Virtual Navigator Automatic Registration Technology in Transcranial Application

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Aims and objectives

Ultrasound (US) fusion imaging is an emerging imaging technique which permits real-time US with parallel imaging of a pre-acquired second imaging dataset, e.g. a Computed Tomography (CT), Magnetic Resonance Image (MRI), and/or PET/CT [1,2]. In recent years, this technique has become increasingly used in both diagnosis and image-guided interventional procedures. The main focus of interest so far has been on abdominal US, e.g. the liver and the kidneys [3-11]. More recently, fusion imaging gained interest in diagnostic Neurosonology [12] and Neurosurgery applications [13, 14].

An important precondition for a stepwise distribution of the technique from specialized centers into widespread everyday practice is the ease of its use, i.e. the fact to have a user friendly system interface and workflow pathways which simplify the complexity of the procedure. To achieve this goal, a simple, fast and reliable matching algorithm is required, which subsequently secures a precise real-time fusion between US and the second imaging modality. Current registration procedures usually comprise a manual definition of matching points within the pre-acquired MR or CT imaging volume dataset and the patient or the patient's US image. The second precondition, to keep the matching throughout the examination despite spontaneous movements of the patient, has recently been solved by the introduction of a Motion Control Sensor [15-17]. This sensor, which is attached to the patient, instantly and continuously corrects any voluntary or involuntary movement of the patient, avoiding therefore any loss of matching after the registration procedure.

The present work is a feasibility study and a performance analysis of a newly developed automatic registration algorithm for MRI and US real-time fusion imaging using in vivo registration for the transcranial color-coded duplex US application.

Methods and materials

Study population

Six (6) subjects (5 males, 1 female; age range: 36-86, mean age: 54 years) were included into the study. All had a patent transtemporal bone window for transcranial insonation and routine US yielded normal findings for all intracranial accessible basal cerebral arteries. In all individuals, pre-registered MRI images, obtained for reasons not related to the study, were available. The study was performed in the Ultrasound Lab of the Dept. of Neurology (Charité- Universitätsmedizin Berlin, Germany) using a commercially available US system (MyLabTwice, Esaote S.p.A. Italy), equipped with Virtual Navigator (VN) option [18] and additional local installation of the automatic registration algorithm.

Methods of examination

Brain MRI was acquired from all subjects using a 1.5 Tesla scanner (Siemens Magnetom Avanto, Erlangen, Germany), equipped with a 12-channel head coil. The following sequences were available and used for registration: 1) 3D T1-weighted MP-RAGE for anatomical comparison; 2) 3D-multislab time of flight magnetic resonance angiography with a cranial saturation band, in order to suppress venous and highlight the arterial signal. The parameters of the two sequences were: 1) repetition time (TR)=2650 ms, echo time (TE)=28/113 ms; echo train length=5; flip angle=150°, 50 axial slices with a matrix size=256x256, interpolated to 512x512, field of view (FOV)=250x250 mm^2, slice thickness 2.5 mm; 2) TR=1900 ms, TE=3.37 ms, TI=1100 ms, flip angle=15°, 176 contiguous, axial slices with voxel size=1x1x1 mm, matrix size=192x256, FOV=192x256 mm and slab thickness=176 mm.

Fusion imaging was carried out with the above US system using a Phased Array Probe (Operating Bandwidth: 1-4 MHz, PA240, Esaote) and a reusable tracking bracket with sensor mounted (639-039, CIVCO Medical Solutions, Kalona, Iowa, USA). MRI DICOM data sets were transferred to the US VN via DVD import or via network based import from the hospital PACS System (Centricity®, GE medical Systems) prior to the matching procedure.

The setup was with the patient lying in a supine body position with the head positioned within the electromagnetic field; source tip positioned to the right of the subject and pointed towards the head in order to achieve the highest homogeneity of the created field in the US scanning area. Two additional small receivers were used. One, attached to the patient forehead with a plaster strip (Figure 1) provided motion correction information, correcting any of the patient movements after completed registration procedure [15]. The second was mounted on the US probe fixed by a support, providing position and

orientation of the US probe in relation to the transmitter in the created 3D space. Before starting, an accuracy check of the electromagnetic field was performed: the same point coordinates were measured twice by a dedicated registration pen with the electromagnetic sensor mounted in two different spatial orientations. An accuracy of 0.2 cm or less was considered acceptable.

Color-coded US vessel images of the subject's circle of Willis were obtained by transtemporal insonation through the right temporal bone window. Matching of US and the MRI data set was achieved in two workflow steps: manual pre-registration (one plane/ one point) and secondly automatic registration of intracranial arteries. The details of both steps are explained in the following paragraphs.

1- Manual pre-registration (one plane - one point)

Similar axial planes - displaying the circle of Willis - were chosen on the US scan and the TOF-MRI dataset. One-point registration resulted in a first rough data set matching, i.e. moving the US images already yielded in real-time by the US probe with simultaneous navigation within the MRI data volume. Then one identical point well visible on US and MRI - the beginning of the M1-segment of the middle cerebral artery (MCA) - was marked on both image modalities. A subsequent one-plane registration resulted in an improved matching, however only adjusting the spatial error in the X, Y, and Z coordinates of this point (Figure 2). The procedure so far leaves a residual registration error with its magnitude depending on the accuracy of the manual matching point identification of the sonographer but this is a necessary precondition for the following automated registration algorithm [19].

2- 3D Automatic registration algorithm

The algorithm removing the residual shift of the two volumes is based on automated matching of the three-dimensional vessel tree visible in both modalities. For this, a prior automatic segmentation of the vessels was performed. In the TOF MRI sequence, the segmentation was started from the previously marked point as an initial seed (Fig. 3). Vessels were recognized based on their gray level similar to the one of the initial point, and their connection to it for at least a volume chosen by the operator (threshold of the segmentation). The three-dimensional US Color Doppler volume of the brain vascular tree (Fig. 4) was acquired and calculated with a similar mathematical approach with the data set acquired by a short 5 seconds sweep through the sonographic region of interest and the previously defined point as the initial seed. The following automated correction and registration step comprised a 3D-matching of both volumes using only the extracted, filtered and resampled vessels, through the downhill simplex algorithm, which maximizes the common part of the vessel trees.

For each subject, the accuracy of the registration was evaluated qualitatively and quantitatively. The former was achieved assessing the correspondence of the arteries visible on TOF MRI and on CD US (Fig. 5). The latter consisted on measuring the residual distance of the following anatomical points visible with the two modalities in the axial plane (Fig. 6): Anterior Cerebral Artery (A1-ACA), two points in the Middle Cerebral Artery (proximal point: M1-MCA and distal point: M2-MCA), and Internal Carotid Artery (ICA)-Siphon.

We tested if the residual error, measured for all the subjects in the four anatomical points, was statistically different from zero with the signed rank Wilcoxon test.

Images for this section:

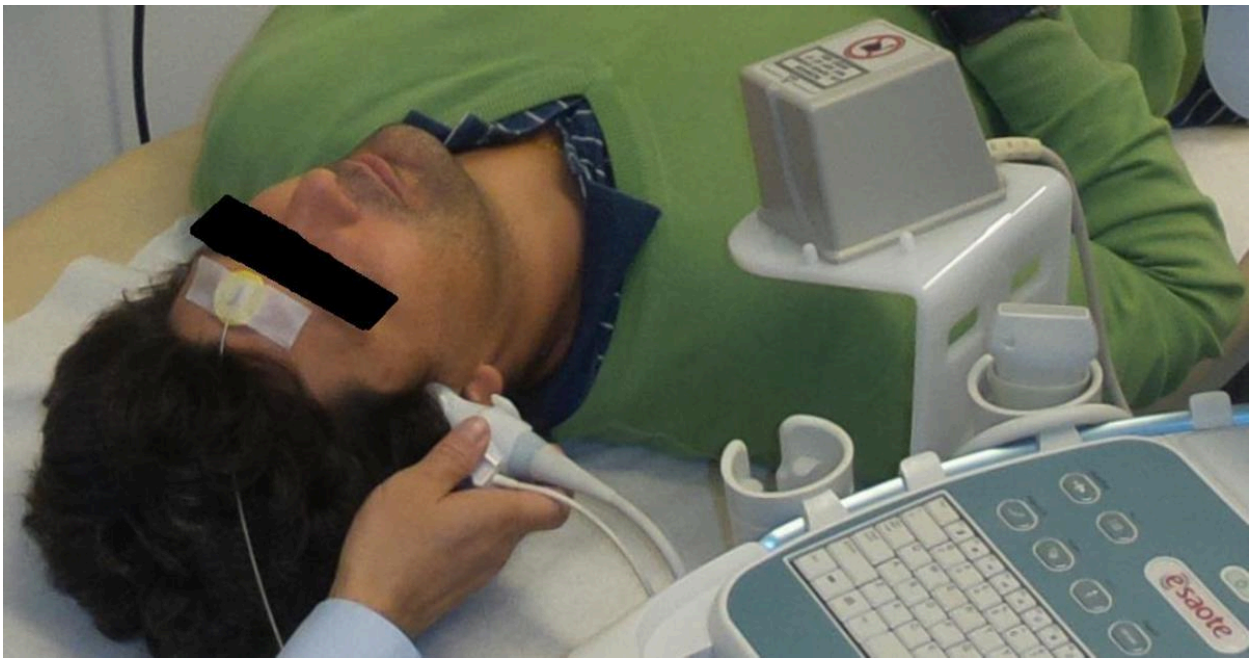

Fig. 1: Virtual Navigator Real-time Fusion Imaging setup with: electromagnetic transmitter placed over its support fixed to the examination bed, electromagnetic receiver applied on Phased Array probe PA240, Motion Control Sensor on the examined subject's forehead

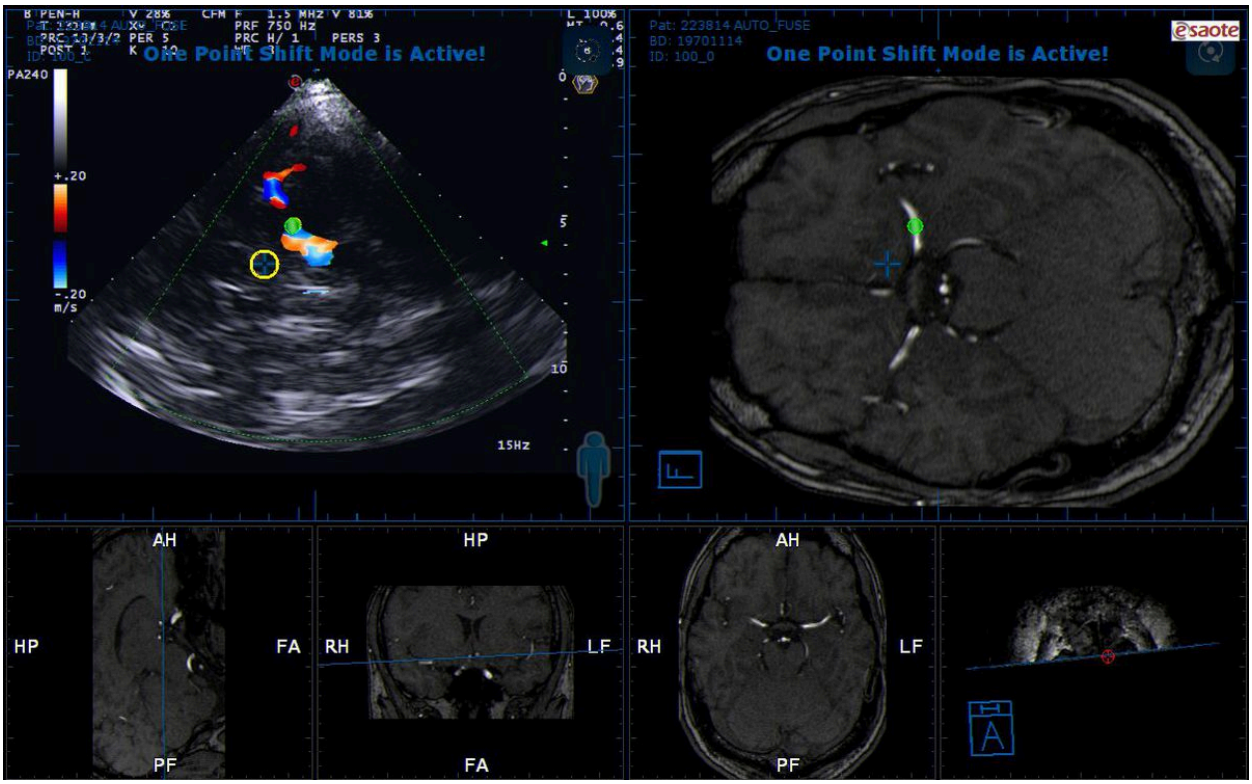

Fig. 2: Visual control of the correspondence of anatomical structures on US and MRI in axial view during the manual pre-registration step. The green point resembles the chosen marker point within the proximal M1-MCA vessel segment on both modalities.

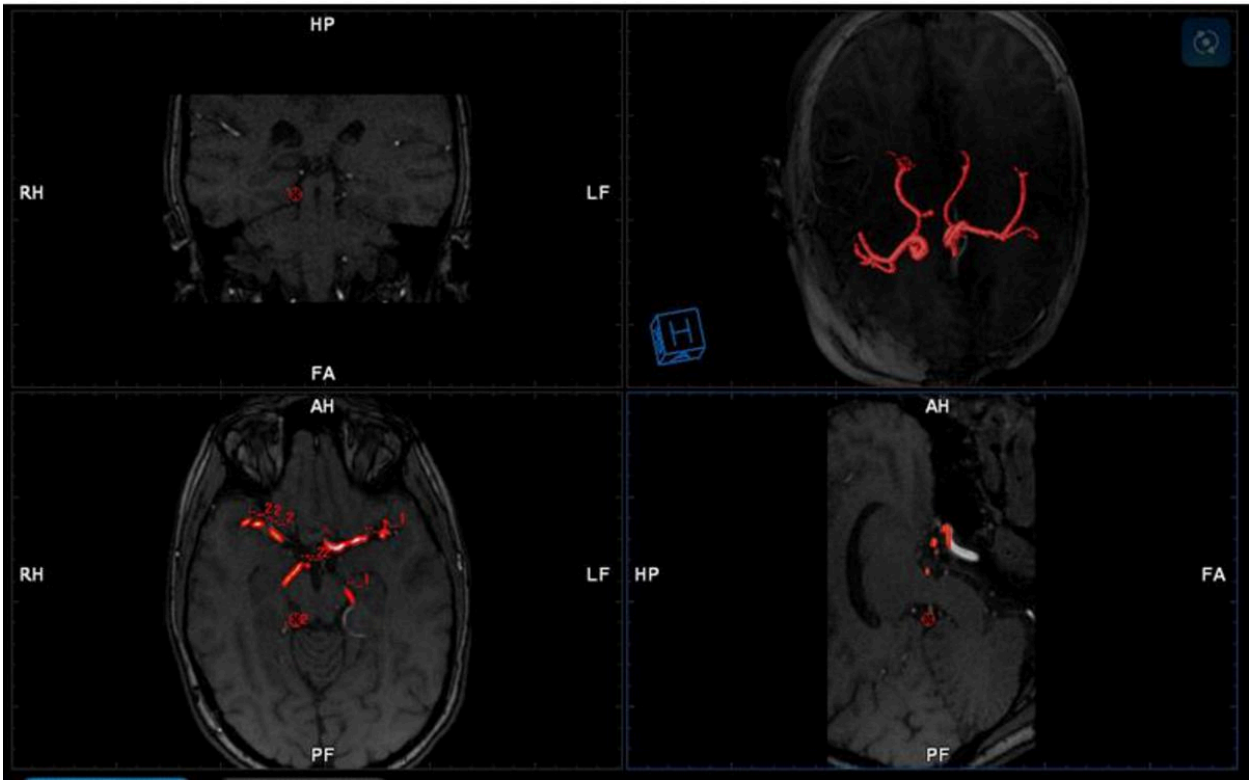

Fig. 3: Automatic registration algorithm - view of the calculated 3D vascular tree derived from the TOF MRI data after the segmentation process.

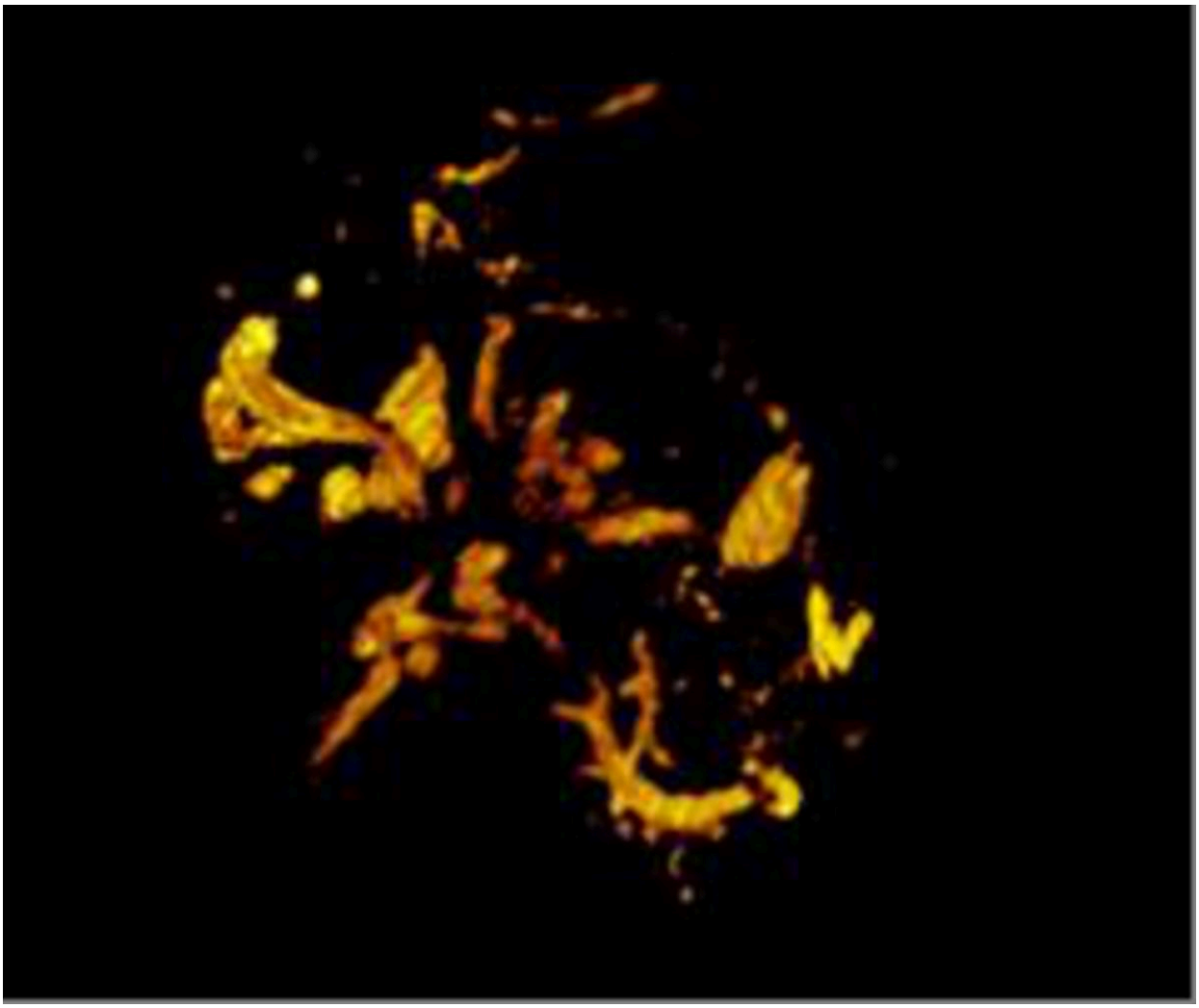

Fig. 4: Automatic registration algorithm - view of the calculated 3D vascular tree derived from a short US Color Doppler sweep through the region of interest.

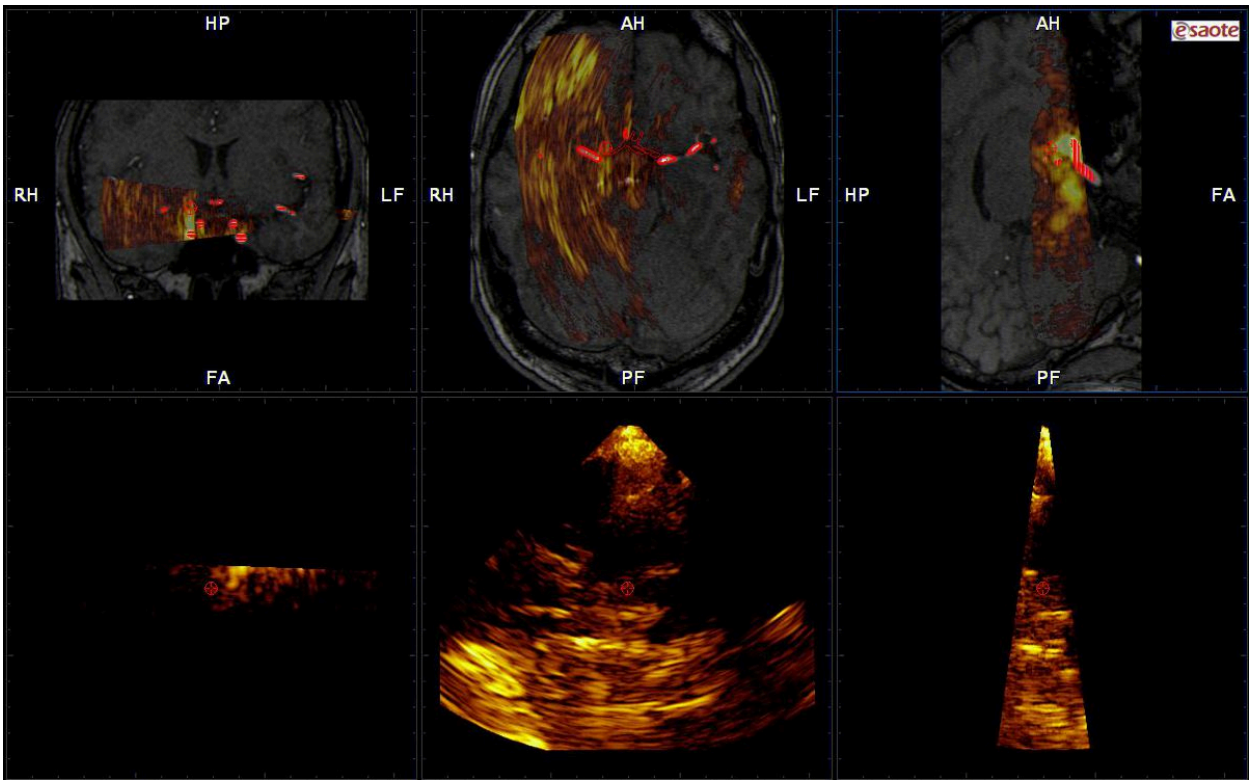

Fig. 5: Automatic registration algorithm - view on the superimposed and already matched images of US and TOF MRI derived images after the completed automatic registration

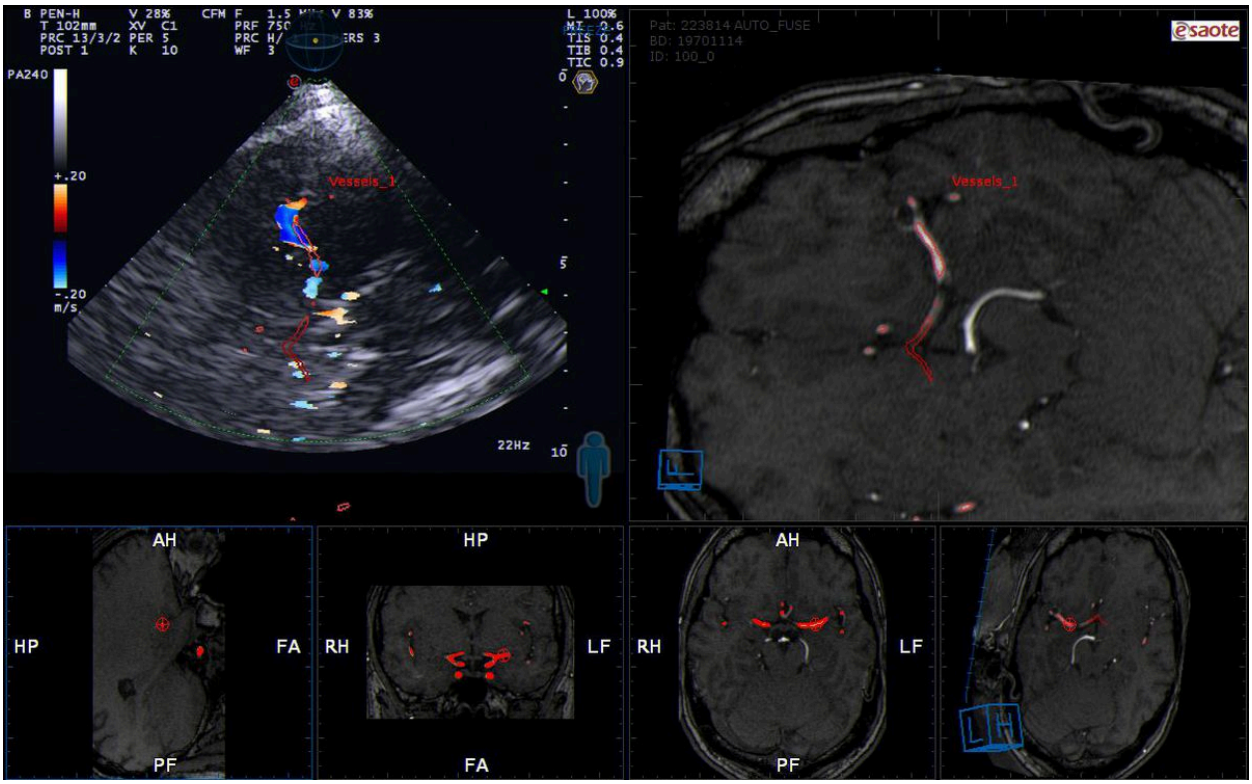

Fig. 6: Qualitative evaluation of the automatic registration accuracy by comparison of both image modalities and automated overlay of MR-derived vessel borders onto the live US image.

Results

All six chosen individuals could successfully be matched and subsequently be evaluated. After implementing the workflow, the time needed to complete the registration steps was less than 5 minutes before routine insonation with matched imaging could be started (time for MRI data download not included). The accuracy of registration performance was tested ipsilateral to the US acquisition, which was obtained through the right temporal US window (Fig. 7).

In more than 70% of the cases the residual error in the points of reference was 0 mm, in one case it was 3 mm and in the remaining cases it was 1 mm. The error was not significantly different from 0 mm (p-value=0.8). In the following Table the detailed measurements for each subject are reported (in mm), the axial imaging plane is considered, with ipsilateral measurements.

Subject	Gender	Age	M1-MCA	M2-MCA	A1-ACA	ICA-Siphon
1	male	36	0	1	1	0
2	male	66	0	3	0	0
3	female	47	1	0	1	0
4	male	86	0	0	0	0
5	male	47	0	1	0	0
6	male	44	0	0	not visible	0

Images for this section:

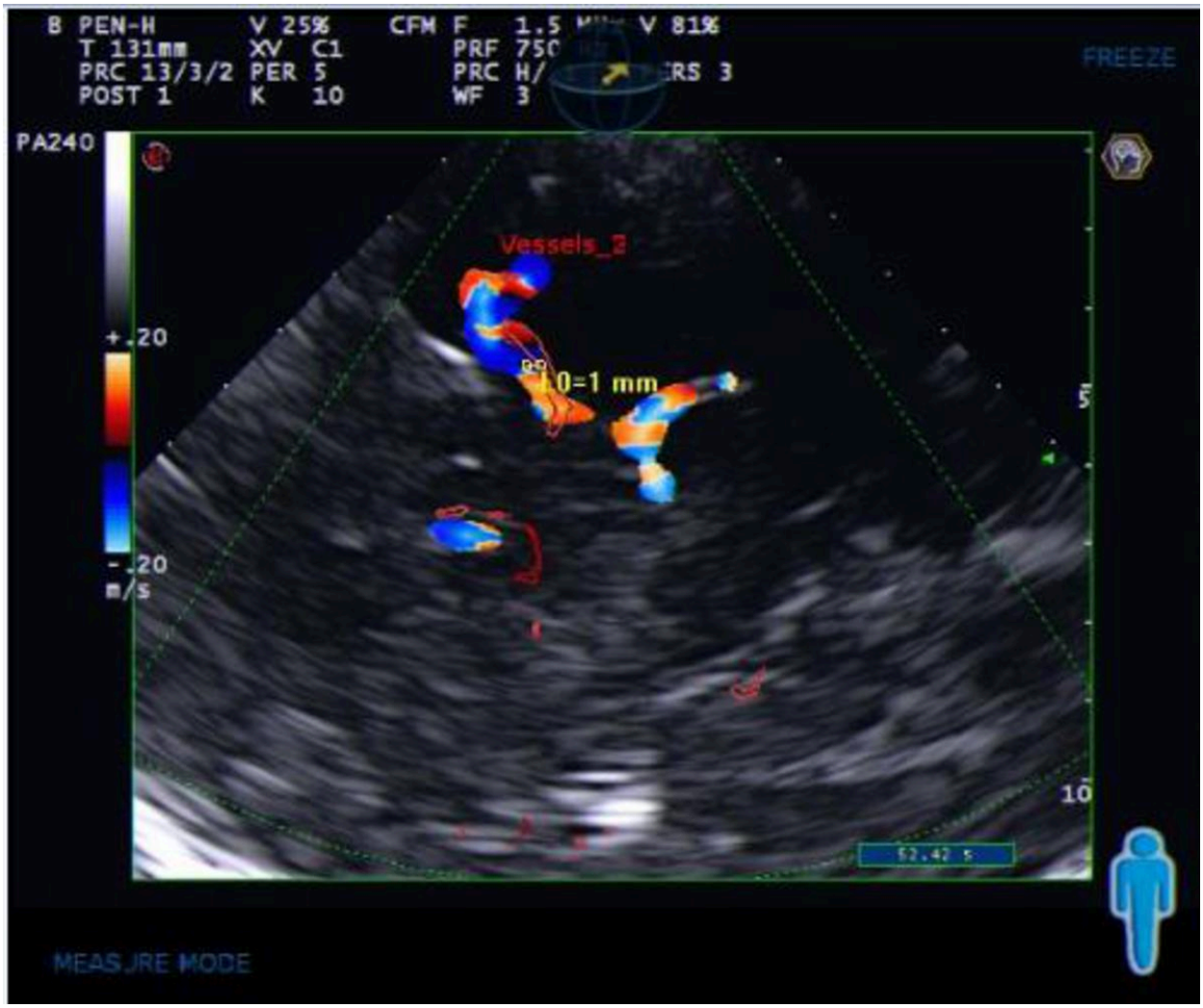

Fig. 7: Quantitative evaluation of the automatic registration accuracy. Distance between M1-MCA visible on US and the same point visible on MRI. Proximal M1-MCA differing by approximately 1 mm

Conclusion

Our data demonstrate the feasibility of introducing a new automated registration protocol into the workflow of transcranial US fusion imaging. The automatic registration algorithm worked in all examined subjects with excellent positional correspondence in major intracranial color Doppler-accessible vessel segments, i.e. the A1-ACA, M1-MCA, M2-MCA and ICA-Siphon.

Time needed for the registration procedure was, despite the use of a study workflow only, not longer than the time needed for the conventionally used external or internal marker matching procedure. A detailed analysis was not performed because of the feasibility character of the current study and would only be sensible after optimized workflow integration into the matching procedure.

By achieving the two above aims, the Virtual Navigator Automatic Registration Algorithm has the potential to become a valuable solution for routine insonation and follow-up sessions performed even by different operators with different skills. An expert, but especially a non-expert examiner, who is not trained in manual matching procedures, will be able to perform precise tuning on the small vessels of the described region. Indeed, the residual error after the automatic registration was zero in most of the cases, otherwise of 1 mm and it was of 3 mm in one case only. As a reference, the average dimensions of the considered vessels were between 0.5 and 2.5 mm for A1-ACA, M1-MCA and M2-MCA with the ICA-Syphon in average had a diameter of 2.5mm at the insonation level considered. Although not formally analyzed and addressed in the current pilot study, this approach not only increases the confidence of image alignment achieved by the technique but also has the potential to improve repeatability of matching results for follow-up investigations.

Compared to previously described registration procedures, which consisted in initial fiducial markers registration, with the registration procedure described in this work [20], there is no need to point external landmarks on the patient's face, and neither to move the receiver from the Registration Pen to the Probe.

Choosing a transcranial insonation paradigm, the proposed Automatic registration algorithm showed to be able to work in rather difficult insonation conditions. Transcranial US insonation frequencies are low, resulting in a low correspondingly spatial resolution. The success in these conditions suggests an even easier application if used for insonation of other - e.g. extracranial - organ systems, provided that vascular tree is visible within the field of interest. Also, the chosen automated segmentation of MR derived vessels

should be transferrable to other organ systems and even other image modalities, e.g. a vascular tree derived from a CT-angiographic data set.

Currently, the 2-step approach of initial manual pre-registration is required to reduce the demand on computational power for the second, automated part of the registration process. Future optimization might be able to proceed even without this extra step within the workflow.

A number of potential pitfalls related to the proposed algorithm has to be considered:

- First, visualization of one or more vessel bifurcations is mandatory to reach an optimal matching result. Consequently, the presented solution will not work in typical muscle, skeletal, prostate, gynecological US due to the lack of clear vascularization landmarks. Currently, the automated vessel recognition and segmentation only work on white vessel signals, so a TOF MRI sequence or a post-contrast T1-MRI sequence are required.
- Second, vessel size between the two datasets varies. MR-derived vessels are small - US Color Doppler derived vessel are larger, depending on system settings like persistency, smoothing level, possible movements and general acquisition errors of the US system. This discrepancy can lead to an incorrect alignment of the US and MRI vessel as the mathematical overlapping region between the vessel tree has more than one "best fit". Although not formally analyzed, this phenomenon is most likely the reason for the observed discrepancies between our analyzed data sets.
- Third, the matching outcome was satisfactory for vessels analyzed in the area ipsilateral to the side of the registration procedure. However, if contralateral vessels were measured, the observed error increased up to more than 5mm (data not shown). Possible reasons of this mismatch could be:
- Persistence of the Color Doppler signal - the longer the distance the US has to travel, the higher the persistence effect and subsequent time-space delay, the larger the mismatch.
- Geometric deformation of the US beam caused by beam configuration of the Phased array probe but also caused by bone window induced beam distortion increases with rising insonation depth.
- Depth dependent loss of Color Doppler signal intensity and spatial resolution with subsequent depth dependent varying quality of automated vessel tree calculation. This phenomenon might potentially be overcome by the use of intravenously applicable US Contrast media (provided that there are no contraindications).
- Technical accuracy - the electromagnetic antenna technology is guaranteed by the manufactures for 1.5mm maximum accuracy only.
- Inhomogeneity of the electromagnetic field - application of the technique in a typical hospital setting implies the nearby presence of metallic structures, other electric instruments or wires which might all result in field distortion.

This phenomenon will even to some degree be caused by the insonated object itself, lying in the field and by the moving insonation probe too.

Looking at small structures like the circle of Willis, where the analyzed vessel structures have original diameters of 0,5-2mm, the above minor effects might sum up to the observed errors.

In conclusion, automatic registration seems to be a feasible, fast and precise method to obtain fusion imaging between US and MRI images in Neurosonology application. This method holds the potential of offering a faster and easier way to obtain a precise registration. Further studies are necessary to confirm our preliminary results.

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