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CBCT volumetric coverage extension using a pair of complementary circular scans with complementary kV detector lateral and longitudinal offsets

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Abstract
Onboard cone-beam CT (CBCT) has been widely used in image guided radiation therapy. However, the longitudinal coverage is only 15.5 cm in the pelvis scan mode. As a result, a single CBCT scan cannot cover the planning target volume in the longitudinal direction for over 80% of the patients. The common approach is to use double- or multiple-circular scans and then combine multiple CBCT volumes after reconstruction. However it raises concerns regarding doubled imaging dose at the imaging beam junctions due to beam divergence. In this work, we present a new method, DSCS (Dual Scan with Complementary Shifts), to address the CBCT coverage problem using a pair of complementary circular scans. In DSCS, two circular scans were performed at 39.5 cm apart longitudinally. In the superior scan, the detector panel was offset by 16 cm to the left, 15 cm to the inferior. In the inferior scan, the detector panel was shifted 16 cm to the right and 15 cm to the superior. The effective imaging volume is 39.5 cm longitudinally with a 45 cm lateral field-of-view (FOV). Half beam blocks were used to confine the imaging radiation inside the volume of interest. A new image reconstruction algorithm was developed, based on the Feldkamp–Davis–Kress cone-beam CT reconstruction algorithm, to support the DSCS scanning geometry. Digital phantom simulations were performed to demonstrate the feasibility of DSCS. Physical phantom studies were performed using an anthropomorphic phantom on a commercial onboard CBCT system. With basic scattering corrections, the reconstruction results were acceptable. Other issues, including the discrepancy in couch vertical at different couch longitudinal positions, and the inaccuracy in couch table longitudinal movement, were manually corrected during the...
reconstruction process. In conclusion, the phantom studies showed that, using DSCS, a 39.5 cm longitudinal coverage with a 45 cm FOV was accomplished. The efficiency of imaging dose usage was near 100%. This proposed method could be potentially useful for image guidance and subsequent treatment plan adaptation.

Keywords: radiation therapy, computed tomography, image reconstruction, image guided radiotherapy

(Some figures may appear in colour only in the online journal)

1. Introduction

Image guided adaptive radiation therapy represents one of the most important future directions in radiation oncology as it provides clinicians the opportunity to adjust treatment plan according to patient’s 3D anatomy and tumor response during the treatment course either online or off-line (Yan et al., 1997, Ding et al., 2007; Richter et al., 2008). As the most popular on-board volumetric imaging modality, cone-beam CT (CBCT) has been widely used in image guided radiation therapy (IGRT). However, the use of CBCT images for adaptive radiation therapy is limited by insufficient volumetric coverage (Ding et al., 2007; Grimmer et al., 2009, Baek and Pelc, 2010). Current on-board CBCT uses a single gantry circular rotation at the fixed treatment couch position. With a 40 × 30 cm² detector panel, the head scan mode has a 17 cm longitudinal coverage and a 25 cm field-of-view (FOV), while the pelvis scan mode has a 15.5 cm longitudinal coverage and a 45 cm FOV by using a 16 cm detector panel lateral shift (On-Board Imager (OBI), 2009). An internal survey of the patients treated at the author’s institution since 2007, totaling 243 head-and-neck patients and 320 pelvis cancer patients, suggests that current CBCT cannot cover the planned treatment volume (PTV) for 95% of the head-and-neck patients and 81% of the pelvis patients (figure 1). The percentage is high because many of the diseases were advanced with lymph nodes involvements. An example is shown in figure 2 for a patient with cervical cancer.

For plan adaption purposes, it is necessary to extend the CBCT longitudinal coverage to at least cover the PTV. The common approach is to use double- or multiple-circular scans and then combine multiple CBCT volumes after reconstruction (Zheng et al., 2012). For pelvis scans, two scans would provide a 31 cm longitudinal coverage. This approach is straightforward and easy to implement however it raises concerns regarding doubled imaging dose at the imaging beam junctions due to beam divergence. Such a double-exposed region can be as large as 5–7 cm long. Ding et al. estimated the imaging dose from a typical head-and-neck CBCT and reported that doses to soft tissues, such as eye, spinal cord, and brain can be up to 8, 6, and 5 cGy, respectively while the dose to the bone, due to the photoelectric effect, can be as much as 25 cGy with default CBCT technique of 125 kVp, 80 mA and 25 ms (Ding and Coffey, 2009). The multiple-circular technique could potentially double these numbers and is therefore difficult to manage. Although newer acquisition settings have been reported to lower the imaging dose, doubled imaging dose at the imaging junction remains a significant health concern, in particular if daily imaging is desired (Palm et al., 2010). Tan et al. recently proposed a method using helical CBCT scan trajectories (Tan et al., 2012). This method works well for head scan mode that requires no detector lateral shift. However, if the detector panel needs to be laterally displaced in order to extend the FOV for a pelvis scan, the helical scan method would require the gantry to be rotated 4 times (360 degrees per
rotation) in order to provide a 51 cm longitudinal coverage. The quadruple scan time is not clinically desirable (Tan et al 2012). Other methods involves registering the planning CT with the CBCT with rigid or deformable image registration, then either using the planning CT to patch the CBCT, or using the deformed planning CT to replace the CBCT (Zhen et al

Figure 1. Longitudinal field size probabilities of head-neck and pelvis cancer patients.

Figure 2. A patient with cervical cancer requires a 36 cm CBCT longitudinal coverage but the single pelvis CBCT (in green) can only provide a 15.5 cm longitudinal coverage. The blue contour is the planning target volume (PTV). The dashed lines suggest the treatment beam longitudinal field size. The image in purple is the planning CT.
Because there is no reliable way to verify image registration accuracy, such methods could not guarantee the accuracy of the made-up image and therefore cannot be confidently applied for clinical use.

In this study we report a new solution, DSCS (Dual Scan with Complementary Shifts), to address the on-board CBCT’s longitudinal coverage problem. The solution consists of two major steps: 1) use two complementary circular CBCT scans with the kV detector displaced both longitudinally and laterally in the opposite directions, and 2) perform image reconstruction using a modified Feldkamp–Davis–Kress (FDK) algorithm (Feldkamp et al 1984). DSCS is able to provide a 39.5 cm longitudinal coverage and a 45 cm FOV. The data acquisition time is about 2 min, comparable to 1 min standard CBCT scan.

2. Materials and methods

2.1. Background: detector shifts in CBCT

Figure 3 illustrates the possible cases of detector shifts in CBCT scans. Figure 3(a) shows the head scan mode where the detector panel is centered in the beam-eye-view and the x-ray source rotates around the scanned object in the axial plane. In a pelvis scan protocol lateral shift, shown in figure 3(b), is used to extend the FOV to 45 cm. In this example, the detector panel is shifted to the right in the beam-eye-view. Figure 3(c) uses longitudinal detector shift (in the superior direction) and figure 3(d) uses both longitudinal (superior) and lateral (right) shifts. Longitudinal detector shift has probably never been used in any clinical systems, or discussed in any publications. If used, a longitudinal shift would increase the cone angle and result in more image artifacts because of the approximation.
in the FDK-type reconstruction algorithms, and will not provide a longitudinal coverage extension in a single scan.

The detector panel width and the use of lateral shift determine the FOV (field-of-view) size, while the CBCT detector panel height and the FOV size determine the CBCT longitudinal coverage. The detector panels are 40 cm in width and 30 cm in height in on Varian on-board imaging system (Varian Medical Systems, Palo Alto, California) OBI systems. In head scan mode in which the detector is laterally centered, the FOV is 25 cm and longitudinal coverage is 17 cm. In pelvis scan mode in which the detector panel is laterally shifted by 16 cm, the FOV is 45 cm and the longitudinal coverage is 15.5 cm. These values were geometrically limited by the detector panel size, the selection of scan mode, and the mechanical design of the machine.

2.2. DSCS—Dual scans with complementary detector panel shifts

DSCS uses two complementary circular scans with the opposite longitudinal and lateral shifts of the kV detector. The superior scan $S_1$ has the detector panel shifted to the right and inferior while the inferior scan $S_2$ to the left and superior (figure 4(a)). Figure 4(b) reveals how each of these two scans covers exactly the half of the scanned object for a total 40 cm longitudinal coverage. The two scans can be thought as a pair of x-ray sources plus a pair of detectors, shown in figure 4(c), rotating together circularly along the central axis. Figure 4(c) assumes a perfect matching of the image beams at the longitudinal junction and therefore there is no longitudinal overlapping between these two scans at any rotation angle. However, this may not always be dependable due to system’s mechanical uncertainty. In order to account for this, as shown in figure 4(d), a 0.5 cm longitudinal beam overlapping is intentionally introduced. In addition, the detector is shifted by 16 cm laterally, which is the same lateral shift as used in a standard pelvis scan. Note that there is an 8 cm lateral overlapping which is used by the FDK half-fan reconstruction algorithm to reduce the image artifacts.

There are several important notes for the proposed DSCS technique. Two scans are able to completely cover a 39.5 cm long object if a 0.5 cm longitudinal beam overlapping is used. Two x-ray source positions are 39.5 cm apart in the longitudinal direction. In scan 1 (blue color), the detector panel is shifted 14.75 cm to the inferior and 16 cm to the left. Its superior edge is aligned with the source position of this scan. In scan 2 (red color), the detector panel is shifted 14.75 cm to the superior and 16 cm to the right. Its inferior edge is aligned with the source position of this scan. The beam collimators or blades are used to create a half-beam block, shown in figure 3(c). As a result, there is no imaging radiation superior to the source position in $S_1$, or inferior to the source position in $S_2$. The imaging radiation is 100% confined inside the effective imaging volume. Each scan can occur in either gantry rotation direction. Practically, one scan uses the forward gantry rotation (clockwise), while the other one uses the backward rotation (counterclockwise), in order to reduce the total scan time.

2.3. Image reconstruction

2.3.1. Existing reconstruction algorithms for detector lateral shifts. CBCT reconstruction for the laterally displaced detector panel has been well studied (Cho et al 1996, Wang 2002). Let $P_\phi(u, w)$ be the CBCT projection data at gantry angle $\phi$, and $u$ and $w$ be the horizontal and vertical pixel coordinates on the flat panel detector. If the detector panel is laterally shifted to the negative $u$ side, the actual detector will be only available for $u \in [-L, +\sigma]$, where $L$ is the maximal distance from the detector left edge to the detector center, and $\sigma$ is the overlapping or the amount of the detector that remains available to the other side of the center. Note that $L$
+ σ is the detector width, which is 40 cm for a Varian onboard CBCT system. The image \( f(\vec{r}) \) can be reconstructed as:

\[
f(\vec{r}) = \frac{1}{4\pi^2} \int \frac{v^2}{\sqrt{v^2 + u^2 + w^2}} \tilde{P}_\phi(\mu(\vec{r}), w(\vec{r})) d\phi
\]  

(1)

where \( \vec{r} \) is the position of a voxel in the object, \( v \) is source-detector distance (SDD), \( \tilde{P}_\phi(u, w) \) is the filtered projection data, and

\[
\tilde{P}_\phi(u, w) = \int_{-L}^{L} \frac{v}{\sqrt{v^2 + u^2 + w^2}} P_\phi(u', w) g(u - u') du'
\]  

(2)

where \( g(u) \) is the ramp-filter function (Feldkamp et al 1984), and

\[
\tilde{P}_\phi(u, w) = \lambda(u) \begin{cases} 
P_\phi(u, w) & \text{if } -L \leq u \leq \sigma \\
0 & \text{if } \sigma < u \leq L 
\end{cases}
\]  

(3)

where \( \lambda(u) \) is the weighting function of the laterally shifted detector, and

\[
\lambda(u) = \begin{cases} 
\frac{1}{2} \left[ \cos \left( \frac{\pi}{2} \frac{u + \sigma}{\sigma} \right) + 1 \right] & \text{if } -L \leq u < -\sigma \\
1 & \text{if } -\sigma \leq u \leq \sigma \\
0 & \text{if } \sigma < u \leq L 
\end{cases}
\]  

(4)

Equation (1) can be further simplified as:

Figure 4. DSCS uses a pair of complementary circular scans with the opposite detector longitudinal and lateral shifts. (a) Two complementary circular scans with the opposite kV detector shifts. (b) A different view to show the x-ray sources in the opposite positions. (c) 2D view with a perfect beam junction. (d) With an intentionally introduced 0.5 cm longitudinal beam overlapping.
\[
f(\vec{r}) = \int k(\vec{r}, \vec{s}) \bar{P}_\phi d\phi
\]
by dropping the constant term \(1/4\pi^2\) and the \(u\) and \(w\) variables, and defining the distance correction function as:
\[
k(\vec{r}, \vec{s}) = \frac{\nu^2}{\nu^2 + u(\vec{r}, \vec{s})^2 + w(\vec{r}, \vec{s})^2}
\]

### 2.3.2. DSCS without longitudinal overlapping.

Figure 5(a) shows a simplified illustration where the entire object is covered by source #1 at a project angle \(\phi\) and source #2 at \(\phi + \pi\). Note that the region marked as top half corner is not illuminated by source #1 but by source #2 only. The idea is to perform image reconstruction separately for the top and bottom half by following the FDK reconstruction algorithm with a modification in that, if a projection is missed because the point in the object is not illuminated by a scan at a projection angle, it will be replaced using the projection in the complementary data provided by the second scan. In figure 5(a), the point \(\vec{r}\) in the top half corner is not illuminated by \(\vec{s}_{1, \phi}\) because the projection line from \(\vec{s}_{1, \phi}\) to \(\vec{r}\) fails to reach inside the detector panel. To replace the unavailable projection data, the complementary projection from \(\vec{s}_{2, \phi}\) to \(\vec{r}\), which is available, will be used where \(\phi\) is the fan angle, and the source angle \(\phi' = \phi + \pi + 2\phi\).

By using the complementary projection in \(S_2\) to replace the missing projection data in \(S_1\), the top half can be reconstructed as:
\[
f_{\text{top}}(\vec{r}) = \int k(\vec{r}, \vec{s}_{1, \phi}) \left( M(\vec{r}, \vec{s}_{1, \phi}) \bar{P}_{1, \phi} + (1 - M(\vec{r}, \vec{s}_{1, \phi})) \bar{P}_{2, \phi'} \right) d\phi
\]
where \(M\) is the coverage function and defined as:
\[
M(\vec{r}, \vec{s}) = \begin{cases} 
1 & \text{if } \vec{r} \text{ is covered by the scan } \vec{s} \\
0 & \text{if } \vec{r} \text{ is not covered}
\end{cases}
\]

It is important to note that the projection data provided by \(S_2\) at a source angle \(\phi'\) is only near-complementary, but not exact-complementary, to the missing projection data in \(S_1\). An exact complementary projection needs to come from the exactly opposite fan angle and cone angle (Parker 1982, Wang 2002). In DSCS, the projection generated in \(S_2\) at \(\phi'\) is at the exactly opposite fan angle but not the opposite cone angle. Figure 5(a) shows the cone angle-mis-matching problem, which could cause image artifacts in the corner regions.

It is also important to note that, in order to have the near-complementary projection data from \(S_2\), it is required to use a lateral shift in \(S_2\) opposite to the lateral shift used in \(S_1\). Only if \(S_2\) uses an opposite lateral shift, it can provide the near-complementary projection at \(\phi' = \phi + \pi + 2\phi\) for a projection data that is not available in \(S_1\) at a source angle \(\phi\).

The bottom half can be reconstructed similarly as:
\[
f_{\text{bot}}(\vec{r}) = \int k(\vec{r}, \vec{s}_{2, \phi}) \left( M(\vec{r}, \vec{s}_{2, \phi}) \bar{P}_{2, \phi} + (1 - M(\vec{r}, \vec{s}_{2, \phi})) \bar{P}_{1, \phi'} \right) d\phi
\]

### 2.3.3. DSCS with longitudinal overlapping.

When a longitudinal overlapping is used in order to account for the detector panel mechanical inaccuracy (figures 4(d) and 5(b)), the corner
Equations (7) and (8) are still applicable without modification. The authors have noticed that redundant projection data from both scans are available in the region of longitudinal overlapping. These data redundancies may be exploited in the future with a more comprehensive reconstruction algorithm in order to provide potentially better image quality or a smoother transition between the top and bottom halves.

### Table 1. CBCT scan parameters used in the physical phantom measurements.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Setting</th>
</tr>
</thead>
<tbody>
<tr>
<td>Scan mode</td>
<td>Pelvis (half-fan), 360° gantry rotation</td>
</tr>
<tr>
<td>Longitudinal couch position</td>
<td>Couch extended 39.5 cm further into the gantry in scan 2 than in scan 1</td>
</tr>
<tr>
<td>Detector panel lateral shifts</td>
<td>-16 cm for scan 1, +16 cm for scan 2</td>
</tr>
<tr>
<td>Detector panel longitudinal shifts</td>
<td>+15 cm for scan 1, -15 cm for scan 2</td>
</tr>
<tr>
<td>Beam parameters</td>
<td>125 kVp, 80 mA, 20 ms</td>
</tr>
<tr>
<td>kV blade positions</td>
<td>Set to collimate the kV beam to the detector panel</td>
</tr>
<tr>
<td>Filter</td>
<td>1 mm copper flat filter*</td>
</tr>
</tbody>
</table>

regions become smaller. Equations (7) and (8) are still applicable without modification. The authors have noticed that redundant projection data from both scans are available in the region of longitudinal overlapping. These data redundancies may be exploited in the future with a more comprehensive reconstruction algorithm in order to provide potentially better image quality or a smoother transition between the top and bottom halves.

### 2.4. Implementation and verification

The DSCS scans were digitally simulated using a 3D Shepp–Logan phantom. Note that only the photon attenuation was simulated. Photon scattering, table sag, beam spectrum, beam hardening, and image noises were not considered in the digital simulation. The proposed DSCS scan was then implemented on a Varian TrueBeam LINAC in its research mode. The scan parameters, listed in table 1, were programmed with XML files. An anthropomorphic phantom was used in the physical phantom measurements. The lateral dimension of the anthropomorphic phantom is about 40 cm. The phantom projection image data were transferred to a Windows
PC where the image reconstruction took place. The projection images were preprocessed using iTools, a CBCT reconstruction research tool developed by Varian Medical Systems, for photon scattering correction, beam-hardening correction, and air scan normalization. The program was developed using C programming language and NVIDIA’S CUDA library, and ran on a Windows PC with a GeForce 560Ti (2G onboard RAM) GPU card. The total reconstruction time for a $512 \times 512 \times 400$ volume with 2 mm slice thickness was less than 2 min.

3. Results

3.1. Digital phantom simulation results

The reconstruction results of a digital phantom are shown in figure 6. The image qualities were reasonably good. The center slice had the most severe image artifacts, which was expected.
because the center slice was a result of both scans at the widest cone angle. The image artifacts were caused by 1) inaccuracy of the approximation nature of the FDK reconstruction algorithm at wider cone angles, 2) under-sampling of the half-fan scans (a 360° rotation with detector lateral shifted is equivalent to a 180° rotation, i.e. a half rotation), and 3) cone angle mismatching in the complementary projection data as previously mentioned in 2.3.2.

In figure 6, five ROIs were marked. The mean CT numbers and standard deviations were computed and listed in table 2. ROIs 1–4 were located at wider cone angles and reconstructed using the projection data from both scans. ROIs 1 and 2 were located in the overlapping region and therefore had the worst image quality and the largest standard deviation. ROI 3 was also in

Figure 7. Vertical profiles at 4 different locations. Profiles 1–3 were generated along the vertical lines passing through the ROIs 1–3 shown in figure 6(a). The profile center was generated along the vertical center.

Figure 8. Reconstructed images of an anthropomorphic phantom. Display window is [−1500 1500].
the overlapping region, but its standard deviation was lower. However, its CT number was the least accurate. ROIs 4 and 5 were only covered by one x-ray source. Therefore, the image qualities in these regions were the same as one obtained with a regular pelvis scan. Note that although ROI 4 was reconstructed from the projection data by one source only, it was not included in the regular pelvis scan reconstruction volume because it was outside the standard 15.5 cm longitudinal coverage.

Figure 7 shows the vertical intensity profiles for 4 different locations. The off-center profiles had higher noises in the overlapping regions. The intensity profiles in the image volumes outside the overlapping regions were the same as those obtained in a regular FDK reconstruction.

3.2. Anthropomorphic phantom results

Reconstructed images of a body phantom are shown in figure 8. A 39.5 cm longitudinal coverage with a 45 cm FOV was accomplished with DSCS. The image artifacts were less visible than those present in the digital phantom simulation, probably due to the lower image intensity gradients in the physical phantom.

4. Discussion

DSCS was not designed to improve image quality. It was designed to provide better longitudinal coverage and to eliminate double imaging dose at the imaging junction. The imaging dose outside the effective imaging volume has been minimized but not completely eliminated due to photon scattering. In fact, the image quality varied longitudinally across the imaging volume. It became worse in the longitudinal center region due to the wider cone angle and the approximate FDK reconstruction algorithm. The corner regions, illuminated by both scans and reconstructed using the projection data from both scans in a complementary way, had the worst image quality.

Image reconstruction is very challenging in this study due to wider cone angles, detector longitudinal shifts and overlapping, detector lateral shift and overlapping, and couch table sagging. Our current reconstruction algorithm, which was based on a modified FDK reconstruction algorithm (Cho et al 1996, Wang 2002), can handle these issues. However, the inherent inaccuracy with the FDK algorithm would result in inferior image quality in these situations.
We believe that completely different image reconstruction algorithms are necessary to provide a definite improvement in image quality, for example, iterative reconstruction algorithms as they naturally support DSCS. For now, iterative CBCT reconstruction is still too slow to be clinically acceptable, but the situation could improve in the near future.

There are two additional issues associated with the physical phantom scans: mechanical accuracy of the imaging system and the use of bowtie filter. It could be DSCS’s disadvantage that the geometry accuracy of the reconstructed image volume depends on the machine mechanical accuracy. The couch longitudinal position difference is 39.5 cm between the two scans as opposed to the 15.5 cm difference in the double scan approach (Zheng et al 2012). As a result, there are two potential problems: 1) the accuracy of the couch longitudinal movement, and 2) the couch vertical discrepancy (i.e. table sagging) at the two longitudinal positions that are 39.5 cm apart. We have observed up to 3 mm couch vertical and longitudinal errors in the phantom experiments. In this study these errors in the reconstruction were manually corrected by realigning the two independently reconstructed half volumes.

As shown in the physical phantom results, the image is very noisy because bow-tie filter was not used. This is because the current bow-tie filter is unilateral (figure 9(a)) and therefore it can only filter one of the two scans as it cannot be flipped to filter the other scan. It is necessary to design a new bi-lateral bowtie filter to filter both scans as shown in figure 9(b).

It is feasible to use DSCS method in different ways to further reduce imaging dose. Not all clinical cases require a 40 cm longitudinal coverage but any longitudinal coverage greater than 15.5 cm in pelvis mode requires two scans. For a required coverage of, e.g. 25 cm, we can use longitudinally narrower beams with DSCS to reduce the total imaging dose. As a result, scatter dose is also reduced, thereby potentially resulting in better image quality. The double-scan configuration will still be similar to that shown in figure 4(d) but just with narrower beams in longitudinal.

For DSCS, FOV is not dependent on the longitudinal coverage, unlike standard clinical CBCT scans in which a wider FOV would result in shorter longitudinal coverage due to beam divergence. The FOV independency can be seen in figures 4(c) and (d). The implication is that it is possible to shift the detector panel laterally further to obtain a larger FOV. However, for the 40 × 30 cm² kV detector panel, the maximal lateral shift achievable is 16 cm on current Varian LINACs. If the panel can be shifted by 19 cm, the FOV can then be extended from 45 cm to a 50.3 cm. An extended FOV is useful to scan large patients.

5. Conclusion

This proof-of-concept study has demonstrated that the proposed DSCS method is able to effectively extend the CBCT longitudinal coverage to 39.5 cm. No wasted imaging dose is present outside the effective imaging volume. This method could be potentially useful for image guidance and subsequent treatment plan adaptation.

Acknowledgements

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