

Computational workflow for 2D PTV: validation and application to patient-specific intracranial aneurism phantoms made by additive manufacturing

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Keywords: Intracranial aneurysms, Hemodynamic analysis, Particle Tracking Velocimetry (PTV), Computational workflow, *In Vitro*, Biofluid mechanics, Pulsatile flow

ABSTRACT

Intracranial aneurysms are pathological dilatations of cerebral arteries associated with potentially fatal rupture events. Although several studies highlighted the importance of hemodynamic factors and aneurysmal geometry in aneurysm formation and rupture, the mechanism is not fully understood. This study introduces a complete and validated computational workflow designed to transform raw 2D PTV images into Eulerian fields for biofluid analysis, which is used to investigate the hemodynamic behavior within patient-specific intracranial aneurysm phantoms assembled by additive manufacturing. A pulsatile flow regime is implemented under both resting and exercise conditions specific for aneurysms in the Internal Carotid Artery (ICA).

The results show that, for each case evaluated, the overall flow topology remains generally consistent across the investigated conditions, while significant variations in velocity magnitude, vortex dynamics and Wall Shear Stress (WSS) distribution occur due to changes in flow rate. The exercise condition leads to increased velocity magnitudes, particularly in the outer regions of the primary vortex, resulting in changes in WSS that depend on the flow organization. Furthermore, temporal analysis reveals the migration of vortex structures throughout the cardiac cycle. While Reynolds and Womersley numbers influence vortex dynamics, aneurysm geometry (size and Aspect Ratio) affects vortex stability and aneurysmal flow behavior.

1. Introduction

The World Health Organization reports that cardiovascular diseases (CVDs) are the leading cause of death globally, accounting for approximately 17.9 million deaths in 2019. Alongside heart

attacks and strokes (which represent 85% of these deaths), aneurysms are a significant contributor to this statistic. According to the Brain Aneurysm Foundation, an estimated 6.8 million people in the United States, of roughly 1 in 50 individuals, have an unruptured brain aneurysm. Approximately 2–3% of the population harbors an unruptured intracranial aneurysm, and approximately 30% of the patients have multiple aneurysms (Bhokal *et al.*, 2018). Despite these clear risks, predicting which aneurysms will grow or rupture is a major clinical challenge. Currently, there are no definitive techniques, and decisions often rely on general morphological factors like size, for which, in several critical cases, there may be no clear consensus on the need for intervention. This uncertainty has pushed researchers to try and answer key questions such as what causes these aneurysms to form, why do they appear in specific locations, and how can one accurately predict their rupture risk for effective and patient-specific intervention (Lasheras, 2007). Currently, it is widely recognized that aneurysms are not simply congenital defects but acquired lesions influenced by the interaction between the hemodynamic environment of the vessel and its wall. This shift in understanding underscores the importance of analyzing hemodynamic factors, which have a direct influence on endothelial cell responses and their role in vascular remodeling (Oliveira *et al.*, 2021). Studying these flow parameters is crucial not only for understanding pathophysiological mechanisms but also for identifying key markers to improve clinical decisions. In this context, *in vitro* studies have become an important resource, enabling researchers to replicate complex vascular geometries and investigate mechanical and biochemical signals that govern disease (Lust *et al.*, 2021; Liu *et al.*, 2025; He *et al.*, 2024). One of the main goals of these types of studies is to allow for the quantification of hemodynamic parameters such as blood velocity, pressure gradients, and wall shear stress (WSS), which are critical for understanding, predicting, and managing vascular diseases (Soldozy *et al.*, 2019; He *et al.*, 2022). To meet this challenge, particle tracking velocimetry (PTV) has proved to be a relevant analysis technique. Among other imaging techniques, PTV may provide higher spatial resolution than correlation-based methods (like PIV), which is a distinct advantage in the high-gradient, complex flow fields common in vascular models (Cierpka *et al.*, 2013; Gu *et al.*, 2023).

On the other hand, it also presents some practical and computational difficulties. Advanced 3D systems (such as Tomo-PTV) offer high-quality data but require specialized lasers and high-speed cameras, making them expensive and unachievable for many labs (Mäkiharju *et al.*, 2021). This maintains a need for accessible and simpler PTV setups, which are frequently used to provide the “ground truth” data for validating computational fluid dynamics (CFD) models, now common in cardiovascular research (Li *et al.*, 2020; Hoi *et al.*, 2006).

The main difficulty in 2D PTV is not just image capture, but the data processing workflow required to handle the noisy, low-particle-density images common in bioflows in complex anatomical geometries. Additionally, PTV's raw output, being Lagrangian, leads to a fundamental data problem. This data, while useful, is not sufficient for a complete hemodynamic analysis. Many of the most critical hemodynamic parameters, such as vorticity and pressure fields, are field derivatives that require a continuous, gridded Eulerian velocity field (Shadden and Arzani, 2015; Espa *et al.*, 2019). Thus, a complementary step in this data analysis is the conversion of scattered Lagrangian data into an Eulerian field. Therefore, the main goal of this work is to present a complete and validated computational workflow designed to transform raw 2D PTV images into Eulerian fields for biofluid analysis. The analysis is demonstrated in patient-based phantom, obtained by additive manufacturing, in various conditions representative rest and moderate to intensive physical activity. The flow is also analyzed considering both steady-state and pulsatile flow conditions. Besides providing insight into particular flow characteristics in real patient cases, the analysis performed here aims at providing data to feed and validate complex numerical models, which may aid the physicians to take more informed decisions with less risks and less invasive techniques.

2. Experimental methodology

2.1 Fabrication of patient-based phantoms

Different cases of intracranial aneurysms were considered to provide realistic anatomic flow geometries. All cases are of patients with saccular aneurysms located in the ophthalmic segment of the internal carotid artery (ICA). Original anatomical geometries were derived from 3D

rotational angiography images and segmented in 3D Slicer to create digital STL models. Vascular biomodels were fabricated using an Anycubic Photon D2 3D printer, which is based on digital light processing (DLP) technology. The material used was Anycubic Clear, a translucent, photopolymerizable resin. After printing, the biomodels underwent a postprocessing workflow, including washing in isopropyl alcohol, clearing of internal channels with compressed air, and post-curing under UV radiation. The final surface finish was achieved through mechanical polishing on a benchtop metallographic polisher. This sequence resulted in translucent biomodels with a homogeneous and smooth surface, enabling clear visualization of the particulate fluid flow during experiments. A critical requirement for PTV is the matching of the phantom's refractive index (RI) with that of the working fluid. To visually confirm the optical quality, tests were conducted by passing fluids with different RIs through the models. These tests confirmed that the close RI match between the resin ($RI \approx 1.5$) and the working fluid, DMSO dimethyl sulfoxide ($RI \approx 1.48$), resulted in a clear, minimally distorted view, validating the suitability of this fabrication approach.

2.2 Experimental arrangement and methodology

Once the patient-specific phantoms were fabricated, they were integrated into an experimental setup designed to simulate circulatory flow under steady-state conditions and acquire 2D PTV data. The experimental arrangement used is schematically represented in Figure 1.

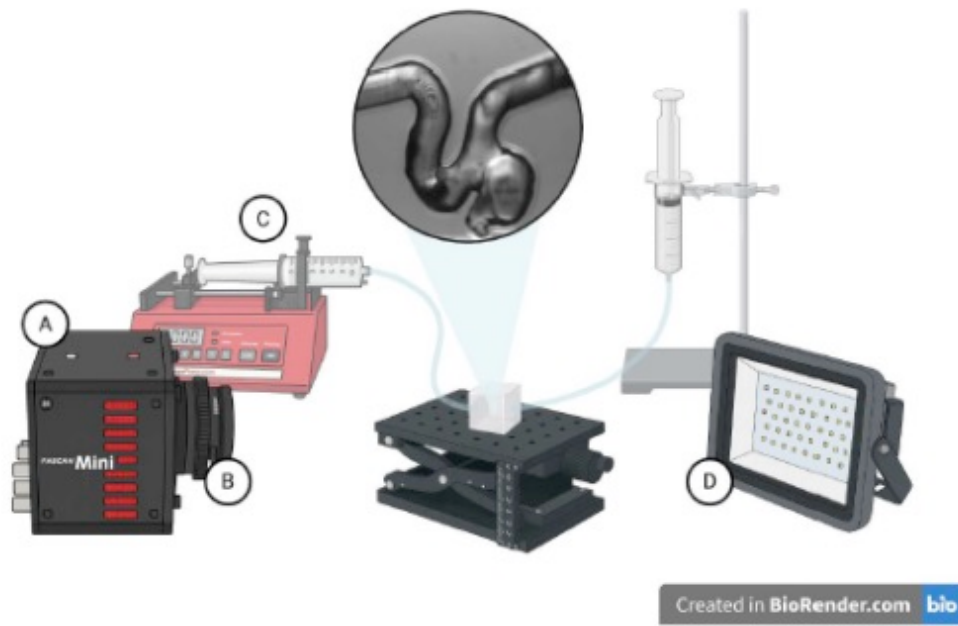


Figure 1: Schematic representation of the setup for Particle Tracking Velocimetry (PTV), showing: (A) high-speed camera, (B) lens, (C) syringe pump, and (D) LED floodlight illumination source.

The experimental apparatus was mounted on a custom-built track, which held the phantom support, a high-speed camera (A), and illumination source in fixed alignment. A syringe pump (C) (NE-4000, New Era Pump Systems) with two syringes was used to impose a controlled, steady flow rate. The syringes were connected through a valve and were attached to the phantom's inlet via flexible tubing, fixed through pipette tips. Fluid exiting the phantom's outlets was collected in a third syringe, which was then used to refill the pump's fixed syringes. For image acquisition, a high-speed camera (A) (Photron Fastcam Mini AX) was positioned in front of the phantom. The distance between camera and phantom was defined according to the lenses focal distance. Images were captured using one of three lenses (B): a Nikon AF Micro Nikkor 60mm f/2.8D or one of two Leica objectives, HI PLAN I 10x/0.22 and N PLAN EPI 20x/0.40, which allowed for magnification of up to 20X. Illumination was provided by a LED floodlight (D), with a diffuser placed between the light source and the model to ensure uniform lighting and minimize glare. The camera was connected to a computer running FASTCAM Viewer 4 (PFV4) software. Regarding imaging parameters, the camera's frame rate was set to 6000 fps, calculated to achieve an average particle

displacement of 5-15 pixels per frame based on expected maximum flow velocities (1m/s) and optical magnification.

Pure dimethyl sulfoxide (DMSO) was selected as the working fluid due to its close RI match, transparency, and appropriate dynamic viscosity ($\eta \approx 2.24$ cP). Safety precautions were observed due to its nature as a solvent. Suitable tracer particles must satisfy three criteria: good optical contrast, neutral buoyancy, and the ability to accurately follow flow streamlines. Flow-following fidelity is quantified by the Stokes number (Stk), a dimensionless number comparing the particle's characteristic response time to the characteristic time scale of the flow; accurate tracking requires $\text{Stk} \ll 1$. Based on these, 50 nm blue polystyrene (PS) micro particles (microParticles GmbH) were selected. Their blue colour and RI (=1.59) provided excellent contrast; their density (1.05 g/cm^3) closely matched that of DMSO (1.10 g/cm^3), ensuring good suspension; and their calculated Stokes number ($\text{Stk} \approx 0.017$, based on expected maximum velocity $U = 1 \text{ m/s}$ and average vessel diameter $D = 4.25 \text{ mm}$) was well below the target threshold of $\text{Stk} < 0.1$ for accurate flow tracing. Finally, the particle seeding concentration was optimized through testing. A low concentration was crucial for the PTV algorithm to distinguish individual particles, with an ideal concentration determined to be approximately 0.03% w/v. This concentration falls within the medium range (approximately 1 to $< 2 \times 10^{-4}$ particles per pixel) known to minimize uncertainty while avoiding excessive particle overlap (Dal Sasso *et al.*, 2018). Since the experiments were conducted under steady state conditions, physiologically relevant inlet flow rates were determined. As patient-specific data was unavailable, a literature review established a population flow rate for the internal carotid artery (ICA). To account for patient-specific geometry, a flow-diameter scaling model with an exponent $n = 2$ was chosen (Valen-Sendstad *et al.*, 2015), using reference values $Q_{avg} = 245 \text{ ml/min}$ and $D_{avg} = 4.15 \text{ mm}$. The parent artery mean diameter (D_i) for each case was measured in 3D Slicer over the C4-C6 segments. Using the scaling relationship:

$$Q_i = Q_{avg} \left(\frac{D_i}{D_{avg}} \right)^n \quad (1)$$

a baseline physiological flow rate corresponding to a “rest” state was calculated for each case. To explore the effect of physiological state, a second “exercise” state was defined as an 8.2% increase over the baseline “rest” flow rate, based on literature values for increased cerebral blood flow during moderate exercise (Bowker *et al.*, 2010; Poulin *et al.*, 1999). These calculated physiological flow rates (Q_{phys}) were converted into the actual values programmed into the syringe pump (Q_{pump}). This required scaling for dynamic similarity (maintaining Reynolds number, which is the main dimensionless number in flow regime characterization, giving the ratio between inertial and viscous forces) in the 0.5-scale phantoms ($Q_m = Q_{phys}/2$) and accounting for the dual-syringe setup (Q_{pump} per syringe = $Q_m/2$). Thus, the final value programmed into the pump was one-quarter of the calculated physiological flow rate ($Q_{pump} = Q_{phys}/4$). It is worth mentioning that, despite being a Newtonian fluid, DMSO has average bulk density and dynamic viscosity close to those of whole blood. Some authors (*e.g.* Hellstrom *et al.*, 1996; Sato *et al.*, 2011; Sadamoto *et al.*, 2016; Ogoh *et al.*, 2019) further address larger increase in blood flow due to physical activity, ranging between 17% to 22%, depending on the intensity and type of exercise performed. In this study, an increase of 22% is also considered to simulate the exercise condition.

Table 1 summarizes the main geometric characteristics of the patient-based aneurysms under study. Table 2 depicts the physiological flow rate scaled using Equation 1 for each case studied.

Table 1: Characteristic dimensions of the patient-based aneurysms under investigation.

Case	Position	Size category (-)	Neck (mm)	Height (mm)	Largest diameter (mm)
1	Left ICA	Small	2.28	1.29	1.89
1	Left ICA	Medium	5.06	5.68	9.64
6	Right ICA	Small	4.58	2.33	4.78
7	Left ICA	Large	7.27	14.75	16.79

Table 2: Characteristic diameter for each case and physiological flow rate scaled using Equation 1. Here, Q_{rest} and Q_{exe} represent the flow rate determined from Equation 1 associated with rest and exercise conditions.

Case	Average diameter (mm)	Q_{rest} (mL/min)	Q_{exe} (mL/min)
1	3.72	196.99	240.32
6	4.76	321.87	392.68
7	4.28	260.04	317.25

Preliminary results were first obtained at steady state conditions. Then a pulsatile flow was also addressed, which was achieved with the adequate programming of the pump. The model used was based on data reported in literature (Ford *et al.*, 2005) and widely used in similar studies (*e.g.* Najafi *et al.*, 2020). In Najafi *et al.* (2020), the flow waveform in the intracranial aneurysm during the cardiac cycle was measured in 17 healthy individuals, providing a set of characteristic points. The waveform was reconstructed using a cubic spline interpolation applied to the data available, providing a continuous and smooth representation of the normalized flow waveform over the cardiac cycle. By multiplying the normalized flow waveform by the case-specific mean flow rate and the heart rate (HR) of the tested condition (HR = 65 bpm for the resting condition and HR = 140 bpm for the exercise condition) the curve was scaled in both amplitude and time (*e.g.* Hellstrom *et al.*, 1996; Sato *et al.*, 2011; Sadamoto *et al.*, 2016; Ogoh *et al.*, 2019). This procedure was performed using a MATLAB script, which also adapted the signal to the programming constraints of the syringe pump, like a discrete flow profile with a limited number of points. Sampling was performed in a non-uniform manner, giving priority to regions where the flow exhibits greater temporal variation. These points were selected considering the first and second derivatives of the waveform to increase the accuracy of signal preservation.

Finally, the discretized profile was converted into the corresponding values of mean flow rate and delivered volume for each time interval of the cardiac cycle, as shown in Figure 2. These values were then set into the software spreadsheet, which automatically generated the programming file.

To investigate the flow behavior during the different phases of pulsatile cycle, the syringe pump was connected to the high-speed camera to send a trigger signal that started data acquisition exactly at the beginning of the cycle.

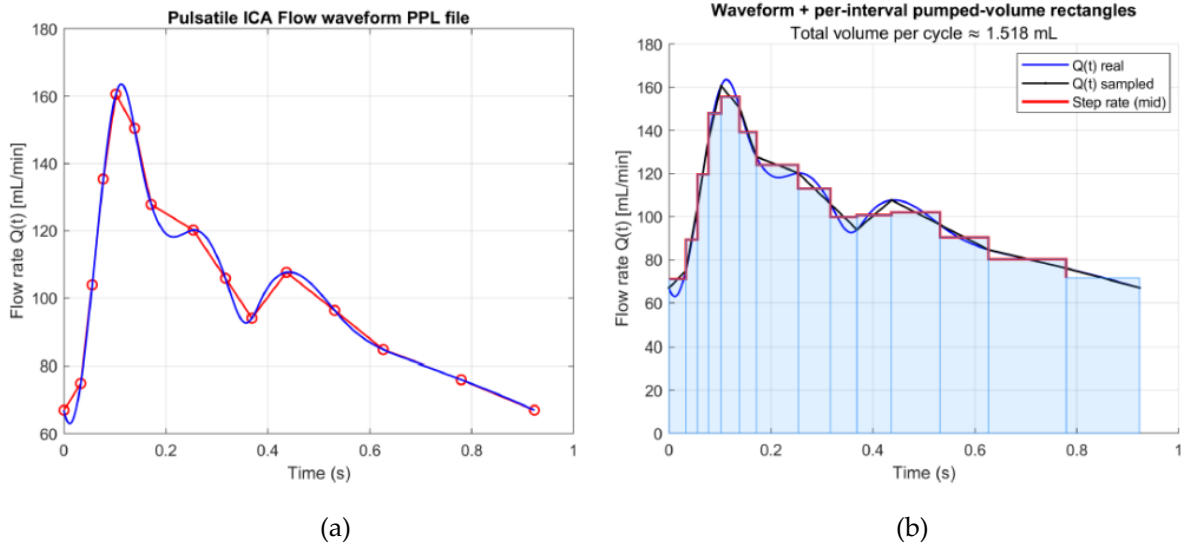


Figure 2: (a) Sampling of pulsatile flow profile to match the requirements imposed by the pump. (b) Comparison of the continuous and discrete flow profile and representation of the rectangle which indicates the volume of fluid pumped for each phase.

To further examine the temporal characteristics of the vortices, an additional function was implemented to track the motion of the vortex core throughout the different phases of the cycle. For this purpose, a MATLAB function was developed containing three main stages: detection of potential vortices, determination of the best mutant candidate and validation of the real vortex. Since the center of a vortex is generally characterized by lower velocity magnitudes, the detection phase keeps all the coordinates in which the intensity of the velocity vector is below a specific threshold. However, a velocity minimum does not always correspond to the true vortex center. Therefore, the best mutant candidate was considered. By generating a window around each candidate, the cell which maximizes the local vortical density index was selected for validation, improving robustness to velocity field noise. The validation phase analyzes the local flow around each candidate by evaluating the orientation of the velocity vectors relative to the candidate center. Vectors were considered valid when they were tangential to the radial direction and shared a

coherent rotational direction (clockwise or counterclockwise). The vortex center was then identified as the point with the highest normalized density of these coherent tangential vectors, ensuring the detection of true circular flow structures.

2.3 Data processing

The acquired videos were converted in images which were treated following two main steps: Lagrangian particle detection and tracking (done using Fiji/ImageJ and the TrackMate plug-in) and generation of an Eulerian field (based on an in-house code developed in Matlab®). Images were first pre-processed performing a background subtraction followed by contrast enhancement. Then, a PTV analysis was performed in two steps, namely, particle detection and particle linking. The resulting Lagrangian tracks were imported into MATLAB for Eulerian field reconstruction. This step transformed the scattered particle velocities into a dense, structured grid suitable for gradient analysis. Detailed description of this entire process and its validation can be found in Ré *et al.* (2026) In the present work, this validated workflow was employed in the context of patient-based phantoms. Hence, after obtaining clean, dense Eulerian velocity field, quantitative flow parameters were calculated. Velocity gradients were computed using MATLAB's gradient function and Wall Shear Stress was estimated along the domain perimeter using a 2D shear stress approximation for Newtonian fluids. The Reynolds number was determined for each case to monitor the consistency of the flow regime.

3. Results and Discussion

For each case considered, both the frontal and the lateral planes were analyzed to allow visualization of the three velocity components. Each plane of every model was analyzed under rest condition and dynamic exercise condition, as aforementioned in section 2. To highlight the most relevant results, rest cases are compared to dynamic conditions, which correspond to an increase in flow rate of 22% compared to the rest condition. Analysis was performed at steady-state and pulsatile flow conditions. Hence, the velocity interpolations obtained for case 1 are reported in Figure 3 for the frontal plane and Figure 4 for the lateral plane, for both rest and activity

conditions. The resulting WSS reconstruction is shown in Figures 5 and 6 for the frontal and lateral plane, respectively. Regarding the temporal analysis, the vortex displacement is illustrated in Figure 7 for the frontal plane and Figure 8 for the lateral plane.

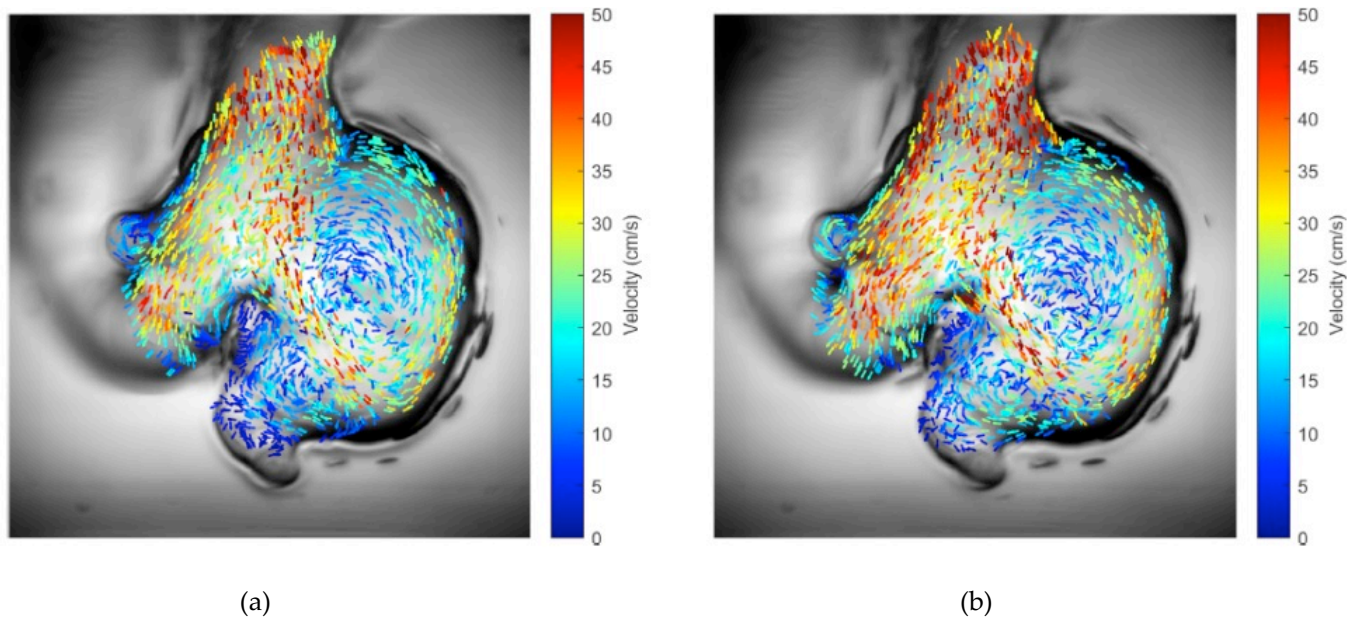


Figure 3: Interpolation of the velocities in the domain for a pulsatile flow for: (a) rest condition and (b) exercise condition, referred to case 1 - frontal plane.

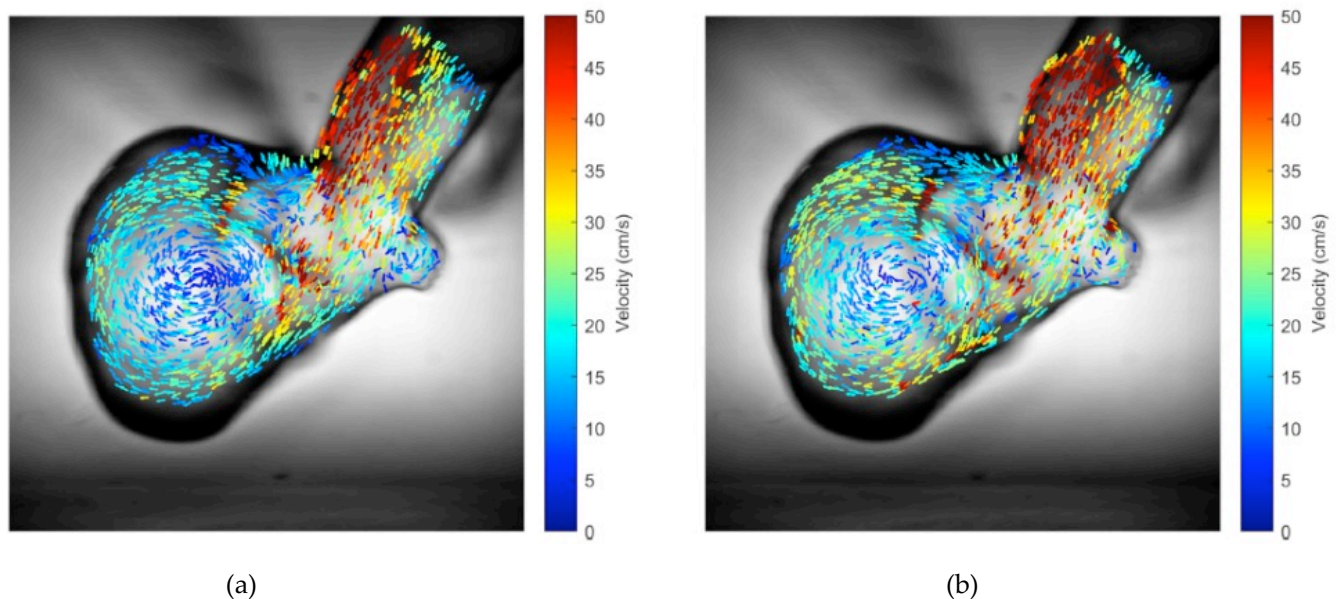


Figure 4: Interpolation of the velocities in the domain for a pulsatile flow for: (a) rest condition and (b) exercise condition referred to case 1 - lateral plane.

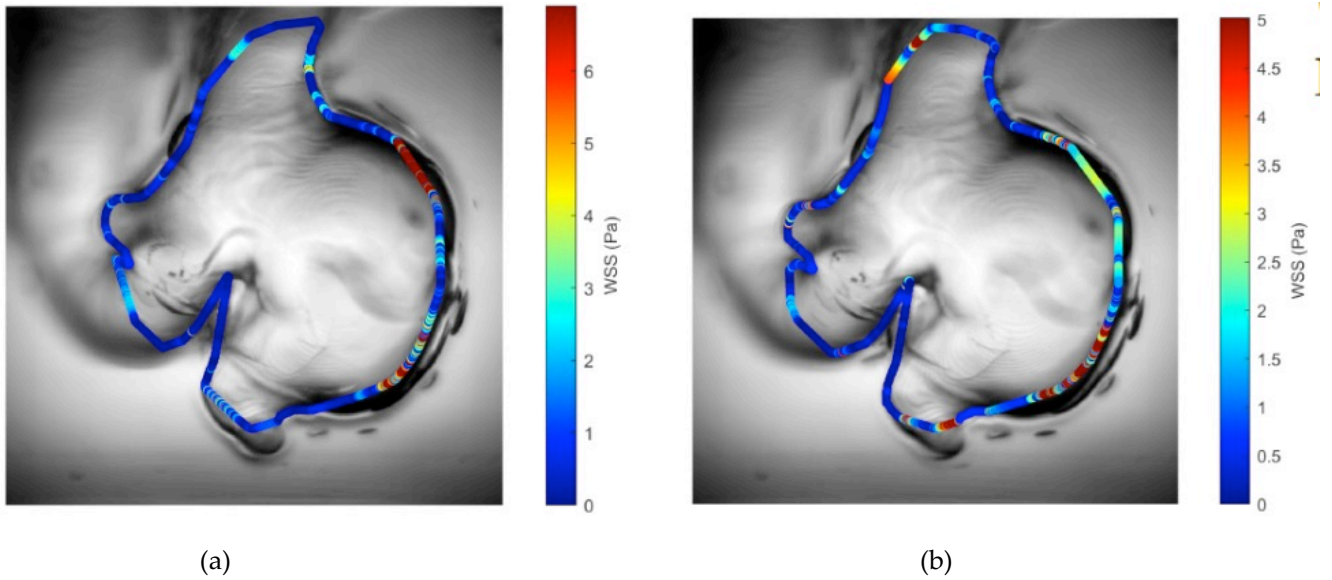


Figure 5: Reconstruction of WSS using pulsatile flow regime under: (a) rest and (b) exercise conditions for case 1 - frontal plane.

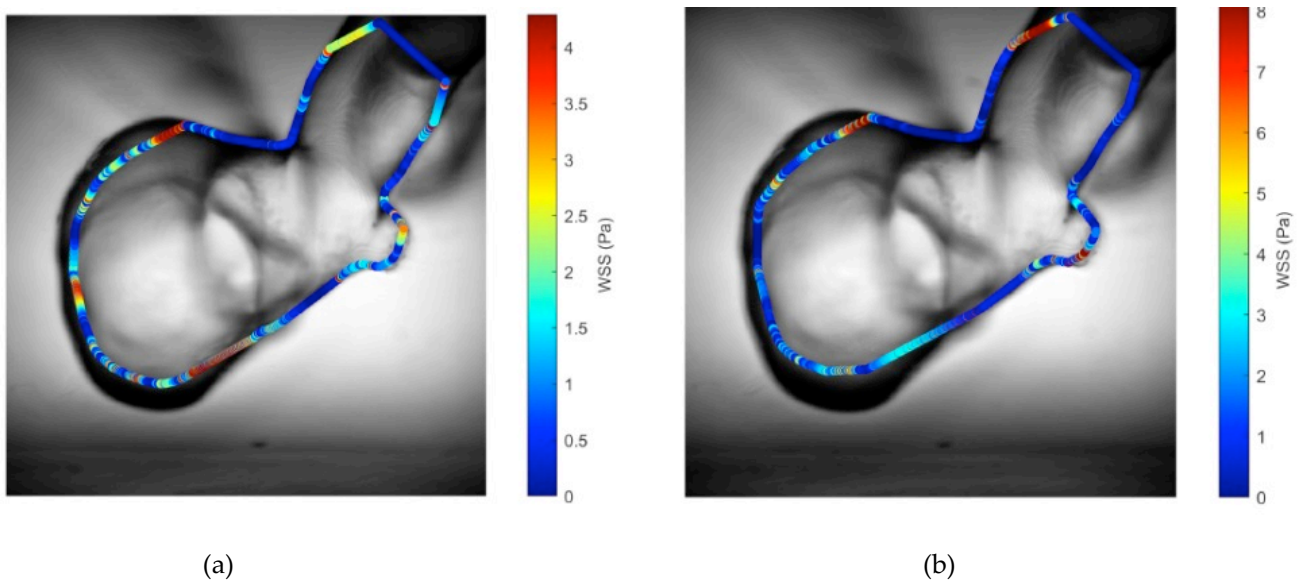
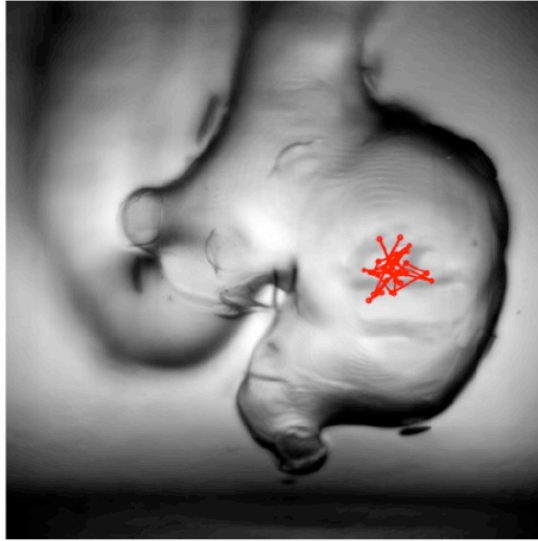
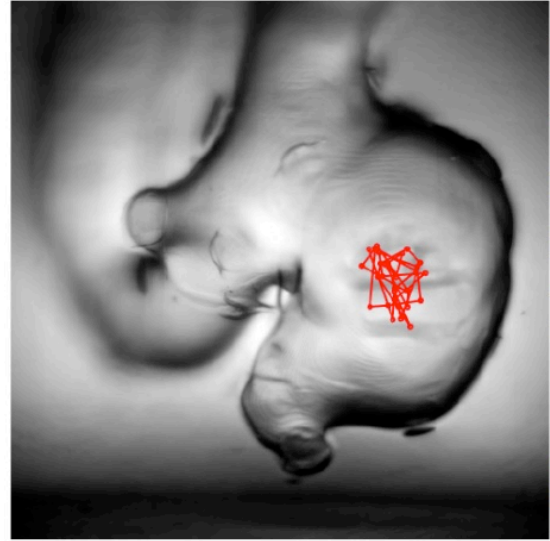


Figure 6: Reconstruction of WSS using pulsatile flow regime under: (a) rest and (b) exercise conditions for case 1 - lateral plane.

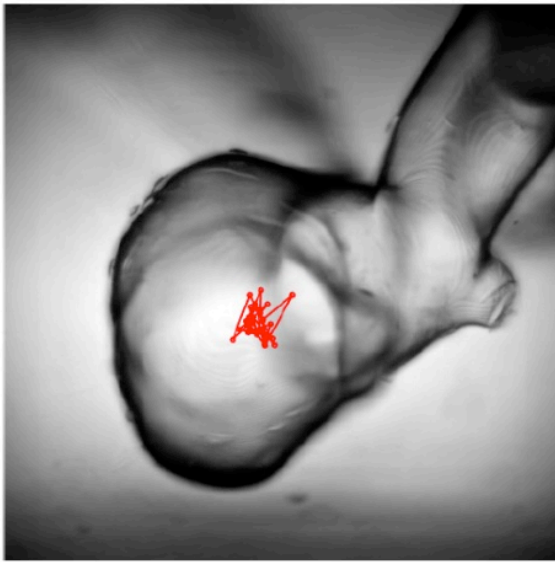


(a)

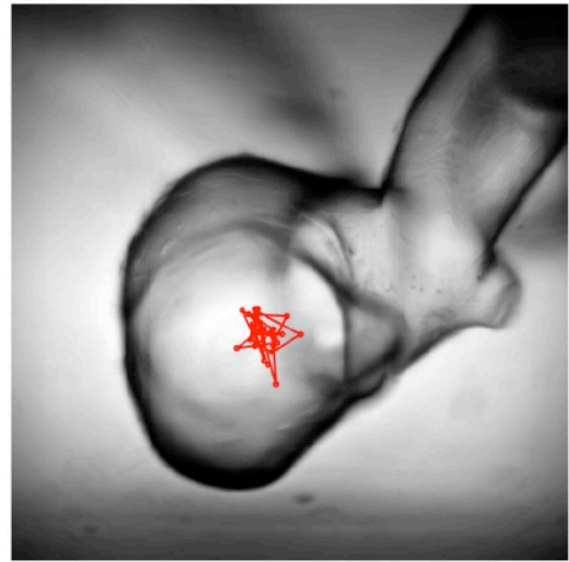


(b)

Figure 7: Tracking of the vortex movement across the various phases of the cardiac cycle in case of: (a) rest and (b) exercise conditions for case 1 – frontal plane.



(a)



(b)

Figure 8: Tracking of the vortex movement across the various phases of the cardiac cycle in case of: (a) rest and (b) exercise conditions for case 1 – lateral plane.

Following Table 2, it is now interesting to analyze case 6. However, in this case, only the results of rest condition are available because the pump cannot reach the maximum flow rate required to represent exercise conditions, like those reported for case 1. Hence, the velocity interpolation and WSS reconstruction are shown in Figure 9 and Figure 10 for the frontal and lateral planes,

respectively. The results from unsteady analysis are reported in Figure 11 for the frontal plane, while the vortex identification is not possible in the lateral plane due to the incomplete experimental reconstruction.

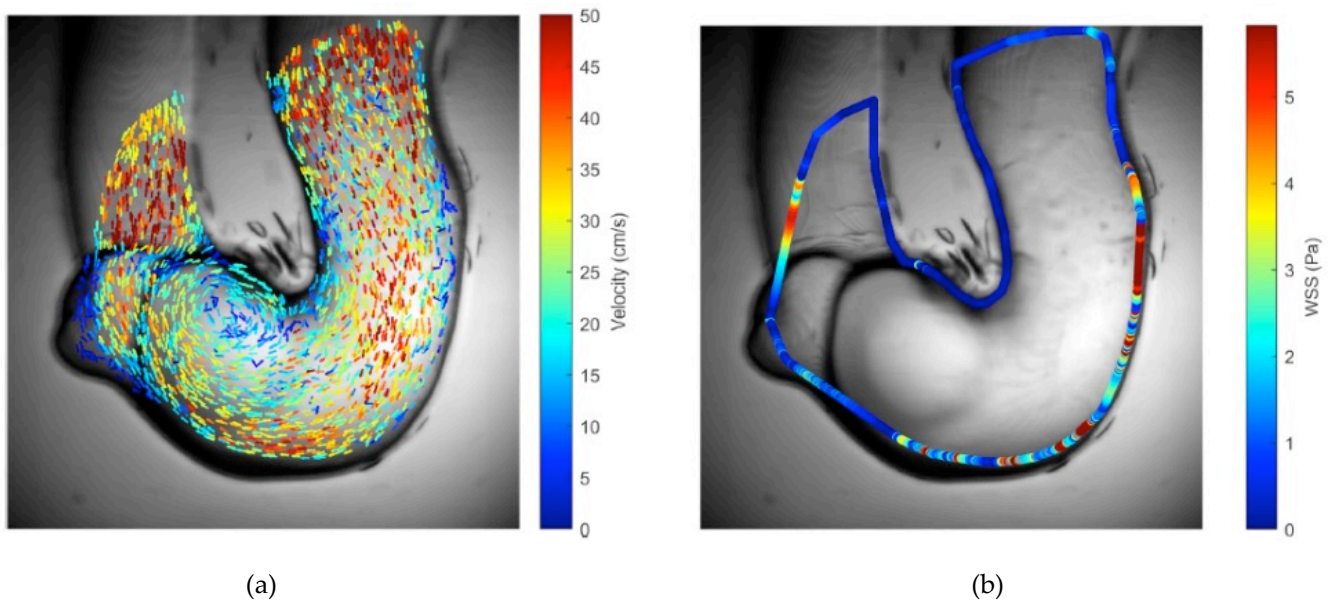


Figure 9: (a) Interpolation of the velocities in the domain for a pulsatile flow under rest condition condition and (b) reconstruction of WSS for case 6 - frontal plane.

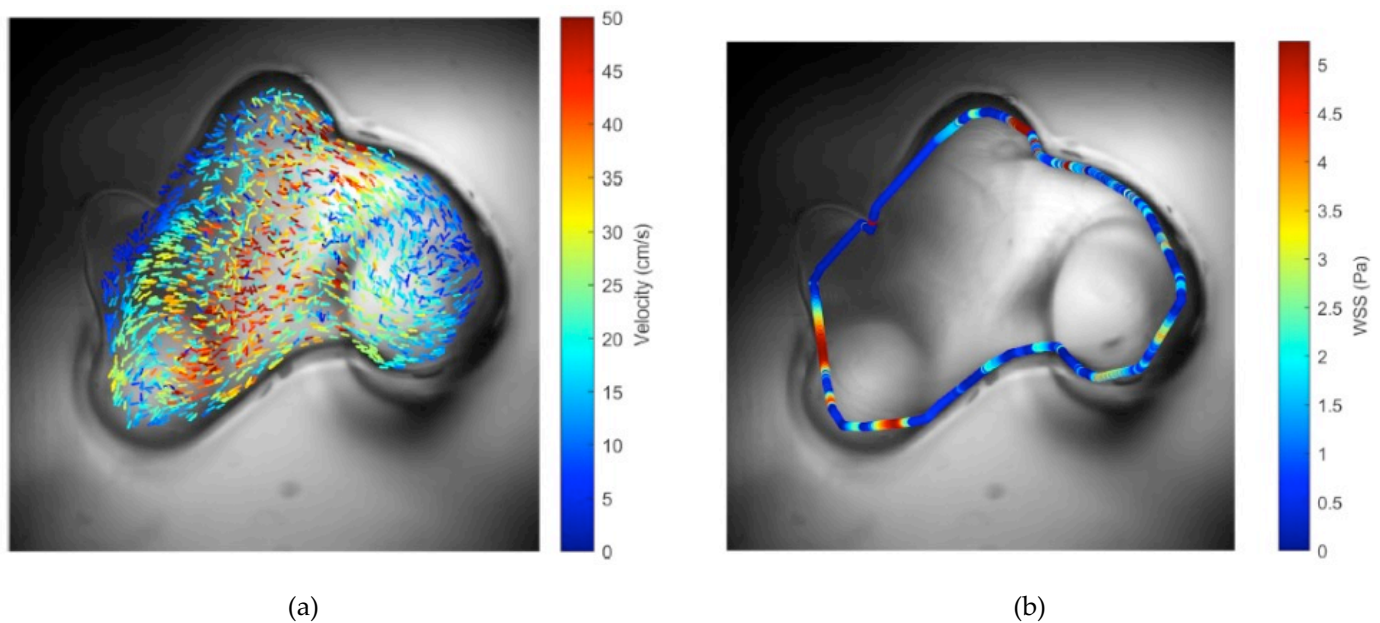


Figure 10: (a) Interpolation of the velocities in the domain for a pulsatile flow under rest condition condition and (b) reconstruction of WSS for case 6 - lateral plane.

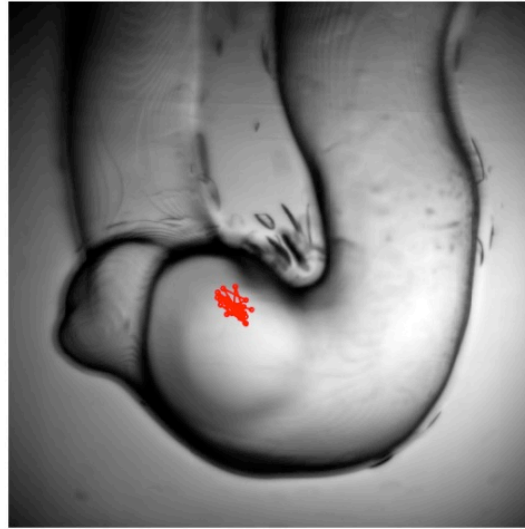


Figure 11: Tracking of the vortex movement across the various phases of the cardiac cycle in case of rest condition for case 6 – frontal plane.

On the other hand, regarding case 7, for each condition tested, the velocity interpolations are reported in Figure 12 for the frontal plane and Figure 13 for the lateral plane. The resulting WSS reconstruction is shown in Figures 14 and 15 for the frontal and lateral plane, respectively. As for the temporal analysis, the vortex displacement is illustrated in Figure 16 for the frontal plane and in Figure 17 for the lateral plane.

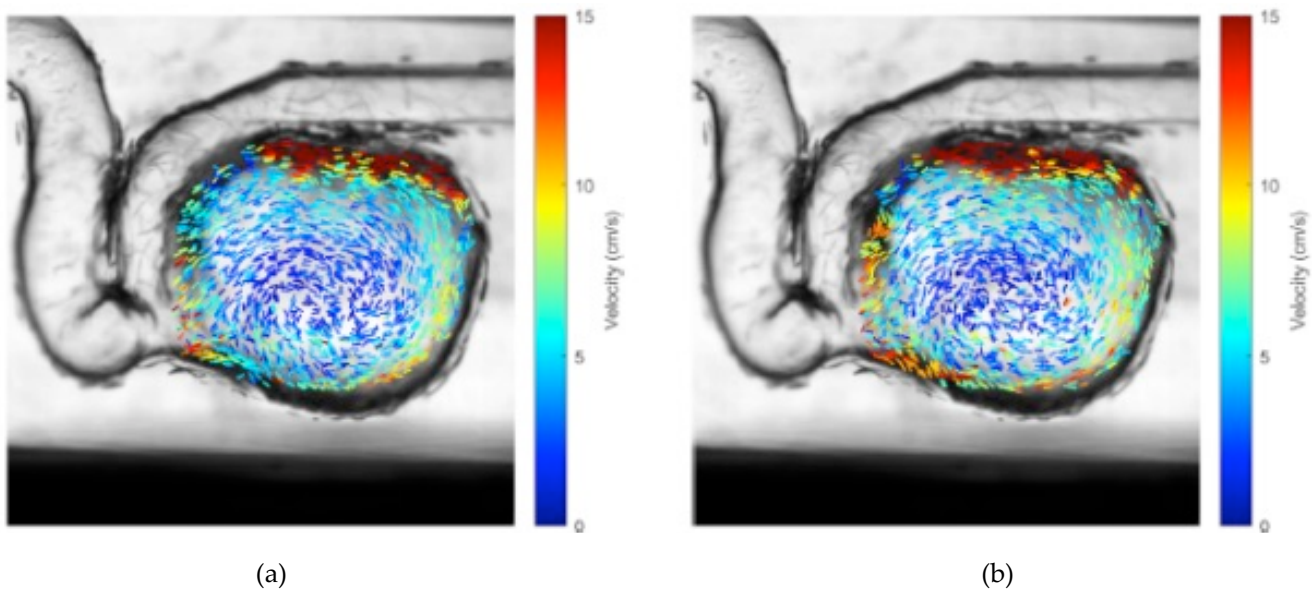


Figure 12: Interpolation of the velocities in the domain for a pulsatile flow for: (a) rest condition and (b) exercise condition, referred to case 7 - frontal plane.

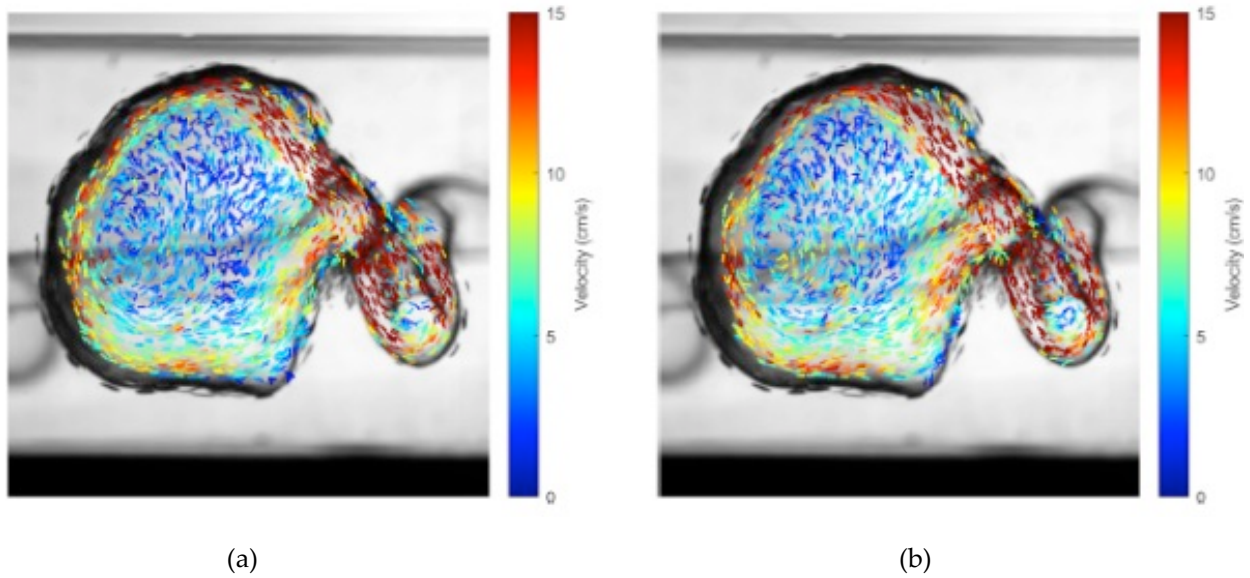


Figure 13: Interpolation of the velocities in the domain for a pulsatile flow for: (a) rest condition and (b) exercise condition, referred to case 7 - lateral plane.

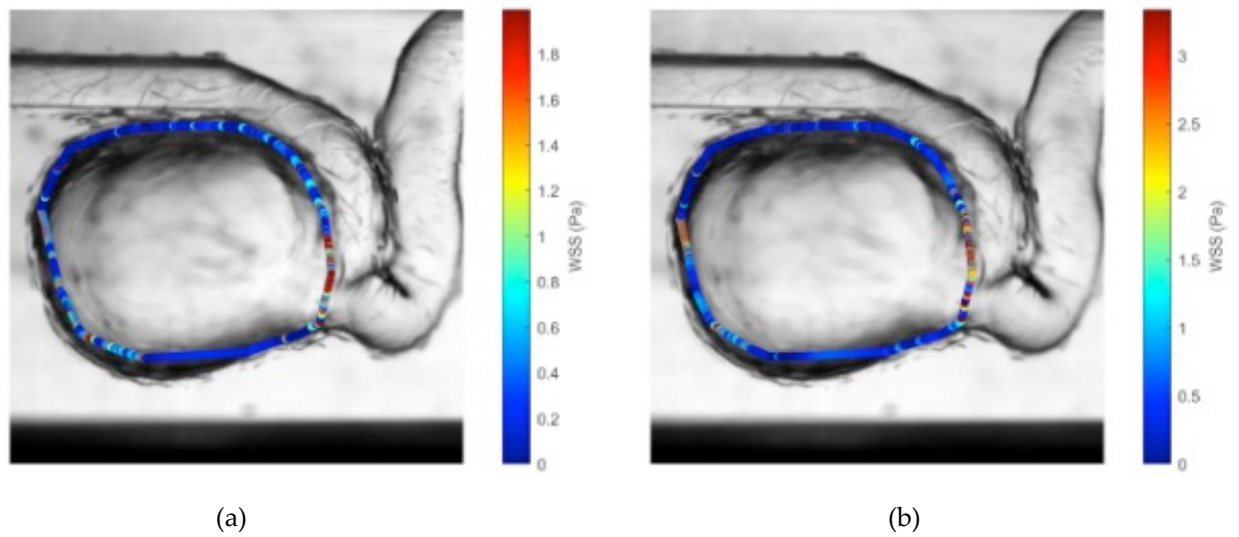


Figure 14: Reconstruction of WSS using pulsatile flow regime under: (a) rest and (b) exercise conditions for case 7 - frontal plane.

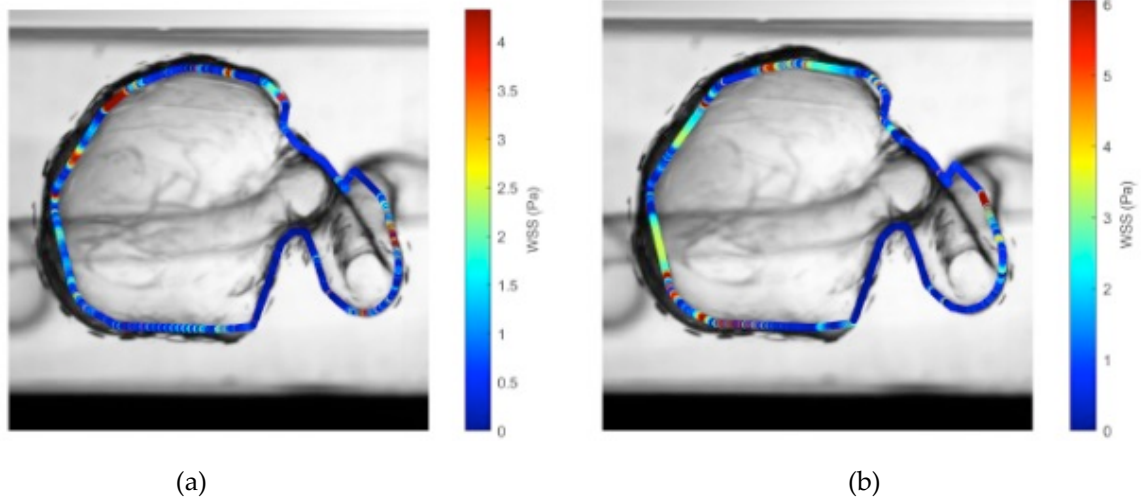


Figure 15: Reconstruction of WSS using pulsatile flow regime under: (a) rest and (b) exercise conditions for case 7 - lateral plane.

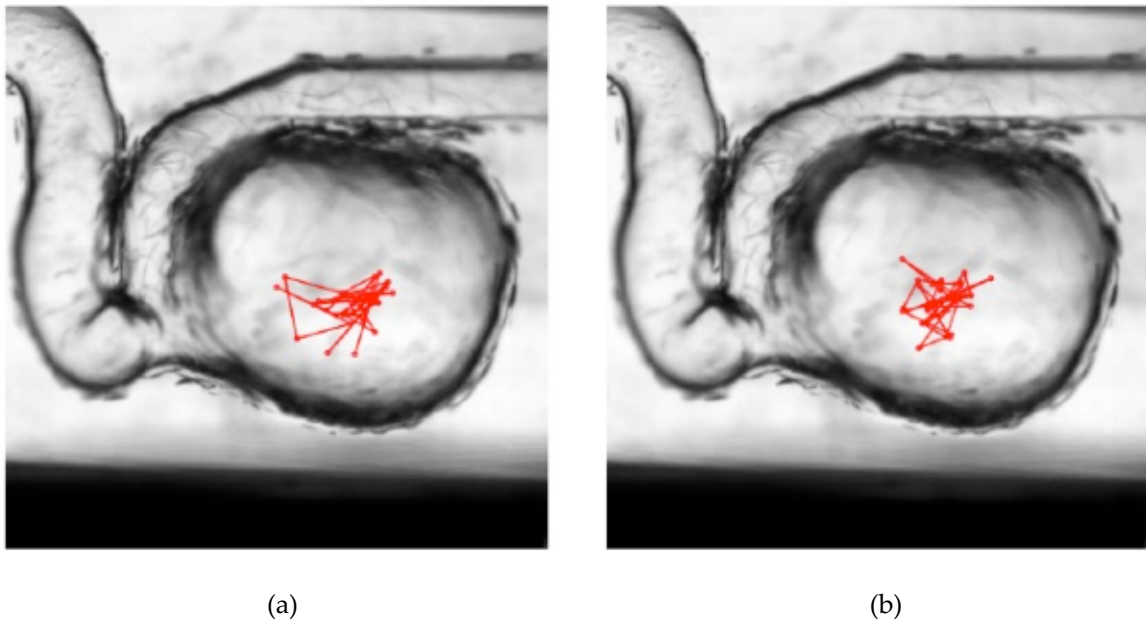


Figure 16: Tracking of the vortex movement across the various phases of the cardiac cycle in case of: (a) rest and (b) exercise conditions for case 7 – frontal plane.

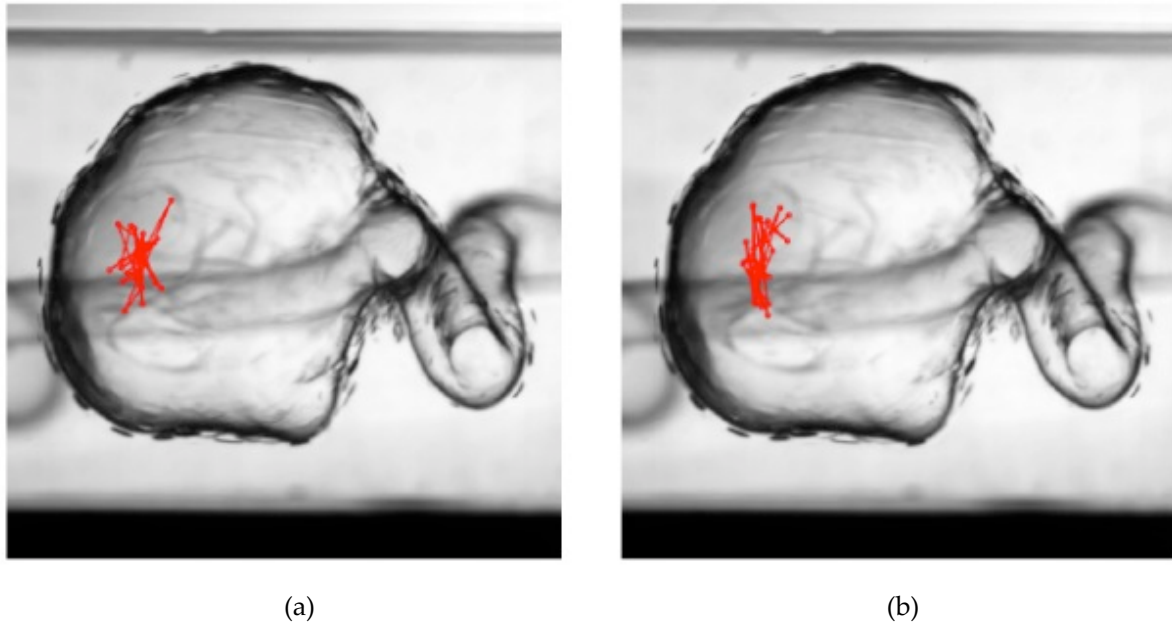


Figure 17: Tracking of the vortex movement across the various phases of the cardiac cycle in case of: (a) rest and (b) exercise conditions for case 7 – lateral plane.

The presented results highlight specific behaviors of the velocity field and WSS, which are supported by the computed hemodynamic parameters reported in Table 3 and Table 4.

Table 3: Percentage variation in various hemodynamic parameters from the resting to the exercise conditions for Case 1 and Case 7, in frontal and lateral planes.

	Case 1		Case 7	
	Frontal	Lateral	Frontal	Lateral
Mean velocity	16.4%	12.2%	26.9%	13,9%
Max velocity	10.2%	-16.9%	7.3%	-2.1%
Reynolds number	19.2%	17.0%	24.2%	17.9%
Womersley number	46.8%	46.8%	46.8%	46.8%
Vel percentile 10	19.7%	15.8%	29.3%	16.6%
Vel percentile 90	14.8%	8.9%	23.4%	13.4%
Mean WSS	21.7%	10.5%	23.5%	17.9%
Max WSS	-7.6%	10.9%	65.0%	-1.1%

Table 4: Various hemodynamic parameters calculated for resting conditions for Case 6, in frontal and lateral planes.

	Case 6	
	Frontal	Lateral
Mean velocity	30.9%	25.65%
Max velocity	76.04%	83.00%
Reynolds number	788	727
Womersley number	4.14	4.14
Vel percentile 10	12.71%	10.34%
Vel percentile 90	48.77%	44.38%
Mean WSS	1.7%	1.94%
Max WSS	47.17%	31.43%

From the calculated parameters, the increase in mean velocity under exercise conditions is evident, consequently leading to an increase in the Reynolds number. However, an increase in maximum velocity is not always observed, due to the inability to detect and remove outlier velocity components during the post-processing. For this reason, the 90th percentile velocity represents a more reliable metric, consistently confirming the increase under exercise conditions. Similarly, the trend of maximum WSS does not appear to be directly correlated with the flow rate, as it is more strongly influenced by the flow topology. While flow topology, like primary and secondary vortical structures, remains generally consistent despite variations in flow rate, velocity magnitude and WSS distribution demonstrate a strong sensitivity to these variations.

Exercise conditions generally lead to higher velocities, especially along the outer regions of the primary vortex, where high velocity components tend to persist over a longer portion of the recirculating path. This behavior directly affected the WSS distribution, often producing higher mean values and a broader spatial extension of stressed regions. However, when the increase in velocity produces a thicker high velocity layer, the WSS is reduced in that area due to the decrease in shear rate.

These considerations suggest that geometry plays a dominant role in determining the overall flow organization, while flow conditions modulate its intensity. This observation is consistent with literature, which reports that pulsatility mainly affects vortex dynamics and migration rather than the global flow organization itself. In particular, Womersley number (which represents the ratio between the pulsatile flow frequency and the viscous effects) affects the temporal evolution and migration of vortical structures during the cardiac cycle, while Reynolds number and flow rate influence vorticity intensity and flow disturbances (Chowdhury *et al.*, 2025). This consideration is proven by temporal analysis, which emphasizes that increases in Reynolds and Womersley numbers under exercise conditions are associated with more dynamic vortex behavior and larger temporal variations of the velocity field, without altering the overall topology of the flow structures.

The literature also discussed the influence of geometric parameters on hemodynamic variables. The Bottleneck factor (BF), defined as the ratio between the maximum aneurysm diameter and neck diameter, is often associated with the amount of blood entering the aneurysmal sac and participating in recirculation . Lower BF values generally facilitate stronger inflow penetration and more intense recirculating flow. Another parameter is the Aspect Ratio (AR), defined as the ratio between aneurysm height and neck width. High AR values are commonly associated with slower aneurysmal flow, lower WSS and increased flow stagnation (Chowdhury *et al.*, 2025). These parameters are reported in Table 5 for the analyzed cases.

Table 5: Geometric parameters, namely Bottleneck Factor (BF) and Aspect Ratio (AR) calculated for the various cases analyzed in this study.

	Case 1	Case 6	Case 7
Bottleneck factor (BF)	1.91	1.04	2.31
Aspect Ratio (AR)	1.12	0.51	2.03

In this study, the influence of geometry is evident in Case 7. This aneurysm is characterized by lower velocity magnitude within the aneurysmal sac compared to the parent vessel, and the vortex displacement under exercise conditions is lower than rest. This behavior can be explained

by the larger dimension and the higher AR that may damp the effect of unsteady inertial forces and stabilize the recirculating structures inside the sac.

Comparing vortex motion across the three analyzed cases, Case 6 exhibits the smallest displacement, being characterized by smaller aneurysm dimensions and the lowest AR. Case 1 presents intermediate dimensions and AR values, while Case 7, representing a giant aneurysm with the highest AR, shows the largest vortex displacement. These observations suggest a possible correlation between aneurysm size, AR, and vortex motion. However, since only three cases were analyzed, further investigations involving additional aneurysmal geometries would be necessary to validate this hypothesis.

4. Conclusions

This study addresses the use of a complete and validated computational workflow designed to transform raw 2D PTV images into Eulerian fields for biofluid analysis to investigate the hemodynamic behavior within patient-specific aneurysm phantoms using pulsatile flow regimes, considering both resting and exercise conditions. This PTV method can be used for the non-invasive in vitro investigation of biomechanical parameters in intracranial aneurysms, as well as for the reconstruction of velocity field and Wall Shear Stress (WSS). A clear sensitivity to changes in flow rate between resting and exercise conditions was observed. Pulsatile flow conditions were also experimentally implemented, allowing the investigation of a more realistic unsteady flow condition.

The results showed consistency in flow patterns across the tested conditions and coherence passing from velocity fields to WSS distributions. The qualitative behavior observed in the images was generally confirmed by the computed hemodynamic parameters. This study confirmed the influence of hemodynamic parameters on flow behavior: while geometry determines the overall organization of the flow field, flow conditions modulate its intensity.

There are however some limitations of this approach to overcome. The main limitations derive from the twodimensional nature of the measurements, forcing the selection of a single plane for flow visualization. However, the 2D nature is what keeps the method simple and cost effective.

Moreover, the phantom models are characterized by rigid walls, representing a simplification of the real physiological condition, as well as the impossibility to evaluate the cellular response and vascular wall adaptation to hemodynamic stimuli. In addition, the working fluid used exhibits Newtonian behavior, whereas blood is intrinsically non-Newtonian.

These limitations also affect the post-processing phase, preventing the reconstruction of a fully three-dimensional velocity field and limiting vortex tracking to a two-dimensional analysis. Despite these limitations, to overcome in future work, the methodology presented here showed validated results able to provide additional information in terms of flow and hemodynamic characteristics that may be used to validate complex numerical models and aid physicians to further understand the flow characteristics and further relate them with physiological and morphological aspects explaining aneurysms rupture.

Acknowledgments

Authors are grateful to the Fundação para a Ciência e a Tecnologia (FCT), for partially financing the project Waste2NFs, 2022.06207.PTDC (<https://doi.org/10.54499/2022.06207.PTDC>). The authors also acknowledge the partial financial support by national funds through UIDP/50009/2025 and UIDB/50009/2025. Finally, A.S. Moita also acknowledges FCT for partially financing her contract through CEECINST/00043/2021/CP2797/CT0005, doi: 10.54499/CEECINST/00043/2021/CP2797/CT0005.

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