Passive needle guide tracking with radial acquisition and phase-only cross-correlation

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• Purpose: Acceleration of a passive tracking sequence based on phase-only cross-correlation (POCC) using radial undersampling.
• Methods: The phase-only cross-correlation (POCC) algorithm allows passive tracking of interventional instruments in real-time. In a POCC sequence, two cross-sectional images of a needle guide with a positive MR contrast are continuously acquired from which the instrument trajectory is calculated. Conventional Cartesian imaging for tracking is very time consuming; here, a higher temporal resolution is achieved using a highly undersampled radial acquisition together with a modified POCC algorithm that incorporates the point-spread-function. Targeting and needle insertion is performed in two phantom experiments with 16 fiducial targets, each using 4 and 16 radial projections for passive tracking. Additionally, targeting of eight deep lying basivertebral veins in the lumbar spines is performed for in vivo proof-of-application with four radial projections for needle guide tracking.
• Results: The radially undersampled POCC sequence yielded in the phantom experiments a lateral targeting accuracy of 1.1 ± 0.4 mm and 1.0 ± 0.5 mm for 16 and 4 radial projections, respectively, without any statistically significant difference. In the in vivo application, a mean targeting duration of 62 ± 13 s was measured.
• Conclusion: Radial undersampling can drastically reduce the acquisition time for passive tracking in a POCC sequences for MR-guided needle interventions without compromising the targeting accuracy.

KEYWORDS
interventional magnetic resonance imaging, interventional radiology, MR-guided interventional procedures, MR-guided needle intervention, MR-guided prostate biopsy

1 | INTRODUCTION

Image-guided interventions benefit from fast and reliable instrument detection and simultaneous visualization of the relevant anatomy along the instrument pathway.\textsuperscript{1-4} MRI is ideally suited for percutaneous needle interventions where image-guidance is required to steer a needle through the skin to a target organ.\textsuperscript{5} Clinical applications of percutaneous MR-guided
interventions range from analgesic injections for pain palliation,6,7 biopsy procedures in prostate,8-10 to laser11 and cryoablation thermal treatments12 for local tissue necrosis.

Although open-bore low-field MR systems permit better patient access,13 percutaneous interventions require MR images with high spatial resolution, signal-to-noise ratio (SNR), and contrast-to-noise ratio (CNR) to accurately visualize and detect the target structures ideally together with the interventional instrument. Therefore, MR-guided percutaneous interventions are preferably performed in high-field closed bore MR-systems. High field systems with their closed-bore magnets severely limit patient access so that it becomes more difficult to effectively guide the instrument under real-time MR imaging. To overcome this limitation, MR-compatible assistance systems can be used that hold and orient the needle-type instruments in the MRI bore to increase the positioning accuracy and to reduce the total procedure time compared with freehand needle placement.14-17 Unfortunately, most of these assistance systems require sophisticated control mechanisms and additional hardware, which makes them cumbersome to integrate into clinical practice.

Simple mechanically steerable assistance systems can help to overcome these limitations. The systems are designed from material/plastics which interfere neither with the static magnetic nor the radiofrequency field.18,19 Fast and reliable passive tracking can be achieved with an MR-safe needle guide filled with contrast agent.8,20-22 The orientation of the needle guide can be reliably and efficiently determined with a sequence using a phase-only cross-correlation (POCC) algorithm. Therefore, the POCC sequence acquires two parallel tracking images that are oriented approximately perpendicular to the symmetry axis of the needle guide using either a sequential8,20,21 or simultaneous22 excitation and acquisition scheme. The needle guide appears as a ring structure in these cross-sectional images, and its position can be automatically detected with the POCC algorithm; here, prior knowledge of the ring appearance in the image can be exploited for detection via template matching. Subsequently, a targeting image is aligned parallel to the needle guide in real time with a projected needle trajectory. The tracking and targeting images are acquired continuously, so that the POCC sequence follows the needle guide movements to permanently visualize the theoretical needle trajectory while it is aligned with the target structure. A disadvantage of the POCC sequence is the relatively long acquisition time of the tracking images which limits the velocity with which the needle guide can be moved, and which can lead to prolonged targeting times.

In this technical feasibility study, the conventional Cartesian sampling for the tracking images is replaced with a faster, undersampled radial k-space acquisition, and the POCC detection accuracy is studied as a function of the number of radial k-space projections. To improve needle guide detection, a modified synthetic image for POCC template matching is introduced that takes into account the point-spread function of the k-space sampling trajectory. The new method is implemented into a real-time tracking sequence, and its effect on the targeting precision is compared in phantom and in vivo experiments to results from a previous study.19

2  METHODS

2.1  Passive tracking sequence

The POCC sequence8,20-22 automatically follows the movements of a cylindrical passive needle guide with a central opening for needle insertions (Figure 1A). The sequence acquires two parallel Cartesian T1-weighted fast low-angle shot (FLASH) images oriented perpendicular to the symmetry axis of the needle guide which is filled with a solution or gel with short T1 and appears in the image as a ring with positive contrast (Figure 1B). The location of the ring in the image is detected with the POCC algorithm with subpixel precision.23,24 Therefore, an ideal synthetic image $M_{\text{ideal}}(x, y)$ of the needle guides cross-section at the center of the image is calculated from the actual imaging parameters (ie, field of view [FOV] and base resolution) and the physical dimensions (ie, inner and outer diameter). With the assumption that the acquired image $I(x, y)$ is a shifted version of the ideal synthetic image, that is, $I(x, y) = M_{\text{ideal}}(x + x_0, y + y_0)$, the POCC is computed rapidly in the k-space representation as a simple multiplication

$$
P\text{POCC}(k_x, k_y) = \frac{\tilde{I}(k_x, k_y)}{||\tilde{I}(k_x, k_y)||} \cdot \tilde{M}_{\text{ideal}}(k_x, k_y)^* e^{-i(k_x x_0 + k_y y_0)},
$$

where the tilde denotes the Fourier transform. Note, that Equation (1) can be derived using the Fourier Shift theorem. The shift $(x_0, y_0)$ of the ring from the center is determined after an inverse Fourier Transform of $P\text{OCC}(k_x, k_y)$ which leads to a high signal (Delta function) at $(x_0, y_0)$. Imperfections of the synthetic image $M(x, y)$ and in the acquired image $I(x, y)$ can blur this maximum so that the ring coordinates are extracted via maximum search of the POCC signal distribution—here, subpixel precision can be achieved with a center-of-mass algorithm.20,24

After $(x_0, y_0)$ are determined in both tracking slices a targeting image with a different contrast (eg, balanced steady state free precession, bSSFP) is aligned with the needle guide to better visualize the target structure. The POCC sequence alternates between tracking and targeting acquisitions and allows to follow needle guide movements in real-time.

The temporal resolution of the POCC sequence depends considerably on the acquisition time of the two tracking
images. For a typical TR of 4 ms and 192 acquired phase encoding steps, the acquisition time for one POCC cycle is $T_{A_{POCC}} = 2.3$ s, that is, acquisition time needed for two tracking images and one targeting image. Even with dual-band imaging, which accelerates the acquisition by sampling both tracking slices simultaneously, the acquisition takes approximately $T_{A_{POCC}} = 1.5$ s. A substantially faster acquisition of the tracking information could be achieved with a radially undersampled k-space data acquisition scheme.

2.2 Radial undersampling of tracking slices

Radial MRI acquisition schemes can be drastically undersampled by acquiring only a few angular increments of the radial k-space. Undersampling is known to produce radial artifacts, known as streaks, which get more prominent with a decreasing number of radial projections. However, the appearance of these streaks can be estimated for a known object using the point-spread function (PSF) of the k-space sampling trajectory $s_{PSF}(x, y, N_P)$ as a k-space filter. Therefore, it is possible to calculate the synthetic image for a given radial sampling scheme as the convolution between the ideal image and the PSF of the k-space trajectory:

$$M_{PSF}(x, y, N_P) = M_{ideal}(x, y) \ast s_{PSF}(x, y, N_P). \tag{2}$$

Here, $s_{PSF}(x, y, N_P)$ can be simply calculated as the Fourier Transform of the gridded k-space trajectory after density compensation. The projection-specific image $M_{PSF}(x, y, N_P)$ is used for POCC calculation (Equation 1) for radially undersampled tracking images (Figure 2, top row).

2.3 Implementation of sequence and reconstruction

The POCC sequence was implemented based on a customized spoiled gradient echo acquisition and excitation in combination with an evenly spaced radial k-space trajectory. Gradient delay errors were corrected with the method proposed by Speier et al. All imaging experiments were performed on a 3T clinical MRI system (Siemens Prisma, Siemens Healthineers, Erlangen, Germany). For signal reception the system’s open loop coil (outer diameter: 11 cm) and the integrated spine array were used.

Real-time image reconstruction was implemented using the vendors’ image reconstruction environment (Image Calculation Environment, ICE, Siemens). Radial data were gridded to a Cartesian raster using a Kaiser-Bessel kernel, and an image was then calculated via Fourier transformation. The magnitude of this image was then back transformed to the k-space representation to $\tilde{I}(k_x, k_y)$ to perform POCC.
calculation. Note, that this guaranteed that no division by zero can occur in Equation 1.

2.4 | POCC detection accuracy

A passive needle guide (Poly(methyl methacrylate) (PMMA), outer diameter: 13 mm, diameter of central opening: 5 mm, Figure 1A) was filled with a contrast agent solution (Magnevist®/H2O: 1/200, Bayer Schering Pharma AG, Berlin, Germany) and placed roughly at the magnet isocenter in the \(xz\)-plane. Phantoms containing fat and water, and a bottle filled with a standard calibration solution \((1.25 \, \text{g NiSO}_4 \times 6\text{H}_2\text{O}+5 \, \text{g NaCl})/1000 \, \text{g H}_2\text{O})\) were positioned in the vicinity and images were acquired (TR/TE = 3.93 ms/1.85 ms, \(\alpha = 25^\circ\), base resolution = 256, bandwidth = 1150 Hz/px, slice thickness (SL) = 10 mm, slice distance = 30 mm, 30 repetitions). The number of radial projections \(N_p\) was varied \((N_p = 2, 4, 6, \ldots, 32, 64, 128, 192)\) to determine the maximally usable acceleration factor. Data were acquired for two different slice orientations either perpendicular (Figure 2, 2nd and 4th row) to or at an angle of approximately \(15^\circ\) (Figure 2, 3rd and 5th row) with respect to the needle guide’s symmetry axis to assess the tracking stability against misalignments.

The POCC distribution (Equation 1) was calculated for both slices using either the ideal image \(M_{\text{Ideal}}\) or the projection-specific synthetic images \(M_{\text{PSF}}\) (Equation 2) which considers the point-spread function (PSF) of the k-space trajectory for each number of projections \(N_p\). The location of the POCC maximum \((x_0, y_0)\) was determined for each image with subpixel precision and the mean POCC maximum was calculated for each projection \(N_p\) (Figure 3A,B).
Additionally, the mean deviation $\Delta r$ of $(x_0, y_0)$ from a reference coordinate (mean POCC location in the images acquired with $N_p = 192$) was calculated for each projection number $N_p$ (Figure 3C,D).

### 2.5 Targeting accuracy of the POCC sequence

Two identical agarose gel phantoms (agarose concentration 2%) with 16 embedded fiducial targets in each phantom were constructed. Needle guide movements inside the magnet bore were performed with an assistance system via extension rods and the system’s loop coil (outer diameter: 11 cm) was used for signal reception. Targeting experiments were performed with $N_p = 4$ (time for one POCC tracking cycle $T_{APOCC} = 0.64$ s, that is, with a framerate of approximately 1.5 image/s) or $N_p = 16$ ($T_{APOCC} = 0.73$ s, framerate: 1.4 images/s) (radial GRE tracking: TR/TE = 3.93/1.85 ms, FOV: $300 \times 300$ mm$^2$, matrix: 256$^2$, $\alpha_{GRE} = 25^\circ$, slice thickness = 10 mm, pixel bandwidth = 1150 Hz/px; bSSFP targeting: TR/TE = 3.93 ms/1.85 ms, FOV: 206 $\times$ 203 mm$^2$, matrix: 552 $\times$ 832 (interpolated), $\alpha_{GRE} = 13^\circ$, thickness = 0.5 mm, pixel bandwidth = 500 Hz/px, averages: 5). A Student’s t-test was performed to compare the results for $N_p = 4$ and $N_p = 16$ statistically.

For in vivo proof-of-concept testing the assistance system was placed on the abdomen of a healthy volunteer in supine position (cf. Figure 4A) and basivertebral veins in the lumbar spines were defined as fiducial targets (cf. Figure 4B) as described in and targeting was performed with $N_p = 4$ (radial GRE tracking images: TR/TE = 4 ms/2 ms, FOV: $380 \times 380$ mm$^2$, matrix: 256$^2$, $\alpha_{GRE} = 25^\circ$, slice thickness = 10 mm, pixel bandwidth = 1085 Hz/px; bSSFP targeting image: TR/TE = 4 ms/2 ms, FOV: $380 \times 380$ mm$^2$, matrix: 205 $\times$ 256, partial Fourier: 6/8, $\alpha_{bSSFP} = 45^\circ$, slice thickness = 5 mm, pixel bandwidth = 1085 Hz/px, phase resolution = 80%, $T_{APOCC} = 0.65$ s, framerate: 1.5 image/s). For each target, the needle guide was initially moved at the center position of the assistance system and then, the theoretical
needle pathway was aligned with the target in sagittal and transversal view, respectively (Figure 4C and Supporting Information Video S1). The targeting duration was recorded for each target together with the estimated target diameter and compared to the previously published results for the sequence without any acceleration. All methods were carried out in accordance with relevant guidelines and regulations, healthy volunteer scanning was approved by the institutional review board (Ethikkommission) of the University Medical Center Freiburg (No. 160/2000), and informed written consent was obtained before imaging.

3 | RESULTS

3.1 | POCC detection accuracy

The mean maximum values of the POCC distribution calculated with the projection specific mask image $M_{PSF}$ are in general higher than those calculated with $M_{Ideal}$ for the perpendicular and the angulated slice for all projections $N_p$ (Figure 3A,B). The mean deviation $\Delta r$ of the POCC location from the reference coordinates is smaller than 0.3 mm for all $N_p$ for both slice orientations (Figure 3C,D).

3.2 | Targeting accuracy of the POCC sequence

In the phantom experiments, targeting and needle insertion were successful for each fiducial target. A lateral targeting accuracy of $1.1 \pm 0.4$ mm and $1.0 \pm 0.5$ mm was achieved for 16 and 4 radial projections, respectively (Table 1). No statistical significant difference was found for the lateral distance under a Student’s $t$-test ($P$-value $> .45$).

In the in vivo experiment the theoretical needle trajectory was successfully aligned with each of the eight targets (cross-sections of basivertebral veins) using $N_p = 4$. The measured mean targeting duration was $62 \pm 13$ s for a mean target diameter of $8.2 \pm 0.7$ mm. The target diameter and the targeting duration are summarized in Supporting Information Table S1.

4 | DISCUSSION

In this work, a highly accelerated POCC sequence was presented for MR-guided needle interventions utilizing radial undersampling. To detect a passive needle guide the sequence acquires two radially undersampled tracking images with a POCC template matching algorithm that uses synthetic images with streaking artifacts from the undersampled radial trajectories. It was found that POCC detection was possible even at the very extreme case of $N_p = 2$. Compared to conventional synthetic images, projection-specific images yield higher POCC maximum values. The deviations between POCC locations and the ground truth coordinates is in general below 0.3 mm even for $N_p = 2$, which translates to about 1 mm at the target (pixel size: 1 mm, targeting slice distance: 30 mm, target distance: 60 mm).

Several reconstruction techniques have been proposed in recent years which use iterative reconstruction methods and which allow for high undersampling factors to visualize very fast motions in the millisecond range while minimizing streaking artifacts. However, these algorithms require

| TABLE 1 | Summary of the results for the phantom targeting experiments |
|----------|-----------------------------|
| Projections $N_p$ | 4 | 16 |
| Distance [mm] | $1.0 \pm 0.5$ (0.4-2.1) | $1.1 \pm 0.4$ (0.5-1.7) |
| Target diameter [mm] | $7.3 \pm 0.4$ (6.7-8.1) | $7.0 \pm 0.4$ (6.3-7.6) |
| Targeting duration [s] | $38 \pm 8$ (24-51) | $42 \pm 9$ (30-68) |

Note: Mean values, SD, and range for the lateral distance of the needle pathway to target center, target diameter, and targeting duration.
high computational power and are, therefore, reconstructed offline\textsuperscript{30,31}, which makes them unsuitable for real-time instrument visualization so far.

To demonstrate the feasibility of accelerated tracking, phantom experiments were performed similar to previous studies.\textsuperscript{19,22} In this work, needle guide tracking was always possible and the targeting image update rate (TA\textsubscript{POCC} = 0.64 s for \( N_p = 4 \) and TA\textsubscript{POCC} = 0.73 s for \( N_p = 16 \)) was approximately four times faster than a conventional sequential Cartesian acquisition (TA\textsubscript{POCC} = 2.62 s) which allows to monitor movements during breathing. In the experiments the targeting accuracy (1.1 ± 0.4 mm for \( N_p = 16 \) and 1.0 ± 0.5 mm \( N_p = 4, P > .45 \)) was similar to or even better than previous studies (1.5-1.7 mm\textsuperscript{19-22}). However, in an \textit{in vivo} situation the heterogeneous mechanical properties of tissue which might cause needle deflections thus reducing the targeting accuracy.\textsuperscript{32}

In the \textit{in vivo} experiment, the accelerated POCC sequence could target deep-lying basivertebral veins. The needle trajectory was successfully aligned with the targets in 62 ± 13 s. Compared to our previous study\textsuperscript{19} using a similar setting (120 ± 30 s), the undersampled tracking sequence provides a twofold faster targeting procedure which might be beneficial during targeting of abdominal treatments, for example, biopsies in liver or kidneys, where peristaltic and breathing motion are present. Here, the placement of the assistance system above the patient might have further advantages: the signal acquired with a loop coil beneath the assistance system is sufficient for needle guide tracking with only four radial projections while providing good visualization of the targeted region. Unfortunately, tracking with as low as two projections was not stable with this setup, because the signal was too low in the tracking slice further away from the coil. However, with the short TRs used in this sequence the additional two projections prolong the acquisition by only 2xTR = 8 ms which does not significantly contribute to the total duration of the whole tracking cycle.

Compared to active tracking, passive tracking approaches do not suffer from RF heating.\textsuperscript{33} However, the proposed tracking sequence—in particular for a very small number of projections—requires a sufficiently high signal from the needle guide which is only achievable when the receive coil is in close proximity. Therefore, in applications like prostate biopsies, where the needle guide is inside the rectum of the patient, undersampled tracking might not be feasible with four or eight projections.

Targeting with the metallic needle inside the needle guide is currently not possible because susceptibility artifacts of the needle distort the ring-shaped cross-section required for POCC tracking. Tracking with the needle in the guide could be advantageous to detect needle deflections which could be achieved by taking into account the susceptibility artifacts of the needle for template matching.\textsuperscript{34} and detection of the needle itself could be realized with template matching with simulated needle artifacts\textsuperscript{35} or artifact suppressed imaging.\textsuperscript{36,37} Furthermore, this would allow monitoring the insertion depth of the needle, since this is not possible in the current implementation. Alternatively, spin-echo-based sequences or multispectral imaging\textsuperscript{37} may be utilized to minimize needle artifacts and monitor the needle insertion.

5 | CONCLUSIONS

The POCC sequence allows passive tracking of a needle guide with radial undersampling with only two radial projections if the point-spread function of the sampling trajectory is taken into account during POCC template matching.

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SUPPORTING INFORMATION

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

**TABLE S1** Target diameter and targeting duration for the in vivo proof-of-concept experiment. Mean values and standard deviations are summarized in the bottom line

**VIDEO S1** Exemplary targeting maneuver for one target

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