Effects of using Different Reconstruction Algorithms on Coronary Motion Artifacts at Various Heart Rates during Coronary CT Angiography: A Phantom Experiment

Rika Fukui, Yuzo Yamamoto, Koji Tanigaki, and Shigeru Suzuki

Objective: We assessed coronary motion artifacts at various heart rates (HRs) using coronary computed tomography angiography (CCTA) and a phantom; the resulting data were reconstructed using half-scan reconstruction algorithms (HSRA), multi-sector reconstruction algorithms (MSRA), and a novel vendor-specific motion correction algorithm (MCA) introduced to eliminate coronary motion artifacts. Materials and Methods: Using retrospective electrocardiographic (ECG)-gated helical CCTA scans of a cardiac phantom that included branching coronary artery models filled with iodine contrast medium and pulsating at HRs of 50 to 100 beats per minute (bpm), we reconstructed images using HSRA, MSRA, and HSRA combined with MCA during both systole and diastole. On axial images, 2 readers graded image quality focused on coronary motion artifacts at 50 to 100 bpm in 9 segments of the models using a scale from 1 (poor) to 5 (excellent). We then compared the average scores among the 3 algorithms using Kruskal-Wallis and post-hoc tests. Results: At 50 to 60 bpm, there were no significant differences in image quality among the 3 algorithms (P > 0.05). At 70 to 100 bpm, the image quality using MSRA was comparable or better than that of HSRA, and HSRA combined with MCA provided a comparable or better image quality compared with the other 2 algorithms. Conclusion: Coronary motion artifacts are comparable or significantly reduced using HSRA combined with MCA, compared with MSRA.

KEY WORDS: coronary CT angiography, heart rate, image quality, motion artifact, motion correction algorithm, reconstruction
There are some reports that the additional use of MCA to HSRA is effective for reducing coronary motion artifacts\textsuperscript{13-16}. On the other hand, to our knowledge, no studies have examined the effectiveness of MCA to MSRA for objects within an HR above 75 bpm\textsuperscript{17}. Using a pulsating cardiac phantom, we assessed coronary motion artifacts at various HRs reconstructed using HSRA, MSRA, and HSRA combined with MCA.

II. Materials and Methods

1. Original cardiac phantom

As shown in Fig. 1, we constructed an original pulsating cardiac phantom containing silicone models of branching coronary vessels (PFCP-1; Fuyo Corporation, Tokyo, Japan) to simulate the morphology of the human left coronary artery tree; the silicone vessels had inner diameters of approximately 1.5 mm (most distal site) to 6 mm (most proximal site), and the model contained a silicone chamber to simulate the left ventricle (LV). The chamber was directly connected to a cardiac driver pump controlled by a computer that simulated ECG signals and arbitrarily determined the HR of the LV model. We used these ECG signals to perform ECG-gated CT scans. The chamber model was directly connected to a cardiac driver pump controlled by a computer that simulated ECG signals and arbitrarily determined the HR of the LV model. We used these ECG signals to perform ECG-gated CT scans. The pump pulsed the LV model and simulated the \textit{in vivo} time-volume curve of the LV at various HRs. Fig. 2 shows information regarding the simulated ECG signals and time-volume curve of the LV model at each HR. Silicone models of branching vessels were fixed in the urethane gel present on the surface of the LV model. The CT attenuation value of the urethane gel on the image obtained under a tube voltage of 120-kVp was approximately 30 Hounsfield units (HU). This arrangement of the vessel model allowed the automated tracking of vessels using a dedicated workstation and postprocessing with MCA in the CCTA studies.

We submerged the models of the LV and coronary vessels in degassed water, filling a plastic chamber, and placed the chamber on a CT scanner bed. The long axis of the LV model was arranged parallel to the \( z \) axis of the CT gantry. The LV and coronary vessels were filled with diluted iopamidol (Iopamiron; Bayer Healthcare, Osaka, Japan), an iodine contrast medium. The CT attenuation value of the contrast medium on the image using a 120-kVp tube voltage was about 250 HU within the LV model and about 650 HU within the coronary vessel models. We classified the coronary vessel models into 9 segments, which we numbered as 5 to 13 in the left coronary artery, according to the 15-segment model of the American Heart Association\textsuperscript{18}. Fig. 3 shows this in a volume-rendered CCTA image of the phantom. The coronary vessel models were divided into 9 segments, numbered as 5 to 13 in the left coronary artery, according to the 15-segment model of the American Heart Association\textsuperscript{18}. Fig. 3 shows this in a volume-rendered CCTA image of the phantom. The inner diameters of the segments were as follows: Segment 5, approximately 6 mm; Segments 6 and 11, 5 mm; Segments 7, 9, and 12, 4 mm; Segment 8, 3 mm; and Segments 10 and 13, 3.5 mm. For the following ECG-gated CT scans, the LV model was pulsed at a constant HR of 50 to 100 beats per minute (bpm) at 10-bpm intervals.

2. ECG-gated CT scan

Using a single-source 64-detector CT scanner (Discovery CT750HD; GE Healthcare, Milwaukee, WI, USA), we acquired a scout image to determine the scan range for the following CT scan to cover the entire LV and coronary vessel models. We then performed retrospective ECG-gated helical scanning using the following scan parameters: tube voltage, 120 kVp; tube current, 590 mA; rotation time, 350 msec; pitch, 0.16; collimation, \( 64 \times 0.625 \) mm; scan field of view (FOV), 36 cm; and CTDi\textsubscript{vol}, 107.19 mGy. The ECG-gated CT scan was performed once at each HR.
3. Image postprocessing

After scanning, we reconstructed multiphase axial images using a composite of 50% adaptive statistical iterative reconstruction (ASiR; GE Healthcare) with 50% filtered back projection (FBP) and the following scan parameters: slice thickness, 0.625 mm; slice interval, 0.625 mm; display FOV, 200 mm; and convolution kernel, Standard. We then transferred the reformatted images to a workstation (Advantage Workstation VolumeShare 4.6; GE Healthcare) for postprocessing.

In principle, after the automated tracking of each coronary vessel on image sets reconstructed with HSRA at adjacent cardiac phases (within 80 msec before and after the target phase), MCA (SnapShot Freeze; GE Healthcare) performs the following operations: (1) characterizes vessel motion (both path and velocity) by analyzing per-vessel and per-segment motion at the voxel level to determine the actual positions of the vessels at the prescribed target phase, and (2) compensates for any residual motion by correcting any motion shift of the voxels from the actual position at that phase using information from adjacent cardiac phases. Thus, the algorithm effectively compresses the temporal window for reconstruction. MCA corrects the different degrees of motion of each voxel in CCTA to reduce not only per-vessel, but also per-segment motion artifacts.

We reconstructed cardiac axial images using HSRA (SnapShot Segment; GE Healthcare), MSRA (SnapShot Burst Plus; GE Healthcare), and HRSA combined with MCA at the cardiac phases of 30% (systole) and 70% (diastole) of the R-R interval at each HR.

4. Image evaluation

On the workstation, 2 independent radiology technologists with more than 10 years of experience reviewed all the reconstructed axial CCTA image datasets at a window width of 900 HU and a window level of 300 HU in 9 segments of the coronary vessel models (Segments 5 to 13), as shown in Fig. 3, and graded the image quality by focusing on coronary motion artifacts according to the coronary segment. The image quality was semi-quantitatively ranked using a 5-point scale as previously described. A score of 5 signified no motion artifacts and clear delineation of the segment, 4 signified minor artifacts and mild blurring of the segment, 3 signified moderate artifacts and moderate blurring without structural discontinuity, 2 signified severe artifacts and doubling or discontinuity in the course of the seg-
ment, and 1 signified an image that could not be evaluated and vessel structures that could not be differentiated. Fig. 4 shows examples of the various image quality scores. An image quality of grade 3 or higher was considered sufficient for routine clinical diagnostic purposes and was defined as interpretable.

5. Statistical analysis

All continuous variables were expressed as the mean ± standard deviation. For the statistical analyses, we used statistical software (JMP 12; SAS Institute Inc., Cary, NC, USA). We averaged the image quality scores for all the segments from both readers and used the Kruskal-Wallis test to compare the scores among images obtained with HSRA, MSRA, and HSRA combined with MCA at each HR. When a given comparison demonstrated a significant difference, we used the Steel-Dwass test as a post-hoc test to compare the scores between any 2 reconstruction algorithms. P<0.05 was considered to be statistically significant.

In addition, we quantified the inter-reader agreement in the scores between the 2 readers using Cohen’s linear weighted k-statistics.

III. Results

Tables 1 and 2 and Fig. 5 summarize the average of the per-segment scores for image quality as determined by both readers for images obtained using HSRA, MSRA, and HSRA combined with MCA at each HR. The inter-reader agreement for the scoring of image quality was moderate (κ value = 0.48).

At 50 to 60 bpm, there were no significant differences in the average scores among the 3 algorithms for both cardiac phases (P > 0.05). At 70 to 100 bpm, the average scores obtained using MSRA were comparable or higher than those obtained using HSRA for both cardiac phases. At 70 to 100 bpm, the average scores obtained using HSRA combined with MCA were comparable or higher than the other 2 algorithms for both cardiac phases.

The average scores obtained using HSRA were less than 3 for all HRs during systole and at 70, 90, and 100 bpm during diastole. The average scores obtained using MSRA were higher than 3 for all HRs during diastole, but they were less than 3 for all HRs except 60 bpm during systole. Meanwhile, the average scores obtained using HSRA combined with MCA were higher than 3 for all HRs during diastole and for all HRs except 90 and 100 bpm during systole.

Table 3 summarizes the numbers of the interpretable segments (image quality score of 3 or higher) according to HR. Under most conditions, the number of the interpretable segments in HSRA with combined MCA was not less than that in MSRA.

IV. Discussion

Coronary motion artifacts reduce the reliability of diagnosis in CCTA. Coronary motion artifacts tend to be most pronounced at higher HRs because the temporal resolution of HSRA is insufficient to produce a motion-free image. To overcome this problem, MSRA retrospectively composes images from different cardiac cycles, which theoretically could improve the effective temporal resolution significantly. MSRA usually achieves a better temporal resolution than HSRA, in which the images are reconstructed from a single cardiac cycle. In clinical settings, misregistration can also occur when the coronary artery does not return to the same position for each cardiac cycle because of arrhythmia or a changing HR. This may blur the images, reducing the image quality. On the other hand, MCA is free from the risk of such misregistration and the effects of de-
Reconstructed axial CCTAs of the coronary vessel models illustrate the use of a semiquantitative 5-point scoring of image quality.

(a) No motion artifacts and clear delineation of the segment (score 5).
(b) Minor artifacts and mild blurring of the segment (score 4).
(c) Moderate artifacts and moderate blurring without structural discontinuity (score 3).
(d) Severe artifacts and doubling or discontinuity in the course of the segment (score 2).
(e) Image not evaluable and vessel structures not differentiable (score 1). Image quality graded 3 or higher was defined as interpretable.

| Table 1 | Image quality scores as evaluated by 2 readers during systole |
|---------|-------------------------------------------------------------|
| Reconstruction algorithm | 50 bpm | 60 bpm | 70 bpm | 80 bpm | 90 bpm | 100 bpm |
| HSRA | 2.6 ± 1.4 | 2.8 ± 1.2 | 2.0 ± 0.7 | 2.1 ± 0.9 | 1.3 ± 0.6 | 1.6 ± 0.6 |
| MSRA | 2.7 ± 1.2 | 3.4 ± 0.8 | 1.7 ± 0.8 | 2.7 ± 0.6 | 1.9 ± 1.0 | 2.6 ± 1.4 |
| HSRA with MCA | 3.4 ± 1.1 | 3.6 ± 1.3 | 3.4 ± 0.7 | 3.3 ± 0.8 | 2.7 ± 1.3 | 2.6 ± 1.0 |
| \( P \) value (among 3 algorithms) | 0.1008 | 0.1082 | <0.0001* | 0.0018* | 0.0107* |
| \( P \) value (HSRA versus MSRA) | NA | NA | 0.4632 | 0.0399† | 0.2101 | 0.0607 |
| \( P \) value (HSRA versus HSRA with MCA) | NA | NA | <0.0001† | 0.0007† | 0.0018† | 0.0093† |
| \( P \) value (MSRA versus HSRA with MCA) | NA | NA | <0.0001† | 0.0195† | 0.1108 | 0.9994 |

Data are the mean ± standard deviation. bpm: beats per minute, HSRA: half-scan reconstruction algorithm, MSRA: multi-sector reconstruction algorithm, MCA: motion correction algorithm, NA: not available.
*\( P \) value was less than 0.05 when compared among with HSRA, MSRA, and HSRA combined with MCA at the same heart rate.
†\( P \) value was less than 0.05 when compared between 2 reconstruction algorithms at the same heart rate.

| Table 2 | Image quality scores as evaluated by 2 readers during diastole |
|---------|-------------------------------------------------------------|
| Reconstruction algorithm | 50 bpm | 60 bpm | 70 bpm | 80 bpm | 90 bpm | 100 bpm |
| HSRA | 3.9 ± 1.2 | 3.9 ± 1.0 | 2.9 ± 0.9 | 3.4 ± 0.8 | 2.9 ± 1.0 | 2.6 ± 1.0 |
| MSRA | 3.9 ± 1.1 | 4.1 ± 0.6 | 3.1 ± 0.8 | 3.8 ± 0.7 | 3.8 ± 1.1 | 3.3 ± 1.0 |
| HSRA with MCA | 4.1 ± 1.0 | 4.5 ± 0.6 | 4.1 ± 0.6 | 3.7 ± 0.8 | 3.9 ± 0.8 | 3.2 ± 1.2 |
| \( P \) value (among 3 algorithms) | 0.8644 | 0.1127 | 0.0002* | 0.2428 | 0.0128* | 0.0823 |
| \( P \) value (HSRA versus MSRA) | NA | NA | 0.7993 | NA | 0.0445† | NA |
| \( P \) value (HSRA versus HSRA with MCA) | NA | NA | 0.0009† | NA | 0.0219† | NA |
| \( P \) value (MSRA versus HSRA with MCA) | NA | NA | 0.0012† | NA | 0.99998 | NA |

Data are the mean ± standard deviation. bpm: beats per minute, HSRA: half-scan reconstruction algorithm, MSRA: multi-sector reconstruction algorithm, MCA: motion correction algorithm, NA: not available.
*\( P \) value was less than 0.05 when compared among with HSRA, MSRA, and HSRA combined with MCA at the same heart rate.
†\( P \) value was less than 0.05 when compared between 2 reconstruction algorithms at the same heart rate.
creased temporal resolution.

There are two methods of ECG gating: prospective and retrospective ECG gating. In prospective ECG-gated scans, raw data acquisition is performed using an axial scan and half scan reconstruction is then performed. On the other hand, in a retrospective ECG-gated scan, a helical scan is used. In general, the image quality is better with an axial scan than with a helical scan if the effects of motion artifacts can be neglected. One reason is that the effective slice thickness of axial scans is smaller than that of helical scans, since the slice sensitivity profile of the former has steeper edges\textsuperscript{22}. In addition, axial scans are free from spiral artifacts, such as cone beam artifacts and rod artifacts\textsuperscript{22}.

In fact, higher diagnostic abilities of prospective ECG-gated axial scans have been reported for patients with HRs of less than 65 bpm, compared with retrospective ECG-gated helical scans\textsuperscript{23,24}. For patients with high HRs, retrospective ECG-gated helical scans are generally preferable to prospective ECG-gated axial scans, since the former has a better temporal resolution.

The current study using helical scans showed that MCA can reduce motion artifacts comparably or significantly better than MSRA, even for patients with high HRs. MCA is applicable not only to helical scan data, but also to axial scan data. As mentioned above, the quality of images with axial scans is superior to that obtainable using helical scans when motion artifacts can be neglected.

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### Table 3 Numbers of interpretable segments according to heart rate (HR)

| Phase | HR   | Numbers of interpretable segments |
|-------|------|-----------------------------------|
|       |      | HSRA     | MSRA     | HSRA with MCA |
| Systole | 50 bpm | 5 (56)  | 5 (56)  | 7 (78) |
|        | 60 bpm | 6 (67)  | 7 (78)  | 7 (78) |
|        | 70 bpm | 1 (11)  | 1 (11)  | 8 (89) |
|        | 80 bpm | 4 (44)  | 6 (67)  | 7 (78) |
|        | 90 bpm | 0 (0)   | 2 (22)  | 4 (44) |
|        | 100 bpm | 0 (0)   | 5 (56)  | 4 (44) |
| Diastole | 50 bpm | 7 (78)  | 8 (89)  | 8 (89) |
|        | 60 bpm | 8 (89)  | 9 (100) | 9 (100) |
|        | 70 bpm | 8 (89)  | 5 (56)  | 9 (100) |
|        | 80 bpm | 7 (78)  | 9 (100) | 8 (89) |
|        | 90 bpm | 5 (56)  | 8 (89)  | 9 (100) |
|        | 100 bpm | 5 (56)  | 7 (78)  | 6 (67) |

Data are numbers of interpretable segments (image quality score ≥ 3); percentages of interpretable segments are in parentheses (%). bpm: beats per minute, HSRA: half-scan reconstruction algorithm, MSRA: multi-sector reconstruction algorithm, MCA: motion correction algorithm.
be neglected, and MCA can reduce motion artifacts sufficiently even in the patients with HRs of 70–80 bpm. This situation could lead to the adaptation of prospective ECG-gated scans for patients with higher HRs, although prospective ECG-gated scans are currently only used for patients with an HR under 65 bpm. In fact, Li et al. suggested that prospective ECG-gated scans can be used at 65–75 bpm using MCA based on the results of MSRA and MCA studies in patients with 75 bpm or less\(^1\). Our phantom data suggests that prospective ECG-gated scans can be used even up to 80 bpm.

Radiation exposure is one of the problems associated with CCTA. Especially, MSRA tends to increase radiation exposure because it requires retrospectively ECG-gated scans made using a lower pitch\(^2\). On the other hand, prospective ECG-gated axial scans are capable of exposure reduction, and the radiation dose is reportedly reduced by 77%–88% for prospective ECG-gated axial scans, compared with retrospective ECG-gated helical scans\(^3, 26–28\). Therefore, the expanding adaptations of prospective ECG-gated scanning are useful not only for improving image quality but also for reducing the radiation dose.

In general, improving the rotation speed contributes to a reduction in coronary motion artifacts. In fact, Seitaro et al. reported that a 256-slice CT scanner with a rotation speed of 270 ms can reduce coronary motion artifacts in patients with an HR exceeding 60 bpm more effectively than a 64-slice CT scanner with a rotation speed of 420 ms\(^2\). Since MCA and a higher rotation speed reduce coronary motion artifacts based on different principles, the combination of a higher rotation speed and MCA should lead to further reductions in coronary motion artifacts.

Our study had several limitations. First, our phantom did not include models simulating the right ventricle and right coronary vessels. The influence of the HR on motion artifacts differs between right and left coronary vessels\(^7\). In this study, MCA reduced motion artifacts even at a high HR, so this technique is expected to be useful for reducing motion artifacts in the right coronary vessels. Second, our phantom did not reproduce the LV wall motion heterogeneity, although heterogeneous myocardial velocities exist within individual myocardial segments in clinical settings\(^2\). The heterogeneity of LV wall motion can disturb the accurate simulation of coronary motion artifacts. Third, the CT attenuation values of the LV model (250 HU) and the coronary vessel models (650 HU) were far from those used in clinical settings. Forth, we used only a single set of imaging parameters for tube voltage, tube current, pitch, and blending ratio of ASiR. These factors may affect the ability of MCA to correct for motion artifacts. Fifth, we analyzed the effect of HR on the occurrence of motion artifacts but not the effect of HR variability, which has been found to be potentially substantial\(^8\). In principle, HSRA using data for one cardiac cycle may be less affected by HR fluctuations than MSRA. Further experiments using various imaging parameters and phantom conditions, including HR variability and CT attenuation within coronary vessel models, are needed. Sixth, we only evaluated image quality and interpretability visually. Though often used in previous studies, visual scoring of image quality is subjective. However, we did not perform an appropriate quantitative evaluation of coronary motion artifacts in this phantom experiment. Finally, at least currently, our results are thought to be vendor-specific. To our knowledge, MCAs are not commercially or clinically available for use in CCTA with CT scanners from other vendors.

V. Conclusion

The use of a vendor-specific MCA resulted in a comparable or significant reduction in coronary motion artifacts in images obtained under HR conditions of 50 to 100 bpm, compared with those obtained using MSRA, in this phantom study.

Disclosure statement

The authors have no conflicts of interest to declare.

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