Analysis of 4-D CT Images Degraded by Motion Artifacts

Yi-Qiang YANG, Nobuyuki NAKAMORI, and Yasuo YOSHIDA
Department of Electronics and Information Science, Kyoto Institute of Technology.
Matsugasaki, Sakyo-ku, Kyoto-shi, 606-8585, Japan
(Received October 5, 2002, in final form March 14, 2003)

Abstract: 4-D computed tomography (CT) scanner provides a dynamic volumetric imaging and good contrast detectability with a high speed for collecting projections. In spite of these merits, some motion effects such as respiration, cardiac motion, and patient restlessness produce artifacts that appear as blurring, doubling, and distortion in the reconstructed images, and may lead to inaccurate diagnosis. To reduce these negative effects on reconstructed CT images, we have developed an approximative thorax phantom to analyze the motion effect in 4-D CT for testing the performance of the algorithm. The semi-anthropomorphic raw projections are obtained by based on this phantom with less computer resources. We simply simulate the motion artifacts and apply a knowledge-based method to improve the reconstructed images.

Key words: computed tomography (CT), thorax phantom, motion artifact.

1. Introduction

It is well known that the organ motion in medical image always affects negatively the quality of reconstructed image and may lead to inaccurate diagnosis. Moreover, cardiac diseases have been a big killer in the world. It is important to early screen and accurate diagnosis of cardiac diseases. 4 dimensional computed a tomography (4-D CT) prototype system has been developed recently. It is used cone-beam CT as the base, which has a large area 2-D detector with a high speed rotating gantry. Therefore, projection data can be collected in short acquisition time. By a single rotation, a volume data set with higher spatial resolution over a wide z-axis range can be obtained. Using continuous rotation, time-series volume data sets and three-dimensional (3-D) dynamic images can be acquired. In spite of these merits, during the relatively long data acquisition time, some motion effects such as respiration, cardiac motion, and patient restlessness hardly make the projection data inconsistent. As the result of reconstructing from projection with motion effects, there appear some blurring, doubling, and distortion in the reconstructed images, which may lead to inaccurate diagnosis.

Some effective algorithms have been developed to correct the motion artifacts. For instance, the algorithm based on gating scheme has been widely studied [2], [3]. The auxiliary signal such as electrocardiogram (ECG) is used to identify a quiescent period in the periodic motion so that the acquisition of projection data may be synchronized to be centered within the quiescent period. Ritchie et al. developed a pixel-specific filtered back projection algorithm. They modeled the cardiac motion by experimental data which simplifies the clinical data. And, the in-plane motion artifacts in CT image caused by respiration are well corrected using this method.

In the study for dynamic image of cone-beam CT, an experimental phantom is necessary to analyze the motion effects on reconstruction images. For this purpose, we developed a simple thorax phantom. We describe the definition of the phantom and give an application example using a knowledge-based method in the sections as follows.
2. The thorax phantom model

It is difficult to mathematically model a real thorax of body. In an experiment, a phantom is used to analyze the motion effects and investigate the performance of algorithms to restore the motion-degraded images. To analyze the motion effects on the reconstruction images and compare to the experimental data, it is rational to use a simple phantom model. In this study, we have developed a phantom by superposing simple shape such as ellipsoids and cylinders. The phantom includes the important parts of the thorax such as lungs, heart, and spine which are usually visualized in the CT images. We represent the lungs by two ellipsoids, the heart by one ellipsoid, and the spine by a long cylinder. The semi-anthropomorphic raw projection data for cone beam CT are obtained based on the simple phantom.

2.1 The description of the phantom

The individual parts of the phantom are set at the appropriate positions according to translation and rotation calculation. Figure 1 shows two simple components: ellipsoid and cylinder. The coordinate-transform process is shown in Fig. 2. The point located at (0,0,0) in the reference coordinate system is first translated to the point \((x, y, z)\), then rotated to \(\phi\) around axis \(Z\) and \(\theta\) around \(Y\). Then, the new coordinate system is formed. The object at any spatial location can be easily represented according to the new coordinate system. The phantom can be formulated as

\[
\Pi = \{(\Omega_k, \pi_k)\},
\]

\[
\Omega_k = \text{Trl} (x_k, y_k, z_k) \text{Rot} (Z, \phi_k) \text{Rot} (Y, \theta_k) \text{Scl} (a_k, b_k, c_k),
\]

\[
\pi_k = \{\text{Cylinder or Ellipsoid}\},
\]

where \(k\) represents the \(k\)-th component of phantom which is composed by \(N\) components. \(\text{Trl} (x_k, y_k, z_k)\), \(\text{Scl} (a_k, b_k, c_k)\), \(\text{Rot} (Z, \phi_k)\), and \(\text{Rot} (Y, \theta_k)\) are transform operations individually defined as

1. translation

\[
\text{Trl}(x_k, y_k, z_k) = \begin{bmatrix}
1 & 0 & 0 & x_k \\
0 & 1 & 0 & y_k \\
0 & 0 & 1 & z_k \\
0 & 0 & 0 & 1
\end{bmatrix},
\]

2. scaling

\[
\text{Scl}(a_k, b_k, c_k) = \begin{bmatrix}
a_k & 0 & 0 & 0 \\
0 & b_k & 0 & 0 \\
0 & 0 & c_k & 0 \\
0 & 0 & 0 & 1
\end{bmatrix},
\]
In this study, we composed the thorax phantom by 6 elementary components. The values of these parameters are shown in Table I. Using these parameters, we can compute the 3D thorax phantom. The 3D phantom is shown in Fig. 3. The reconstruction conditions are: The cone angle and fan angle are 13°. The matrix size of projection is 128 X 128. The projections are collected with 360 views in one scan. The reconstruction algorithm used is Feldkamp’s filtered back-projection [4].

3. Numerical experiment and results

The numerical simulation is conducted to evaluate the 3D thorax phantom. A function is firstly defined to simply represent cardiac motion. The function is given as

\[ f(t) = \frac{1}{2} \sin(\omega t) + \sin(\omega t), \]  \hspace{1cm} (2) \]

where \( \omega = 2\pi f \) is a heart beat rate, \( t \) is the time at every projection angle. Figure 4 shows an example of the function \( f \) within one scanning cycle for heart-beat rate of 72 bpm (beat per minute). The knowledge-based algorithm used later assumes the cardiac status as a function of the cardiac phase. Here, the function (see Eq. (2)) is introduced to represent the cardiac cycle. Its ascent and descent parts of the function simply correspond to the diastole and systole period of a heart, respectively. In the simulation, we assume that there is no translation motion and the supposed heart (\( k=6 \)) fluctuates based on

\[
\begin{align*}
A &= a_0 + \Delta a \times f(t) \\
B &= b_0 + \Delta b \times f(t) \\
C &= c_0 + \Delta c \times f(t)
\end{align*}
\]

We compute the sinograms of the phantom in two cases (static and motion). Figure 5 shows these sinograms. Figures 5(a) and (b) are the sinograms along the x axis and (c) and (d) are along the y axis. For distinguishing the sinograms of two cases, we compute the absolute difference images of them. Figure 6 shows these difference images. They denote that the projections collected under the conditions of the existing motion are inconsistent and leads to the artifacts in reconstructed images. We give an example to correct motion artifacts for the developed phantom using a knowledge-based algorithm [5]. This approach effectively utilizes temporal
and spatial information of the beating heart, selects segments of proper projection data when the heart is with similar volumes, and rearranges them in one scanning cycle. We obtain the reconstructed images at every phase of the heart beating. We examined the motion effects on CT images by changing heart-beat rate. In this paper, we describe the results as an example for the following condition. We assume that the heart beat rate is 72 bpm. The fluctuation amplitude ($\Delta [a : b : c]$) is set to 25% of the maximum of the length along the respective axis.

Table 1: The parameters of the thorax phantom

<table>
<thead>
<tr>
<th>Shape</th>
<th>$a_k$</th>
<th>$b_k$</th>
<th>$c_k$</th>
<th>$x_k$</th>
<th>$y_k$</th>
<th>$z_k$</th>
<th>$\theta_k$</th>
<th>$\phi_k$</th>
<th>$\mu_k$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cylinder (k=1)</td>
<td>0.69</td>
<td>0.9</td>
<td>0.9</td>
<td>0.0</td>
<td>0.0</td>
<td>0.0</td>
<td>0.0</td>
<td>0.0</td>
<td>2.0</td>
</tr>
<tr>
<td>Cylinder (k=2)</td>
<td>0.64</td>
<td>0.85</td>
<td>0.9</td>
<td>0.0</td>
<td>0.0</td>
<td>0.0</td>
<td>0.0</td>
<td>0.0</td>
<td>-0.96</td>
</tr>
<tr>
<td>Cylinder (k=3)</td>
<td>0.135</td>
<td>0.135</td>
<td>0.9</td>
<td>-0.43</td>
<td>0.0</td>
<td>0.0</td>
<td>0.0</td>
<td>0.0</td>
<td>0.035</td>
</tr>
<tr>
<td>Ellipsoid (k=4)</td>
<td>0.33</td>
<td>0.30</td>
<td>0.58</td>
<td>0.0</td>
<td>-0.40</td>
<td>0.0</td>
<td>110.0</td>
<td>165.0</td>
<td>-0.025</td>
</tr>
<tr>
<td>Ellipsoid (k=5)</td>
<td>0.33</td>
<td>0.30</td>
<td>0.58</td>
<td>0.0</td>
<td>0.40</td>
<td>0.0</td>
<td>250.0</td>
<td>165.0</td>
<td>-0.025</td>
</tr>
<tr>
<td>Ellipsoid (k=6)</td>
<td>0.19</td>
<td>0.21</td>
<td>0.32</td>
<td>0.05</td>
<td>-0.3</td>
<td>0.16</td>
<td>0.0</td>
<td>0.0</td>
<td>0.05</td>
</tr>
</tbody>
</table>

Fig. 5: The sinogram of the phantom.

Fig. 6: The absolute difference between the static and motion cases.

Fig. 7: The cardiac motion map.

And the x-ray source rotation period is set to 1s. We divide the heart beat into three phases and utilize five source rotation periods for the projection data. The knowledge-based cone beam CT method used here can be summarized as follows: 1. Obtain the relationship between the cardiac status and the projection angle, and make cardiac motion map. The cardiac motion map is a representation of all the information contained in Eq. (2) within only one rotation period of source, i.e., it is also a superposition of shifted versions of Eq. (2) within one scanning cycle; 2. Find cardiac state values $v_i$, $(1 \leq i \leq 3)$, where $v_0$ and $v_3$ are the minimum and maximum of the cardiac state. These states are defined by $[v_0, v_1]$, $[v_1, v_2]$, $[v_2, v_3]$, respectively; 3. Search for the angular interval of every cardiac state; 4. Select and rearrange the cone beam CT projection data $P_i$; 5. Reconstruct each slice in the $i$th cardiac state from $P_i$, using Feldkamp's filtered back-projection algorithm [4]. Next, we describe in detail...
the correction process for the semi-anthropomorphic cone beam CT raw data. Firstly, a cardiac motion map is made by based on the supposed cardiac cycle function. It is shown in Fig. 7. Secondly, we find cardiac state values \( v_i \), \( i=0,1,2,3 \) in the cardiac motion map, i.e., the supposed cardiac volumes are divided into three states: 0-31\%, 31-81\%, and 81-100\%. \( v_0 \) and \( v_3 \) are the minimum and maximum of the state, respectively. We then search for the angular interval corresponding to state \( v_i \) and rearrange cone-beam CT projection data. Finally, the supposed cardiac volume corresponding to a certain state \( v_i \) is reconstructed from the selected projection data using Feldkamp filtered back-projection (FBP) algorithm. The reconstructed volume is of \( 128 \times 128 \times 128 \) volume size. Figures 8 and 9 show the slice images at three axis coordinates of the reconstructed volume. In these images, Figure 8(a) and 9(a) and (e) are the results by using only Feldkamp FBP, the others are the results corresponding to \( v_i \) by using knowledge-based algorithm. It shows that there is minimum volumetric change in the state 1 and maximum volumetric change in the state 3. Because the percentage interval of the state 3 is the smallest, the

4. Conclusion

In this study, a simple thorax phantom has been
developed to analyze motion effect for testing algorithm for cone-beam CT. The results of numerical experiment show that the thorax phantom for cone-beam CT can be computed without much computer resources.

The results show that the motion artifacts based on a simple state function is removed partly by using multi-state knowledge-based algorithm and the cardiac states is well observed.

Acknowledgement

The authors would like to thank Dr. Masahiro Endo and Dr. Takanori Tsunoo of National Institute of Radiological Sciences, Mr. Kazumasa Sato of Sony Corp. for their helpful information.

References


