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FULL PAPER

Enhanced susceptibility-weighted angiography (ESWAN) of cerebral arteries and veins at 1.5 Tesla

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Objective: Enhanced susceptibility-weighted angiography (ESWAN) is a three-dimensional (3D) multi-echo gradient-echo sequence which consists of both magnitude and phase images. This study aims to demonstrate the feasibility of ESWAN for the depiction of both cerebral arteries and veins at 1.5 T by comparing with time-of-flight (TOF) MR angiography (MRA) and MR venography (MRV).

Methods: 13 healthy volunteers underwent both ESWAN and 3D-TOF-MRA examinations. Among them, nine volunteers underwent an additional two-dimensional-TOF-MRV examination. With regard to the ESWAN sequence, both maximum intensity projection (MIP) and minimum intensity projection (mIP) images were reconstructed and compared with MIP reconstructions of the TOF MRA and the TOF MRV.

Results: Concerning the depiction of the constituent segments of the Circle of Willis, as well as A1, A2, A3 (segments of the anterior cerebral artery), M1, M2

(segments of the middle cerebral artery), P1 and P2 (segments of the posterior cerebral artery), the value of the ESWAN MIP was comparable to that of the TOF MRA without regard to visualization of branches, vessel homogeneity and wall irregularities or slight stenosis. ESWAN-mIP visualized more deep cerebral veins than TOF MRV in this study.

Conclusion: By use of either mIP reconstruction of a long echo data set or MIP reconstruction of a short echo data set, ESWAN allows simultaneous visualization of both cerebral veins and proximal segments of intracerebral arteries at 1.5 T.

Advances in knowledge: ESWAN acquires multiple images at different echo times corresponding to different T_2^* weightings, wherein a short echo TOF-MRA data set and a long echo susceptibility-weighted imaging-MRV data set are obtained simultaneously.

MR angiography (MRA) and MR venography (MRV) demonstrate different vascular abnormalities and provide complementary diagnostic assessments of brain diseases. In general, many brain diseases involve both the arterial and venous systems, and detailed information is needed for both to direct clinical therapy. For example, MRA visualizes the feeding vessels of a tumour, and MRV identifies the draining veins of a tumour.¹ Time-of-flight (TOF) MRA, which uses the inflow of fresh, unsaturated blood so that the vessels will appear bright compared with the surrounding tissues, provides relevant details of arterial vasculature. Owing to its significant amount of deoxygenated haemoglobin, venous blood has the feature of paramagnetic properties and consequently a high susceptibility effect. When compared with surrounding tissue, this feature of venous blood will lead to a relatively fast loss of phase coherence of excited spins in veins and a short T_2^* relaxation time. Therefore, heavily T_2^* weighted sequences were better suited for imaging of venous vessels. The susceptibility-weighted imaging (SWI) was first mentioned by Reichenbach et al² as a new technique, where

the loss in phase coherence calculated from phase images was used to weight magnitude images accordingly in order to enhance pixels with higher magnetic susceptibility. Here, SWI refers to the use of magnitude or phase images, or an integration of the two, obtained with a three-dimensional (3D), velocity-compensated, gradient-echo sequence. The SWI-based MRV techniques enable visualization of both large (diameter of approximately 1 mm) and small (in the sub-millimeter range) veins in the brain without the use of an exogenous contrast agent.² Both TOF-based MRA and SWI-based MRV sequences are quite time consuming. The increased scan time for undergoing both MRA and MRV examinations may result in patient discomfort and motion, especially when large volume coverage and high resolution are required.

In this study, a 3D, multi-echo gradient-echo sequence with partial flow-compensation is presented for the simultaneous acquisition of MRA and MRV. The sequence is called enhanced susceptibility-weighted angiography (ESWAN) by GE Medical Systems, Milwaukee, WI. ESWAN acquires

multiple images at different echo times corresponding to different T_2^* weightings, wherein a short echo TOF-MRA data set and a long echo SWI-MRV data set are obtained simultaneously. With multi-echo acquisition, 3D R_2^* maps may also be calculated beside the phase masks using the ESWAN sequence. The R_2^* and phase-value quantification may be used to evaluate the regional increase of the non-haeme cerebral iron concentration in patients with neurodegenerative diseases, such as Parkinson's disease, Alzheimer's disease and multiple sclerosis.^{3,4} ESWAN may be used to investigate oxygen saturation by measuring the phase and R_2^* throughout the entire brain, which is important for stroke and venous thrombosis. The aim of this study was to assess the potential of ESWAN in the depiction of both cerebral arteries and veins using an echo-selective "sum-of-squares" combination, along with maximum intensity projection (MIP) and minimum intensity projection (mIP), in comparison with the widely used non-contrast techniques of TOF-angiography.

METHODS AND MATERIALS

Subjects

After institutional review board approval, informed written consent was obtained from each volunteer. A total of 13 volunteers (5 males and 8 females aged 19–55 years; mean age, 31 years) were subjected to both ESWAN and 3D-TOF MRA examinations. Among them, nine volunteers underwent additional two-dimensional (2D)-TOF MRV examinations.

MRI technique

The volunteers were scanned with a 1.5 Tesla scanner (HD pro-peller™; GE Medical Systems) using an eight-channel phased-array head coil. ESWAN, a 3D multi-echo gradient-echo T_2^* weighted angiography with 11 echoes, was applied under the following conditions: echo time of the first echo (TE1), 5 ms; echo separation, 5.1–5.2 ms; repetition time (TR), 77 ms; flip angle, 30°; receiver bandwidth of 62.5 kHz; slice thickness, 2 mm; matrix, 416 × 356; field of view (FOV), 240 × 192 mm; the number of excitations (NEX), 0.7; flow compensation; parallel imaging array spatial sensitivity encoding technique (ASSET) with an acceleration factor of 2; number of slices, 64; and scan time, 7:54 (min:s). The 3D-TOF MRA was performed under the following conditions: TR/TE, 25/3 ms; flip angle, 20°; receiver bandwidth of 20.83 kHz; slice thickness, 1.2 mm; matrix, 320 × 192; FOV, 200 × 176 mm; NEX, 1; flow compensation; number of slices, 124; number of slabs, 4; zero fill interpolation processing (Zip) 512; Zip 2; and scan time, 6:58 (min:s). 2D-TOF MRV was carried out under the following conditions: TR/TE, 24/4.9 ms; flip angle, 60°; receiver bandwidth of 15.63 kHz; slice thickness/overlap, 1.5/–0.5 mm; matrix, 256 × 160; FOV, 240 × 168 mm; NEX, 1; flow compensation; the number of slices, 180–186; and the scan time, 8:25–8:42 (min:s).

Data post-processing

MR images were post-processed and reconstructed using a computer workstation (SUN ADW 4.2 workstation; GE Medical Systems).

Enhanced susceptibility-weighted angiography data

Both phase and magnitude data were acquired separately for processing.

Enhanced susceptibility-weighted angiography-magnitude maximum intensity projection

MIP images were reconstructed simply with the first echo magnitude images (TE of 5 ms) to depict the arterial vasculature (Figure 1a). The 10-mm thick MIP images in the axial, coronal and sagittal planes were generated and used for analysis.

Enhanced susceptibility-weighted angiography-magnitude minimum intensity projection

First, a new set of heavily susceptibility-weighted magnitude images were integrated by calculating a weighted sum of the images obtained at long echo times (the 5th, 6th, 7th, 8th, 9th, 10th and 11th echoes acquired with TE between 25 and 56 ms). Second, mIPs of the new integrated ESWAN-magnitude images with a thickness of 10 mm in the axial plane were acquired and used for analysis (Figure 1b).

Enhanced susceptibility-weighted angiography-phase minimum intensity projection

The brain extraction algorithm was used to extract the brain region. The unwanted background field effects were removed from the phase images with a high-pass homodyne filter. To do this, a low-pass version of the original image is created using a filter matrix of 32 × 32. Unsuitable echoes (the first, second, third and fourth) were skipped, a complex division of the original images and the low-pass filtered images turned into high-pass filtered phase images, which then could be used in construction of a 3D phase mask by setting the phase values above a threshold of 0° to unity, whereas all the phase values below the threshold and $\geq \pi$ were linearly scaled between zero and unity.⁵ After sum-of-square weighted averaging, the phase images at reserved long echoes (the 5th, 6th, 7th, 8th, 9th, 10th and 11th echoes) were integrated into a set of heavily susceptibility-weighted phase mask. Finally, mIPs of the new phase mask images with a thickness of 10 mm in the axial plane were generated and used for analysis (Figure 1c).

Three-dimensional R_2^* maps

With the acquisition of more than two echoes, more accurate quantification of R_2^* may be expected with the exponential fitting of multi-TE signal decay. In this study, 3D R_2^* maps were obtained by exponential fitting of the last seven echoes with minimized root-mean-squared error (Figure 1d).

Three-dimensional-time-of-flight MR angiography

10-mm thick MIP images in the axial, coronal and sagittal planes were generated to visualize arterial vasculature and used for analysis.

Two-dimensional-time-of-flight MR venography

10-mm thick MIP images in the axial plane were reconstructed to visualize deep cerebral veins and used for analysis.

Data analysis

The respective data sets were analysed collaboratively by two neuroradiologists using a standardized evaluation form. Ratings were arrived at by consensus. The quality of vessel identification was graded using a three-point scale (zero, not identified; one, identified but not continuously; and two, identified continuously). The delineation of the arteries and their segments was analysed on

Figure 1. (a–d) Enhanced susceptibility-weighted angiography data set from a 32-year-old male volunteer. (a) MR angiography maximum intensity projection over 48 slices of integrated magnitude images show that these high-resolution scans depict the usual M4 arteries with edge definition. (b) MR venography minimum intensity projection over 48 slices of integrated magnitude image and (c) the corresponding phase mask show that the venous anatomy is different from the arterial pattern. (d) R_2^* map in the basal ganglia obtained by exponential fitting of the last seven echoes, and the regions of globus pallidus have a much higher averaged R_2^* value than other regions.

ESWAN-magnitude MIPs compared with TOF-MRA MIPs. The numbers of the revealed deep cerebral veins (corresponding to a rating of one or two) on the slice at the level of the foramen of Monroe were compared between the ESWAN-phase mIPs and the TOF-MRV MIPs. Both ESWAN-magnitude mIPs and ESWAN-phase mIPs can be used for the assessment of the venography results. In this study, we chose the ESWAN-phase mIPs owing to the less spatial inhomogeneities and better contrast.

Statistical analysis

According to each major intracerebral artery segment and Circle of Willis constituent segment, the comparison of scores was performed between ESWAN-magnitude MIP and 3D-TOF-MRA MIP using non-parametric two-related-samples Wilcoxon test. A comparison of the numbers representing the quantity of the revealed deep cerebral veins on the slice at the level of the foramen of Monroe between ESWAN-phase mIP and 2D-TOF-MRV MIP was performed using non-parametric two-related-samples Wilcoxon test. A p -value <0.05 was considered to indicate statistically significant difference.

RESULTS

The scores of each major intracerebral artery segment [A1–A5 segments of the anterior cerebral artery (ACA), M1–M5 segments

of the middle cerebral artery (MCA) and P1–P4 segments of the posterior cerebral artery (PCA)] and the Circle of Willis constituent segments of the 13 volunteers were measured on both imaging modalities (ESWAN-magnitude MIP and 3D-TOF MRA). The different vessel segments which were subjected to analysis are detailed in Figures 2 and 3.

3D-TOF MRA was significantly superior than ESWAN-magnitude MIP in visualization of A4 ($Z = -4.243$; $p < 0.0001$), A5 ($Z = -4.123$; $p < 0.0001$), M3 ($Z = -4.123$; $p < 0.0001$), M4 ($Z = -3.162$; $p < 0.05$), M5 ($Z = -4.344$; $p < 0.0001$), P3 ($Z = -2.236$; $p < 0.05$) and P4 ($Z = -2.00$; $p < 0.05$) segments. Concerning the depiction of the proximal segments of ACA (Figure 4a vs Figure 4b), MCA and PCA (Figure 5a vs Figure 5b), the ESWAN-magnitude MIP was comparable with TOF MRA with limitations in the visualization of branches, homogeneity and wall irregularities or slight stenosis. With regard to the visualization of the Circle of Willis, the value of ESWAN-magnitude MIP was comparable with that of TOF MRA without concerning vessel homogeneity (Figure 6a vs Figure 6b).

Nine volunteers underwent both ESWAN and 2D-TOF MRV examinations. ESWAN mIP and 2D-TOF-MRV MIP images were

Figure 2. Median value of the enhanced susceptibility-weighted angiography (ESWAN) magnitude-maximum intensity projection (MIP) vs the time-of-flight (TOF)-MR angiography (MRA) MIP for the depiction of the intracerebral arteries. A1-A5 correspond to the A1-A5 segments of the anterior cerebral artery; M1-M5 correspond to the M1-M5 segments of the middle cerebral artery; P1-P4 correspond to the P1-P4 segments of the posterior cerebral artery. The black bars represent the quality of the depiction of each labelled artery by ESWAN, and the white bars indicate the vessel delineation on TOF MRA using the following three level scores on the vertical axis: 0, not identified; 1, identified but not continuously; 2, identified continuously. Significant differences are asterisked ($p < 0.05$).

reconstructed accordingly. ESWAN mIP and 2D-TOF-MRV MIP resulted in 98 and 51 visualized deep cerebral veins, respectively. A statistically significant difference was found between the two sequences ($Z = -2.38$; $p < 0.05$). In this study, ESWAN mIP (Figure 7a–b) visualized more deep cerebral veins than 2D-TOF MRV MIP (Figure 7c).

Figure 1d shows the R_2^* maps calculated from the last seven echoes in the slice at the level of the basal ganglia. The mean R_2^* in globus pallidus (GP) was much higher than other regions as expected, given the high concentration of iron deposition in GP.

DISCUSSION

Deoxyhaemoglobin in the venous blood acts as an intrinsic contrast agent, causing T_2^* related losses in magnitude⁶ and a shift relative to the surrounding tissues in the phase image.^{7,8} The iron in oxyhaemoglobin is shielded by oxygen, thus the T_2^* and susceptibility effects are only observed in venous blood. This condition provides a natural signal contrast between venous and arterial blood. Hence, MRI with a long TE tends to distinguish arteries from veins. Apart from the T_2^* shortening, the bulk susceptibility shift effect of the deoxygenated blood causes a frequency shift of the protons, which leads ultimately to

Figure 3. Median value of the enhanced susceptibility-weighted angiography (ESWAN)-magnitude maximum intensity projection (MIP) vs the time-of-flight (TOF)-MR angiography (MRA) MIP for the depiction of the Circle of Willis. The black bars represent the quality of the depiction of each labelled artery by ESWAN, and the white bars indicate the vessel delineation on TOF MRA using the following three level scores on the vertical axis: 0, not identified; 1, identified but not continuously; 2, identified continuously. No significant difference was found ($p < 0.05$). ACOA, anterior communicating artery; LA1, left A1 segment; LICA, left internal carotid artery; LP1, left P1 segment; LPCA, left posterior communicating artery; RA1, right A1 segment; RICA, right internal carotid artery; RP1, right P1 segment; RPCA, right posterior communicating artery.

Figure 4. Direct comparison of a sagittal enhanced susceptibility-weighted angiography (ESWAN)-magnitude maximum intensity projection (MIP) (a) vs a sagittal time-of-flight-MR angiography MIP (b) regarding the delineation of the anterior cerebral artery (ACA). The difference in the depiction of the distal segments of ACA, in this case especially, and the insufficient delineation of the A4 and A5 segments by ESWAN is clearly visible. 1, A2 segment of ACA; 2, A3 segment of ACA; 3, A4 segment of ACA; 4, A5 segment of ACA.

a phase difference between venous blood spins and tissue spins. For a two-compartment model containing only blood and tissue, this phase difference is a periodic function of the TE. Partial volume effects can cause a signal cancellation in reconstructed magnitude-contrast images if the protons in the venous blood and in the tissue are out of phase for an appropriately chosen TE.⁹ Veins are dark because of T_2^* losses, whereas arteries are bright because of TOF inflow enhancement. Simultaneous MRV and MRA single-echo approach is inappropriate for 1.5 T and lower fields, whereas it is especially efficient at high and ultra-high fields, because a short echo can be used to reduce flow dephasing in the arteries without affecting the venous contrast. Selecting the proper TE is particularly important in applying the single-echo approach. In fact, TE selection is a compromise because a long TE increases the venous contrast but results in flow-related losses in rapidly flowing arteries, whereas a short TE has a lower degree of SWI.¹⁰ To overcome the defects of the

single-echo approach, a double-echo approach was proposed by Deistung et al.¹¹ The second echo is added to the conventional 3D-TOF MRA pulse sequence for SWI-based MRV data acquisition. Therefore, a short-echo TOF-MRA data set and a long-echo SWI-MRV data set are obtained within a single protocol without prolonging the scan time. This multi-echo acquisition technique can also be applied at 1.5 T. The TE needs to be doubled to achieve a similar susceptibility weighting at 1.5 T compared with that at 3 T.¹² Hence, a long idle duration can be found between radio frequency excitation and data acquisition in the SWI sequence at 1.5 T. This increased idle time in the SWI sequence allows acquisition of more echoes within a TR at 1.5 T.

ESWAN is a 3D multi-echo gradient-echo pulse sequence with partial flow compensation. Multiple phase and magnitude images at 11 echo times were acquired within a single scan. If the first TE was set at 5 ms, then the TE of the eighth echo would be

Figure 5. A direct comparison of an axial enhanced susceptibility-weighted angiography (ESWAN)-magnitude maximum intensity projection (MIP) (a) vs an axial time-of-flight-MR angiography MIP (b) regarding the delineation of the middle cerebral artery (MCA) and the posterior cerebral artery (PCA). The difference in the depiction of the distal segments of MCA and PCA, in this case especially, and the insufficient delineation of the M3 and P4 segments by ESWAN is clearly visible. 1, M1 segment of MCA; 2, M2 segment of MCA; 3, M3 segment of MCA; 4, M4 segment of MCA; 5, P2 segment of PCA; 6, P3 segment of PCA; 7, P4 segment of PCA.

Figure 6. Direct comparison of an axial enhanced susceptibility-weighted angiography (ESWAN)-magnitude maximum intensity projection (MIP) (a) vs an axial time-of-flight-MR angiography-MIP (b) of the Circle of Willis demonstrates a comparable quality of delineation of the constituent segments of the Circle of Willis in this case. 1, Anterior communicating artery; 2, right internal carotid artery; 3, left internal carotid artery; 4, right posterior communicating artery; 5, left posterior communicating artery; 6, right P1 segment; 7, left P1 segment.

beyond 40 ms. In this way, the sensitivity of the ESWAN sequence to susceptibility effects is enhanced by the longer echo times (four echoes acquired with TE between 40 and 56 ms). In our study, ESWAN-MRV images also contain contributions from data acquired at shorter echo times (three echoes acquired with TE between 25 and 35 ms), which leads to a higher signal-to-noise ratio (SNR) in resulting susceptibility images. Although the first several echoes all possess the TOF inflow effect, the flow compensation is only true for the first echo in ESWAN. Accordingly, the ESWAN-MIP images were reconstructed simply with the images of the first echo to depict the arterial vasculature. In this study, we performed the ESWAN sequence, which differs from both the SWI sequence introduced by Reichenbach *et al*² and another 3D multi-echo gradient-echo sequence, the so-called “SWAN”.^{13,14} ESWAN can also create the phase-contrast mask, but unlike the SWI sequence, the ESWAN-magnitude image is not multiplied by the ESWAN phase mask, which is a crucial step to enhance visibility of venous structures in the SWI technique. The ESWAN sequence uses multiple magnitude or phase images with different echo times for the image generation. The first echo applies the arterial inflow effect, whereas longer echoes are responsible for susceptibility effects. The echo-selective “sum-of-

squares” combination technique used in the ESWAN sequence enables any combination of different echoes that might improve the visualization of the TOF effect by keeping only the first echo magnitude images or highlighting the susceptibility effects by eliminating the short echoes. This is significantly different from the averaging data acquired with all echo times used in the SWAN sequence. Furthermore, SWAN lacks phase information, which can be obtained and also processed with the echo selective combination technique in the ESWAN sequence. Concerning the depiction of the proximal segments of the middle, anterior and posterior cerebral arteries, the results of this study showed that ESWAN-magnitude MIP was comparable to TOF MRA. In particular, the visualization results of the Circle of Willis showed excellent agreement between ESWAN-magnitude MIP and TOF MRA. Both ESWAN-magnitude mIP and ESWAN-phase mIP visualized more deep cerebral veins than TOF MRV. As the artefacts in the phase mask caused by the off-resonance effect become more severe at a long TE in regions with severe field inhomogeneity,¹⁵ the echo combination technique could improve the SNR in comparison with a single-echo SWI method by balancing the venous contrast and off-resonance artefacts. Our findings concerning the deep cerebral veins are well in accordance

Figure 7. Enhanced susceptibility-weighted angiography (ESWAN) was superior to time-of-flight (TOF) MR venography (MRV) regarding the depiction of the deep cerebral veins: (a) ESWAN-magnitude minimum intensity projection (mIP), (b) ESWAN-phase mIP and (c) TOF-MRV maximum intensity projection. 1, Anterior septal vein; 2, internal cerebral veins; 3, thalamostriate veins; 4, vein of the lateral ventricle.

with the reported capability of high susceptibility-weighted sequences for the depiction of venous vasculature.^{2,5,9,13} However, further studies comparing the sensitivity and accuracy of several commercially available susceptibility-weighted MRI techniques might be needed to confirm the exact role of ESWAN in clinical routine. ESWAN scan time is significantly faster compared with the sum of both 3D-TOF MRA and 2D-TOF MRV, resulting in significantly decreased patient discomfort and specific absorption rate. The use of simultaneous acquisition of MRA and MRV could eliminate the possible misregistration between the arterial and venous vasculatures, which could be induced by patient motion during acquisition, and it would greatly benefit clinicians to distinguish the exact spatial relationship between the arterial and venous vasculatures at or near the lesions. Such exact co-registration between MRA and MRV could also be beneficial in interventional or stereotactic procedures. ESWAN MIP could depict the Circle of Willis and the proximal segments.

ESWAN MIP could depict the Circle of Willis and the proximal segments of big intracerebral arteries in this group. However, compared with 3D-TOF MRA, the arterial edge in ESWAN MIP is less sharp, and flow-related losses can be found in rapidly flowing proximal arteries, whereas the distal arteries are small and have fewer branches. One of the main reasons is that the slice thickness of ESWAN (2 mm) is much larger than that of 3D-TOF MRA (1.2 mm). Experimental results show that the aspect ratio for the highest venous contrast of a vein perpendicular to the main field is $R:w:h = 1:1:4$,¹⁶ where R refers to the diameter of the vein and w and h are the width and height of the voxel size, respectively. With regard to imaging small venous vessels 0.5 mm in size, a resolution of 0.5×0.5 mm in the x - y plane with a slice thickness of 2 mm will yield the optimal venous contrast for transverse acquisitions. Du and Jin¹² found that the venous contrast is greater with the high-resolution acquisition than with the standard resolution and that small arteries are better depicted in high-resolution MRA. Therefore, arteries can be better depicted without sacrificing the venous contrast at the expense of time by using higher spatial resolution and applying thinner thickness to maintain an appropriate voxel aspect ratio. The voxel aspect ratio can also be altered during image reconstruction. As a result, thinner slices can be available during acquisition, and several slices can be combined to generate a new data set with a resolution of $0.5 \times 0.5 \times 2$ mm and an aspect ratio of 4:1. Another reason is that flow compensation was not applied to all three spatial directions. The incompletely flow-compensated echo also showed spatial displacement of the hyperintense arterial signal, which led to a blurred depiction of arteries. Deistung et al¹¹ showed that insensitivity to oblique laminar flow is achieved in the dual-echo (TOF-SWI) approach using a second complete 3D, first-order, flow-compensated echo. When imaging both the arteries and veins, applying complete 3D flow compensation in acquisition with the multi-echo technique is probably more necessary. Moreover, multiple overlapping thin-slab acquisition improves arterial delineation in TOF MRA because of reduced flow saturation. However, ESWAN implemented only one thick slab acquisition.

This study is the first report on the multi-echo acquisition of MRA and MRV at 1.5 T. Visual assessment of these MRI data by

radiologists has become an important part of patient care in terms of improving diagnosis, prognosis and monitoring, especially for neurological diseases. However, time is a constraint for the construction of a general protocol that covers many different imaging sequences. Consequently, an increasing focus has been given to combining multicontrast and multiparameter protocols in one sequence acquisition. ESWAN, a commercial software used for SWI, is usually installed on GE 1.5 T and 3.0 T scanners. ESWAN has high sensitivity for venous vasculature, blood products and changes in iron content despite lacking the last step of multiplying the filtered phase with magnitude, which is a patented technique. As mentioned in the Introduction section, ESWAN offers the potential to measure multiparameter protocols (phase and R_2^*) throughout the brain. Unfortunately, this feature was only shown in the Results section and not analysed or discussed because it is another complicated issue and is beyond the scope of this article.

The present study has several limitations. First, we used the clinical routine imaging parameters of ESWAN with no regard for the questions about fair comparisons between ESWAN and TOF sequences. Given the scanning time and patients' comfort, we applied parallel imaging and partial Fourier to ESWAN for speeding up the scanning process. As ESWAN data are acquired with a 3D, long range, single slab, their SNRs are quite high. So, we can reduce the scanning time by sacrificing part of the SNR. But if parallel imaging and/or partial Fourier are applied to 3D-TOF MRA or to 2D-TOF MRV, sufficient SNR may not be guaranteed. From the parametric perspective, it might be unjust to say that ESWAN can save quite a lot of time compared with the sum of both 3D-TOF MRA and 2D-TOF MRV. Second, the simple evaluation scores for artery imaging were adopted that gave no information on visualization of branches, vessel homogeneity and wall irregularities or slight stenosis. The agreement between ESWAN and TOF MRA in the depiction of cerebral arteries could be overestimated. Besides, our study was conducted to assess the potential of ESWAN in the delineation of main segments of intracerebral arteries. Indeed, in clinical practice, the detailed evaluation of small vessels and of minor vessel wall changes seems to be more important. The possible clinical implication of ESWAN MRA has to be examined in further studies with a combination of clinical cases. Third, the rating of dural sinuses delineation was not included in our study, because the signals of these structures were complex, which could be low, isointense or high on ESWAN sequences. We may speculate that the heterogeneous signal within the sinuses could be related to a higher blood-flow velocity considering the inherent TOF effect of ESWAN. Moreover, the edges of sinuses were obscured by susceptibility artefacts from adjacent bones. But the exclusion of the dural sinuses in the rating may highlight the advantage of ESWAN and the disadvantage of TOF MRV.

CONCLUSION

Preliminary results of this study demonstrate the feasibility of the simultaneous acquisition of MRA and MRV using a 3D multi-echo gradient-echo pulse sequence with partial flow compensation at 1.5 T. Multiple MRV data sets acquired at different TEs are complementary in depicting deep cerebral veins, which are difficult to detect with conventional MRA techniques

such as TOF or phase contrast. However, the visualization of MRA using ESWAN should be further improved. The contrast in arteries could be increased without overly altering the venography by using a complete 3D flow compensation, a high-resolution acquisition and a thin slab.

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