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Accuracy and Precision of Plane Wave Vector Flow Imaging for Laminar and Complex Flow In Vivo

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Abstract—In this study, a comparison between velocity fields for a plane wave 2-D vector flow imaging (VFI) method and a computational fluid dynamics (CFD) simulation is made. VFI estimates are obtained from the scan of a flow phantom, which mimics the complex flow conditions in the carotid artery. Furthermore, the precision of the VFI method is investigated under laminar and complex flow conditions in vivo. The carotid bifurcation of a healthy volunteer was scanned using both fast plane wave ultrasound and magnetic resonance imaging (MRI). The acquired MRI geometry of the bifurcation was used for fabricating an anthropomorphic flow phantom, which was also ultrasound scanned. The same geometry was used in a CFD simulation to calculate the velocity field. Results showed that similar flow patterns and vortices were estimated using CFD and VFI in the phantom. Velocity magnitudes were estimated with a mean difference within 15 %, however, it was 23 % in the external branch. For the in vivo scan, the precision in terms of mean standard deviation (SD) of estimates aligned to the cardiac cycle was highest in the center of the common carotid artery (SD 4.7° for angles) and lowest in the external branch and close to the vessel wall (SD 15.0° for angles).

I. INTRODUCTION

The combination of high-frame rate ultrasound imaging methods and 2-D vector flow imaging (VFI) [1]–[4] presents several advantages compared to conventional velocity imaging: Rapid blood flow patterns can be followed at hundreds or thousands of frames per second, the velocity estimates are angle-independent, and quantitative estimates are continuously available everywhere in the image. This enables visualization of complex blood flow patterns, e.g. the formation of vortices, which often form around bifurcations and valves. The VFI methods must provide precise and accurate velocity estimates for both laminar and complex flow when measured in vivo. However, when investigating the performance of a VFI method in vivo, two challenges arise: (i) independent methods such as spectral Doppler and magnetic resonance angiography (MRA) do not have sufficient spatial and temporal resolution to accurately capture complex flow patterns in vivo, which makes a comparison to VFI difficult; and (ii) data should be acquired for at least two to three heart beats to evaluate the precision in terms of repeatability of VFI estimates, but this generates more transducer element data than most scanners can store.

In relation to (i), an alternative approach to investigate complex flow is to perform a realistic phantom study. Using an anthropomorphic geometry for the phantom fabrication, complex flow patterns may be measured in a realistic vessel geometry. The same geometry and boundary conditions can be used in a simulation of the velocity field using computational fluid dynamics (CFD). The CFD simulated velocities can then be compared to the VFI estimates obtained from the phantom.

In relation to (ii), all element data can be stored in scans for up to 10 s by using the experimental scanner SARUS [5], which enables evaluation of the precision of the estimates from several heart beats.

The method for VFI in this paper was based on plane wave imaging and a combination of transverse oscillation (TO) and directional beamforming (DB), which was previously validated in simulations [6]. The objective of this paper was to investigate the accuracy and precision of the method under laminar and complex flow conditions in the carotid bifurcation of a healthy volunteer.

The accuracy of the method was investigated by comparing ultrasound VFI estimates obtained from an anthropomorphic phantom measurement to a CFD simulated velocity field derived from the same geometry. The precision was investigated in vivo by calculating the repeatability of the VFI estimates at several estimation points in the bifurcation.

II. METHODS

A 27 year old healthy volunteer participated after informed consent, and the study was approved by the National Committee on Biomedical Research Ethics (Protocol No. H-1-2014-fsp-072). An in vivo VFI scan was conducted during a session in the morning, and an MRI scan was performed in the evening.

A. In vivo VFI scan

A plane wave VFI scan was performed using a linear array transducer (BK 8L2, BK Ultrasound, Herlev, Denmark), which was connected to the SARUS scanner [5]. The carotid bulb was
scanned with the best possible view of the bifurcation, and a
duplex sequence consisting of flow and B-mode emissions
was employed (4.1 MHz center frequency, 10 kHz PRF,
plane waves at -15°, 0°, 15°) [6]. Element RF data were
acquired for a total of 10 s. Echo-canceling of beamformed
data was performed using an energy-based filter with a manual
threshold [7].

A two-step procedure was employed for velocity estimation:
First, the TO method was used to estimate an initial flow angle
θ, which was calculated from the axial and lateral velocities
v_3 and v_z, respectively. Second, three DB lines were focused
around θ at each estimation point to find a refined angle ̂θ
using the largest normalized cross-correlation estimate [6]. The
velocity magnitude was estimated along the direction ̂θ using
a cross-correlation estimator.

B. MRI scan

A 1.5 T whole body scanner (Magneton Avanto, Siemens,
Erlangen, Germany) was used with a head and a neck matrix
coil. The carotid artery was MRI scanned to acquire anatom-
ical images of the common carotid artery (CCA) and carotid
bifurcation. The repetition time of the sequence was 42 ms,
echo-time 3 ms, pixel resolution 1.1 x 1.1 mm, and slice
thickness 5 mm. The anatomical images were used to create a
3-D geometry representing the carotid artery of the volunteer.
A CFD model and a flow phantom were produced from the
3-D geometry as described in the following sections.

C. Flow phantom fabrication and scan

An anthropomorphic flow phantom matching the flow do-
main of the original scanned vessel was fabricated using stereolithography [8]. After 3-D printing of the vessel core, it
was fixated in a container and cast in polyvinyl alcohol (PVC)
cryogel to obtain a tissue-mimicking surrounding medium.
The elastic properties of the cast material were controlled by
varying the number of freeze-thaw cycles. After completion
of three cycles, the core was removed manually to obtain the
wall-less phantom.

A flow system (CompuFlow 1000, Shelley Medical Imaging
Technologies, Toronto, Canada) was connected to the phantom
and circulated a blood-mimicking fluid, which had a viscous-
ity of 4.1 · 10^{-3} Pa·s and a density of 1030 kg/m³. The
CompuFlow system was set to generate a standard carotid
artery waveform. The transducer was placed on the phantom
to scan the bifurcation. The SARUS scanner was used for data
acquisition with the same sequence and parameters as for the
in vivo scan.

D. CFD model

The flow simulation in the CFD model was governed by
solving the Navier-Stokes equations. The properties of the
emulated fluid matched those of the blood-mimicking fluid
used for the phantom study. It was assumed that the fluid was
Newtonian, entrance effects were discarded, the vessel was
rigid, and the flow had reached a steady state of pulsation.
The inlet condition was constructed using the Womersley-
Evans model [9], [10]. In details, the individual components
of the mean spatial velocity curve, which was obtained from
the measured plane wave VFI data at a point in the CCA, were
found by Fourier decomposition. Thereafter, superposition of
the first ten harmonics provided a smooth spatial velocity
variation. This result was then used to reconstruct the velocity
profile in both time and space using the Womersley-Evans
model. An outlet pressure of 0 Pa was set for the two exit
branches, while a no-slip condition with ̅v = 0 m/s was set at
the walls of the flow domain. The CFD simulation was carried
out in Comsol Multiphysics (v5.2a, Comsol AB, Stockholm,
Sweden).

E. Evaluation

Prior to the comparison between the flow field of the
CFD simulation and the phantom VFI scan, the geometries
were aligned. Rotation and translation of the geometries were
performed manually after visual inspection. A quantitative
comparison of velocities acquired from CFD and VFI were
made at selected planes in the bifurcation. Furthermore, the
vorticity was calculated using the rotational frequency:

\[ \omega = \nabla \times \vec{v} = \left( \frac{\partial v_x}{\partial z} - \frac{\partial v_z}{\partial x} \right), \]

where \( \nabla \) is the nabla operator. The mean rotational frequency
was calculated for an area covering a (part of a) vortex.

For the precision study, a medical doctor evaluated the
in vivo scan by selecting evaluation points in the center of
the CCA, close to the vessel wall of CCA, in the internal
carotid artery (ICA), and in the external (ECA). An automatic
procedure calculated the mean cardiac period and coherently
aligned the velocity magnitude and angle estimates according
to the cardiac period. The standard deviation \( \sigma(t) \) at time \( t \)
was calculated among the aligned estimates. The mean standard
deviation (SD) was

\[ SD = \sqrt{\frac{1}{N_f} \sum_{t=1}^{N_f} \sigma(t)^2}, \]

where \( N_f \) is the number of frames encompassing a cardiac
cycle. The mean SD of velocity magnitude estimates, (2) was
calculated relative to the peak velocity magnitude \( v_0 \).

III. RESULTS

A. Comparison between phantom VFI and CFD simulation

A frame from the comparison between CFD simulation
and the VFI scan is shown in Fig. 1. The frame is within
systolic deceleration (0.32 s) and shows similar flow patterns
for CFD and VFI. Both the CFD and VFI showed a large
vortex or recirculation zone, which extended from the ICA
into the CCA. There were small differences in the extent of
the recirculation zone and velocity magnitudes, but similar
flow patterns were provided. The jet of streamlined forward
flow found close to the inner vessel wall of the ICA in the
CFD image was also measured using VFI. A small vortex in
the ECA was measured using VFI as indicated by the green area in the lower branch in Fig. 1 (right). However, this vortex was not visible in the CFD.

A more quantitative comparison between CFD and VFI is shown in Fig. 2, where the velocity magnitude profiles at selected locations in the CCA, ICA, and ECA are shown at peak systole ($t = 0.16$ s) and systolic deceleration ($t = 0.32$ s).

The mean difference between CFD and VFI was of -1.4 % for the velocity profile in the CCA at peak systole, while it was 23.6 % for the ECA. During systolic deceleration (Fig. 2 bottom row), lower velocities were estimated in both the CCA and ECA (mean difference was -12.3 % and -13.3 %, respectively), when using VFI compared to CFD. However, the velocity magnitudes of the backflow in the ICA from -5 mm to 0 mm were larger for VFI, which is also shown in Fig. 1.

The rotation of flow in the recirculation zone inside the white box in Fig. 1 was quantified by calculating the mean rotational frequency as a function of time, and the result is shown in Fig. 3. The two peaks of the rotational frequency for VFI at 0.14 s and 0.3 s revealed that a vortex or recirculation evolved at two time instances, while a single recirculation built up after 0.15 s in the CFD simulation and lasted until 0.3 s. For the remaining cardiac period (from 0.3 s to 0.84 s) there is a good agreement between CFD and VFI for the rotational
frequency.

**B. Precision in vivo**

The precision of velocity magnitude and angle estimates was calculated for evaluation points in the center of the CCA, close to the vessel wall of the CCA, in the ICA, and the ECA. The points are indicated on the VFI frame shown in Fig. 4. The highest precision was obtained in the center of the CCA (3.0 % for velocities and 4.7° for angles) and in the ICA (5.5 % and 6.5°), where the highest velocities were estimated and flow was mainly unidirectional. Slightly lower precisions were obtained in the ECA (5.1 % and 13.3°) and close to the vessel wall of the CCA (7.3 % and 15.0°). Lower velocities and a larger variation in the angles were found for these evaluation points.

The rotational frequency was calculated for the vortex indicated by the white square in Fig. 4 and for all frames during the 10 s scan. After alignment of estimates to the cardiac cycle, the mean SD of the rotational frequency was calculated to 6.2 %. This indicated a high cardiac cycle-to-cycle repeatability of the velocity estimates and flow patterns in a zone covering the vortex.

**IV. DISCUSSION AND CONCLUSION**

A comparison between flow in a carotid bifurcation phantom using VFI and simulated CFD velocities was made. Similar flow patterns were obtained, and the mean difference between VFI and CFD was within 23 % for velocities. However, the alignment of the transducer scan plane in the phantom and the CFD geometry was not perfect, and the assumptions of rigid walls and inlet and outlet boundary conditions may represent limitations in the CFD model.

The study showed that the precision in vivo was highest in the center of the common carotid artery and lowest in the external branch (standard deviation for angles 4.7° and 15.0°). This indicated that quantitative velocity estimates could be obtained in the carotid artery of the healthy volunteer.

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