FULL PAPER

Bilateral Multiband 4D Flow MRI of the Carotid Arteries at 7T

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Purpose: Simultaneous multislab (SMSb) 4D flow MRI was developed and implemented at 7T for accelerated acquisition of the 3D blood velocity vector field in both carotid bifurcations.

Methods: SMSb was applied to 4D flow to acquire blood velocities in both carotid bifurcations in sagittal orientation using a local transmit/receive coil at 7T. $B_1^+$ transmit efficiency was optimized by $B_1^+$ shimming. SMSb 4D flow was obtained in 8 healthy subjects in single-band (SB) and multiband (MB) fashion. Additionally, MB data were retrospectively undersampled to simulate GRAPPA $R = 2$ (MB2_GRAPPA2), and both SB datasets were added to form an artificial MB dataset (SumSB). The band separation performance was quantified by signal leakage. Peak velocity and total flow values were calculated and compared to SB via intraclass correlation analysis (ICC).

Results: Clean slab separation was achieved yielding a mean signal leakage of 13% above the mean SB noise level. Mean total flow for MB2, SumSB, and MB2_GRAPPA2 deviated less than 9% from the SB values. Peak velocities averaged over all vessels and subjects were $0.48 \pm 0.11$ m/s for SB, $0.47 \pm 0.12$ m/s for SumSB, $0.50 \pm 0.13$ m/s for MB2, and $0.53 \pm 0.13$ m/s for MB2_GRAPPA2. ICC revealed excellent absolute agreement and consistency of total flow for all methods compared to SB2. Peak velocity showed good to excellent agreement and consistency for SumSB and MB2 and MB2_GRAPPA2 method showed poor to excellent agreement and good to excellent consistency.

Conclusion: Simultaneous multislab 4D Flow MRI allows accurate quantification of total flow and peak velocity while reducing scan times.

KEYWORDS
4D flow MRI, 7 Tesla, carotid artery, multiband, SMS
4D flow MR imaging\(^1\) is a technique that enables noninvasive measurements of the time-resolved 3D blood velocity vector field. Previous studies showed successful applications in the carotid arteries\(^2\) and quantification of hemodynamic parameters. In combination with other MR imaging methods that characterize or quantify the properties of the vessel and vessel wall,\(^3\) both plaque anatomy and hemodynamics could be investigated using a single modality.

This is particularly interesting for the carotid bifurcations, because 20% of all strokes are attributable to carotid stenosis.\(^4\) Not only the identification of carotid plaques but also the risk stratification for disease progression are essential for treatment. Indications exist that the progress of local atherosclerosis is linked to hemodynamics,\(^5\) particularly wall shear stress, but further investigations are required.

However, accurate quantification of velocity and derived hemodynamic parameters requires high spatial and temporal resolution,\(^6,7\) which leads to long acquisition times and requires high signal-to-noise ratio (SNR). Thus, addressing the scan times and SNR is key for improved hemodynamic quantification in carotid 4D flow.

Apart from optimizing sequence and acquisition parameters, the SNR for 4D flow imaging targeting the carotid arteries can be increased by using a dedicated receive coil that is closely positioned to the target region\(^2\) and by applying 4D flow MRI at higher field strength.\(^8,9\) Although most 4D flow MRI scans are obtained at 1.5T and 3T, recently promising results have been presented at 7T showing an approximately 2.2-fold increase in SNR in the aorta\(^8\) and 2.6-fold increase in the brain\(^9\) compared to 3T studies.

Motivated by these findings, the present work investigates the feasibility of performing 4D flow MRI targeting both carotid bifurcations at 7T using a local transmit/receive coil. This approach will not only benefit from receiving the signal through a local, close-fitting radiofrequency (RF) coil at 7T, but also transmitting the RF pulses by a local carotid coil is appealing.\(^10,11\) To cover the carotid artery (CA), the external carotid artery (ECA), and the internal carotid artery (ICA) in a single acquisition while benefiting from the high signal of blood flowing into the slab (inflow effect), a thicker (50 mm) slab oriented in axial or axial-oblique orientation has been used in previous studies at lower fields.\(^2,12\)

Due to the intrinsically limited and distinct field-of-excitation along the head-foot direction of the local transmit/receive (TX/RX) coil, the current approach allows a sagittal acquisition using a thinner slab that is adjusted to the left-right dimensions of the vessel geometries while still benefitting from the inflow effect. A clear drawback of this approach is, however, that both carotid arteries need to be acquired sequentially, which increases total scan times and, therefore, acceleration of the acquisition is required.

Simultaneous multislice imaging (SMS),\(^13,14\) also termed multiband (MB) imaging, is an ideal acceleration technique to address this issue because it allows exciting and acquiring multiple slices at the same time with only a mild effect on SNR.\(^13\) In most applications, SMS is applied to acquire multiple slices. Its potential has been demonstrated for example in applications such as functional MRI, diffusion MRI, arterial spin labeling, and even cardiac MRI and others.\(^15-18\) In addition, SMS has been applied to slice-selective phase-contrast (PC) velocity imaging,\(^19\) recently. However, although the extension of SMS toward slab-selective excitation, that is, Simultaneous Multislab (SMSb) imaging, has been shown for example in time-of-flight angiography,\(^20\) it has not yet been introduced for 4D flow MR imaging to the best of the authors’ knowledge.

In the present work, we investigate the performance of SMSb applied to 4D flow MRI in the carotid bifurcations and we make use of 7T in combination with a local TX/RX coil. By exciting two sagitally oriented slabs, the acquisition covers both bifurcations. Eight healthy subjects were scanned using 4D flow MB imaging and quantification of peak velocity and total flow has been performed in left and right common CAS, as well as left and right ICAs and ECAs. Results were compared with those obtained from conventional acquisitions exciting only single slices (single-band, SB).

2  |  METHODS

2.1  |  Setup and hardware

Eight subjects were scanned (mean/median age: 36.9/28.9 years, range: 51.4 years, 6 women) after providing written consent according to an approved institutional review board protocol. All scans were performed at a 7T whole body system (Magnetom 7T, Siemens, Erlangen, Germany), which is equipped with a dedicated 16 channel $B_1^+$ shimming system (CPC, Hauppauge, New York, USA). A prototype carotid coil (Virtumed LLC, Minneapolis, Minnesota, USA) similar to the one reported by Krafft et al\(^10\) with eight transceiver channels was used to specifically target the carotid bifurcation as illustrated in Figure 1. The coil consists of two halves, each containing four overlapping loop elements of size 5 cm × 3.5 cm that are geometrically decoupled. The coil is oriented with its longer side along the A-P direction and the center was positioned such that it is located close to the carotid bifurcation on both sides of the neck. The coil was relocated in case that localizer images showed a sub-optimal position of one or both elements. A vendor-provided 3-lead ECG was attached to the subject’s chest and the leads were repositioned in case the detection of the cardiac cycle was not acceptable.
2.2 Adjustments and calibration scans

Each session started with the acquisition of axial localizer images using an unoptimized setting of the transmit phases. Subsequently, 3D multislab time-of-flight angiography MR images covering the bifurcations were obtained in axial orientation. Imaging parameters were as follows: 4 slabs containing 32 slices, 0.5 mm slice thickness, 5 mm overlap, TR/TE = 20 ms/3.9 ms, field of view (FOV) = 160 × 144 × 49 mm³; (0.5 mm)³ isotropic resolution, 2-fold acceleration (GRAPPA).

The position of the carotid bifurcation was then identified based on the time-of-flight images for subsequent $B_1^+$ mapping (Figure 1D,E), which was performed using a cardiac triggered gradient echo sequence. Three parallel slices 8 mm apart with predominantly coronal orientation were positioned through both carotid bifurcations. The sequence consisted of eight measurements, and for each measurement, only a single but successively alternating transmit (TX) channel was active during transmission whereas all eight channels were active during signal reception. In addition, one measurement with all TX channels and one with no TX channel enabled were obtained (parameters: TR/TE = 50 ms/3.8 ms, FOV = 160 × 160 mm², matrix = 128 × 128, thickness = 4 mm, 1:08 min). Two oval-shaped regions of interest (ROI) were manually drawn on each of the three slices with each ROI covering one bifurcation and its surrounding tissue, yielding six ROIs in total. $B_1^+$ maps were calculated based on such measurements using the algorithm described by Van de Moortele et al.\textsuperscript{21}

Subsequently, the transmit phases were optimized using Matlab (The Mathworks, Natick, Massachusetts) to maximize the transmit efficiency $\eta_t = \frac{\sum_{k=1}^{8} |R_{1k}(r)|}{\sum_{k=1}^{8} |B_{1k}(r)|}$ of all 8 TX channels within the six ROIs, which maximized the constructive interference of the $k = 1 \ldots 8$ transmit fields $B_{1k}^+(r)$ within the ROIs. The resulting $B_1^+$ field was mapped using the actual flip-angle imaging (AFI) technique\textsuperscript{22} with the following parameters: TR1/TR2/TE = 20 ms/120 ms/2.5 ms, FOV = 256 × 208 × 140 mm³, resolution = 2 × 2.5 × 5 mm³, GRAPPA = 2, nominal flip angle = 50 degrees. $B_1^+$ optimizations were followed by $B_0$ shimming that was manually performed using the scanner’s adjustment routine.

2.3 Multiband 4D flow MRI acquisition

Three 4D flow MRI scans were performed without parallel imaging acceleration with identical imaging protocols as listed in Table 1. The imaging slabs were planned in sagittal
orientation with readout in head-foot direction as shown in Figure 1F. The first two scans, both SB scans (SB\textsubscript{R} and SB\textsubscript{L}), excited a single 19.2 mm thick slab covering the right or left carotid bifurcation. The third scan was an MB acquisition of two slabs (MB\textsubscript{2}) with identical parameters as before except that the two 19.2 mm thick slabs were excited simultaneously to cover both bifurcations. For this MB\textsubscript{2} scan, the two complex RF pulses of SB\textsubscript{R} and SB\textsubscript{L}, each consisting of a 1 ms long sinc-shaped pulse with bandwidth-time-product of 4, were replaced by a 1 ms long MB pulse. The MB pulse was generated by summing up the individual complex pulses. The same \(B^*_1\) shim setting with optimized \(B^*_1\) efficiency was used as calculated before. Note, that we thus followed a slab-joint shimming approach\textsuperscript{23} for which the pulses were first added prior to applying the \(B^*_1\) shim setting.

CAIPIRINHA\textsuperscript{24} was enabled for the RF pulse of the right carotid artery by applying a \(\pi\) phase shift for every other phase encoding (PE) k-space line, which resulted in an imaging slab of the right carotid artery shifted by FOV/2 in the PE direction.

Two calibration datasets (Cal\textsubscript{L} and Cal\textsubscript{R}) were acquired in addition to the MB and SB flow datasets. Both Cal\textsubscript{L} and Cal\textsubscript{R} datasets had identical parameters as the SB flow datasets (Table 1) but were acquired in a single cardiac phase (ie, not time resolved) with total acquisition time of 1:30 minutes.

### 2.4 Reconstruction

Image reconstruction of all datasets was performed offline in Matlab. Four different datasets were reconstructed and analyzed as outlined subsequently: (1) the two SB acquisitions, (2) an artificial MB dataset consisting of the sum of the two SB datasets (SumSB), (3) the acquired MB\textsubscript{2} dataset, and (4) MB\textsubscript{2} with retrospective GRAPPA \(R = 2\) acceleration (MB\textsubscript{2,GRAPPA2}). Acquisition parameters for these datasets are listed in Table 1. The SumSB dataset was analyzed and compared to the SB dataset because this comparison can use identical carotid vessel ROIs for both SB and SumSB, and thus, deviations between their hemodynamic results are linked to the reconstruction process rather than the imaging and segmentation process. MB and SumSB data were reconstructed as follows. Raw k-space data of both MB datasets and the calibration datasets were first Fourier transformed along the slab dimension into hybrid space. Subsequently, each slice, each encoding direction, and each cardiac phase was reconstructed individually using the calibration data and a split-slice GRAPPA reconstruction kernel.\textsuperscript{25} The separated slices of each band were then grouped to form two individual time-resolved 4D flow slabs in hybrid space covering the left and right artery. Prior to transforming the data along readout (RO) and PE directions into image space, data from band 2 (right artery) were Fourier transformed back into k-space. Subsequently, data were multiplied by a linear phase along the slab direction to correct for a circular spatial shift of slab 2 along the slab direction. Note, this step is necessary, as the two slabs only fold on top of each other if the distance between them is exactly a multiple of the slab thickness. In addition, a linear phase is applied in the PE direction for slab 2 to correct for the FOV/2 shift due to applying CAIPIRINHA. After these correction steps both slabs are transformed into image space.

Both SB datasets were also reconstructed using the aforementioned reconstruction pipeline resulting in two datasets each: the passing signal \(P_{1 \rightarrow 1}\) and \(P_{2 \rightarrow 2}\) that is passed through the reconstruction pipeline into the identical slab as well as the leaking signal \(L_{1 \rightarrow 2}\) and \(L_{2 \rightarrow 1}\) that denotes the signal that leaks from one slab into the other.\textsuperscript{26}

For the fourth reconstructed dataset (MB\textsubscript{2,GRAPPA2}), the MB acquisition was accelerated retrospectively by omitting every second PE line and applying in-plane GRAPPA\textsuperscript{27} with an acceleration factor of \(R = 2\). The reconstruction was performed sequentially: first the reconstruction separated the two accelerated slabs using the split-slice GRAPPA algorithm, then, the folding due to in-plane GRAPPA was unwrapped.

### 2.5 Analysis

#### 2.5.1 Leakage

A main source of quantification error can consist in signal leaking from one slab to the other. Therefore, the leakage was qualitatively and quantitatively assessed based on \(L_{1 \rightarrow 2}\) and \(L_{2 \rightarrow 1}\). Qualitatively, the leakage maps and maximum intensity projections of the maps along the temporal and the spatial dimension (slab dimension) were created and visually inspected. Quantitatively, the mean value and the 99th percentile value (L99) of all leakage magnitude maps considering all slices, all encoding directions, and all cardiac phases

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**TABLE 1**  Acquisition parameters of the 4D flow scans

| Parameter                      | Value                                |
|--------------------------------|--------------------------------------|
| Field of View                  | \(160 \times 100 \times 19.2 \text{ mm}^3\) |
| Spatial resolution             | 0.8 mm isotropic                     |
| Temporal resolution            | 81.6 ms                              |
| Cardiac phases*                | 9-13                                 |
| Echo time                      | 3.9 ms                               |
| Pulse repetition/echo spacing  | 6.8 ms                               |
| Acquisition bandwidth          | 460 Hz/Px                            |
| Nominal flip angle**           | 15 degrees                            |
| Venc                           | 100 cm/s in all three directions      |

*Cal\textsubscript{R} and Cal\textsubscript{L} were acquired without temporal resolution.

**Note that the actual flip angle varies over the field of view.
were calculated per subject. To set the leaking signal into relation with the noise level, these values were normalized by the mean noise level magnitude of the passing signal taken from the first five lines of the images.

2.5.2 | Image processing

All reconstructed 4D flow MRI datasets were first preprocessed using an in-house developed Matlab-based preprocessing tool\(^\text{28}\) to correct for noise\(^\text{29}\) and to calculate the pseudo-complex phase-contrast angiogram (PC-MRA).\(^\text{30}\) The preprocessed data were imported into a novel analysis tool specifically developed in Matlab for intra- and extracranial 4D flow MRI data analysis.\(^\text{31}\) The data analysis workflow is illustrated in Figure 2. The PC-MRA was used for the automatic segmentation of the left or right carotid arteries using a thresholding method described by Schrauben et al.\(^\text{32}\) In addition, manual corrections were performed using thresholding, region growing, and manual ROI-based deletion using native Matlab functions. The segmented volume was then used as the basis for the detection of center points using a thinning algorithm.\(^\text{32}\) Quadratic spline functions allowed the extraction of individual numbered centerlines representing separate vessels, which were named individually as left CA, left ICA, left ECA, right CA, right ICA, and right ECA. Equidistant analysis planes were automatically positioned perpendicular along each centerline with a 1 mm distance between them.

A secondary automatic segmentation determined a refined vessel contour for each analysis plane, which we inspected visually and corrected manually if needed. The maximum of peak velocity and median total flow of all analysis planes was collected for each vessel and used for further statistics.\(^\text{31}\) All initial segmentations were performed on the left and right SB datasets and saved for analysis of SumSB, MB2, and MB2\_GRAPPA2. If the location of the carotid artery was sufficiently different from SB, the segmentations were redone for the MB2 acquisition and applied to MB2 and MB2\_GRAPPA2. Vessel centerlines and definitions were also saved for SB (and for MB2 if vessel location was too different from SB) and was applied for analyzing all other acquisitions. In addition, the blood flow was visualized in Matlab using streamlines at the time point of the largest average velocity within the volume.

2.5.3 | Statistics

All calculated peak velocities and total flow of a certain vessel were determined by taking the median peak velocity or flow of several analysis planes perpendicular along the centerline of the specific vessel.

To assess agreement and investigate a potential bias of the proposed MB method, a Bland-Altman analysis was performed for peak velocity and total flow within left and right CA, ICA, and ECA measured with MB2, MB2\_GRAPPA2, SumSB, and resulting values were compared with those obtained by SB acquisitions. Intraclass correlation (ICC) estimates and their 95% confidence intervals (CI) were calculated for each method compared to SB. ICC methods described by McGraw et al\(^\text{33}\) were based on consistency and an absolute-agreement, 2-way mixed-effects model. The following definitions were used to interpret results: ICC values less than 0.5 are indicative of poor reliability, values between 0.5 and 0.75 indicate moderate reliability, values between 0.75 and 0.9 indicate good reliability, and values greater than 0.90 indicate excellent reliability.\(^\text{34}\)

3 | RESULTS

All MB and SB datasets were successfully acquired in all subjects. The total scan time of the MB datasets was on average 15 minutes 27 seconds with similar acquisition time for each of the SB acquisitions. Figure 3A shows the individual, relative $B_1^+$ maps of the 8 TX channels of the left and the right half of the carotid coil obtained in subject 5, demonstrating that the elements of the left half barely excite the right CAs and vice versa. A similar effect has been observed for the receive profile $B_1^-$. To estimate the profile extension along the head-foot direction, Figure 3B shows the sum of the $B_1^+$ magnitude of all channels (ie, phases are neglected) and Figure 3C plots the profile along head-foot direction after integration over the map displayed in Figure 3C in left-right direction. The resulting profile shows a FWHM of 12 cm along the head-foot direction.

Figure 4 illustrates exemplary flip angle (FA) maps for a target FA of 50 degrees in the same subject after $B_1^+$ shimming in a transverse slice covering both carotid arteries. A steep FA gradient showing reduced FA with increasing distance to each coil can be observed in Figure 3B.C. The gradient in combination with the variable location of the bifurcation within the different subjects resulted in variations of the FA within the bifurcation between 5 degrees and 30 degrees with a mean/SD value of 19.5 degrees/7.9 degrees.

3.1 | MB reconstruction and leakage

Exemplary magnitude and velocity images of the original, aliased MB and the SB datasets, as well as the reconstructed MB and SumSB datasets are illustrated in Figure 5. Qualitatively, successful separation of the two slabs can be appreciated in the second column of Figure 5B providing similar velocity values to the SB data obtained from two separate scans. This
finding is further supported by the last column, which shows reconstructed data of the left carotid artery obtained from the summation of both SB datasets. In Figure 6A, maximum intensity projections (MIP) of the signal magnitude along the slab direction and along the cardiac phase are shown for the leaking and passing signal. Corresponding phase images during systole are illustrated in the same subject for a single slice crossing the carotid bifurcation. Figure 6C,D summarize the quantified L99 and mean leakage normalized by the mean noise level of the passing signal for all subjects. Averaged over all subjects, the mean leakage was 1.13, that is, 13% above the mean noise level of the passing signal, and L99 amounted to 1.86 on average.

3.2 | Hemodynamics

Figure 7 displays Bland-Altman plots comparing MB2, MB2 GRAPPA2, and SumSB with original SB results. Plots A-D illustrate the agreement of total flow in each of the vessels for all subjects and E-H the agreement of peak velocities. Table 2 summarizes the Bland-Altman analysis results. Both total flow and peak velocity differences compared between SumSB and original SB show, as expected, values close to zero (Table 2). For total flow, all MB methods show a bias of less than 9% in the mean total flow value from the value obtained by the SB acquisition. The largest bias was observed for MB2 (total flowMB = 2.50 ± 1.34 ml/cycle, total flowMB GRAPPA2 = 2.40 ± 1.34 ml/cycle, total flowSumSB = 2.37 ± 1.40 ml/cycle, total flowSB = 2.39 ± 1.36 ml/cycle).

The largest bias for peak velocity was observed in the MB2 GRAPPA2 dataset that was retrospectively accelerated with parallel imaging (bias of 11% from mean peak velocity). Peak velocity measured with MB2 was on average 5.96% higher than obtained by the SB dataset, and for SumSB peak velocity was on average 2.1% lower compared to SB (peak velocityMB = 0.50 ± 0.13 m/s, peak velocityMB GRAPPA2 = 0.53 ± 0.13 m/s, peak velocitySumSB = 0.47 ± 0.12 m/s, peak velocitySB = 0.48 ± 0.11 m/s). In summary, the MB methods show larger bias as well as limits of agreement compared to SB with peak velocity being more sensitive to error than total flow.

In addition, ICC assessment showed excellent absolute agreement as well as consistency for total flow for all MB methods compared to SB (both lower and upper bound >0.9,
FIGURE 3  (A) Relative $B_1^+$ maps of the eight transmit channels in coronal view obtained by the mapping technique described in reference.\textsuperscript{21} (B) Sum of the $|B_1^+|$ of all channels that are displayed in subfigure (A). (C) The sum of $|B_1^+|$ illustrated in (B) was integrated along the left-right direction and is displayed as a line plot demonstrating the transmit profile variation along the head-foot direction.

FIGURE 4  (A,B) Flip angle (FA) map and corresponding magnitude image obtained in subject 5 after $B_1^+$ shimming at the level of the carotid bifurcations. The nominal FA was set to 50 degrees in the protocol. (C) A left-right FA profile plot along the dashed line in b) shows a strong FA gradient from either side of the neck towards the center.
For peak velocity, the results were more diverse. Absolute agreement as well as consistency of SumSB correlated with SB was excellent, correlating MB2 with SB gave good to excellent agreement (lower bound above 0.75 and upper bound above 0.9). The retrospectively accelerated MB2 dataset showed poor to excellent agreement for peak velocity (lower bound below 0.5 and upper bound above 0.75) and good to excellent consistency (lower bound above 0.75, upper bound above 0.9). All tested ICCs for both total flow and peak velocity as well as all methods compared to SB accepted the null hypothesis that ICC is similar to 1.

The blood flow was visualized using streamlines at the timepoint of fastest velocity during the cycle, which is illustrated for all eight subjects exemplary for the MB2 acquisition in Supporting Information Figure S1. The 0.8 mm isotropic resolution allowed resolving the small features of deranged flow especially at the bifurcation, such as the helical flow pattern in the left CA of subject 4.

### DISCUSSION

The presented work demonstrates a proof of concept for performing SMSb 4D flow MRI acquisitions targeting the carotid arteries at 7T. We demonstrated excellent agreement and consistency of total flow compared to single-band...
We also showed good to excellent absolute agreement and consistency of peak velocity compared to single band, except for the retrospectively accelerated multiband dataset. The study benefited from multiple technical aspects. First, from the high magnetic field strength of 7T, which enhances SNR by more than 2-fold as compared to 3T.35 Second, the presented work also benefited from using a local, tight-fitting TX/RX coil maximizing the receive signal due to the short distance between the elements and the carotid bifurcation. Additionally, the coil provides a distinct $B_1^+$ profile that allows using a sagittal slab orientation while making use of the inflow effect. Third, by using multiband acquisitions a nominal acceleration (neglecting the calibration scan) of a factor of two can be achieved. We successfully combined all these technical aspects allowing velocity and flow quantification simultaneously in both carotid bifurcations. Similar blood flow values were obtained using multiband compared to single-band high-resolution 4D flow MRI acquisition.

Whereas previous studies have investigated the anatomy and the relaxation times of the carotid vessel wall at 7T,36 the local hemodynamics of the carotid arteries using 4D flow MRI had not been studied at 7T, to the best of the authors’ knowledge. To investigate the carotid wall, resolutions of 0.6 to 0.7 mm isotropic have been used for anatomical scans at lower field strength, which may not be sufficient for characterizing plaque.37 A few 4D flow MRI studies targeting the carotid arteries have been performed at lower field strength with reduced or similar spatial resolution of $(0.7 \times 0.7 \times 1.0)$ mm$^3$, $(1.1 \times 0.9 \times 1.4)$ mm$^3$,12 and $(1.2 \times 1.8 \times 1.8)$ mm$^3$ at 3T.2 In a study by Cibis et al39 a high isotropic, noninterpolated resolution of $(0.625 \times 0.625 \times 0.625)$ mm$^3$ has been applied at 3T but with reduced temporal resolution of 138 ms. Although the present work did not aim to achieve the highest possible resolution, high spatial resolution, and thus, high SNR is required for accurate quantification of hemodynamics, which can be obtained by imaging at ultrahigh fields.35 The combination of high-resolution 4D flow imaging, which is required

**FIGURE 6** (A) Maximum intensity projection (MIP) through the slab and through time of the leaking signal magnitude scaled by a factor of 5 (left) and the passing (right) signal magnitude of the subject. (B) Leaking and passing signal phase of the central slice covering the left carotid bifurcation of the same subject. (C,D) 99th percentile leakage normalized by the mean noise level (C) and mean leakage normalized by the mean noise level (D) for all 8 subjects.
to obtain less error in quantitative hemodynamic parameters, and high-resolution imaging of the vessel wall could be highly beneficial for not only characterizing plaques, but also the local hemodynamic forces.

The use of a sagittal slab with readout oriented along the head-foot direction is less beneficial in carotid 4D flow as it yields lower signal magnitudes within the vessels due to increased saturation of the inflowing blood. Although the orientation is less essential for clinical applications that apply contrast agents, which increases SNR and reduces the inflow-sensitivity, previous noncontrast enhanced 4D flow acquisitions at lower field strength targeting the carotid bifurcations utilized axial or axial oblique slices oriented perpendicular to the carotids to maximize the inflow effect.

**FIGURE 7** Bland-Altman diagram (horizontal black line = mean difference, horizontal red lines = lower and upper limits of agreement [LOA]) to investigate bias in total flow (A–D) between MB data and original SB data for all volunteers at all three locations (left or right carotid artery [CA], left or right internal CA, and external CA). Panels E–H show the Bland-Altman plots for peak velocity comparison between MB methods and original SB acquisitions. The provided values are determined by the median value of multiple cut-planes spaced 1 mm apart along each vessel. Solid black lines indicate the mean difference of total flow/peak velocity, the red lines the limits of agreement (1.96-fold standard deviation)

**TABLE 2** Summarized results of Bland-Altman analysis for total flow and peak velocity

|                          | Mean Difference | Average Mean (method 1, method 2) | LOA upper | LOA lower |
|--------------------------|-----------------|----------------------------------|-----------|-----------|
| Total Flow in ml/cycle   |                 |                                  |           |           |
| MB2 vs SB                | 0.12            | 2.45                             | 0.90      | −0.66     |
| MB2-GRAPPA2 vs SB        | 0.01            | 2.39                             | 0.73      | −0.71     |
| SumSB vs SB              | −0.02           | 2.38                             | 0.41      | −0.45     |
| Peak Velocity in m/s     |                 |                                  |           |           |
| MB2 vs SB                | 0.028           | 0.49                             | 0.126     | −0.071    |
| MB2-GRAPPA2 vs SB        | 0.052           | 0.50                             | 0.164     | −0.060    |
| SumSB vs SB              | −0.003          | 0.47                             | 0.048     | −0.054    |

The mean difference between the two methods compared as shown in the Bland-Altman plot, is given by the mean of (method 1 – method 2) of all measurements over all subjects. The mean of method 1 and method 2 averaged over all subjects is denoted as Average Mean (method 1, method 2); LOA, limits of agreement; MB, multiband; SB, single band.
This approach, however, required thick slabs of 50 mm or more to cover the entire volume of interest. In the presented approach, we made use of the distinct excitation profile (cf. Figure 4) of the local TX/RX coil, which enabled not only larger coverage along the head-foot direction but also allowed to reduce slab thickness by a factor of 2.5. Because the axial orientation requires an approximately quadratic field of view (about 100 × 100 mm²), the change in orientation directly translates into a 2.5-fold reduction of scan time, if readout is oriented in head-foot direction. An alternative approach at 7T would be to choose a coronal slab covering both bifurcations simultaneously. Because the extent of the bifurcation of the internal and external carotid arteries are typically oriented in anterior-posterior direction rather than left-right direction, the slab thickness needs to be increased. Averaged over all subjects, we computed a required increase in slab thickness of 40% when choosing a coronal slab. Furthermore, for the coronal orientation, the resulting velocity data might be more biased by swallowing artifacts, however, the effect of swallowing artifacts on the quantification accuracy of hemodynamics has not been investigated to the best of the authors’ knowledge. In some subjects or patients, the carotid arteries show a tortuous anatomy, and in such cases the slab thickness may need to be increased, which would reduce this scan time benefit.

The applied SMSb technique benefits substantially by the arrangement and setup of the RF coil, which can be appreciated by the low signal levels leaking from either side. Due to the small diameter of the four coil loops only the proximal carotid bifurcation is excited, whereas $B_1^+$ barely reaches the distal side. However, the small size of the individual loops of 3.5 cm × 5 cm in combination with the $B_1^+$ shimming approach also resulted in a steep left-right gradient of the FA showing an almost linear decrease with increasing distance from the RF coil. The resulting $B_1^+$ shim setting was optimized based on ROIs covering the left and right bifurcation. Although choosing an undersized ROI may yield variations of the $B_1^+$ pattern between subjects, our approach yielded a consistent $B_1^+$ pattern in all subjects. However, due to the variable location of the subject’s carotid bifurcations with respect to the body surface, a highly variable FA between 5 and 30 degrees across the different subjects was measured despite the similarity of the FA patterns across subjects. An RF coil with larger element size will likely reduce such variations, however, it may also affect the leakage in SMSb imaging and potentially reduce patient comfort as well as flexibility in positioning the coil.

In addition to a modified RF coil such FA variations can also be addressed using a different RF shimming approach that optimizes the resulting $B_1^+$ homogeneity, which, however, would yield reduced $B_1^+$ efficiency, and therefore, would require higher RF power to compensate for the losses. In this work, we therefore rather aimed at maximizing the $B_1^+$ efficiency because conservative power limits of 0.3W for the 6 minutes averaged power limit would have restricted the maximum achievable flip angles.

In this work, the $B_1^+$ distribution was optimized jointly for both slabs and the shim setting was applied after the summation of the two RF pulses for the slabs. This approach results in a common RF waveform for all TX channels that is multiplied by channel-dependent complex factors to adjust for the

### Table 3

Results of ICC Calculation using Single-Rating, Absolute-Agreement and consistency, 2-Way Random-Effects Model

|                          | Lower Bound | Upper Bound | Value (r) |
|--------------------------|-------------|-------------|-----------|
| **Intraclass Correlation (ICC), 95% confidence interval** |             |             |           |
| **Degree of Absolute Agreement** |             |             |           |
| SumSB vs SB              |             |             |           |
| Total flow               | 0.9778      | 0.9930      | 0.9875    |
| Peak Velocity            | 0.9544      | 0.9854      | 0.9741    |
| MB2 vs SB                |             |             |           |
| Total flow               | 0.9160      | 0.9739      | 0.9532    |
| Peak Velocity            | 0.7606      | 0.9482      | 0.8947    |
| MB_GRAPPA2 vs SB         |             |             |           |
| Total flow               | 0.9346      | 0.9790      | 0.9628    |
| Peak Velocity            | 0.3587      | 0.9323      | 0.8244    |
| **Degree of Consistency** |             |             |           |
| SumSB vs SB              |             |             |           |
| Total flow               | 0.9775      | 0.9929      | 0.9873    |
| Peak Velocity            | 0.9540      | 0.9853      | 0.9739    |
| MB2 vs SB                |             |             |           |
| Total flow               | 0.9227      | 0.9751      | 0.9559    |
| Peak Velocity            | 0.8546      | 0.9519      | 0.9158    |
| MB_GRAPPA2 vs SB         |             |             |           |
| Total flow               | 0.9333      | 0.9786      | 0.9621    |
| Peak Velocity            | 0.8196      | 0.9396      | 0.8947    |

Cells colored in light green indicate excellent (>0.9) agreement or consistency. Good agreement (0.75-0.9) is indicated by light blue and poor agreement (<0.5) by light red. The degree of freedom was 47.
phase/gain of the individual channels (here only the phases were adjusted). This approach can be realized using a $B_1^+$ shimming system that does not require full parallel transmit (PTX) capability allowing to play out N independent complex RF waveforms on the N TX channels.\textsuperscript{23} Nevertheless, optimizing the $B_1^+$ shims separately for each of the two bands prior to the summation of the TX channels is expected to result in improved homogeneity while potentially yielding similar FA. Furthermore, since the left carotid coil barely excites the neck’s right half and vice versa, MB pulses could be replaced by two different SB pulses transmitted at two different frequencies. In this case, the first SB pulse is applied to the left coil elements to excite the left carotid bifurcation and the second pulse excites the right bifurcation through the right coil. For the latter approach as well as for the MB approach with separate $B_1^+$ shim solutions for each side a PTX system is required.\textsuperscript{23}

The study has a few limitations, which still prevent a broad usage of this method in clinical application. First, currently the presented neck coil is a prototype coil that limits broad application. Furthermore, while the limited field-of-excitation is beneficial for obtaining strong inflow-signal, the limited FOV of the coil shown in Figure 5B (particularly the phase images) might affect the performance of background phase correction techniques that are used for correcting biases due to eddy currents or concomitant fields. The question if existing correction techniques such as the one reported by Bernstein et al\textsuperscript{41} can be applied straightforwardly to SMSb acquisitions and in how far the leakage impacts such methods remains to be investigated. In this work we refrained from correcting for eddy currents or concomitant fields, though both steps are typically performed in 4D flow MRI to achieve reduced errors in velocity quantification. However, the study was conducted at 7T and applied in the neck, which is close to isocenter, therefore, concomitant fields (which scale quadratically with distance from isocenter and inversely with $B_0$) are expected to be minor. Furthermore, concomitant fields and eddy currents are expected to affect all investigated acquisitions (MB2, SB, SumSB, MB2 GRAPPA2) in the same way apart from temporal changes due to warming of the gradient coil that may affect residual background phases, because MB2 and the two SB acquisitions used identical gradient waveforms during acquisitions and only the RF pulses differed between the three scans. Potential residual changes of the background velocities between MB2 and SB acquisition have been quantified exemplarily in subject 1 by calculating the difference of the background velocities between both acquisitions for the entire background tissue in the volume as well as for selected spatial regions. Differences in the background velocities of static tissue did not exceed 0.1 cm/s.

Another limitation is given by the acquisition order of MB2 and SB scans that were kept constant across subjects. Thus, potential systematic changes in peak velocity or mean flow, which may occur when the subject relaxes during the course of the study would reflect a systematic bias between the resulting values obtained from MB2 and SB datasets. Therefore, to analyze the sole impact of the reconstruction on the resulting hemodynamic values, the SumSB dataset has been investigated.

Due to the overall long scan time including three 15-minute long MB/SB scans, the additional calibration scans for $B_1^+$ shimming, and the anatomical scans, we were not able to acquire an additional 4D flow dataset in axial view, which would have allowed for a direct comparison between the two views. Such a comparison may also provide insight into the question of whether the change in orientation would yield results less biased by swallowing effects, which are known to affect image quality in the neck.\textsuperscript{42}

In conclusion, the presented technique of using a local TX/RX neck coil in combination with SMSb 4D flow MRI at 7T represents an effective step towards studying local hemodynamics within the carotid arteries. This technique could be combined with high-resolution anatomical acquisitions to simultaneously characterize the vessel walls, the composition of carotid plaques, as well as the local hemodynamics. Both anatomical and hemodynamic scans require high spatial resolution and will strongly benefit from the higher SNR and improved parallel imaging performance provided by 7 Tesla.

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**SUPPORTING INFORMATION**

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

**FIGURE S1** Streamlines for all 8 volunteers shown for MB2 datasets at the time point of largest mean velocity within the right and left carotid artery. Velocities are color coded on a scale between 0 and 0.8 m/s with red indicating fastest velocities. Flow patterns can be recognized especially in subject 3 in the right bifurcation and subject 4 in the left bifurcation.

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