Spatial Resolution Enhancement Based on Detector Displacement for Computed Tomography

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Abstract
In this paper, the impact of resolution enhancements of X-ray 2D projections on the spatial resolution of the reconstructed 3D computed tomography volume data is investigated. The resolution enhancement of an X-ray 2D projection is achieved by sub-pixel displacement of the X-ray detector which is mounted on linear stages for x- and y-positioning. Multiple projections at the same rotation angle but with different detector positions are applied to reconstruct a high resolution (HR) projection. Due to the HR projections which contain additional information, the spatial resolution of the 3D volume data is enhanced. For assessing the improvement quantitatively, the modulation transfer function (MTF) has been evaluated based on the ASTM-E 1695-95 standard. In order to reduce the reconstruction time of the 2D HR image out of the captured projections with detector displacement, the reconstruction task has been decomposed into sub-problems and implemented on a multi-GPU computer system. It is shown that multi-GPU implementations have accelerated the HR reconstruction process sufficiently such that no extra delay for the 3D CT reconstruction is generated.

Keywords: Spatial Resolution, Detector Displacement, Multi-GPU Acceleration

1 Introduction
In recent years, CT images with higher resolution are of increasing interest. To overcome the inherent spatial resolution limitation of imaging systems, the most straightforward way is to reduce the pixel size. However, the challenge is to reduce the pixel size but simultaneously suppress the increasing shot noise. The other way of resolution enhancement is to increase the sampling rate. This can be achieved by quarter-detector-offset [4] or flying focal spot (FFS) [5]. The third way of resolution enhancement is based on the focal spot size, e.g., by modeling finite focal spot (MFFS) [6] or adopting aperture collimator [7]. As the detector resolutions dominate the overall resolution of the 3D volume data in industrial cone-beam CT scanners for the cases where the voxel size is above the focal spot size, we propose a method based on the above mentioned detector displacement. The proposed method is easy to implement in an existing CT setup and without modifying the standard filtered backprojection. Besides, the proposed HR reconstruction scheme is robust to temperature drift and random focal spot shift. Furthermore, we have realized a parallel software implementation of the proposed HR reconstruction on a multi-GPU system.

2 Method
2.1 Reconstruction Method
In the scientific community of super resolution 2D image reconstruction, the imaging system model can be treated as

\[ Y = AX + \epsilon, \]

where matrix \( A \) is the system matrix which degrades the HR image \( X \) and \( \epsilon \) represents an additive noise which indicates the electronic properties of the sensor or detector [9], [10]. In general, \( A \) is composed of subsampling operator \( D \), blurring operator \( B \) and motion operator \( M \). The subsampling operator produces the aliased low resolution (LR) images from the HR image grid. The motion operator describes the individual displacement during the image acquisition process which may contain global or local translation, rotation and etc. The blurring operator models the point spread functions (PSF) of the optical system, including PSF generated by the camera lens and by the atmospheric turbulence.

For the applications of X-ray CT scans, gray-scaled X-ray projection for the \( i \)th ray denoted by \( y_i \) can be modeled as the sum of the noise-free photon-counting statistics \( z_i \) which obeys a Poisson distribution and additive white Gaussian noise in Equation (2)

\[ y_i = z_i + \epsilon_i, \]

where \( z_i \sim \text{Poisson}(\bar{z}_i) \) and \( \epsilon_i \sim N(0, \sigma^2) \). \( z_i \) denotes the number of incident X-ray photons on the detector for the \( i \)th ray. \( \epsilon_i \) is the standard deviation of the electronic noise [8].

Considering Equations (1) and (2), we formulate the model (3) to address the restoration problem for X-ray projections as

\[ Y_i = A_iX + \epsilon_i, \]
where \( i \in \{1, \ldots, M\} \) denotes the index of multiple LR projections \( Y_i \). The system matrix \( A_i \) is projection-dependent which can be formulated as \( A_i = DBM_i \). Knowing the geometry setup of the CT system, multiple projections taken at the same angle can be considered approximately from the same view. In other words, the detector displacement process can be modeled by a motion operator \( M_i \) which is actually a 3 by 3 translation matrix and the blurring operator \( B \) can be modeled, e.g., by a Gaussian blur.

The restoration of high resolution images from low resolution images is generally an ill-posed problem. An iterative reconstruction method for enhancing the spatial resolution of X-ray images based on the above mentioned mathematical model according to Equation (3) has been implemented using multiple GPUs. The captured multiple X-ray images at the same rotation angle are applied to reconstruct a HR projection based on Equation 4

\[
J(x) = \sum_{i=1}^{n} f_i(x) + g(x),
\]

(4)

where \( i \) is the X-ray image index. \( f_i(x) \) is a generalized formulation of the data fidelity term and \( g(x) \) is the regularization term which is commonly used for ill-posed problem to enforce the convergence of the energy function. The above mentioned energy function can be decomposed for multiple GPUs and the subproblems or domain for each GPU with index \( j \) shown in Equation (5) will be solved individually by the scaled conjugate gradient (SCG) method.

\[
J_j(x) = \sum_{i=1}^{n} f_{i,j}(x_j) + g(x_j).
\]

(5)

Each GPU is required to broadcast a part of the local update to the neighboring GPUs to preserve the global convergence. Making use of these 2D HR projections, the spatial resolution of the 3D CT volume is able to be upgraded.

2.2 Measurement Setup

The mechanism of enhancing the resolution by detector shift can be well understood from Figure 2 and Figure 1. The basic idea is to shift the low resolution detector rightwards, downwards and leftwards by sub-pixel to increase the sampling rate and to produce HR X-ray projections like applying a high resolution detector.

It should be noted that all the experiments are performed on Nikon HMX ST 225 CT scanner which is equipped with a Varian PaxScan@4030E detector. In addition, the detector is mounted on a controllable linear stages shown in Figure 3 Newport M-IMS400CCHA and Newport M-IMS300V, for x- and y-positioning, respectively.

![Figure 1: Schematic depiction of the detector shift by 0.5 pixel.](image1)

![Figure 2: Simplified scheme for interpretation of the resolution enhancement mechanism.](image2)

![Figure 3: CT measurement setup for MTF evaluation with mounted linear stages.](image3)
3 Simulation-Based Experimental Evaluation

In order to evaluate the proposed reconstruction method, we analysed both the X-ray projection images and the 3D CT volume data. For X-ray projection images, we evaluated and analysed the proposed method on subsampled projections with ground truth and verified the resolution enhancement on the experimental X-ray projections with 0.5 pixel detector shifts. In order to quantify the resolution enhancement on the 3D CT volume data, we implemented a series of HR CT scans for a manufactured aluminium cylindrical phantom of 20mm diameter with high manufacturing precision at different CT magnifications. In addition to the HR CT scans, we carried out conventional CT scans and calculated the MTF of the HR CT scans and conventional CT scans according to the ASTM-E 1695-95 standard. To analyse the performance of the proposed method with regard to the resolution enhancement without the influence of detector placement inaccuracies, we subsampled the projections of the conventional CT scans by a factor of two via averaging the four neighbouring pixels and compared the MTF achieved with the ground truth.

3.1 Resolution Analysis of the 2D X-ray Projection Images

Figure 4 shows the resolution improvement of a printed circuit board (PCB) by adopting the proposed method in comparison to bicubic upsampling. In particular, we firstly subsampled one X-ray image by binning with horizontal and vertical offsets to generate four LR images without displacement deviation. It is clearly shown that the detailed structure, for instance, bond wires, can be better recognized in the 2D image reconstructed by the proposed method. A further analysis is shown in Figure 5 as the residual images between the upscaled one and ground truth are demonstrated. The relative errors of both methods are approximately Gaussian distributed. However, the bicubic upsampling leads to a more heavy-tailed distribution.

Figure 4: Comparison of the HR X-ray image with bicubic interpolated image for a printed circuit board. (The resolution difference can only be visible in the PDF file.)

In order to compare the bicubic upsampling with the proposed method, a series of experiments are carried out by shifting the detector with 0.5 pixel or 0.25 pixel as shown in Figure 6. Figure 6 and Figure 7 exhibit that employing 16 X-ray images enables us further to observe even more detailed structures. In the mean while, the edges of the structure are sharpened.

3.2 Resolution Analysis of the 3D CT Volume Data

Aiming at evaluation of the proposed method with respect to the resolution enhancement of 3D CT volume data, we scanned the resolution test target shown in Figure 8. The difference between bicubic and proposed method is evident. To quantify
the resolution enhancement, the MTF of the CT system is calculated based on the ASTM-E 1695-95 standard. An aluminium cylindrical phantom of 120mm length and 20mm diameter was taken as the phantom shown in Figure 3. The roundness of the cylinder is measured by the phantom manufacturer with a tolerance of $2\mu m$ and the roughness is $R_a = 0.226\mu m$ and $R_z = 1.307\mu m$.  

The edge of the phantom is nearly fully overlapped with the y-axis. The widely used software VGSTUDIO for 3D visualization of CT volume data [? ] offers support for computing the MTF according to the ASTM-E 1695-95 standard. Therefore, we imported the reconstructed volume into VGSTUDIO and fitted a cylinder on the CT volume of the phantom. The angle between a fitted cylinder and the x-z plane is measured as 90.02°. The measurement parameters are listed in Table 1. The detector is shifting only when the rotation table has moved to the predefined angle and the detector is resting. Therefore, all the four projections are captured with the same rotation angle shown in Figure 9. Scripts are written to interact with the Nikon CT scanner via DBus and Inspect-X IPC.

![Figure 8: Comparison of bicubic and proposed method on CT slice of a resolution test target.](image)

**Table 1: CT setup for the MTF evaluation experiments**

<table>
<thead>
<tr>
<th>Voltage</th>
<th>Current</th>
<th>Projection Numbers</th>
<th>Filter</th>
<th>Exposure Time</th>
<th>Time of Conv. Scan</th>
<th>Time of HR Scan</th>
</tr>
</thead>
<tbody>
<tr>
<td>200kV</td>
<td>50\mu A</td>
<td>1000</td>
<td>Alu 2.5mm</td>
<td>1s</td>
<td>c.a. 1000s</td>
<td>c.a. 9000s</td>
</tr>
</tbody>
</table>
In order to assess the performance of our proposed method, we generated four sets of low resolution projections by binning the pixels of the projection images of a conventional CT scan with horizontal and vertical offsets of high resolution pixels so that we have obtained four sets of subsampled LR projection images by different offsets. We reconstructed the HR projections from the four sets of LR projections and applied a filtered backprojection as shown in Figure 11. The measured MTF at three different magnifications are indicated in Figure 12. As we can see, the proposed method improves the resolution of the subsampled data and even approximates to the ground truth. The MTF is also given in Table 2. Due to these results we can claim that the proposed method for image resolution enhancement by using four projection images with a horizontal and vertical detector displacement of half a pixel leads to a resolution of the reconstructed 3D volume which is approximately equivalent to that of a 3D volume obtained by a high resolution detector with half the pixel size of the so called conventional case. Based on these results, we evaluated the MTF of the CT scan using the native detector resolution and compared that with the MTF of a CT scan with the proposed resolution enhancement. All the measurements were conducted after 30 mins X-ray tube warming up. The results are depicted in Figure 13 and Table 3. We can achieve roughly 25%–30% resolution enhancement by shifting the detector with half of the detector pixel size. In addition, we measured the temperature drift during the CT scan (see Figure 14, 15 and 16) because the HR scan could be effected by temperature changes due to focal spot shift. The temperature deviates up to 0.75°C during the image capturing time of 9000 seconds and our proposed method is robust enough to handle the detector displacement inaccuracy and random focal spot shift.
Figure 12: Comparison of MTF on subsampled data with ground truth.

Table 2: The MTF evaluation for subsampled data at different magnifications

<table>
<thead>
<tr>
<th>Magnification</th>
<th>Conventional Reconstruction</th>
<th>Proposed Method</th>
<th>Ground Truth</th>
</tr>
</thead>
<tbody>
<tr>
<td>4.0</td>
<td>7.56 lp/mm</td>
<td>9.22 lp/mm</td>
<td>11.40 lp/mm</td>
</tr>
<tr>
<td>7.0</td>
<td>12.25 lp/mm</td>
<td>14.58 lp/mm</td>
<td>15.18 lp/mm</td>
</tr>
<tr>
<td>10.56</td>
<td>15.27 lp/mm</td>
<td>17.70 lp/mm</td>
<td>19.24 lp/mm</td>
</tr>
</tbody>
</table>

Figure 13: Comparison of MTF on upsampled data without ground truth.

Table 3: The MTF evaluation for upsampled data at different magnifications

<table>
<thead>
<tr>
<th>Magnification</th>
<th>Conventional Reconstruction</th>
<th>Proposed Method</th>
</tr>
</thead>
<tbody>
<tr>
<td>4.0</td>
<td>11.40 lp/mm</td>
<td>14.38 lp/mm</td>
</tr>
<tr>
<td>7.0</td>
<td>15.18 lp/mm</td>
<td>19.09 lp/mm</td>
</tr>
<tr>
<td>10.56</td>
<td>19.24 lp/mm</td>
<td>24.25 lp/mm</td>
</tr>
</tbody>
</table>

Figure 14: Temperature measurement at magnification 4. Left column: Conventional scan; Right column: SR scan.
To accelerate the computation rate, we implemented the HR reconstruction method using multi-GPU systems. The pseudocode is shown in Table 5. Each X-ray projection will be split into several parts with necessarily small overlap and each part will be processed by one GPU device based on a nonstationary iterative optimization method, conjugate gradient. At the end, all the parts of the X-ray image will be merged together. The overlapped region has to be broadcasted between neighboring GPU devices to preserve the global convergence. In the experiments, multiple Nvidia GeForce GTX 1080 GPUs equipped with 11GB of VRAM and 2 × Intel(R) Xeon(R) Gold 6148 CPU@2.4GHz have been used. We listed the time consumption in Table 4 for different-sized projections with different number of GPU devices. For four 2000 × 2000 X-ray projections, each HR reconstruction can be accomplished within 1.5 seconds. We found out that using more than four GPU devices for this computational complexity will slow down the reconstruction process due to the communication overhead among neighboring GPUs.

<table>
<thead>
<tr>
<th>Time/Reconstruction</th>
<th>Original Size / Upsampled Size</th>
<th>Number of X-ray Images</th>
<th>Number of GPUs</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.72s</td>
<td>1000×1000 / 2000×2000</td>
<td>4</td>
<td>1</td>
</tr>
<tr>
<td>0.64s</td>
<td>1000×1000 / 2000×2000</td>
<td>4</td>
<td>2</td>
</tr>
<tr>
<td>0.41s</td>
<td>1000×1000 / 2000×2000</td>
<td>4</td>
<td>4</td>
</tr>
<tr>
<td>2.57s</td>
<td>2000×2000 / 4000×4000</td>
<td>4</td>
<td>1</td>
</tr>
<tr>
<td>1.47s</td>
<td>2000×2000 / 4000×4000</td>
<td>4</td>
<td>4</td>
</tr>
<tr>
<td>2.42s</td>
<td>3000×2000 / 6000×4000</td>
<td>4</td>
<td>4</td>
</tr>
</tbody>
</table>

4 Conclusion

In this paper, we proposed a resolution enhancement method for the 2D X-ray projections and the 3D volume data in computed tomography by detector displacement. Multiple projections at the same rotation angle but with different detector positions are captured and reconstructed to a high resolution (HR) 2D projection. The reconstructed HR projections which contain more detailed information are used for a standard filtered backprojection such that a 3D volume with higher spatial resolution is obtained. The resolution enhancement of both the 2D X-ray image and 3D volume data have been investigated experimentally. For the resolution analysis of the 3D volume data, the ASTM-E 1695-95 standard is applied. The proposed method for image
resolution enhancement by using four projection images with a horizontal and vertical detector displacement of half a pixel leads to a resolution which is approximately equivalent to the resolution of the 3D volume data obtained by a high resolution detector with half the pixel size. Therefore, further research for improvements concerning the obtained resolution by the proposed method is not required.

Based on these results we evaluated the MTF of the CT scan using the native detector resolution and compared that to the MTF of a CT scan with the proposed resolution enhancement. We can achieve roughly 25%–30% resolution enhancement by the detector displacement with 0.5 pixel horizontally and vertically. In order to reduce the time of the high resolution scan procedure, a multi-GPU implementation to reconstruct the high resolution projection images has been implemented. For four 2000 × 2000 X-ray projections, each high resolution projection image of 4000 × 4000 pixels can be computed within 1.5 seconds which is in the time range of capturing the all four X-ray images even for short exposure time such that no delay in the 3D CT reconstruction is generated by the proposed method. Future work will focus on the alternative evaluation procedures for the spatial resolution.

Table 5: Pseudocode for Multi-GPU Implementation

1. Initialize parameters for scaled conjugate gradient (SCG) optimizer.
2. Calculate the sparse system matrix $A_i$ for the $i$th detector position.
3. for each rotation angle:
   /* Reading projections and assign them to GPU devices.*/
   3.1 Read multiple projections which are captured by detector shifts at the same rotation angle.
   3.2 Split each projection $y_i$ into multiple regions with overlaps on the boarder and assign the individual region to the corresponding GPU device.
   /*Executing resolution enhancement reconstruction based on Equation (4) at each GPU device in parallel.*/
   3.3 do:
      Calculate the gradient of Equation (4) and update local variables and $X_i$ based on SCG.
      if the local update is successful:
         Broadcast the locally updated overlapped $X_i$ to the neighbors to make the convergence consistant.
   while iteration < Iter
   3.4 Merge the output of each GPU device to construct an integrated projection.

References