart is described by the changes, or the derivatives, with respect to the respiratory position, of the Euler angles φ, θ, and ψ.

The sensitivity of the method to the accuracy of segmentation was evaluated by comparing the slope values, given by the phantom, to the results, obtained by the proposed method, using thoracic and heart projections

Phantom cross-section, which would be highly influenced by the translational heart displacement due to respiration, was selected, and reconstruction with and without motion correction was performed. The tomograms were compared visually.

3. Results

Fig. 3 visualizes the values and the interpolating lines for the coordinates of the center of mass of the heart objects, reconstructed from the isolated heart- and thoracic projections. The slope values for the two cases, along x-axis were correspondingly 0 ± 1 and -0.4 ± 0.8, along y-axis (the heart translation direction) were -10 ± 0.5 and -10.7 ± 1, and along z-axis they were 0.7 ± 0.7 and 1.3 ± 1.2. The reference slope values, given by the phantom along the three axes, were 0.15, -10.25 and 0 respectively. The normalized to the maximum possible heart displacement differences between the estimated motion and the motion given by MCAT did not exceed 6%. The similar normalized differences for the Euler angles did not exceed 5%.

The phantom cut, the original and the reconstructed slices, before and after respiratory motion correction are shown on Fig. 4.

4. Discussion and conclusion

The calculated slope values and their standard errors

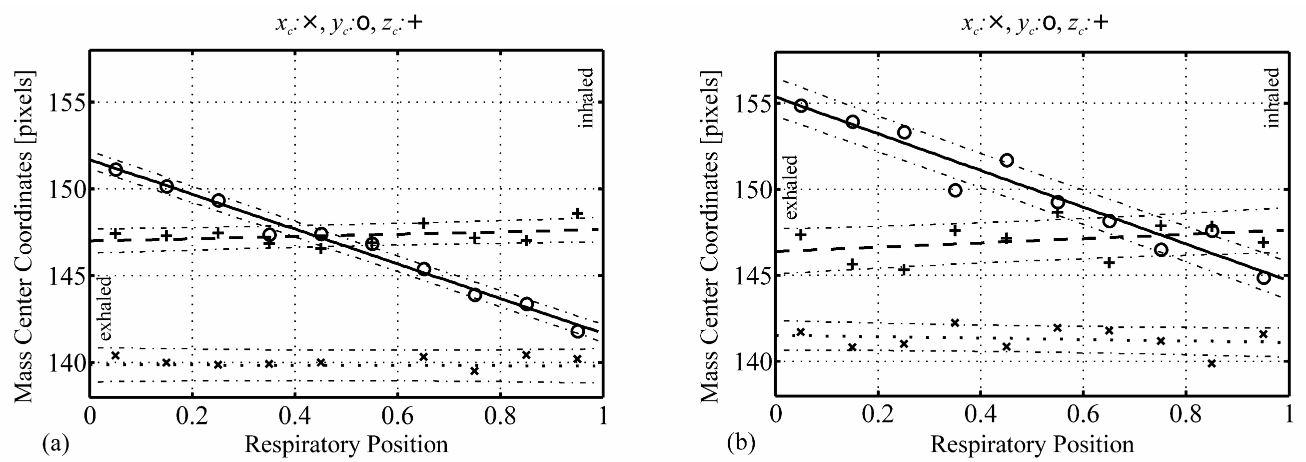

Fig. 3. The computed values and the interpolating lines for the coordinates of the center of the mass of the heart objects, reconstructed from isolated heart- (a) and thoracic (b) projections.

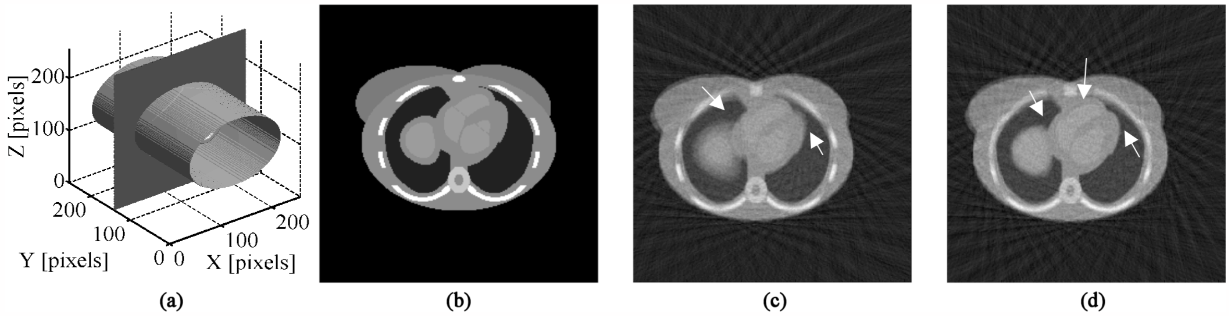

Fig. 4 Reconstruction of MCAT phantom slice for translational respiratory displacement of the heart. (a) presents the orientation of the cut through the volume; (b) shows the phantom slice; the tomographic reconstructions before and after respiratory motion correction are shown by (c) and (d), respectively.

are similar for any coordinate, concerning MCAT and the isolated heart case, and slightly different for the normal case. This is due to the lower quality of the segmentation process for these images. The entire rotational motion of the heart is of few degrees, thus the final relative differences between the estimated and the MCAT rotation are non-significant.

The new proposed approach, tested on simulated data given by MCAT, yields a reasonably good estimate of the motion of the heart due to respiration. The estimated motion based on heart projections is consistent with the reference motion given by MCAT. The complexity of the thoracic projections with all the anatomical details can, lead to errors in the segmentation without however significant consequences on the accuracy of motion estimation.

The template model for heart outlining was intuitively chosen. Although it provides a good approximation of the heart shadows, we believe that it can be further improved.

The motion-corrected reconstruction, using FDK, improves the image quality. In Fig. 4c, the moving organ is shown defocused due to respiratory movement during the sampling cycle. When the correction is applied, the effect is "transferred" to peripheral structures in the image, while the organ of interest comes sharply into focus (Fig. 4d).

The main advantage of the method is that an object-specific model of breathing is obtained from the same data, required for the CBCT itself, which eliminates the need of preliminary sampling. The method is independent of the reproducibility of the respiratory cycle (and the cardiac cycle, in the case of the heart) – the respiratory position registration and the image processing is done retrospectively.

This study demonstrates the feasibility of the method based on simulations. It will be further continued in testing of the method under real conditions.

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