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ORIGINAL PAPER

Surface Point Cloud Ultrasound with Transcranial Doppler: Coregistration of Surface Point Cloud Ultrasound with Magnetic Resonance Angiography for Improved Reproducibility, Visualization, and Navigation in Transcranial Doppler Ultrasound

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Abstract

Transcranial Doppler (TCD) ultrasound is a standard tool used in the setting of recent sub-arachnoid hemorrhage (SAH). By tracking velocity in the circle-of-Willis vessels, vasospasm can be detected as interval velocity increase. For this disease process, repeated TCD velocity measurements over many days is the basis for its usefulness. However, a key limitation to TCD is its user dependence, which is itself largely due to the fact that exact information about probe positioning is lost between subsequent scans. Surface point cloud ultrasound (SPC-US) was recently introduced as a general approach combining ultrasound and three-dimensional surface imaging of patient + probe. In the present proof-of-principle demonstration, we have applied SPC-US to TCD and co-registered the skin surface with that from MRA images to provide a roadmap of the vasculature in 3D space for better speed, accuracy, reproducibility, and potential semi-automation of TCD. Collating the acronyms, we call the combined approach SPC-US-TCD. TCD of the M1 was obtained while three-dimensional photographic images were obtained with the Structure Sensor camera. MRA imaging was also obtained. SPC-US-TCD and corresponding MRA 3D reconstruction images were co-registered in MeshMixer using the skin surfaces for alignment. A cylinder the width of the TCD probe was placed over the fused images and aligned with the direction and orientation of the TCD probe to demonstrate the acoustic beam. In the fused images, the acoustic beam intersects the right M1 segment of the middle cerebral artery (MCA). The angle of insonation is well demonstrated and measurable in various planes. Distance measurements made in Blender localized the TCD probe position based on three skin surface landmarks, and tabulated orientation based on three angles along the corresponding directions. SPC-US-TCD provides valuable information that is otherwise not present in TCD studies. By co-registering SPC-US-TCD data with that from cross sectional vessel imaging, precise probe location relative to external skin surface landmarks as well as 3D vessel location relative to TCD probe placement offers the potential to provide a roadmap that improves exam reproducibility, speed of acquisition, and accuracy. The goal of future work is to demonstrate this improvement statistically by application to multiple patients and scans.

Keywords Transcranial doppler · Ultrasound · Angiography · Subarachnoid hemorrhage · Vasospasm · Surface point cloud ultrasound · MRA · CTA

Introduction

Transcranial doppler (TCD) ultrasound, first introduced in 1982 [1], has become a standard and increasingly used tool employed in intensive care settings to monitor for cerebral vasospasm following subarachnoid hemorrhage (SAH) [2–5]. Vasospasm is a serious and common complication of SAH, occurring in 20–30% of SAH cases [6, 7] and resulting in a 10–20% rate of permanent disability or death [8].

In addition to cerebral vasospasm following SAH, TCD has shown promise in evaluating intracranial vessels for

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diagnosis of ventriculoperitoneal shunt malfunction [9], for the care of sickle cell disease [10], stroke, carotid artery occlusion, stenosis of intracranial vessels, and brain death [11].

TCD measures the velocity of blood flow in the cerebral arteries via the Doppler equation:

$$V = \frac{c}{\cos\theta} \times \frac{f}{f_0} \quad (1)$$

where V is the flow velocity ($^{cm}/_{sec}$), c is the velocity of sound (on average 1540 $^{m}/_{sec}$ in human tissue), f is the doppler frequency shift (kHz), f_0 is the original transmitted frequency (typically around 2 kHz is used to minimize sound attenuation by bone), and θ is the angle of insonation relative to the vessel's long axis [12–14].

Now V can give us the degree of vasospasm, or cerebral arterial constriction. Vessel stenosis results in a pressure gradient across the region of stenosis, which by the simplified Bernoulli equation is proportional to flow velocity:

$$\nabla P = 4 \times V^2 \quad (2)$$

where ∇P is the pressure gradient and a surrogate measure for degree of stenosis or vasospasm [4, 15].

Because TCD requires transmission of sound waves through dense calvarial bone, effective use requires exploiting a few regions where the bone is thin and sound is likely to travel with less attenuation and loss of signal: the cranial windows (in addition to patent fontanelles and/or burr/craniectomy holes if they exist) [16]. The chief cranial windows are the orbital window, the transforaminal window and the temporal window [4], this last window showing particular effectiveness for monitoring velocity through the anterior, middle, and posterior cerebral arteries [1, 16].

As a form of the ultrasound imaging modality, TCD has many of the standard benefits and tradeoffs inherent to ultrasound as often used in body or musculoskeletal imaging. Major benefits are noninvasiveness, lack of ionization radiation, low cost, and ready access at the bedside.

Inherent tradeoffs however are high inter-user variability and dependence on user skill and experience. Additionally, information about probe position and orientation relative to external landmarks usually is lost between scans when two different individuals perform successive examinations. For vasospasm detection after SAH, repeated TCD velocity measurements over many days is the basis for its usefulness.

However, reproducibility is a longstanding area of concern and thus research interest in TCD. Typical reported intra-subject and inter-operator coefficients of variation for MCA velocity measurement are on the order of 10–20% [17–19]. The problem is worsened with decreasing operator experience [20, 21] and for longer periods between repeat scans [22].

Consistency in all aspects of TCD is important for reproducibility; consistent velocity measurements are the key to successful surveillance of TCD-assessed vasospasm. This is because diagnosis of vasospasm depends on differences, namely, interval increases, in velocity between successive scans. Hence, these differences should to the utmost degree be due to inherent differences in flow velocity and not from differences in technique – probe settings, location, or orientation. As above, reliable monitoring for cerebral vasospasm requires consistent insonation angles and beam depths. Insonation angles ultimately depend in large part on precise external placement of the ultrasound probe, information which again is typically lost between successive scans.

One potential advance to help remedy these limitations is transcranial color-coded duplex sonography (TCCS). TCCS provides direct interrogation of vessels via two-dimensional parenchymal and color doppler images. This visualization permits a more systematic search for vessels that may be in a state of vasospasm. It also allows the use of landmarks to identify vessels in a more reliable way than the "blind" approach of spectral TCD [4, 23, 24]. Finally, being able to visualize vessels helps to make the angle of insonation more consistent because to some extent it can be recorded during image acquisition. In fact, TCCS has been shown in some cases to detect vasospasm more accurately than TCD [25, 26].

However, although TCCS can provide *internal* landmarks within the calvarium, it does not provide information about *external* landmarks or probe position with regard to the patient's skin surface. This is a fundamental limitation of ultrasound because the sound waves need to travel through human tissue.

Prior work has shown the promise of coregistration of external patient information with cross sectional angiographic data to improve the accuracy and reproducibility for localizing the major intracranial arteries. In particular, Auer et al. [27] used masks mounted onto patient faces with light emitting diodes attached to the upper jaw and nasion regions. With this external reference system, they co-registered MRA data with TCD. They thereby imaged insonation position with respect to target vessels. The approach yielded a roughly two-fold improvement in localization accuracy and reproducibility [27].

Kantelhardt, Greke et al. [28, 29] used navigation via the Kolibri image guidance system, used in BrainLab. These system use reflective markers with the reference marker fixed to the patient's head by a skin adhesive double-sided tape and a flexible headband and an ultrasound probe equipped with a reflective marker array. There is a calibration step of moving a marker around the patient's face. The markers are tracked by a mounted infrared depth-sensing[sension] camera. Doing so then allowed simultaneous tracking of the reference points attached to the patient and the ultrasound probe. This permitted display of the position of the tracked probe on sagittal, axial, and in-line view reconstructions of the separately obtained CTA [28, 29]. They found that the navigation afforded a nearly four-fold

reduction in time to locate the major intracranial arteries, with relatively high accuracy and reproducibility [28, 29].

However, all of these approaches require some external attachment to the patient for surface point registration. This can be bulky, uncomfortable, and possibly unfeasible, particularly in patients with bandage material and/or who have recently undergone craniotomy as part of surgical treatment, for example, of a ruptured aneurysm. They are also subject to misregistration if part of the external apparatus moves or shifts.

Further, none of these approaches provides reliable (or in fact any) information about the relative position and orientation of probe to skin surface. Hence, they cannot capture intuitively useful information to guide future examinations.

A recently introduced technique called Surface Point Cloud Ultrasound (SPC-US) allows for the capture of this external information [30]. By acquiring three-dimensional photographic data of the patient skin surface + probe during ultrasound image acquisition, the precise three-dimensional position of the probe as well as its orientation are recorded and accompany the "internal" ultrasound images [30]. As shown in the paper, combining surface 3D information with ultrasound images provides a fast and intuitive overall sense for where a patient is being scanned, one that can guide another user quickly to the approximate same location and probe orientation. It also showed that combining the absolute external reference frame of patient surface + probe with the "internal" ultrasound information can eliminate ambiguities and certain latent errors such as wrong side/wrong patient/wrong organ errors as well as reversed probe direction, and thus incorrect flow direction within vessels.

Although SPC-US was initially introduced for body radiology applications of ultrasound, here we apply it to TCD, and call the approach Surface Point Cloud Ultrasound for Transcranial Doppler (SPC-US-TCD). We show that, as for body applications, SPC-US-TCD captures concise and detailed information regarding probe position and orientation. Further, by aligning/co-registering the skin surface portion of SPC-US-TCD images with a 3D reconstruction of a magnetic resonance angiogram (MRA), we can also convey probe position relative to the vessels and project this inward toward the vessel to visualize insonation angle in a manner that, though less quantitatively reliable than TCCS, provides a complementary intuitive overall perspective.

Methods

For this proof-of-principle demonstration of the approach, TCD images were obtained of a volunteer researcher (JNS) by another group member with extensive experience operating TCD on patients in the Columbia University Medical Center Neurointensive Care Unit. Specifically, the M1 vessel was assessed, and its spectral waveform recorded (Fig. 1). Correct probe positioning that detected the right MCA M1 segment was confirmed by the expected spectral waveform, as seen in the right panel of Fig. 1. TCD waveforms were acquired with a Multigon TCD probe and machine (Multigon Industries Inc., Elmsford, NY) running software by Delica (SMT Medical, Wuerzburg, Germany).

Simultaneously, another researcher was scanning the subject with a Structure Sensor camera (Occipital Inc., Boulder, CO). The Structure Sensor, like other 3D cameras, uses infrared light to capture depth information, which is paired with standard 2D camera images to produce images with the depth of real objects in 3D. It generates a point cloud, which is the set of surface points of an object in 3D. A snapshot image of the subject's head and face + TCD probe surface point cloud is displayed on the left panel in Fig. 1.

Within a week of the TCD scan, an MRA was obtained for the same subject. The scan was performed on the 3 T GE 750 W scanner (software vs DV26.0) with the GEM head and neck coil with head elements selected. For 3D multi-slab TOF MRA, 6 overlapping thin slab acquisition was performed with the following parameters (3D TOF SPGR Axial scan, TR/TE = 24/3.4 ms, flip angle = 20 degrees, FOV =24 cm × 24 cm, matrix = 400 × 400, BW = 141 Khz, slice thickness /gap =1 mm/ -0.5 mm, number of locations per slab = 52 (10 slices overlapping per slab), and flow compensation, ARC reduction factor 2 and compressed sense factor 1.5 for a total scan time of 6 minutes and 51 seconds.

The full MRA volume was uploaded as an STL file and manually post-processed to remove non-skin, non-vessel image elements. A maximum intensity projection (MIP) representation of the MRA is shown in Fig. 2.

The resulting STL file was uploaded into MeshMixer, along with the STL file of the surface point cloud data obtained with the 3D camera. Then the 3D camera acquisition and MRA 3D volume were co-registered in MeshMixer using the skin surfaces for alignment. Skin surface alignment was performed manually using translations, rotations, and size scaling until the two skin surfaces matched to a degree judged as optimal. A cylinder the width of the TCD probe was placed over the fused images and aligned with the direction and orientation of the TCD probe to demonstrate the acoustic beam. Fig. 3 shows this alignment and ultrasound beam projection along with the vessels from the MRA. As seen in this figure, the beam intersects with the M1 vessel segment, consistent with the spectral doppler waveform from Fig. 1. Fig. 4 again demonstrates the fused images, shown from two different angles. Of note, these images can be rotated in any direction in real time for better visualization and perspective.

In order to demonstrate the use of our approach in improving reproducibility and precision, we measured probe position and angle. Probe position was determined by measuring distance from the nearest part of the probe to three surface anatomic landmarks. The rationale was that this fully specified the

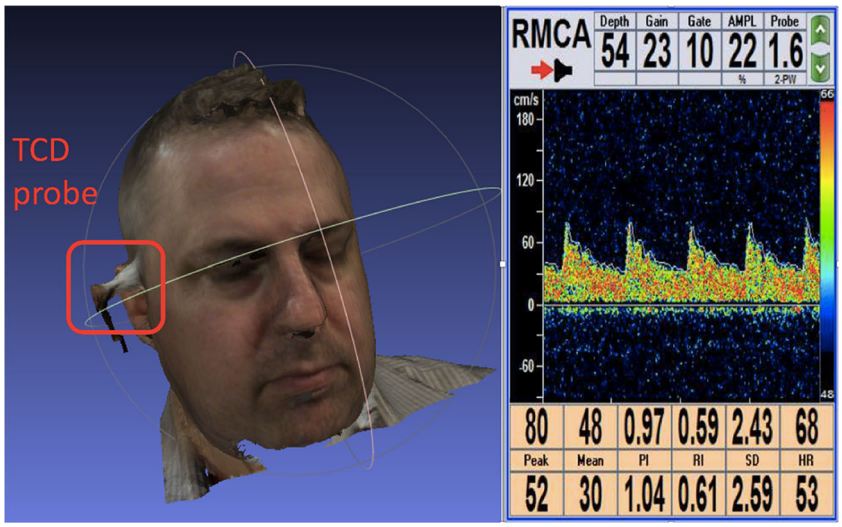

Fig. 1 SPC-US with TCD probe in position while the corresponding MCA M1 segment waveform displayed to the right was acquired

probe position with respect to the subject by providing three spatial coordinates in a new basis (distance to the three landmarks). Specifically, the three distances used were: probe-to-lateral-canthus distance (PC), probe-to-superior helix distance (PSH), and the probe-to-inferior lobule distance (PIL). In order to specify probe orientation, three angles were measured, taken tangential to the skin surface in directions corresponding to PC, PSH, and PIL. All measurements were performed in Blender after importing the surface point cloud 3D camera data as an STL file. The resulting values are enumerated in Table 1. The three distance measurements and two of the angle measurements are shown in Figs. 5 and 6, respectively.

Discussion

We have demonstrated a proof-of-principle application of SPC-US-TCD. The approach offers a new type of visualization of the TCD probe with regard to the patient and patient vessels. This may prove especially useful in the monitoring of disease processes such as vasospasm after SAH which depends on repeated TCD velocity measurements over many days. By recording information about precise spatial location and orientation of the probe with regard to patient skin surface during image acquisition, a more intuitive, broader perspective is afforded. It offers an intuitive sense of where the probe is positioned, which can serve as a guide to future scans. It may lead toward more reliable and automated scans that are less dependent on operator skill.

TCD, like any ultrasound modality, is highly user-dependent. Typical neurointensive care workflow involves frequent "handoffs" in care: a technologist with 10 years experience performing TCD may perform a scan in the morning. A neurointensive care fellow may perform a follow-up scan in the late afternoon. An intern covering the unit may scan overnight. These three operators have disparate skill and

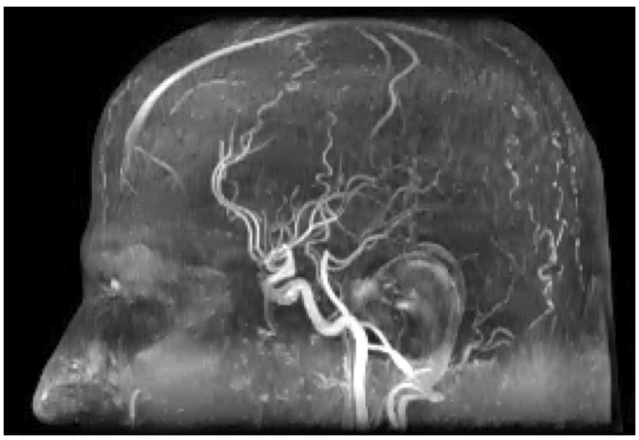

Fig. 2 Lateral maximum intensity projection image of the full image volume of the MRA obtained from the same subject. Note that the skin surface information is present and can be used for co-registration with the skin surface data from Fig. 1

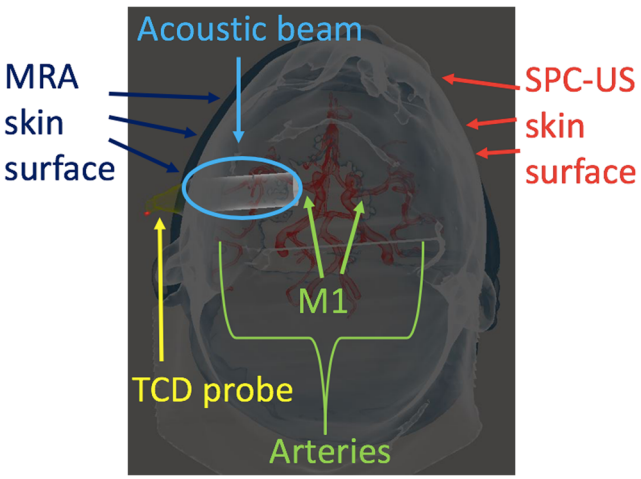

Fig. 3 Schematic of skin surface co-registration. The 3D skin surface point clouds were manually aligned in MeshMixer. Highlighted in this figure is the pair of skin surfaces from MRA and from SPC-US. Also seen is the TCD probe and expected/extrapolated acoustic or insonation beam, which is seen to intersect with the M1 branch of the right MCA

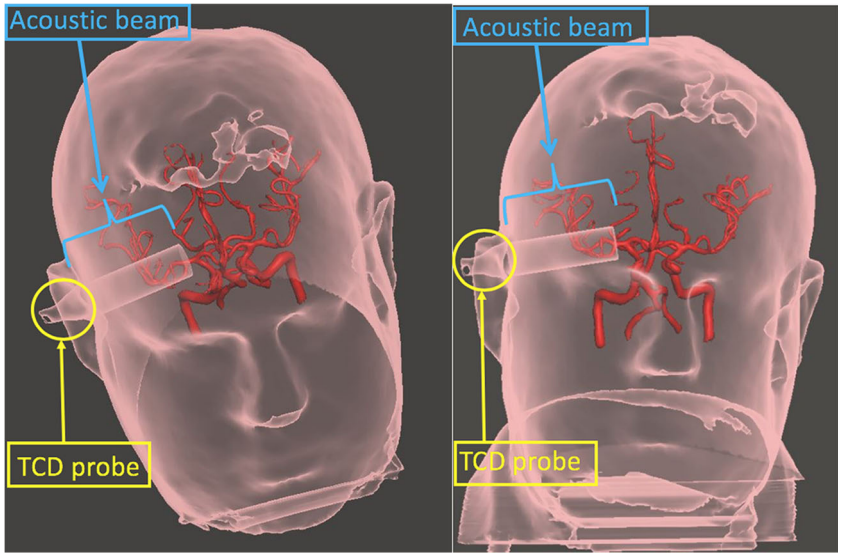

Fig. 4 Two angles illustrating the fused SPC-US-TCD and MRA vessel images, with insonation beam intersecting the M1 segment of the right MCA

experience with TCD. Further, in all likelihood none would have seen the other perform the image acquisition. For example, they would not know where exactly is a particular patient's ideal window for interrogating the MCA. By having a rotatable 3D image of where the probe was on the patient and how it was tilted, the intern without much experience who did not see the experienced technologist scanning should have a higher success rate in less time. We may also anticipate decreased intra-user variability, since access to probe positioning parameters could help the same technologist recall and possibly measure precisely where he or she placed the probe in an earlier scan.

We have demonstrated that this position data can be quantified in order to obtain precise probe localization. Both the position and orientation can be fully specified and recorded. Accurately diagnosing vasospasm on TCD requires reliable velocity difference measurements between successive scans. Making those scans as similar and directly comparable as possible is thus tantamount to reliable detection of vasospasm or vessel narrowing more generally. As a complement to the internal orientation afforded by TCCS, SCP-US-TCD can provide the external orientation.

As an anecdotal initial proof-of-principle, having a streamlined process was not the goal. Post-processing was performed manually to align the SPC-US data and the MRA skin surface. This took roughly 10 min. However, future iterations could be increasingly automated. We would anticipate particular usefulness in automating co-registration via deep learning techniques. One may imagine later implementations projecting a hologram of the patient's vessels into their head, as some prior surgical image tools have done, or using augmented reality glasses to achieve this effect.

Table 1 Summary of external reference measurements

Measurement type	Value
Probe-to-lateral-canthus distance (PC)	6.3 cm
Probe-to-superior helix distance (PSH)	2.1 cm
Probe-to-inferior lobule distance (PIL)	5.3 cm
Angle with skin surface along PC direction (θ_{PC})	96.6 degrees
Angle with skin surface along PSH direction (θ_{PSH})	85.0 degrees
Angle with skin surface along PIL direction (θ_{PIL})	82.5 degees

These use the conversion factor 11.5 cm: 0.11 AU (AU measured on Blender for lateral canthus-to-lateral canthus distance on study subject)

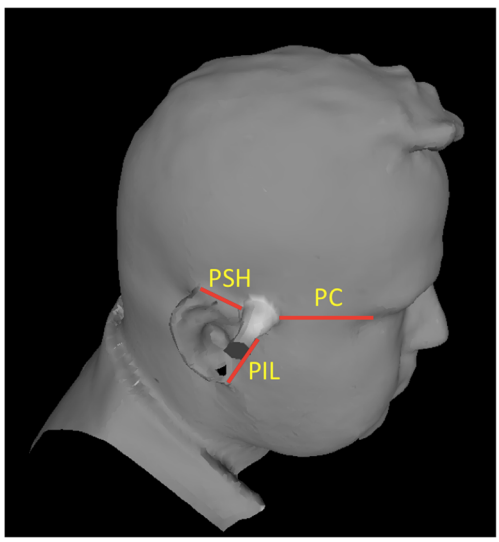

Fig. 5 Linear measurements for accurate localization of the TCD probe. Measurements are illustrated for the probe-to-lateral-canthus distance (PC), the probe-to-superior helix distance (PSH), and the probe-to-inferior lobule distance (PIL)

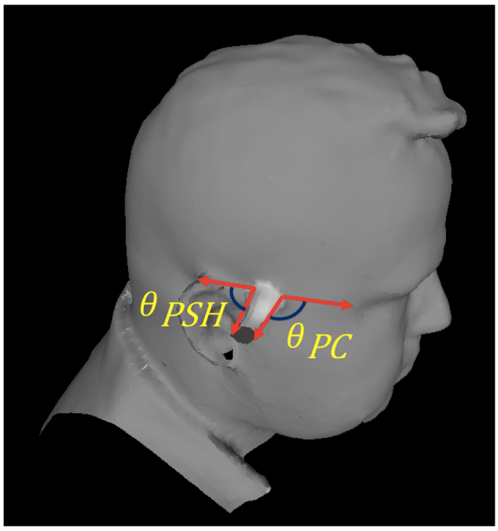

Fig. 6 Two of the three measurements for accurate localization of the TCD probe. Subscripts indicate the direction of the angle measurements, each angle being made locally in the corresponding direction with the skin surface

As alluded to above, CTA is the preferred modality for vessel imaging in most real applications. Standard protocol is for all patients with acute SAH to have at least one CTA performed soon after the pathology is discovered on noncontrast head CT. Typically, the hospital course will involve at least one follow-up CTA. The CTA data can be turned into volumetric image files just as we did for MRA. MRA was chosen here for expediency, since it does not involve ionizing radiation nor contrast and was thus safer to perform with a volunteer.

Future work will endeavor to validate the approach with a cohort of SAH patients using CTAs obtained as part of their clinical care. We anticipate that SPC-US-TCD will permit faster and more reproducible evaluation of the vessels and vasospasm determination than without the approach. Another direction for future research is to apply SPC-US to TCCS. By overlaying TCCS, which depicts vessel flow, with the external landmarks of 3D photographic data and the corresponding co-registered ground truth MRA or CTA vessel anatomy, vessel identities can be confirmed. Additionally, being able to see the absolute position of ultrasound probe relative to cross sectional anatomy of a particular vessel overlaid onto the color doppler representation, along with adjacent non-vascular structures, may help to detect doppler artifacts such as aliasing, color in non-vascular structures, vascular motion artifact, and direction ambiguity [31].

Compliance with Ethical Standards

Conflict of Interest On behalf of all authors, the corresponding author states that there is no conflict of interest.

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