Phase contrast coronary blood velocity mapping with both high temporal and spatial resolution using triggered Golden Angle rotated Spiral k-t Sparse Parallel imaging (GASSP) with shifted binning

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Purpose: High temporal and spatial resolutions are required for coronary blood flow measures. Current spiral breath-hold phase contrast (PC) MRI at 3T focus on either high spatial or high temporal resolution. We propose a golden angle (GA) rotated Spiral k-t Sparse Parallel imaging (GASSP) sequence for both high spatial (0.8 mm) and high temporal (<21 ms) resolutions.

Methods: GASSP PC data are acquired in left anterior descending and right coronary arteries of eight healthy subjects. Binning of GA rotated spiral data into cardiac frames may lead to large k-space gaps. To reduce those gaps, the binning window is shifted and a triggered GA scheme that resets the rotation angle every heartbeat is proposed. The gap reductions are evaluated in simulations and all subjects. Peak systolic velocity (PSV), peak diastolic velocity (PDV), coronary blood flow rate, and vessel area are validated against two reference scans, and repeatability/reproducibility are determined.

Results: Shifted binning reduced the mean k-space gaps of the triggered GA scheme by 14°-22° in simulations and about 20° in vivo. The k-space gap across three cardiac frames was reduced with the triggered GA scheme compared to the standard GA scheme (35.3°± 3.6° vs. 43°± 13.7°, t-test \( P = .04 \)). PSV, PDV, flow rate, and area had high intra-scan repeatability (0.92 \( \leq \) intraclass correlation coefficient [ICC] \( \leq 0.99 \)), and inter-scan (0.78 \( \leq \) ICC \( \leq 0.91 \)) and intra-observer (0.91 \( \leq \) ICC \( \leq 0.98 \)) reproducibility.

Conclusion: GASSP enables single breath-hold coronary PC MRI with high temporal and spatial resolutions. Shifted binning and a triggered GA scheme reduce
1 | INTRODUCTION

Phase contrast (PC) velocity-encoded MRI\(^1\) enables blood velocity and flow measurements of the coronary arteries.\(^2,3\) These measurements are of clinical interest, because they enable the assessment of vascular physiology, to complement conventional anatomic lumen imaging. Among others, coronary PC MRI has been applied to assess coronary flow reserve,\(^4,7\) local coronary endothelial dysfunction,\(^8-10\) and more recently to measure the pressure gradients across a coronary artery stenosis.\(^11\)

Coronary PC MRI is challenging because the coronary arteries are small in diameter (<3-4 mm) and move due to cardiac contraction and respiration. High spatial resolution \(\Delta x \leq 1 \text{ mm}\) is needed for accurate assessment of the mean flow,\(^12\) which includes the determination of cross-sectional areas that may benefit from even smaller voxel sizes.\(^13\) Furthermore, high temporal resolution is needed due to fast cardiac motion and to resolve temporal detail, especially for the right coronary artery (RCA), which is twice as mobile as the left anterior descending artery (LAD).\(^14\) According to Marcus et al. the temporal resolution \(\tau\) should be \(\leq 58 \text{ ms}\) and \(\leq 23 \text{ ms}\) for LAD and RCA, respectively.\(^15\) Additionally, PC MRI requires two acquisitions with identical parameters except for the velocity encoding (VE) gradients to subtract out background phase.\(^1\) As a result, PC MRI is twice as time-consuming as anatomic cine scans, which is especially challenging when using breath-holding to suppress respiratory motion.

Initial coronary PC MRI utilized single breath-held Cartesian segmented spoiled gradient echo techniques enabling acquisitions with \(\Delta x\) of 1.4-1.6 mm and a \(\tau\) of 120-160 ms.\(^2,3\) Keegan et al. pioneered the use of highly efficient spiral readouts to improve the temporal resolution.\(^16-18\) More recently, 3T scanners have been applied to take advantage of the increased signal-to-noise ratio (SNR) enabling higher spatial resolution as well.\(^8,18,19\) Nevertheless, state-of-the-art breath-held spiral techniques at 3T focus on either high spatial resolution\(^8,19\) or high temporal resolution.\(^18\) Brandts et al. use \(\Delta x = 0.8 \text{ mm}\) and show accurate and reproducible measurement of RCA peak velocity and flow volumes.\(^19\) To achieve such a small voxel size they use a long, 26-ms spiral readout window, leading to a temporal resolution of 33 ms and, hence, potentially motion blurring. Moreover, the long readouts may lead to off-resonance induced blurring.\(^20\) Keegan et al. trade spatial resolution for high temporal resolution \((\tau = 20 \text{ ms})\) to assess temporal patterns of coronary blood flow throughout the cardiac cycle.\(^18\) To achieve such a high temporal resolution, they shorten the spiral readout to 12 ms, forcing a rather large voxel size \(\Delta x\) of 1.4 mm.

To achieve both high temporal and high spatial resolution in a single breath-hold, the data acquisition needs to be accelerated. Acquisitions with non-Cartesian trajectories can be accelerated with parallel imaging (PI)\(^21,22\) and additionally with compressed sensing (CS).\(^23\) In cine MRI, the temporal domain is transform sparse and suitable for CS reconstruction.\(^24,25\) In addition to sparsity, incoherent undersampling artifacts are also essential for CS reconstructions.\(^23\) Incoherence in temporal domain can be achieved using golden angle (GA) rotation of radial or spiral readouts.\(^26,27\) \(k\)-t sparse SENSE combines PI and CS in temporal dimension and acceleration factors of 8 for structural cines\(^28\) or 6 for phase-contrast cine\(^29\) have been shown using Cartesian sampling. The combination of \(k\)-t sparse SENSE and GA rotation of either radial or spiral sampling is feasible in cine MRI as well.\(^30-32\)

However, while the acquisition of consecutively GA-rotated radial spokes or spiral arms offers flexibility in terms of temporal resolution versus spatial resolution in dynamic imaging with uniform coverage of \(k\)-space,\(^26\) when binning data into non-consecutively acquired cardiac frames, the \(k\)-space coverage may not be uniform.\(^33-36\) This non-uniformity creates gaps in azimuthal distribution and strongly depends on the number of cardiac frames and the heartrate (HR), which is unpredictable especially for stress studies. To achieve higher \(k\)-space uniformity in cine MRI, Han et al. proposed the use of a segmented golden ratio approach\(^30\) but requires prospective triggering and prevents retrospective data rejection due to motion or arrhythmias.

The objective of this work is to combine \(k\)-t sparse SENSE with GA rotated spiral readouts to achieve single breath-hold coronary PC MRI with both high spatial (0.8 mm) and high temporal (~21 ms) resolutions. GA Spiral \(k\)-t Sparse Parallel imaging (GASSP)\(^37\) is combined with interleaved two-sided VE\(^38\) and electrocardiogram (ECG) based retrospective cardiac gating.\(^39\) To reduce \(k\)-space gaps of binned cine data, two new methods are tested.\(^40\) First, the binning window is shifted by up to the duration of a cardiac frame, and the shift leading to the smallest \(k\)-space gap is picked for image reconstruction. Second, a triggered GA scheme is introduced that applies a new rotation angle to consecutive arms and the
standard GA at the beginning of each heartbeat using prospective ECG-triggering. Compared to the segmented golden ratio approach, this triggered GA scheme is suitable for retrospective rejection of data corrupted by motion or other reasons. Monte-Carlo numerical simulations are performed to determine the optimal rotation angles in the triggered GA scheme, and to quantitatively evaluate the performance of the proposed methods to reduce k-space gaps. The proposed methods are tested at 3T in healthy human subjects, validated against published methods, and intra-scan repeatability, inter-scan reproducibility, and intra-observer reproducibility are determined.

2 METHODS

2.1 Coronary PC MRI with standard GA scheme

Coronary PC cine with both high temporal and high spatial resolutions was implemented by combining GASSP with interleaved two-sided VE, referred to here as standard GA scheme. Spiral arms were acquired in pairs of positive and negative VE with the same azimuthal angle, and then rotated by GA for the next pair (top row of Figure 1A). The GA was $137.508^\circ$. Spiral arm pairs were continuously acquired while the ECG signal was recorded for retrospective data binning.

To achieve high temporal resolution and to mitigate blurring effects due to B0 inhomogeneity the spiral readout window was set to $\sim 14$ ms leading to a repetition time (TR), and hence minimal $\tau$, of $<21$ ms. The spatial resolution was set to 0.8 mm for a $350 \times 350$ mm$^2$ field of view (FOV). This would require 34 spiral arms to obtain a fully sampled k-space, that is, 68 heartbeats to acquire both VE datasets, which is too long for a single breath-hold scan. Therefore, the spiral trajectory was undersampled with a variable density pattern. A numerical algorithm was used to optimally design the spiral trajectory so that the central 15% of k-space were fully sampled with 10 arms similar to Ref. 32. The outer 70% of k-space were uniformly undersampled by a factor of 4.6, and the undersampling factor was linearly increased in between. A total of 500 spiral arm pairs (1000 arms) were acquired, leading to a scan time of $\sim 21$ s. Maximum encoded velocity ($V_{enc}$) was set to $\pm 35$ cm/s. Other parameters of the two-dimensional (2D) gradient echo sequence were echo

![Figure 1](https://example.com/figure1.png)

**FIGURE 1** Rotation angle and VE directions of the triggered GA scheme compared to the standard GA scheme (A) and shifted binning process (B). The VE direction in the standard GA scheme is toggled each TR while it is toggled every heartbeat in the triggered GA scheme. A, the rotation angle in the standard GA scheme is $GA = 137.508^\circ$ throughout, whereas in the triggered GA scheme it is $\theta = 83.8^\circ$ except at the beginning of each heartbeat, when the rotation is set to $GA/2 = 68.754^\circ$ relative to the first spiral arm of the previous heartbeat. B, both schemes are subject to shifted binning process. Retrospective binning with a fixed temporal resolution $\tau = TR$ was used and non-linear stretching was applied due to RR-interval variations. Bins were shifted N times by temporal resolution $\tau$ divided by N ($N = 20$) and for each shift the mean-max gap was determined. The shift resulting in the smallest mean-max gap was selected for image reconstruction.
time (TE) = 3.3-3.5 ms, 1-2-1 binomial spectral spatial water excitation to suppress fat, flip angle (FA) = 20°, no RF spoiling, and slice thickness (ST) = 8 mm.

2.2 Retrospective gating with shifted binning

The continuously acquired spiral data were binned into cardiac frames based on the ECG signal using non-linear stretching. Positive and negative VE data were binned separately. The number of cardiac frames was determined by dividing the acquisition’s average RR-interval, the time between two consecutive R-waves, by the required temporal resolution τ (=TR < 21 ms), and was, therefore, subject specific. As a result, subjects with lower HR had more cardiac frames and less data assigned to each cardiac frame (i.e., higher undersampling factor). Assuming uniformly sampled k-space, the undersampling factor was around 3.0 for HR = 80, but increased to 3.9 and 4.7 for HR = 60 and 50, respectively, but was higher in practice because binning leads to non-uniform undersampling, and hence, gaps in k-space.

In this work, we tested two methods to reduce the k-space gaps. The first method, referred to as shifted binning, shifted the binning windows by up to the duration of τ, trying N = 20 equally spaced shifts and selecting the shift that minimized the largest gaps (Figure 1B). For this, the k-space gaps after binning were quantitatively evaluated after binning the azimuthal angle of each spiral readout to determine first the maximal angular spacing between the sorted spiral arms (max-gap) in each bin (for each cardiac frame and VE direction), and, second the average of the max-gaps (mean-max-gap) over all cardiac frames and both VE directions. This was repeated N times, each time shifting the binning windows by τ/N, leading to N different mean-max-gaps (Figure 1B). The shift resulting in the lowest mean-max-gap was eventually chosen to bin the data, and resulting images were labeled with “best shift.” For comparison, the shift associated with the largest mean-max-gap was applied as well, and labeled with “worst shift.”

2.3 Coronary PC MRI with triggered GA scheme

In addition to shifted binning, the triggered GA scheme was the second method to improve k-space uniformity. The triggered GA scheme uses two different rotation angles and prospective ECG-triggering to reset the rotation angle and to toggle the VE direction at the beginning of each heartbeat. As shown in the bottom row of Figure 1A, spiral arms with the same VE direction, either positive or negative, were continuously acquired within one heartbeat. The rotation angle between consecutive arms within a heartbeat was modified to θ = 83.8°, which was selected based on the numerical simulations below. When an R-wave was detected, the rotation angle was reset so that the first spiral arm in a new heartbeat was rotated by half the GA (68.754°) compared to the first spiral arm of the previous heartbeat. The following arms were rotated again by θ. Concurrently with the angle reset, the VE direction was inverted to toggle its direction every heartbeat. Hence, the first spiral arms of heartbeats with the same VE direction were rotated by the GA.

A total of 1000 spiral arms were acquired for the triggered GA scheme, but due to varying RR-intervals, the total number of spiral arms in each VE direction was usually slightly different than 500. All other acquisition parameters of the triggered GA scheme were identical to standard GA scheme including the shifted binning gating process.

2.4 Simulation of k-space gap after binning

A Monte-Carlo numerical simulation was performed to determine the optimal rotation angle θ in the triggered GA scheme, and to quantitatively evaluate the performance of the shifted binning method and the triggered GA scheme. To quantify k-space uniformity, both mean-max-gap and the largest max-gap (max-max-gap) were determined and called “one-frame mean-max-gap” and “one-frame max-max-gap,” respectively. Additionally, because the reconstruction applies temporal sparsity constraints, k-space gaps in one cardiac frame can be compensated for during the reconstruction by spiral arms from neighboring frames. For this, spiral arms from consecutive frames should be located at the center of the largest gap, which can be achieved if θ ≈ max-max-gap / 2. Therefore, θ was chosen to minimize the absolute of (max-max-gap / 2) − θ. To test if the triggered GA scheme indeed reduces gaps that overlap in consecutive cardiac frames, the mean-max-gap and max-max-gap were also calculated after combining the arms of three consecutive frames, and called “three-frame mean-max-gap” and “three-frame max-max-gap,” respectively.

The simulation covers a wide range of HR from 40 to 100 beats/min with a step size of 1 beat/min. TR = 20.65 ms was used, leading to a duration of 20.65 s for 1000 spiral arms. For each HR, a series of RR-intervals were simulated to cover the 20.65 s duration using a normal distribution with a mean value of the given RR-interval and a SD of 50 ms. A series of spiral rotation angles were generated using standard or triggered GA scheme. These angles were then binned with shifted binning, and mean-max-gap and max-max-gap determined. The simulation was repeated 200 times for each HR and the corresponding values averaged. For the triggered GA scheme, these simulations were repeated for θ ranging from
1°-160° in steps of 3.6°. For the chosen θ (83.8°), the best shift was compared to the worst shift, and the triggered GA scheme to the standard GA scheme.

2.5 | In vivo experiments and reference acquisitions

All human studies were approved by the Johns Hopkins School of Medicine Institutional Review Board and informed, written consent was obtained from all study subjects. In vivo studies were conducted on eight healthy human subjects with no history of heart disease (37±7 y old, 1 female). MR studies were carried out on a 3T scanner (Achieva, Philips Healthcare, Best, The Netherlands) using a 32-channel cardiac receive coil and subject-specific shimming of the excitation radiofrequency field.

All 2D coronary PC scans were acquired within a single end-expiratory breath-hold and planned orthogonally to either proximal or mid segments of the coronary arteries that was straight for at least 20 mm. To facilitate careful planning of the image orientations, targeted 3D DIXON scans were performed to provide water and fat images of both RCA and LAD in 3D double oblique views with TR/TE1/TE2 = 5.6-6.1/1.9-2.2/3.6-3.9 ms, FOV = 200 × 200 × 32 mm³, acquired voxel size = 1 × 0.85 × 3 mm³, and scan duration of ~1.5-6 min depending on the efficiency of the respiratory navigator. These targeted views were defined using the three-point planning tool on the target vessel of a whole heart coronary angiography scan with intermediate resolution acquired in ~1 min. All scans for image planning were gated by respiratory navigator-echo and ECG-triggered to mid-diastole.

The standard and triggered GA scheme were compared to existing methods with either high spatial but low temporal resolution (the “high-spatial” sequence), or high temporal but low spatial resolution (the “high-temporal” sequence). Here, V is the velocity and arg is the argument of complex number.

The high-spatial sequence was ECG-triggered and acquired a fixed (depending on the patient’s HR) number of arms with the same spiral rotation angle in each heartbeat. VE direction was inverted every heartbeat, and 10 spiral arms were acquired for each VE, leading to a total scan time of 20 heartbeats. High in plane spatial resolution of 0.8 mm led to long spiral readouts of ~33 ms and long TR of ~40 ms. Other acquisition parameters were identical to standard GA sequence except for the FOV = 250 × 250 mm². Because of ECG-triggering, the last end diastolic cardiac frames were missed as the sequence was interrupted to wait for and detect the next R-wave. Images and velocity maps were generated on the vendor-provided platform.

In the high-temporal sequence, VE direction was also inverted every heartbeat, with positive and negative arms acquired in 13 heartbeats each. With a high temporal resolution of TR = ~20 ms (spiral readout = ~13 ms), the spatial resolution was sacrificed to fully cover the k-space with 13 arms. Retrospective ECG-gating was used to bin the data into cardiac frames. Other acquisition parameters were identical to the standard GA sequence. Images were reconstructed off-line.

To determine inter-scan reproducibility of the triggered GA sequence, the subjects were removed from the scanner and given a 10-min break, before repeating the examination. For four of the subjects, the triggered GA sequences were also repeated immediately without repositioning to evaluate the intra-scan repeatability.

2.6 | Image reconstruction and flow analysis

Image reconstruction and off-resonance deblurring (except for the high-spatial sequence) were implemented in the graphical programming interface. Retrospective binning of k-space was performed with shifted binning and data were reconstructed by GASSP, minimizing parallel imaging data consistency and temporal total variation (TV) sparsity constraints. In this work, the original GASSP reconstruction for structural CINE images was adapted for PC by joint reconstruction of both VE steps with an additional TV sparsity constraint along the flow dimension. The flow sparsity constraint was only applied to the magnitude of the signal to preserve the phase information of each VE direction. The weight was set to 0.03 for the temporal sparsity constraint, and 0.015 for the additional flow sparsity constraint.

Flow-compensated images were generated by geometric mean, by which positive and negative VE phases were canceled by complex multiplication. Flow-compensated images were used to determine the deblurring frequency at the location of the coronary artery, which was subsequently used to deblur the original two VE images \( I_+ \) and \( I_- \). The deblurred VE images were used to compute deblurred flow-compensated images for vessel segmentation and deblurred velocity maps for velocity quantification. The velocity maps were computed by phase-difference with pixel-wise conjugated multiplication.

\[
V = \frac{V_{enc}}{\pi} \arg \left(I_+ I_-^* \right). \tag{1}
\]

Here, V is the velocity and arg is the argument of complex values.

A semi-automatic flow analysis software was implemented in MATLAB (Mathworks, Natick, MA, USA) that included vessel segmentation and correction for background phase and through-plane motion. All images were Fourier-interpolated to 0.1 mm spatial resolution for more precise segmentation.
First, the vessel area $A_v$ was semi-automatically computed from three consecutive end-diastolic frames by full-width-half-maximum thresholding. Second, the vessel in each cardiac frame was segmented with a circular region of interest (ROI) with the same area $A_v$, by clicking the center of the interpolated vessel as recommended by Brandts et al. Another circular ROI, the myocardial ROI with the same area $A_v$, was manually selected on surrounding myocardial tissue to correct for background phase and through-plane motion, and the mean velocity in myocardial ROI was subtracted from the velocity in the vessel ROI.

Eleven metrics were determined. The vessel area $A_v$ was determined in the first step of the segmentation. Eight velocity metrics were measured: peak systolic and diastolic velocity (PSV, PDV, respectively) together with the trigger delays to those peaks (TSV, TDV); and those velocity metrics were determined in two ways, mean (mean) and maximum (max) velocity spatially within the vessel ROI. And two flow metrics were determined in two ways, mean (mean) and maximum (PSV, PDV, respectively) together with the trigger velocity metrics were measured: peak systolic and diastolic velocity in $[\text{mL/min}]$.

All datasets were tested for normal distribution using the Kolmogorov-Smirnov test. To limit the number of statistical tests, only four of the metrics (mean PSV, mean PDV, flow rate, and $A_v$) acquired with the proposed triggered GA scheme were compared to both high-spatial and high-temporal sequences using paired Student’s t-test with a modified Bonferroni correction to account for the eight tests. Additionally, Bland-Altman analysis was performed for the same four metrics.

The intra-scan repeatability, inter-scan reproducibility, and intra-observer reproducibility were determined by the intraclass correlation coefficient (ICC) for absolute agreement. For intra-scan repeatability, Bland-Altman analysis was performed as well. To test the intra-observer agreement, the same observer repeated the flow analysis at least 1 mo after the first observation, whereas the two analyses of the intra-scan repeatability and inter-scan reproducibility were performed on the same day. A $P$-value smaller than .05 was considered significant for all tests.

3 | RESULTS

The simulation results demonstrate the dependence between rotation angle $\theta$ of the triggered GA scheme, RR-interval, and mean-max-gap and max-max-gap, respectively (Figure 2). The one-frame mean-max-gap increases as functions of both RR-interval and $\theta$ (Figure 2A). A similar behavior but with larger gaps is seen for one-frame max-max-gap (Figure 2B). The lone-frame max-max-gap / $2 - \theta$ averaged over the range of simulated HR in Figure 2C is minimized for $\theta = 83.8^\circ$. Therefore, the rotation angle $\theta$ in the triggered GA scheme was chosen to be 83.8°. The three-frame mean-max-gap and three-frame max-max-gap at $\theta = 83.8^\circ$ are smaller than even the best one-frame gaps, and are also smaller than the three-frame gaps for smaller $\theta$ (Figure 2D,E).

Figure 3A,B illustrate the simulated one-frame and three-frame mean-max-gaps, respectively, as a function of the mean RR-interval of both standard and triggered GA schemes with both best and worst shift. As the RR-intervals increase, larger gaps are observed. Using shifted binning, the one-frame mean-max-gaps of the best shift dropped about 14°-22° for triggered GA and 14°-27° for standard GA scheme (Figure 3A). The one-frame and three-frame mean-max-gaps of the triggered GA scheme increase approximately linearly with the mean RR-interval (608-1503 ms), whereas the ones of the standard GA scheme exhibit an additional modulation pattern. The shifted binning has little effect on the three-frame mean-max-gaps (Figure 3B). The three-frame mean-max-gaps of the standard GA scheme is larger than those of the triggered GA scheme for most RR-intervals and the modulation pattern is even stronger than the one-frame mean-max-gaps.

Complete datasets of both RCA and LAD were obtained from all 8 subjects. However, the RCA of one subject could not be visualized throughout the cardiac cycle (Supporting Information Figure S1, which is available online), leading to no data for that subject. Off-line image reconstruction took ~24 min per dataset. All metrics among the 15 vessels were normally distributed.

Figure 3C,D illustrate the one-frame and three-frame mean-max-gaps, respectively, measured in human subjects. In Figure 3C, the one-frame mean-max-gaps with best shift are smaller compared to those with worst shift. The mean-max-gaps averaged among all cases using the triggered GA scheme ($73.6^\circ \pm 13.3^\circ$, best shift) are significantly smaller (t-test $P = .03$) compared to those of the standard GA scheme ($80.6^\circ \pm 18.0^\circ$, best shift). When combining 3 consecutive frames (Figure 3D), some large gaps were observed in the scatter plot of the standard GA scheme at RR-interval of around 872 ms and 916 ms, which is consistent with the simulation results in Figure 3B. As a result, the three-frame average mean-max-gaps of the standard GA scheme ($42.8^\circ \pm 13.7^\circ$, best shift) is significantly larger (t-test $P = .04$) than the one of the triggered GA scheme ($35.3^\circ \pm 3.6^\circ$, best shift). The difference is especially large for the SD.

Figure 4 shows an example dataset acquired with both GA schemes and worst and best shifts. Using best shift, the mean-max-gaps were reduced by 21° and 19° for standard and triggered GA scheme, respectively. With best shift, the triggered GA scheme has lower mean-max-gaps ($-5.5^\circ$) compared to the standard GA scheme. Swirling artifacts were observed in the right ventricle and the left ventricular wall (yellow arrows) in the standard GA images, for both VE and flow-compensated magnitudes, but not in the triggered GA scheme. Additional examples comparing standard and
FIGURE 2 Surface plots of mean-max-gap (A, D) and max-max-gap (B, E) of the proposed triggered GA scheme with best binning shift were determined for a range of RR-intervals and rotation angles θ using Monte Carlo simulations. The gaps were determined for a single cardiac frame (A, B) or for three consecutive frames (D, E). C, the minimum of |max-max-gaps/2-θ| averaged over all HR is used to determine the rotation angle θ = 83.8° that ensures that neighboring cardiac frames have k-space data in the center of the largest gap, which is beneficial for reconstruction with temporal sparsity constraints. The purple line in the surface plots indicates the chosen θ.
triggered GA schemes are provided as videos for both RCA (Supporting Information Video S1) and LAD (Supporting Information Video S2) demonstrating reduced artifacts with the triggered GA scheme. Supporting Information Video S3 shows magnitude and velocity cine videos acquired with the proposed triggered GA scheme of all 15 coronary arteries.

Figure 5 shows an example dataset of a mid-diastolic frame acquired and reconstructed with all techniques. The flow compensated image of the high-spatial sequence demonstrated in Figure 5A suffers from aliasing artifacts (yellow arrow). The high-temporal sequence, in contrast, shows pixelated pattern on the RCA due to low spatial resolution. The velocity map of the triggered GA method with best binning shift (red frame) has excellent quality across the heart and appears to have less noise/artifacts compared to other methods though the differences are only subtle. Flow rate curves in Figure 5B,C agree well across sequences and shifts. Of note, in this case though not on average (Table 1), the high-spatial sequence underestimates PSV by about half because of the lower temporal resolution.

Figure 6 demonstrates the agreement of flow rate curves of the repeated scans using the triggered GA scheme with best binning shift of all 15 coronary arteries. The eight intra-scan repeats all show excellent agreement with the initial scan. The inter-scan repeats still agreed well in most cases.

The velocity and flow metrics for all four techniques are shown in Table 1, separately averaged among seven RCA and eight LAD scans. After combining all 15 vessels, Supporting Information Figure S2 shows Bland-Altman analysis of mean PSV, mean PDV, flow rate, and Av comparing the proposed triggered GA scheme with the high-spatial and high-temporal reference methods. There is a high agreement for velocity and flow metrics, validating the new technique. Bonferroni corrected t-tests show that the area measured with triggered GA scheme is significantly larger than with both high-spatial and high-temporal sequences, and PDV is significantly lower compared to high-temporal sequence.

Table 2 demonstrates the results of intra-scan repeatability and inter-scan reproducibility. Spatial max PDV, spatial mean PSV and PDV, flow metrics and area are highly correlated (ICC ≥ 0.97, P < .01) for intra-scan repeats. Spatial max PSV is also well correlated (ICC = 0.92, P < .01). TSV and TDV are not correlated (−0.12 ≤ ICC ≤ 0.35, .16 < P < .62). Figure 7 demonstrates the high intra-scan repeatability.
FIGURE 4 Representative LAD PC images of a 39-year-old male showing gap angle and images artifact reduction by applying triggered GA scheme and shifted binning. The max-gaps of the spiral arms are visualized by two black arms form the origin to the end of the corresponding spiral arms, with the angle labeled in degrees at the left top corner of each k-space. The LAD (red arrow) is visualized in cross-sectional view above the left ventricle (LV). Zoomed insets of the LAD are shown at the right-bottom corner in each image. Artifacts in the right ventricle (RV) and LV wall (yellow arrows) in the images acquired with the standard GA scheme are substantially reduced in the proposed triggered GA scheme.

FIGURE 5 Example PC data of the RCA (red arrow) of a 20-year-old male acquired and reconstructed with all the different techniques applied in this work. Magnitude of flow compensated images and velocity maps of a mid-diastolic cardiac frame are shown in (A) and flow rate curves in (B-C). Aliasing artifacts (yellow arrow) are seen in the magnitude images of the high-spatial sequence. Images acquired with the proposed triggered GA method with best binning shift and combined GASSP reconstruction are highlighted with a red frame. The flow rate curves from the four different acquisition sequences (B) and using best and worst binning shift (C), generally agree well with each other. RA, right atrium and RV, right ventricle.
of mean PSV, mean PDV, flow rate, and \( A_v \) metrics using Bland-Altman analysis. For inter-scan reproducibility, the correlation of flow rate, area, PSV and PDV were still high and statistically significant (0.78 ≤ ICC ≤ 0.91, \( P < .01 \)).

The intra-observer reproducibility is demonstrated in Table 3. Area analysis correlation (ICC = 0.98, \( P < .01 \)) is very high. Also the flow (ICC = 0.91, \( P < .01 \)), flow rate (ICC = 0.94, \( P < .01 \)), and PSV and PDV (ICC ≥ 0.94, \( P < .01 \)) correlate very well. The correlation of TDV and TSV are lower, especially for TSV, but are still significant \( P \leq .01 \).

### 4 | DISCUSSION

The proposed triggered GA method combines GASSP and shifted binning to enable single breath-hold coronary PC MRI with both high spatial (0.8 mm) and high temporal (<21 ms) resolutions. Feasibility of the technique has been demonstrated in eight healthy subjects leading to high image quality and highly reproducible quantitative coronary flow measures. The method was validated against two previously published methods with either high spatial or high temporal resolution, and the flow rate and PSV agreed well.

However, the vessel areas were measured larger with the proposed method than either of the other two, and PDV was measured lower compared to the high-temporal sequence. This is supported by the agreement in the flow

| RCA (n = 7) | High-spatial | High-temporal | Standard GA | Triggered GA |
|-----------|--------------|---------------|-------------|--------------|
| PSV max (cm/s) | 21.2 ± 5.4 | 18.8 ± 4.7 | 18.9 ± 4.2 | 19.3 ± 3.8 |
| TSV max (ms) | 87.9 ± 40.0 | 65.9 ± 9.7 | 51.2 ± 20.6 | 69.8 ± 9.3 |
| PDV max (cm/s) | 23.7 ± 5.1 | 20.5 ± 4.5 | 19.8 ± 5.8 | 19.0 ± 5.9 |
| TDV max (ms) | 400.1 ± 35.5 | 398.1 ± 42.0 | 386.8 ± 29.0 | 394.5 ± 33.4 |
| PSV mean (cm/s) | 15.7 ± 5.4 | 15.8 ± 3.8 | 14.5 ± 4.0 | 14.4 ± 3.4 |
| TSV mean (ms) | 76.4 ± 19.4 | 74.4 ± 29.9 | 51.2 ± 20.6 | 69.8 ± 9.3 |
| PDV mean (cm/s) | 15.8 ± 4.8 | 17.8 ± 4.1 | 14.6 ± 5.4 | 14.3 ± 5.2 |
| TDV mean (ms) | 416.9 ± 57.0 | 395.2 ± 35.3 | 381.0 ± 41.7 | 391.5 ± 34.0 |
| flow (ml/cycle) | 0.69 ± 0.18 | 0.76 ± 0.13 | 0.75 ± 0.26 | 0.82 ± 0.18 |
| flow rate (ml/min) | 48.8 ± 16.4 | 52.8 ± 12.7 | 52.5 ± 19.3 | 55.7 ± 15.4 |
| area (mm\(^2\)) | 9.1 ± 3.0 | 9.1 ± 1.9 | 11.6 ± 4.3 | 12.4 ± 3.7 |

| LAD (n = 8) | High-spatial | High-temporal | Standard GA | Triggered GA |
|-----------|--------------|---------------|-------------|--------------|
| PSV max (cm/s) | 6.3 ± 4.1 | 5.0 ± 5.4 | 9.1 ± 3.6 | 7.8 ± 6.8 |
| TSV max (ms) | 137.8 ± 72.9 | 130.1 ± 65.9 | 118.5 ± 59.5 | 106.6 ± 84.4 |
| PDV max (cm/s) | 27.1 ± 8.1 | 22.5 ± 5.4 | 23.2 ± 6.4 | 21.4 ± 5.5 |
| TDV max (ms) | 451.4 ± 57.9 | 435.2 ± 56.3 | 432.5 ± 55.4 | 428.4 ± 51.5 |
| PSV mean (cm/s) | 2.5 ± 2.2 | 3.9 ± 2.7 | 3.8 ± 1.6 | 5.0 ± 4.1 |
| TSV mean (ms) | 142.8 ± 42.0 | 105.1 ± 49.9 | 115.8 ± 66.9 | 96.2 ± 69.9 |
| PDV mean (cm/s) | 16.7 ± 6.1 | 17.4 ± 4.4 | 15.4 ± 4.6 | 14.0 ± 3.7 |
| TDV mean (ms) | 436.4 ± 41.0 | 437.7 ± 47.2 | 440.4 ± 28.7 | 438.7 ± 45.2 |
| flow (ml/cycle) | 0.41 ± 0.16 | 0.44 ± 0.15 | 0.44 ± 0.12 | 0.48 ± 0.16 |
| flow rate (ml/min) | 28.3 ± 12.0 | 31.1 ± 14.4 | 29.9 ± 9.7 | 33.8 ± 14.3 |
| area (mm\(^2\)) | 8.4 ± 2.3 | 9.1 ± 2.5 | 10.1 ± 2.2 | 11.1 ± 3.5 |

Note: Flow velocity metrics are in mean ± SD.

Abbreviations: max = maximum velocity in vessel ROI, mean = average velocity in vessel ROI.

Both standard GA and triggered GA use the best binning shift.

*Due to prospective triggering used in high-spatial sequences, flow data were calculated based on incomplete cardiac cycle.

| TABLE 1 | Coronary flow velocity metrics acquired with reference and proposed methods |

The similarly small area measured with the high-spatial sequence was not expected, and this may be due to remaining image blurring as it was not possible to deblur the images reconstructed and processed on the vendor platform. Lower PDV compared to high-temporal sequence may be due to the aforementioned area underestimation of the high-temporal sequence. This is supported by the agreement in the flow
metric where both vessel area and measured velocities are combined. Additionally, the presented values agree well with those in the original publications that presented the reference methods.18,19

Intra-scan repeatability, inter-scan reproducibility, and intra-observer reproducibility were determined (Tables 1-3). Intra-scan repeatability of coronary flow rate, PSV and PDV (both using spatial mean) were highest with ICC = 0.99, comparing favorably to the high-temporal method that reported ICCs of 0.975, 0.985, and 0.97, respectively.18 Intra-scan repeatability is important when coronary flow/velocity are measured before and during an intervention such as when assessing coronary flow reserve or coronary endothelial function. Inter-scan reproducibility was determined after the participant was given

**FIGURE 6** Flow rate curves of the repeated scans acquired with the triggered GA scheme and reconstructed with the best binning shift of 7 RCA and 8 LAD cases. All 15 cases were repeated after repositioning, while 4 RCA and 4 LAD cases were repeated without repositioning.

**TABLE 2** Intra-scan repeatability and inter-scan reproducibility of coronary flow velocity metrics using proposed triggered GA scheme with shifted binning

|                      | Intra-scan repeatability (n = 8) | Inter-scan reproducibility (n = 15) |
|----------------------|----------------------------------|-------------------------------------|
|                      | First scan | Repeat | ICC       | P       | First scan | Repeat | ICC       | P       |
| PSV max (cm/s)       | 11.8 ± 9.8 | 13.6 ± 7.7 | 0.92 | <.01 | 13.1 ± 8.0 | 12.6 ± 8.0 | 0.88 | <.01 |
| TSV max (ms)         | 79.7 ± 76.9 | 53.0 ± 10.8 | −0.11 | .609 | 89.4 ± 62.9 | 89.6 ± 65.8 | 0.52 | .018 |
| PDV max (cm/s)       | 22.3 ± 5.7 | 21.6 ± 5.4 | 0.98 | <.01 | 20.3 ± 5.6 | 20.9 ± 5.8 | 0.81 | <.01 |
| TDV max (ms)         | 394.8 ± 35.7 | 398.6 ± 29.5 | 0.35 | .164 | 412.6 ± 45.9 | 437.7 ± 98.5 | 0.41 | .054 |
| PSV mean (cm/s)      | 8.9 ± 7.2 | 9.5 ± 7.2 | 0.99 | <.01 | 9.4 ± 6.1 | 8.5 ± 6.8 | 0.84 | <.01 |
| TSV mean (ms)        | 84.7 ± 66.5 | 53.0 ± 10.8 | −0.12 | .619 | 83.9 ± 51.6 | 77.5 ± 43.8 | 0.66 | <.01 |
| PDV mean (cm/s)      | 15.3 ± 4.8 | 15.2 ± 4.7 | 0.99 | <.01 | 14.2 ± 4.3 | 14.4 ± 4.5 | 0.78 | <.01 |
| TDV mean (ms)        | 400.0 ± 35.0 | 393.5 ± 25.0 | 0.33 | .176 | 416.7 ± 45.9 | 440.4 ± 95.2 | 0.36 | .079 |
| flow (ml/cycle)      | 0.65 ± 0.29 | 0.65 ± 0.29 | 0.98 | <.01 | 0.64 ± 0.24 | 0.58 ± 0.24 | 0.86 | <.01 |
| flow rate (ml/min)   | 48.0 ± 22.6 | 48.2 ± 21.5 | 0.99 | <.01 | 44.0 ± 18.2 | 40.6 ± 18.8 | 0.91 | <.01 |
| area (mm²)           | 12.2 ± 3.7 | 12.2 ± 4.1 | 0.97 | <.01 | 11.7 ± 3.5 | 11.1 ± 3.1 | 0.85 | <.01 |

Note: Flow velocity metrics are in mean ± SD.

Abbreviations: max = maximum velocity in vessel ROI, mean = average velocity in vessel ROI, ICC = intraclass correlation coefficients for absolute agreement.
a 10-minute break outside of the scanner, and coronary flow rate had the highest ICC (ICC = 0.91), while PSV and PDV correlated well with ICC from 0.78 to 0.88; similar to the high-spatial method that reported RCA ICCs of 0.89, 0.88, and 0.84, respectively. Intra-observer reproducibility was also very high with ICC ≥ 0.94 for cross-sectional area, PSV and PDV, and coronary flow rate. Interestingly, intra-scan repeatability was higher than the intra-observer reproducibility. This is likely because the intra-scan analysis was conducted on the same day allowing the operator to remember what cardiac frames were selected to determine the area, while the intra-observer analysis was conducted after at least 30 days.

Binning of GA rotated data into cardiac frames may cause large gaps in k-space, and two methods to reduce those have been tested here. First, shifted binning reduced the one-frame mean-max-gap by about 20° in agreement with numerical simulations. Second, the triggered GA scheme reduced the three-frame mean-max-gap compared to the standard GA.
scheme from $42.8^\circ \pm 13.7^\circ$ to $35.3^\circ \pm 3.6^\circ$ and, hence, reduced the overlap of k-space gaps across adjacent cardiac frames. For this, the rotation angle $\theta = 83.8^\circ$ was chosen based on numerical simulations, to compensate for missing data in k-space with data from neighboring cardiac frames acquired at this k-space position. Simulations show that the three-frame mean-max-gap and max-max-gap (Figure 2D,E) increase if $\theta$ is chosen smaller even though the one-frame mean-max-gap and max-max-gap could be reduced with smaller $\theta$, especially for $\theta < 40^\circ$.

The presented simulation results are only valid for acquisitions with the same duration, temporal resolution, and TR but could easily be adjusted to evaluate k-space gaps of other acquisitions that bin GA-rotated readouts into cardiac frames ranging from 2D radial\textsuperscript{31} to 5D flow\textsuperscript{32} sequences.

Despite significantly reducing k-space gaps using the proposed shifted binning and triggered GA scheme, only minor artifacts were removed compared to the standard GA scheme without shifted binning (Figure 4). This indicates that k-t sparse reconstruction performs well even with larger gaps, potentially allowing to shorten the breath-hold duration in combination with the gap reduction in the future.

Because spectral-spatial pulses on our scanner use fly-back gradients,\textsuperscript{53} the slice-thickness is at least ~8-mm thick leading to a rather large slice-thickness vs. in-plane-resolution ratio of 10, requiring careful planning of the slice location in straight segments. Lotz et al have shown as long as motion during the cardiac cycle keeps the vessel within $15^\circ$ of being orthogonal to the imaging plane the error in flow measures remains within 10%.\textsuperscript{54}

5 | CONCLUSIONS

We developed and validated a novel single breath-hold coronary PC MRI technique with both high temporal (<21 ms) and spatial resolution (0.8 mm). To achieve the high resolutions, a GA rotated spiral acquisition was combined with k-t sparse SENSE reconstruction. Gaps in k-space, that occur after binning GA rotated data into cardiac frames, were reduced using shifted binning and a triggered GA scheme, leading to a reduction of undersampling aliasing artifacts and improved image quality. PSV, PDV, coronary flow rate, and cross-sectional area acquired with the proposed triggered GA scheme exhibited excellent intra-scan repeatability and high inter-scan and intra-observer reproducibility.

DATA AVAILABILITY STATEMENT

To promote reproducible science, graphical programming interface\textsuperscript{49} reconstruction code is available at https://github.com/jhu-cardiac-mri/triggeredGASSP/(SHA-1: 86fee6aa3d-7baa885c33da6ad27ffcede0700750) including a subject-approved, anonymized dataset.

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SUPPORTING INFORMATION
Additional Supporting Information may be found online in the Supporting Information section.

FIGURE S1 The RCA of one subject could not be visualized throughout the cardiac cycle and was excluded from analysis. A screenshot demonstrates the planning on the 3D double oblique DIXON image (left). The small caliber RCA was badly visualized and only clearly visible in the mid-section of the RCA where the PC MRI imaging plane was planned. On the right, velocity compensated PC MRI images of a systolic frame (top row) and a diastolic frame (bottom row) are shown for all methods (high spatial, high temporal, standard GA and triggered GA). The RCA cannot be observed in the systolic frames (purple insets) and its signal is low in the diastolic frames (purple arrows). All other image acquisitions were successful with both RCA and LAD in 7 out of 8 subjects, leading to a success rate of 94% (15 out of 16 vessels)

FIGURE S2 Bland Altman analysis of the validation of the proposed triggered GA scheme with best binning shift against the high-spatial (A-D) and the high-temporal (E-H) sequences for PSV (A, E), PDV (B, F), coronary blood volume (C, G), and cross-sectional vessel area (D, H)

VIDEO S1 RCA flow-compensated image and velocity map of a 29-year-old male, comparing standard GA to triggered GA

VIDEO S2 LAD flow-compensated image and velocity map of a 45-year-old male, comparing standard GA to triggered GA

VIDEO S3 Flow-compensated images and velocity maps of all 15 RCA and LAD arteries shown in the same order as the flow rate curves in Figure 6

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