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● *Original Contribution*

A COMPARISON STUDY OF VECTOR VELOCITY, SPECTRAL DOPPLER AND MAGNETIC RESONANCE OF BLOOD FLOW IN THE COMMON CAROTID ARTERY

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Abstract—Magnetic resonance phase contrast angiography (MRA) is the gold standard for blood flow evaluation. Spectral Doppler ultrasound (SDU) is the first clinical choice, although the method is angle dependent. Vector flow imaging (VFI) is an angle-independent ultrasound method. The aim of the study was to compare VFI- and SDU-estimated peak systolic velocities (PSV) of the common carotid artery (CCA) with PSV obtained by MRA. Furthermore, intra- and inter-observer agreement was determined. MRA estimates were significantly different from SDU estimates (left CCA: $p < 0.001$, right CCA: $p < 0.001$), but not from VFI estimates (left CCA: $p = 0.28$, right CCA: $p = 0.18$). VFI measured lower PSV in both CCAs compared with SDU ($p < 0.001$) with improved precision (VFI: left: 24%, right: 18%; SDU: left 38%, right: 23%). Intra- and inter-observer agreement was almost perfect for VFI and SDU (inter-observer correlation coefficient: VFI 0.88, SDU 0.91; intra-observer correlation coefficient: VFI 0.96, SDU 0.97). VFI is more accurate than SDU in evaluating PSV compared with MRA. (E-mail: andreas.hjelm.brandt@regionh.dk)

Key Words: **Vector flow imaging, Magnetic resonance phase contrast angiography, Spectral Doppler ultrasound, Common carotid artery, Peak systolic velocity.**

INTRODUCTION

In vascular ultrasound, the severity of stenosis is often based on alternation of velocity and changed flow patterns. The peak systolic velocity (PSV) obtained with spectral Doppler ultrasound (SDU) is the main criterion for classification of stenosis severity in arteries (Grant et al. 2003; Jahromi et al. 2005). SDU is accepted as a useful ultrasound technique for assessment of PSV in the common carotid artery (CCA), and the need for carotid endarterectomy may be based on SDU assessment (Heijenbrok-Kal et al. 2006; Neschis et al. 2001).

Magnetic resonance phase-contrast angiography (MRA) is considered the gold standard for non-invasive cerebral blood flow measurement (Oktar et al. 2006). However, estimation of PSV with MRA is time consuming, the technique is non-mobile, and the evaluation is not performed in real time, whereas SDU is a dynamic, real-time examination, is easily manageable, is mobile and is the first choice in the clinic.

Spectral Doppler ultrasound estimates only the blood velocity along the ultrasound beam direction (axial direction), where the estimate is angle corrected assuming laminar flow parallel to the vessel boundaries with a constant angle over the cardiac cycle. The assumption that a single beam-to-flow for angle correction is sufficient is not true, because *in vivo* flow rarely is laminar (Hansen et al. 2016a, 2016b; Steel et al. 2003). The PSV estimates obtained with SDU are, thus, limited by the angle dependency and the manual applied angle correction, causing high observer variability (Park et al. 2012; Stewart 2001) and a low inter-observer agreement for carotid stenosis evaluation with SDU has been reported (Corriveau and Johnston 2004; Normahani et al. 2015).

Ultrasound vector flow imaging (VFI) is an angle-independent technique for estimating velocity (Jensen and Munk 1998). In contrast to SDU, VFI estimates the axial and transverse velocity components of blood flow, from

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which the vector velocity can be determined. The technique creates a double-oscillating pulse-echo field by manipulating the apodization function during receive beamforming (Jensen 2001; Udesen and Jensen 2006). VFI has been validated in simulation studies and against MRA on flow in the carotid artery with a strong correlation (Hansen et al. 2009a, 2009b; Udesen and Jensen 2006). VFI is less operator dependent than SDU, because no manual angle correction is applied (Pedersen et al. 2012), and higher intra-and inter-observer agreement for VFI than SDU has recently been reported for portal vein velocity estimation (Brandt et al. 2018).

The aim of the study was to compare PSV of the CCA obtained with VFI and SDU to PSV obtained with MRA, and to determine the intra- and inter-observer agreement for VFI and SDU estimates.

METHODS

Ten healthy volunteers (Table 1) with no history of cardiac, vascular or neurologic disease were included after informed consent and approval by the National Committee on Biomedical Research Ethics (Journal No. H-1-2014-FSP-072). The PSV in both the left and right CCA was measured with SDU and VFI in one session. Within an hour before or after the ultrasound examination, MRA recordings of PSV in both the left and right CCA were obtained. All measurements (MRA, VFI and SDU) were performed with patients in the supine position, with 10 min of rest before each measurement. None of the examiners scan carotid arteries on regular basis, but are experienced ultrasound experts for several different applications.

Table 1. Gender and age distribution among the volunteers

No. of volunteers	10
Gender	
Male	8
Female	2
Age (y)	
Range	25–52
Median ± range	30.5 ± 7.5

Vector flow imaging

A conventional ultrasound scanner equipped with VFI (BK5000, BK Ultrasound, Herlev, Denmark) and a linear probe with a frequency range of 2–8 MHz (8 L2, BK Ultrasound) were used to obtain vector velocity data. Vector velocities were displayed in real time on the B-mode image as color-coded pixels depicted by a 2-D color wheel and as small arrows superimposed on the color map. (Fig. 1) While scanning with VFI, the color box was adjusted to cover the lumen of the carotid artery, and the pulse repetition frequency was adjusted to measure the highest velocities without aliasing. Wall filter and color gain were

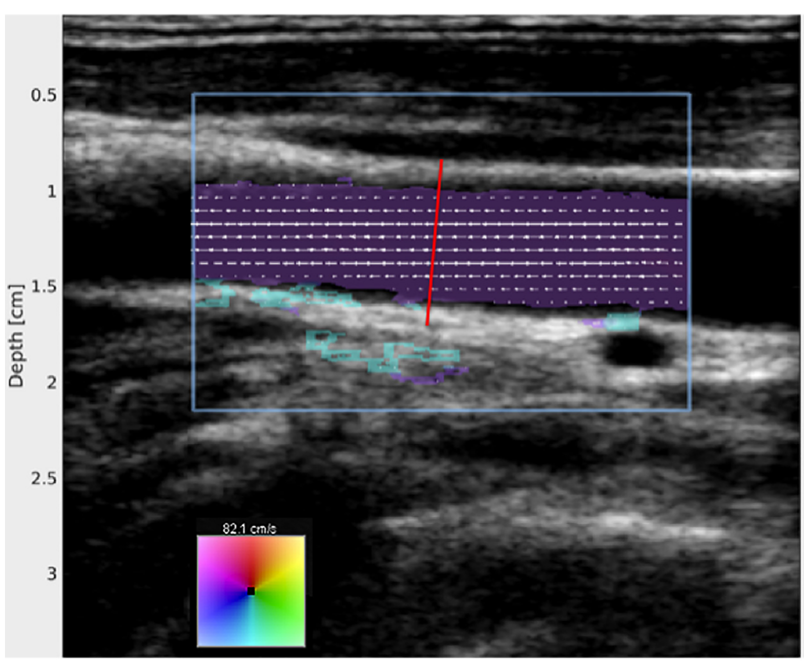

Fig. 1. Setup for estimation of peak flow in the common carotid artery with vector flow imaging. The vector flow imaging peak velocity was found by placing a line perpendicular to the flow direction in the common carotid artery, corresponding to the same position and depth as the range gate placed for the spectral Doppler ultrasound estimation. Direction and velocity magnitude of the blood flow are given by the *color wheel* and indicated by the superimposed *vector arrows*.

set to obtain optimal filling of the vessel without blooming artifacts. VFI measurements were performed by three ultrasound experts (K.L.H., A.H.B., and C.E.) with 10, 3 and 2 y of experience in VFI ultrasound, respectively. Each CCA was evaluated twice in the same session for precision analyses.

Vector flow imaging recordings were processed offline in MATLAB (The MathWorks, Natick, MA, USA) by an in-house developed program (Brandt et al. 2018; Moshavegh et al. 2016). A line was drawn between two points placed on each side of the CCA by the operator. (Fig. 1) The vector velocities were estimated along the line, and the highest vector velocity along the line of the cardiac cycles was selected as the PSV for VFI. The points were drawn at the same position and depth as the range gate for the corresponding SDU estimation, and the same operator (A.H.B.) performed all post-processing of the VFI data.

Spectral Doppler ultrasound

The same three ultrasound experts (C.E., K.L.H. and A.H.B.) performed all SDU measurements. During the scan session, examiners were blinded to the PSV estimation by covering the scale of the spectrum and the estimated velocities. Thus, the examiners were only able to evaluate the spectrogram visually. They had 10, 10 and 5 y of experience in the use of SDU, respectively. SDU data were obtained with the same ultrasound scanner and probe used for the VFI scans using an automatic commercially available standard SDU setup for PSV estimation. (Fig. 2) Hence, SDU peak velocities were obtained directly from the scanner without any post-processing. During scanning with SDU, the range gate was placed in the midlumen of the artery covering a third of the lumen, and angle correction was performed (Tahmasebpour et al. 2005). The pulse repetition frequency was adjusted to the highest velocities without aliasing. Each carotid artery was evaluated twice in the same session for precision analysis, and the PSV, as well as the flow curves, was documented.

Magnetic resonance phase contrast angiography

The MRA measurements were performed within 1 h before or after the ultrasound examination. For one volunteer, MRA measurements were performed 24 h later because of technical issues with the scanner. All MRA scans were performed with a 1.5-T scanner (Magnetom Avanto, Siemens, Erlangen, Germany) and a circular head coil (Neck Coil, Siemens, Erlangen, Germany). A 2-D time-of-flight sequence was performed as a localizer for the carotid artery (repetition time: 42 ms, echo time: 3 ms, flip angle: 20°, field of view: 100 mm, slice thickness: 6 mm, VENC: ± 1.0 to 1–3 m/s, pixel resolution: 1.1×1.1 mm^2 in a matrix of 216×256 pixels, total number of phase intervals fixed at 50), and the flow estimation was found with a through-plane phase-contrast MRA sequence. Two radiologists (C.E. and K.L.H.) performed the MRA examinations. Ultrasound (VFI and SDU) and MRA PSV estimations were performed in the same section of the CCA about 2 cm proximal of the bifurcation. The MRA measurements were oriented perpendicular to the long axis of

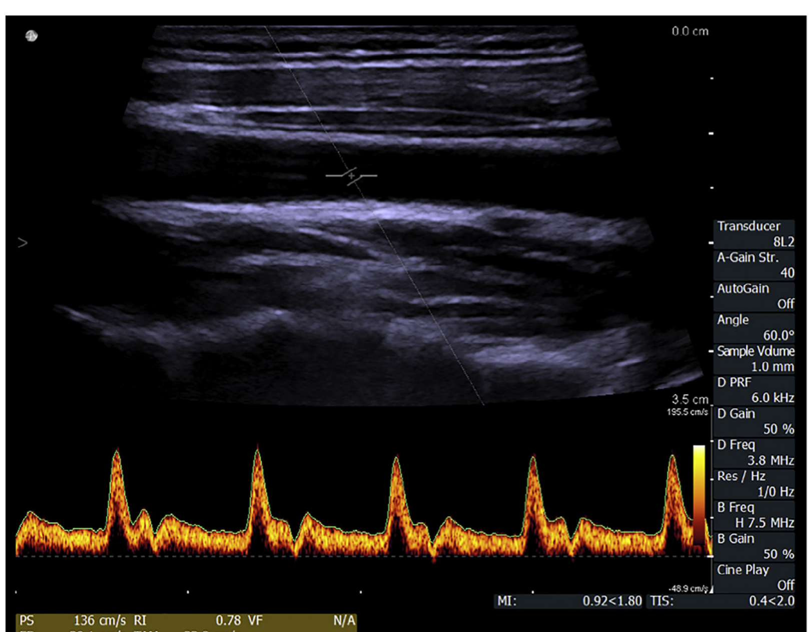

Fig. 2. Standard scanner setup for estimation of peak systolic velocity in the common carotid artery using spectral Doppler ultrasound.

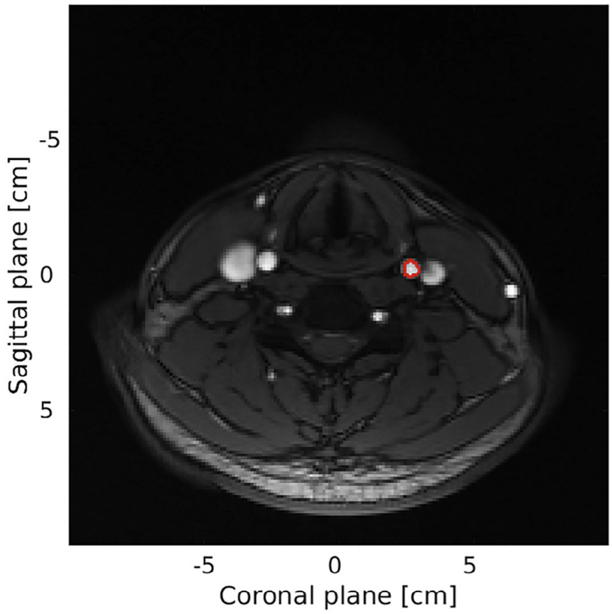

Fig. 3. Transverse 2-D time-of-flight sequence of the neck obtained with magnetic resonance angiography. Within the region of interest (marked in *red*), peak systolic velocity was estimated from a corresponding through-plane phase-contrast magnetic resonance angiography sequence (not shown) as the pixel with the highest signal intensity over all frames obtained.

the carotid artery. Each measurement was performed with electrocardiogram gating, and the MRI measurements were processed offline in MATLAB (The MathWorks, Natick, MA, USA) by an in-house developed algorithm (Holbek et al. 2017). The MRA PSV was interpreted as the velocity value of the pixel with the highest signal intensity in all MRA frames obtained within a region of interest (ROI). (Fig. 3) In contrast to SDU and VFI, only one MRA PSV estimation was performed for each volunteer.

Statistics

The precision p for each method corresponded to two standard deviations (STD) of the difference between replicate measurements a and b of a method x (VFI or SDU PSV estimate), divided by the mean, and was expressed as a percentage:

$$p = \frac{2 * STD\left(x_n^a - x_n^b\right)}{\overline{x}} * 100 \tag{1}$$

Here, n is the replicated experiment number, and $\overline{x}$ is the average of all measurements a and b.

All comparisons for agreement between VFI, SDU and MRA were performed with Bland–Altman plots and linear regression analyses. The second of the two replicated measurements (VFI and SDU) was used for comparisons between VFI, SDU and MRA.

Table 2. Precision for VFI and SDU shown as a mean of the three medical doctors for the left and right common

Method	Precision for replicated measurement (%)
	Left common carotid artery
VFI	24.34
SDU	38.33
	Right common carotid artery
VFI	18.19
SDU	23.24

SDU = spectral Doppler ultrasound; VFI = vector flow imaging.

To examine the limits of agreement (LOA) of the Bland–Altman analyses, the percentage error (PE) was calculated. The PE for each comparison of two methods x and y (VFI and SDU) was calculated as the precision for replicate measurements, that is, 2 STD of the difference divided by the mean of the two methods and expressed as a percentage:

$$PE = \frac{2 * STD\left(x_n - y_n\right)}{\left(\overline{x} + \overline{y}\right)/2} * 100 \tag{2}$$

Here, n is the patient number, and $\overline{x}$ and $\overline{y}$ are the average values obtained for methods x and y. The expected LOA for the Bland–Altman plot of two methods x and y can be calculated as

$$p_{x+y} = \sqrt{\left(p_x^2 + p_y^2\right)} \tag{3}$$

where p is the precision of methods x and y in comparison (Brandt et al. 2016; Moller-Sorensen et al. 2012).

The presence of statistical differences between SDU, VFI and MRA was tested with a paired t-test. A p value < 0.05 was considered to indicate statistical significance. Intra- and inter-observer agreement for SDU and VFI were determined by calculating intra-class correlation coefficients (ICCs), and agreement was interpreted as ≤0 = poor, 0.01–0.20 = slight, 0.21–0.40 = fair, 0.41–0.60 = moderate, 0.61–0.80 = substantial and 0.81–1 = almost perfect (Landis and Koch 1977). MATLAB and SPSS (IBM, Armonk, NY, USA) were used for statistical analyses.

RESULTS

Precision estimates (eqn [1]) for VFI and SDU are listed in Table 2. Mean differences, lower/upper LOA, percentage errors (eqn [2]) and correlation coefficients for the comparisons between VFI, SDU and MRI are listed in Table 3 and illustrated in Figures 4–6. The expected LOA (eqn [3]) for the Bland–Altman method were 45.4% for the left CCA and 29.5% for the right CCA.

Table 3. Mean differences, lower and upper limits of agreement, percentage errors and correlation coefficients for comparisons between VFI, SDU and MRI

	Mean difference (cm/s)	Limits of agreement (cm/s)		Correlation coefficient, *R*	Error (%)
		Lower	Upper		
		Left common carotid artery			
SDU vs. VFI	43.24 (31.46, 55.01)*	−17.54 (−37.94, 2.86)	104.01 (83.62, 124.41)	0.29	63.70
MRA vs. VFI	−4.49 (−8.81, −0.16)	−26.81 (−34.30, −19.32)	17.83 (10.34, 25.32)	0.59	31.20
MRA vs. SDU	−47.72 (−58.44, −37.01)	−103.03 (−121.59, −84.47)	7.58 (−10.97, 26.14)	0.50	59.36
		Right common carotid artery			
SDU vs. VFI	34.99 (−25.99, 43.98)	−11.43 (−27.01, 4.15)	81.41 (−65.82, 96.98)	0.61	48.97
MRA vs. VFI	−6.20 (−10.67, −1.74)	−29.24 (−36.98, −21.51)	16.84 (9.10, 24.57)	0.67	31.05
MRA vs. SDU	−41.19 (−50.83, −31.55)	−90.96 (−107.66, −74.25)	8.58 (−8.13, 25.28)	0.65	54.27

MRA = magnetic resonance angiography; SDU = spectral Doppler ultrasound; VFI = vector flow imaging.
* Ninety-five percent confidence intervals in parentheses.

Spectral Doppler ultrasound-estimated PSV values were obtained at beam-to-flow angles of 42°–77° (mean: 55.6°, STD: 8.8°), and the mean beam-to-flow angle for VFI was 75.0° (STD: 24.0°).

VFI PSVs were significantly different from SDU PSVs in the left CCA ($p < 0.001$), as well as in the right CCA ($p < 0.001$). MRA PSVs did not significantly differ from VFI PSVs (left CCA: $p = 0.29$; right CCA: $p = 0.18$), but did significantly differ from SDU PSVs (left CCA: $p < 0.001$, right CCA: $p < 0.001$).

The differences between VFI- and MRA-estimated PSVs on the right and left sides were not significant ($p = 0.58$), whereas the difference between SDU- and MRA-estimated PSVs on the right and left sides were significant ($p = 0.04$).

Inter- and intra-observer agreement for SDU and VFI was almost perfect (Table 4). The overall intra-observer agreement was 0.97 (95% confidence interval: 0.77–1.00) for SDU and 0.96 (0.75– 0.99) for VFI.

DISCUSSION

This study evaluated VFI and SDU for PSV estimation in the CCA. Both methods were compared with MRA to have an independent method for accuracy determination. VFI-estimated PSV was lower in the left (mean bias: 43.24 cm/s, $p < 0.001$) and right (mean bias: 34.99 cm/s $p < 0.001$) CCA compared with SDU-estimated PSV. VFI measurements were more comparable to MRA estimates (left: $p = 0.28$; right: $p = 0.18$) than SDU estimates (left: $p < 0.001$; right: $p < 0.001$). Furthermore, VFI was more precise than SDU in both the left and right CCA (Table 2), and the correlation between VFI and MRA was slightly higher than the correlation between SDU and MRA (Table 3). This indicates that VFI is more accurate in estimating PSVs in the CCA compared with SDU, when MRA is used as a reference.

The LOA of a Bland–Altman plot in a comparison study should not be wider than the expected LOA and, also, <30% (Critchley and Critchley 1999). The expected LOA of VFI compared with SDU were 45.4% for the left CCA and 29.5% for the right CCA. The percentage errors for both left and right CCA were above the expected and >30%. Therefore, VFI and SDU cannot be considered interchangeable in this study (Table 3), possibly because of the SDU assumption that the beam-to-flow angle is constant over the cardiac cycle. The beam-to-flow angle can be correctly estimated in perfectly laminar flow conditions with SDU, but becomes difficult to determine in complex flow conditions (*e.g.,* in irregular vessels, large vessels and vessels with side branches) (von Reutern et al. 2012). A deviation of ±5° from the true angle can result in a bias of 15% for the SDU PSV estimation (Park et al. 2012) and errors of 10%–100% are associated with the inability of the single-beam Doppler method to measure the true direction of flow (Hoskins 1999a, 1999b). Furthermore, contrary to VFI, SDU ignores the spectral broadening effect, where it is assumed that the PSV can be derived from the maximum Doppler frequency shift. A single velocity gives rise to a range of Doppler frequencies and causes SDU velocity estimation bias at any beam-to-flow angle (Hoskins 1999a, 1999b; Steel et al. 2003; Steinman et al. 2001; Yang et al. 2013). VFI can detect the full angle divergence over a cardiac cycle and can differentiate between laminar and complex flow (Hansen et al. 2016b; Pedersen et al. 2014), which may cause VFI velocity estimates to be more accurate. Furthermore, the flow profile may be determined with a higher level of confidence with VFI than with SDU,

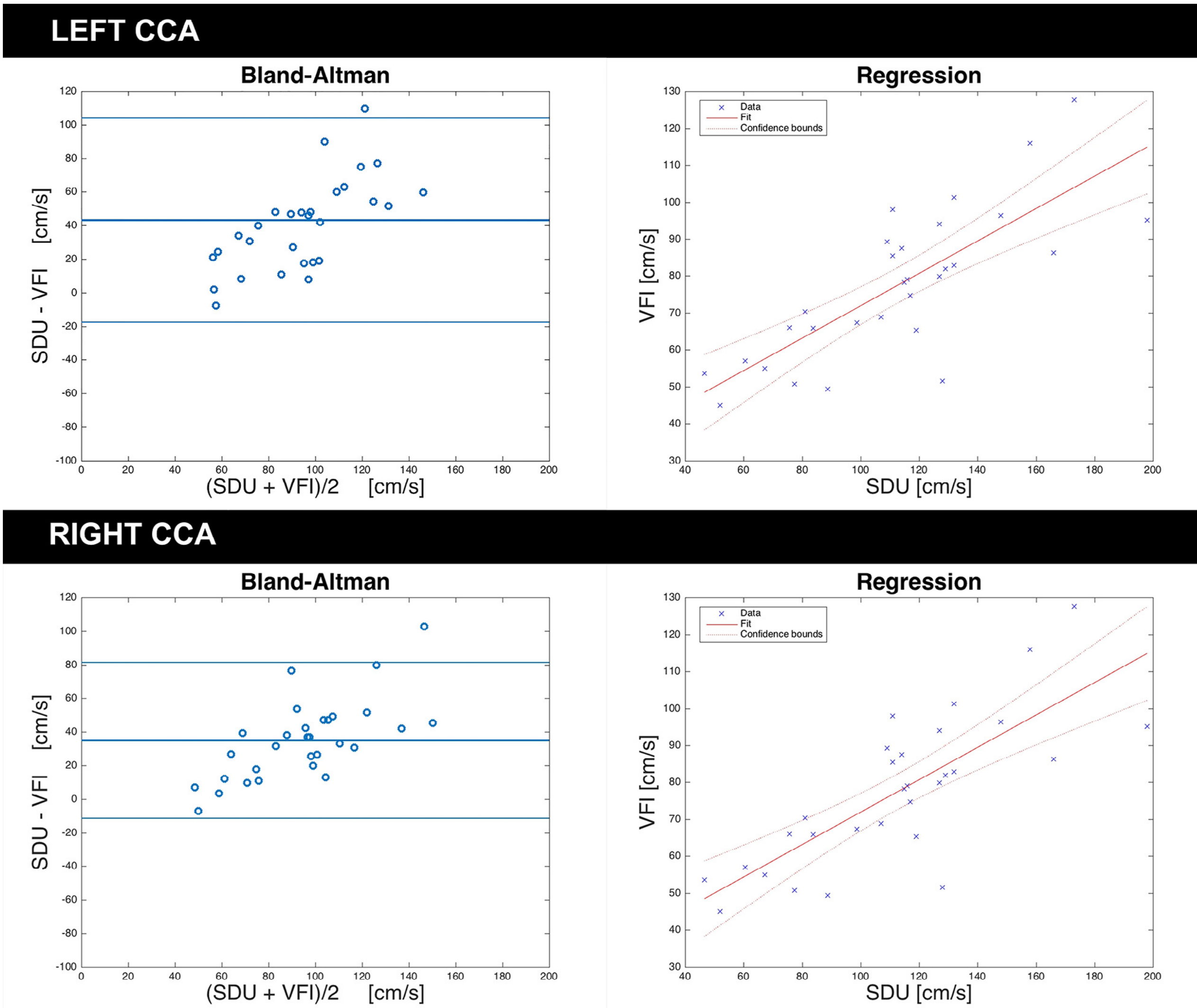

Fig. 4. Evaluation, with Bland–Altman and linear regression plots, of vector flow imaging- and spectral Doppler ultrasound-estimated peak systemic velocities for the right and left common carotid artery. The lines in the Bland–Altman plots (left) correspond to the mean bias and limits of agreement; lines in the linear regression plots (right) correspond to best fit and confidence bounds. Note the larger positive bias found for higher peak systolic velocities for both the left and right common carotid artery.

because small arrows are displayed in real time on the color-coded pixels for flow profile interpretation. VFI was more precise than SDU (Figs. 4–6), so the main bias between the methods may probably be found in the SDU estimation as previously stated (Brandt et al. 2018).

Both VFI and SDU had almost perfect inter-observer agreement and almost perfect overall and individual intra-observer agreement (Table 4). Almost perfect agreement for PSV estimation has previously been reported for SDU and another vector technique approach (Tortoli et al. 2015). VFI may therefore be an alternative for PSV estimation in the CCA. Furthermore, it should be noted that the examiners had more experience with VFI than with SDU, which may indicate that less practice is needed to obtain reliable estimates with VFI. Inter- and intra-observer agreement for SDU PSV was previously found to be lower in patients with stenoses (Corriveau and Johnston 2004; Normahani et al. 2015), and a future study on inter- and intra-observer agreement with VFI in patients with carotid stenosis is recommended.

The precision results for VFI are in line with previous studies determining the reproducibility of vector techniques (Steel et al. 2003; Tortoli et al. 2015). VFI can measure blood velocity with high reproducibility (Brandt et al. 2016, 2018; Hansen et al. 2014), even higher than that for SDU (Brandt et al. 2018; Steel et al. 2003). SDU measurement requires training. Adjustment of the insonation angle, spectral gain and pulse repetition frequency and positioning of the sample volume are all known sources of operator error (Stewart 2001; Szabo 2014). The manually

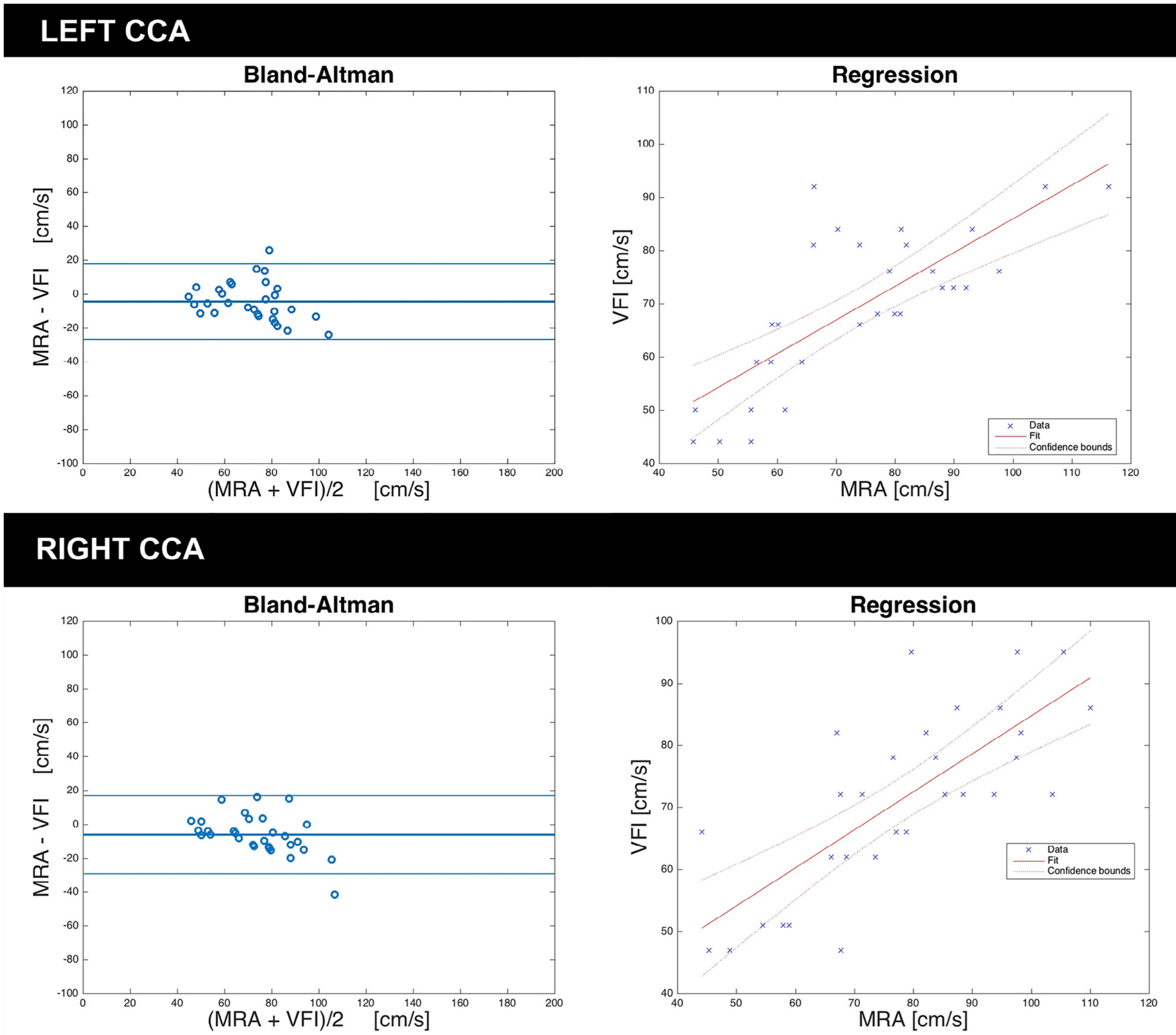

Fig. 5. Evaluation, with Bland-Altman and linear regression plots, of VFI- and MRA-estimated peak systolic velocities in the right and left CCA. The lines in the Bland–Altman plots (left) correspond to the mean bias and limits of agreement; lines in the linear regression plots (right) correspond to best fit and confidence bounds. CCA = common carotid artery; MRA = magnetic resonance angiography; VFI = vector flow imaging.

applied angle correction also causes incorrect alignment parallel to the vessel and is a common error in the velocity estimate (Lui et al. 2005; Zierler et al. 2014). Users with experience in SDU estimation have errors up to 28% in PSV examinations, even on vessel phantoms (Lui et al. 2005). Apart from positioning of the color box and adjustment of the pulse repetition frequency, manual adjustments are redundant with VFI, and VFI may therefore be less operator dependent than SDU, as previously indicated (Brandt et al. 2018; Hansen et al. 2011; Pedersen et al. 2012). Under conditions of turbulent flow in diseased vessels, the direction of flow is seldom parallel to the vessel wall, and therefore, angle correction is impossible. VFI may have an advantage in this setting, which will be pursued in a future VFI study of flow assessment in stenotic carotid arteries.

In evaluation of the CCA, both the left CCA and right CCA are screened for stenosis, because a velocity difference between the left and right sides can indicate a distal stenosis. For all ultrasound examinations performed in this study, the patient was placed in a supine position with the examiner on the right side of the patient. In this standard ultrasound examination setup, the right side was more accessible to examination than the left side. There was no side difference in the MRA examination. Data from both the left and right CCA were obtained with SDU, VFI and

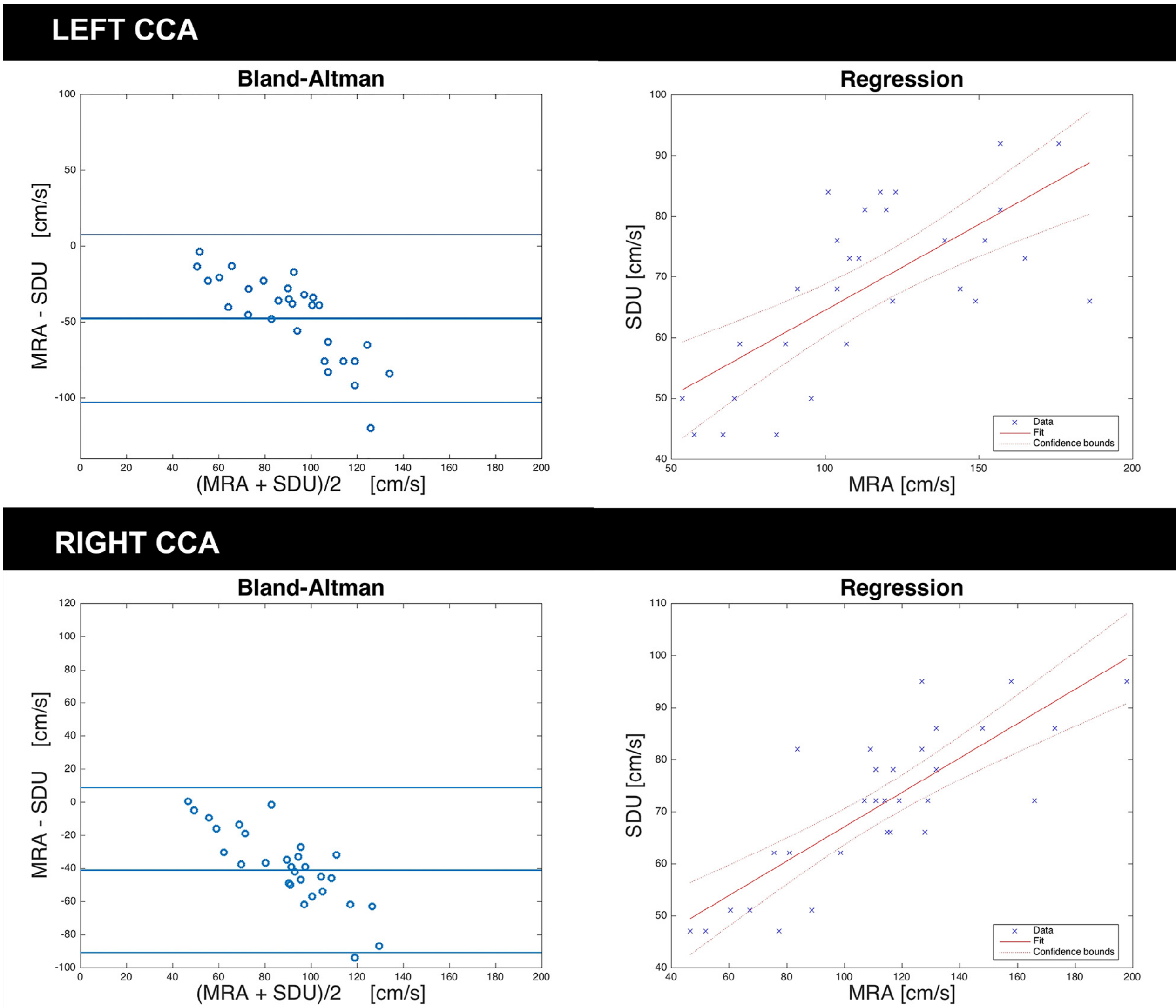

Fig. 6. Evaluation, with Bland–Altman and linear regression plots, of SDU- and MRA-estimated peak systolic velocities in the right and the left common carotid artery. Lines in the Bland–Altman plots (left) correspond to the mean bias and limits of agreement; lines in the linear regression plots (right) correspond to best fit and confidence bounds. Note the trend of increasing SDU overestimation at increasing peak systolic velocity. MRA = magnetic resonance angiography; SDU = spectral Doppler ultrasound.

MRA. Precision for the left side was worse than that for the right side for both VFI and SDU; however, it was more pronounced for SDU. Furthermore, the differences from MRA estimates were significant between the left and right sides for SDU, which was not the case for VFI. A poor scan position may have less impact on the PSV estimate with VFI than that with SDU, indicating that VFI is easier to perform than SDU. A previous study reported VFI to be less operator experience dependent than SDU for portal vein peak velocity estimation (Brandt et al. 2018). This was not the case in this study, probably because our examiners had less variable ultrasound experience compared with the examiners in the previous study, and CCA flow has a higher variance in peak velocity compared with portal venous flow (Brandt et al. 2018).

Stenosis of the CCA has been related to an increased risk of ipsilateral embolic stroke, and it is important to screen patients for symptomatic CCA stenosis with a modality that is accurate, sensitive, reproducible, low in cost and tolerable, such as SDU (Lanzino et al. 2009). SDU is sufficiently accurate for determination of CCA stenosis (Pisimisis et al. 2015); however, MRA is more accurate in determining stenosis where treatment is necessary (Nederkoorn et al. 2003). The correlation between SDU and MRA was moderate, confirming previous studies (Seitz et al. 2001), but higher for VFI and MRA (Table 3). Both

Table 4. Inter- and intra-observer agreement for left and right CCA*

	Intra-class correlation coefficient (95% CI)	
	Left CCA	Right CCA
	Inter-observer agreement	
VFI	0.88 (0.66, 0.97)	0.87 (0.61, 0.96)
SDU	0.89 (0.68, 0.97)	0.93 (0.78, 0.98)
	Intra-observer agreement	
VFI physician 1 (10 y experience with use of VFI)	0.97 (0.91, 0.99)	0.96 (0.86, 0.99)
VFI physician 2 (3 y experience with use of VFI)	0.96 (0.82, 0.99)	0.94 (0.75, 0.98)
VFI physician 3 (2 y experience with use of VFI)	0.94 (0.75, 0.99)	0.98 (0.96, 0.99)
SDU physician 1 (10 y experience with use of SDU)	0.98 (0.92, 0.99)	0.98 (0.93, 1.00)
SDU physician 2 (10 y experience with use of SDU)	0.98 (0.93, 1.00)	0.94 (0.77, 0.98)
SDU physician 3 (3 y experience with use of SDU)	0.95 (0.81, 0.99)	0.98 (0.90, 0.99)

CCA = common carotid artery; CI = confidence interval; SDU = spectral Doppler ultrasound; VFI = vector flow imaging.

* All values are presented as interclass correlation coefficient (ICC) with 95% confidence interval.

SDU and VFI estimated higher PSV values than MRA, which is in line with previous studies (Harloff et al. 2013; Wetzel et al. 2007). For stroke volume estimation of the CCA, MRA and VFI have been found to not significantly differ, which is comparable to PSV estimation of the CCA reported in this study (Hansen et al. 2009a, 2009b). In line with this study, SDU and VFI have previously been found to differ significantly in CCA PSV estimation (Pedersen et al. 2012), and this was also the case for the comparison of conventional Doppler and VFI in the ascending aorta (Hansen et al. 2015).

Some limitations of this study should be acknowledged. The study was conducted on only 10 volunteers. A larger cohort is warranted to confirm the results. Furthermore, PSV estimates for SDU, VFI and MRA are impossible to detect simultaneously, which can add to the bias between the methods. The VFI PSV estimation algorithm only reported the PSV from one pixel of each recorded sequence. An estimation averaged over several pixels and several heartbeats would have provided a more accurate comparison with SDU, as SDU estimates PSV averaged over approximately three heartbeats. However, a limitation of SDU estimation is the fixed time window (8-s cine loop), which might hold different numbers of pulses for different heartbeats; thus, the PSV reported on the scanner might originate from the averaging of different numbers of pulses. Furthermore, no real-time estimate of peak velocity is provided by VFI. A commercial available real-time estimation scheme is warranted for an unbiased comparison with SDU. The range for SDU was placed in the mid-lumen of the CCA as recommended (Tahmasebpour et al. 2005), whereas the VFI sampling line covered the entire lumen of the CCA. There is a small risk that the highest velocities could be found near the vessel border, which could have given an advantage for VFI. The difference in estimation approach between VFI and SDU should therefore be considered a limitation in the study. Angle corrections >60° degrees were performed for 22 of 120 measurements with SDU, which may have added bias to the estimated velocities. Attaining an angel correction at ≤60° degrees for SDU measurements would have improved the study. As for SDU, VFI estimates blood flow in only two dimensions, which is a limiting factor, as flow moves in all three dimensions. To circumvent this limitation, a 3-D velocity vector flow method has been proposed (Holbek et al. 2017). Alignment with MRA recordings was difficult to obtain. A previous study argued that the distance from the bifurcation should be standardized, because mean PSV can increase 9 cm/s for each centimeter of distance from the bifurcation (Meyer et al. 1997). Even though MRA is considered the gold standard, the method is not flawless, because its lower temporal and spatial resolution compared with both VFI and SDU may add to the bias (Wetzel et al. 2007). The CCA is not a predilection site for stenosis as is the internal carotid artery (ICA). However, the CCA is easier to visualize, and because this was a validation study, the CCA was chosen. In a future study, flow examination of the ICA will be examined in patients with varying degrees of stenosis using VFI and compared with conventional methods to establish the value of VFI in a clinical setting.

CONCLUSIONS

This study indicates that VFI is a useful alternative for velocity estimation in the CCA. VFI estimated the same peak velocities as MRA in healthy volunteers with improved accuracy and precision compared with SDU. Furthermore, VFI examination was easier to perform than SDU with comparable inter- and inter-observer agreement.

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