Reduction of motion blurring artifacts using respiratory gated CT in sinogram space: A quantitative evaluation

Wei Lu, Parag J. Parikh, and James P. Hubenschmidt
Department of Radiation Oncology, Washington University School of Medicine, St. Louis, Missouri 63110

David G. Politte and Bruce R. Whiting
Department of Radiology, Washington University School of Medicine, St. Louis, Missouri 63110

Jeffrey D. Bradley, Sasa Mutic, and Daniel A. Low
Department of Radiation Oncology, Washington University School of Medicine, St. Louis, Missouri 63110

(Received 10 May 2005; revised 5 August 2005; accepted for publication 24 August 2005; published 17 October 2005)

Techniques have been developed for reducing motion blurring artifacts by using respiratory gated computed tomography (CT) in sinogram space and quantitatively evaluating the artifact reduction. A synthetic sinogram was built from multiple scans intercepting a respiratory gating window. A gated CT image was then reconstructed using the filtered back-projection algorithm. Wedge phantoms, developed for quantifying the motion artifact reduction, were scanned while being moved using a computer-controlled linear stage. The resulting artifacts appeared between the high and low density regions as an apparent feature with a Hounsfield value that was the average of the two regions. A CT profile through these regions was fit using two error functions, each modeling the partial-volume averaging characteristics for the unmoving phantom. The motion artifact was quantified by determining the apparent distance between the two functions. The blurring artifact had a linear relationship with both the speed and the tangent of the wedge angles. When gating was employed, the blurring artifact was reduced systematically at the air-phantom interface. The gated image of phantoms moving at 20 mm/s showed similar blurring artifacts as the nongated image of phantoms moving at 10 mm/s. Nine patients were also scanned using the synchronized respiratory motion technique. Image artifacts were evaluated in the diaphragm, where high contrast interfaces intercepted the imaging plane. For patients, this respiratory gating technique reduced the blurring artifacts by 9%–41% at the lung-diaphragm interface. © 2005 American Association of Physicists in Medicine.

Key words: motion artifact, CT, respiratory gating, sinogram

I. INTRODUCTION

Respiratory motion during computed tomography (CT) causes artifacts (blurring, streaks, discontinuities, etc.) in thoracic imaging.1,2 These artifacts not only degrade the image quality but also may lead to errors in target delineation and subsequently in dose delivery for radiation therapy. For diagnosis, a helical CT is usually acquired during a single held breath with minimal motion artifacts. Treatment planning CT, however, is usually acquired with the patient breathing freely.3–5 The respiratory motion results in image artifacts that can affect the reconstructed tumor and normal organ shapes.

Three major methods have been proposed to reduce motion artifacts for free-breathing CT scans. One method minimizes the scan time (<100 ms) using a fast scanner.6,7 Half scan reconstruction shortens the effective scan time by using x-ray projections from less than 360° of gantry angles (typically 180° + the x-ray fan angle).8,9 Both approaches improve temporal resolution and reduce motion artifacts. Images generated with shorter scan time, however, have lower signal-to-noise ratios (SNR) than those with conventional scan times.10,11 Another method corrects for motion based on a time-varying expansion model, which models in-plane motion in the chest by scaling functions along two directions. The correction is made in the reconstruction process10,12 or in the sinogram space.11 The usefulness of the correction method is limited by the fact that the model does not sufficiently describe motion in the chest.12,13 Reliable 3D motion models or techniques for estimating the motion are current topic of research.14,15 The third method uses respiratory gating by examining segments of projection data that fall into a gating window over several scans. Cardiac gating with electrocardiogram (ECG) has been useful in CT imaging of the heart.2,16–21 Cardiac motion artifacts are significantly reduced, yielding “frozen” heart images. Instead of using ECG, a spatial overlap correlator has also been used to track the phase of periodic organ (heart) motion to reconstruct gated heart images.13,22 Respiratory gating with a breathing waveform has been used in reducing motion uncertainty for radiotherapy3,23–28 and in reducing motion artifacts for thoracic imaging.7,29 Recent studies have shown that respiratory gated PET significantly improves the accuracy of tumor volume determination by reducing motion blurring artifacts in the thorax.30,31

Besides the above three major methods, there are several other methods for organ motion estimation and compensa-
tion. One method uses spatial redundancy in “opposite rays” to obtain temporal information, but it is limited to in-plane rigid motion. A correlation plus least-square-root method is used to determine the mismatch between adjacent x-ray projections and to estimate patient translation in helical CT, but the motion is limited to in-plane, straight-line, uniform patient movement.

In this study, a technique to retrospectively gate CT projections in the sinogram space was developed. The projections of every CT scan were sorted into different respiratory gated windows. A synthetic sinogram was then constructed by averaging projections from multiple scans inside each window. We then quantitatively evaluated the reductions in motion blurring at the air-phantom interface for wedge phantoms, and at the lung-diaphragm interface for patients.

II. MATERIALS AND METHODS

A. Phantoms and patients

We intended to use the diaphragm to evaluate motion artifacts. Therefore, we constructed a phantom system to simulate respiratory motion of the diaphragm in the craniocaudal (CC) direction (along the longitudinal axis of the scanner). Four wedge phantoms [Fig. 1(a)] were mounted on a computer-controlled linear stage. The phantoms were moved back and forth for approximately 3 cm along the CC direction at four constant speeds: 5, 10, 15, and 20 mm/s with periods of 12, 6, 4, and 3 s. An example synthetic motion is shown in Fig. 2(a) with a speed of 20 mm/s for three cycles. Note that at the beginning of every experiment, the phantoms were moved to a common longitudinal position, which was defined as the reference phantom position (0 cm). The purpose of using constant-velocity motion was to aid in the quantitative analysis of the subsequent reconstructed images. The angles (α) between the phantom motion direction and the slanted surface of the four wedge phantoms were 50°, 60°, 70°, and 80°. To create shallower angles, the four wedges were rotated, providing interfaces at angles of 40°, 30°, 20°, and 10°.

Twelve sets of CT data from six lung cancer patients and three upper-abdomen study patients were collected for this study. One of the lung cancer patients was scanned four times. Patients breathed freely with respiratory motion monitored by quantitative spirometry.

B. CT data acquisition

Transverse slices (1.5 mm thick) were acquired using a 16-slice CT scanner (Sensation 16, Siemens Medical Solutions, Malvern, PA) operated in 12-slice (1.8 cm) settings. The scanner was operated in ciné mode (couch stationary during scanning) with 100 and 15 scans acquired continuously at each couch position for phantoms and patients, respectively. Each scan (360° rotation) required 0.5 s to acquire followed by a 0.25 s dead time. Following our clinical treatment planning CT scanning protocols, we used a 48.0 cm reconstruction field of view, yielding 0.94 mm × 0.94 mm × 1.5 mm voxels. Both phantom motion and patient spirometry were synchronized to the CT scanner acquisition by a photo resistor that monitored the “x-ray On” light. A detailed description of this process can be found in earlier works.

Due to a limitation of our filtered back-projection (FBP) software, only one of the two central slices (the sixth slice) of each 12-slice scan was reconstructed and used in this study. Figure 1(b) shows a CT slice of the static phantoms acquired at a position where the image plane intercepted the slanted surface of every wedge phantom. This position is at 2.4 cm with respect to the reference phantom position [Fig. 2(a)]. We defined this as the desired reconstructed scan position for the phantom studies. Figure 2(b) shows a CT slice of the moving phantoms acquired at the desired position with a velocity of 20 mm/s. The motion caused clearly observable blurring artifacts at the air-phantom interface. Figure 2(c) shows the sinogram for the CT slice. A sinogram depicts the Randon transform of x-ray attenuation coefficients as a function of gantry angle index and detector index. The CT scanner acquires transmission data at 1160 gantry angles for a 360° rotation with 672 detectors for each gantry angle. In this paper, a “projection” refers to the Randon transform at all detectors for a source gantry angle.

C. Respiratory gated image reconstruction in sinogram space

Figure 3 illustrates the concepts of the respiratory gated image reconstruction for phantom study. A respiratory gating window, represented by the two gating thresholds, was centered at the desired reconstructed scan position. Multiple scans had projections that fall into the gating window and are constrained to have the same respiratory direction (inhalation or exhalation). These projections were averaged at each gantry angle to construct a synthetic sinogram. Finally, a gated image was reconstructed from this synthetic sinogram by a fan-beam FBP for full rotation (360°, 0.5 s) (Ref. 36) with a modified ramp filter. The width of the gating window was characterized by the ratio of the motion in the gating window to that which occurred during the 0.5 s acquisition time. In Fig. 3, a 60% gating window for inhalation is shown for phantoms moving at 20 mm/s. The phantoms moved by 10 mm during the 0.5 s CT acquisition, but the range of motion was limited to 6 mm here. There were three inhalation scans shown that had projections falling within the gating window. These projections were averaged in the gantry angle direction to generate the synthetic sinogram.

For the patient study, the couch position where the lung intercepted the diaphragm and where the greatest motion blurring appeared (at the lung-diaphragm interface) was visually selected. Among the 15 scans acquired at this couch position, the scan with the greatest motion blurring, which was usually at mid-inhalation or mid-exhalation, was used as a reference for the blurring reduction evaluation. The spirometer-measured tidal volume was used as a surrogate for respiratory motion. The same gating technique as illustrated for the phantom study above was used to construct a synthetic sinogram. But now the gating window was cen-
tered at the tidal volume of the reference scan and its width was defined as the ratio of the tidal volume change in the window to that occurring during the 0.5 s acquisition time. The narrowest gating window was determined by decreasing the window width from 100% in 5% steps until the resulting

FIG. 1. (a) The wedge phantoms. (b) A transverse CT slice of the wedge phantoms acquired without phantom motion (static). All CT slices in this paper have a size of 512×512 pixels, with a spatial resolution of 0.94 mm×0.94 mm. The X axis represents the lateral direction and the Y axis represents the anteroposterior direction. The vertical line indicates the locations of a CT profile that crosses the air-phantom interface for the 60° wedge. (c) An error function fit to the CT profile [vertical line in (b)], which represents the partial volume effects without motion.

FIG. 2. (a) Synthetic respiratory motion in craniocaudal direction. The motion has a speed of 20 mm/s and a range of 3 cm. (b) A CT slice of the moving phantoms at the desired reconstructed scan position (corresponding to a selected phase of breathing). (c) The sinogram for (b).
D. Quantitative evaluation of the motion blurring

The degree of motion blurring has been qualitatively evaluated by physicians or experts. We propose a quantitative approach for its evaluation.

A CT profile taken along the vertical direction and that crossed the air-phantom interface in the static phantom image Fig. 1(b)] displays a flat air region, a flat phantom region, and a progressively increasing transition from air region to phantom region Fig. 1(c)]. Note that the CT values in the displayed images are shifted Hounsfield values where \( \text{CT}_{\text{air}} = 0 \) and \( \text{CT}_{\text{phantom}} = 1200 \). The transition region depicts the partial volume effect at the air-phantom interface. As shown in Fig. 1(c), an error function was found to be a good fit to the CT profile.

The motion blurring artifact was primarily caused by the different attenuations measured between the 0.25 s period when the CT gantry rotated from one horizontal orientation to the other (half rotation). As illustrated in Fig. 4(a), a change in the height by \( \Delta y \) of the intercepted phantom material in the anteroposterior direction (AP) was a function of the motion \( \Delta z \) in the CC direction and the tangent of the phantom wedge angle \( \alpha \),

\[
\Delta y = \Delta z \times \tan(\alpha),
\]

where \( \Delta z \) is

\[
\Delta z = v \times \Delta t,
\]

where \( v \) is the speed of the phantom motion and \( \Delta t \) is the half rotation time or 0.25 s for the scanner used, therefore,

\[
\Delta y = 0.25 \times v \times \tan(\alpha).
\]

To measure the degree of a motion blurring artifact in a CT image, we examined a CT profile taken along the vertical direction and that crossed the air-phantom interface Fig. 2(b)]. This profile appeared as a two-step function, where the points in the middle region had approximately the mean CT values of air and phantom Fig. 4(b)]. There were two transitions in the CT values: one between the air and artifact and the second between the artifact and the phantom. The shape of the two transitions mimicked that of the single transition for the static phantom in Fig. 1(c)]. Each transition was assumed to be due to partial volume averaging, while the split-
When the blurring was approximately the size of a pixel where CT is the shifted Hounsfield value, the apparent distance of the air to phantom transition into two transitions was strained to be within 1% of each other. The “CT magnitude” and the “slopes” of both fitting functions were constrained to be within 1% of each other. CT was a good estimate of the motion blurring. The effective time span from every linear fit by Eqs. (1) and (2). The resulting Δt values had an average of 0.242±0.006 s, which was close to the expected value of 0.25 s.

A similar process was used to measure the blurring artifact Δd for the eight wedge angles at the four speeds. The measured Δd values were almost always close to the expected Δy values: among the 32 cases, only two had a difference larger than 1 mm. Figure 5 shows the resulting Δd as a function of the motion speed and as a function of the tangent of the phantom wedge angles, respectively. The blurring artifact shows a linear relationship with both variables. A linear fit was applied to each group of data with a constraint that it passed the origin (0, 0). This constraint implied that the motion blurring should be zero when α or v is zero. There are some small deviations from the fits for situations that yielded small blurring (close to or smaller than 1 pixel) because blurring of less than one pixel could not be accurately measured. The linear relationship of Δd with both v and tan(α) and the small difference between Δd and Δy suggest that Δd was a good estimate of Δy, or the degree of motion blurring. The effective time span Δt was determined from every linear fit by Eqs. (1) and (2). The resulting Δt values had an average of 0.242±0.006 s, which was close to the expected value of 0.25 s.

A synthetic sinogram [Fig. 6(a)] was constructed as described in Sec. II C. Among the total acquisition of 100 scans, there were 22 “inhalation” scans that had projections falling into the 60% amplitude gating window. Each projection in this synthetic sinogram was an average of all projections at the same gantry angle position from the contributing scans. This synthetic sinogram displays more smooth variations than the original sinogram [Fig. 2(c)] due to the pro-
jection averaging. A gated CT slice was reconstructed from this synthetic sinogram by FBP [Fig. 6(b)]. Clearly, the motion blurring artifact was reduced compared to that in Fig. 2(b). The average motion blurring $\Delta d$ as measured by Eq. (7) was reduced by 48% in this gated image compared to the nongated image. The quantitative analysis demonstrated that the gating reduced the blurring for all cases (eight angles at four speeds) when the blurring was larger than 1 pixel. Figure 6(c) shows the measured blurring for images of phantoms moving at 20 mm/s and 10 mm/s. In both cases, the blurring was significantly reduced in the gated images. The gated image of phantoms moving at 20 mm/s showed similar blurring artifacts as the nongated image moving at 10 mm/s.

Figure 7 demonstrates that the motion blurring decreases as the gating window width decreases. This was mainly because the net range of motion was decreased. The decrease is not linear to the gating window width; it is greater at a wider gating window or a larger angle. Different wedge angles have different blurring reductions for the same gating window. For example, with a 60% gating window, the blurring was 47% of that in the nongated image for the 80° angle, 55% for the 60° angle, and 66% for the 40° angle. Overall, the proposed gating technique was more useful for cases with greater motion blurring.

B. Patient results

Figure 8 demonstrates the results of the respiratory gating technique on motion blurring reduction for a patient case. Figure 8(a) shows the patient breathing curve during the 15 scans at one couch position where the lung intercepted the diaphragm. Mean-intensity blurring artifacts are clearly observable at the lung-diaphragm interface [Fig. 8(b)]. The narrowest gating window available around the reference scan (first scan) using the acquired sinogram data was determined to be 55%. Scans 1, 6, and 15 had projections falling into this

![Figure 6](image)  
**Fig. 6.** (a) A synthetic sinogram for a 60% gating window and 20 mm/s motion. (b) The gated CT image reconstructed from (a) by FBP. Compared to Fig. 2(b), the blurring artifact is clearly reduced. (c) Comparison of the blurring artifacts between the gated and nongated images at 20 mm/s and 10 mm/s. The gated image at 20 mm/s showed similar blurring artifacts as the nongated image at 10 mm/s.

![Figure 7](image)  
**Fig. 7.** Motion blurring artifacts decreased as the gating window width (in percentage of motion occurred during the 0.5 s acquisition time) decreased. Results for five wedge angles at 20 mm/s are shown.

![Figure 8](image)  
**Fig. 8.** Demonstrates the results of the respiratory gating technique on motion blurring reduction for a patient case. Figure 8(a) shows the patient breathing curve during the 15 scans at one couch position where the lung intercepted the diaphragm. Mean-intensity blurring artifacts are clearly observable at the lung-diaphragm interface [Fig. 8(b)]. The narrowest gating window available around the reference scan (first scan) using the acquired sinogram data was determined to be 55%. Scans 1, 6, and 15 had projections falling into this
Fig. 8. (a) Patient breathing curve for the 15 scans at a couch position where the lung intercepts the diaphragm. A 55% gating window around the reference scan (first scan) is shown. (b) A CT slice of the reference scan. Blurring artifacts are observable at the lung-diaphragm interface. (c) Two error functions fit to the profile that crosses the lung-phantom interface [line in right lung in (b)]. (d) The respiratory gated CT slice. The motion blurring at lung-diaphragm interface is reduced in the right lung compared to (b). (e) Comparison of the blurring at the lung-diaphragm interface between the nongated (b) and gated (d) images. Crosses and curves in (b) and (d) indicate locations for CT profiling.
TABLE I. Motion blurring reductions by the respiratory gating at the lung-diaphragm interface for nine sets of patient data.

<table>
<thead>
<tr>
<th>Patient case</th>
<th>Gating window width</th>
<th>Right lung</th>
<th>Left lung</th>
<th>Both lungs</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>55%</td>
<td>44.8%</td>
<td>-5.2%</td>
<td>32.8%</td>
</tr>
<tr>
<td>2</td>
<td>95%</td>
<td>50.3%</td>
<td>2.6%</td>
<td>27.7%</td>
</tr>
<tr>
<td>3</td>
<td>60%</td>
<td>-4.2%</td>
<td>32.4%</td>
<td>17.8%</td>
</tr>
<tr>
<td>4</td>
<td>50%</td>
<td>15.2%</td>
<td>27.5%</td>
<td>19.2%</td>
</tr>
<tr>
<td>5</td>
<td>70%</td>
<td>24.0%</td>
<td>23.6%</td>
<td>23.9%</td>
</tr>
<tr>
<td>6</td>
<td>75%</td>
<td>18.7%</td>
<td>9.8%</td>
<td>13.6%</td>
</tr>
<tr>
<td>7</td>
<td>95%</td>
<td>26.9%</td>
<td>-0.2%</td>
<td>15.2%</td>
</tr>
<tr>
<td>8</td>
<td>90%</td>
<td>15.1%</td>
<td>1.6%</td>
<td>9.0%</td>
</tr>
<tr>
<td>9</td>
<td>75%</td>
<td>45.9%</td>
<td>36.5%</td>
<td>41.2%</td>
</tr>
</tbody>
</table>

Not significant, p > 0.05.

55% gating window. A typical CT profile that crossed the lung-phantom interface displays a lung region, a diaphragm region, and a middle artifact region with a CT value that was the average of the first two regions [Fig. 8(c)]. Though larger variations appear in the CT profile, the two error functions fit it well. In the reconstructed gated image [Fig. 8(d)], the blurring was observed to be less severe, especially the long mean-intensity artifact at the lung-diaphragm interface in the right lung. To make a quantitative comparison, we obtained CT profiles between a point in the soft tissue region and points on a contour in each lung for both the nongated and gated images. The blurring $\Delta d$ for each profile was measured in the same way as that for the phantom studies with Eq. (7). The $\Delta d$ values for all profiles were plotted in Fig. 8(e), with a blurring range of 0–24 mm. The mean blurring in the gated image was reduced by 44.8% ($p < 0.01$) for the right lung, increased by 5.2% ($p > 0.05$) for the left lung, and decreased by 32.8% ($p < 0.01$) for both lungs.

For 3 out of the 12 sets of patient data, the patients breathed one and a half or fewer breaths during the 11 s period of acquiring the 15 scans at each couch position. It was less likely for two scans to have similar tidal volumes and the same respiratory direction. For these three cases, it was not possible to construct a complete sinogram with a gating window narrower than 100%. Table I summarizes the blurring reductions by the gating technique for the other nine cases. The couch position where the lung intercepted the diaphragm and where blurring was greatest was used for each patient. The narrowest gating window width that provided a complete synthetic sinogram ranged from 55% to 95%. Out of the 15 scans, there were two to four scans falling into this gating window. For patients 1, 3, and 7, for one lung the blurring in the gated image increased slightly relative to the nongated image. This increase was found mainly due to the smoothing of sharp edges as a result of averaging projections and the inability to accurately quantify small blurring. The blurring decreased in the gated image for the other cases. The average blurring in both lungs was reduced for the patients, by 9%–41%. When we decreased the gating window width from 100% to the narrowest, the motion blurring in both lungs decreased similar to the phantom results shown in Fig. 7.

IV. DISCUSSION AND CONCLUSION

Our quantitative evaluation showed that the proposed respiratory gating technique effectively reduced the motion blurring artifacts in phantom and patient studies. One advantage of the gating technique was that it did not require a motion model and was not limited to in-plane motion. As is the case for any gating technique, the performance of the proposed method depends on the reproducibility in anatomic locations as a function of patient breathing. An assumption underlying our analysis was that the anatomical location was reproducible as a function of tidal volume and direction of breathing. Because of hysteresis, it was important to separate inhalation scans from exhalation scans. Visual comparisons between patient images with similar tidal volumes and the same direction suggested that this assumption was most likely true. A quantitative verification is underway at our institution.

The blurring and blurring reduction varied in patients. The blurring reduction also depended on several other factors: (1) the number of scans at each couch position (Generally speaking, the more scans that are acquired, the narrower the gating window could be selected, and the smaller the resulting blurring artifacts, at least until the smoothing effect from projection-averaging dominated. For patients, however, the number of scans was limited due to radiation dose and CT acquisition time.); (2) the patient breathing patterns, including the breathing period, the variation in amplitude, etc.; (3) the relationship between CT acquisition time and breathing period; and (4) the starting gantry angle of a CT acquisition. A set of acquisition parameters adapted to each patient would have improved the efficiency of this technique.

There are several alternative methods for constructing a gated synthetic sinogram from multiple scans at the same couch position. The averaging method reported in this manuscript takes into account projections from all contributing scans and has the advantage of reduced image noise. We investigated several other methods, including (1) “using first available projections,” which fills the synthetic sinogram with as many valid projections as possible from the first scan, then supplementing from subsequent scans until the sinogram is completed or all scans are examined; (2) “using closest projections,” which fills each gantry angle with a projection that has the closest position (tidal volume) to the desired reconstructed scan position (tidal volume); and (3) “using largest block first,” which always fills a sinogram with the scan that contributes the largest gantry angle block for unfilled projections. Our results (not shown) suggested that the averaging method performed best in terms of motion artifact reduction, reconstruction artifacts, and noise reduction.

The phantom was programmed to move at a constant speed to facilitate quantification of the blurring artifact. For patients, the speed of respiratory motion varies during a
breathing cycle, usually smaller at the end of inhalation or exhalation than at mid-inhalation or mid-exhalation. The range of motion is therefore smaller at the two tidal volume extremes for the same time period. This is one reason that gated radiotherapy treatments focus on the ends of respiration.21,22 Our technique can generate gated images at any respiratory period that provides sufficient sinogram projections. For this study, we selected a gating window close to the middle inhalation or exhalation in order to have maximal motion artifacts.

The proposed technique is not only of averaging in the sinogram space, but more importantly of gating in the sinogram space, which reduces the motion range and equivalently the effective scan time for the gated image. When the effective scan time is reduced, the temporal resolution is improved and the motion artifact is reduced. Averaging data from multiple breath cycles at each projection is just an approach used for gating. The gating technique is fundamentally different from image space averaging, which simply averages CT images from multiple breath cycles at a similar phase. The image-space averaged image has an effective scan time greater than the scanner rotation time and is different from image space averaging, which simply averages data from multiple scans. The motion blurring artifact is thus more severe in such image-space averaged image than in the original image. Furthermore, structures with fine detail are blurred out in image-space averaged images. The proposed gating technique requires multiple scans (15 or more) to obtain a complete sinogram at a specified respiratory phase. To keep the patient dose low, only a quarter of the clinical mAs was used in the imaging process. A lower dose is associated with a poorer SNR, however, in lung image this is not critical since lung tissues have high contrast. The averaging of projections improves the SNR for the gated image. On the other hand, if using prospective gating where the scanner is turned on only when the breathing is within the specified respiratory window, unnecessary patient dose would be reduced. It has been shown that prospective ECG gating is twice as efficient as retrospective ECG gating in obtaining complete projections for a specific 10% cardiac cycle window.23 Since we use the tidal volume amplitude as the gating metric, it should be straightforward to implement a similar prospective respiratory gating method. Though only respiratory gating was addressed in this study, it may be possible to combine both ECG and respiratory gating to suppress artifacts caused by both cardiac motion and respiratory motion.24

With the recent advances in CT technology, such as shorter CT acquisition times and partial angle or half scan reconstruction, the time required for acquiring each slice has been shortened, yielding images with improved temporal resolution and fewer motion artifacts. The technique presented in this paper can be combined with the shorter scan times to further reduce the motion artifacts for large amplitude and fast respiratory motions.

ACKNOWLEDGMENT

This work was supported in part by the National Institute of Health Grant No. R01 CA 096679.

10Author to whom correspondence should be addressed. Electronic mail: low@wustl.edu