

Fluid–Structure Interaction of a Stented Elastic Straight Vessel

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ABSTRACT

Stent implantation is a widely adopted endovascular treatment for cardiovascular diseases, yet the associated alterations in local haemodynamics and fluid–structure interaction (FSI) remain insufficiently understood, despite their strong links to complications such as restenosis, thrombosis, and delayed occlusion. This study presents an experimental investigation of the flow field and structural response of a stented elastic straight vessel using two-dimensional two-component (2D/2C) particle-image velocimetry (PIV). A silicone phantom with a wall thickness of 1 mm and an inner diameter of $D_0 = 22$ mm was manufactured from polydimethylsiloxane (PDMS) and mounted in an open-loop flow facility with a refractive index matched working fluid. A commercial self-expanding nitinol stent (sinus-XL) with $d_{stent} = 24$ mm was employed to investigate the effect of stent implantation on the vessel wall dynamics and the bulk flow field. Pulsatile flow conditions covering two Reynolds numbers ($Re_D = 1000$ and $Re_D = 1500$) and three Womersley numbers ($Wo_D = 13, 15$, and 17) were investigated. Analysis of the structural response of the vessel reveals that stent implantation increases the amplitude of pulsatile wall dilatation directly downstream of the stent during both the expansion and relaxation phases of the pulsatile cycle, while leaving the mean inner diameter largely unaffected. Additionally, the stent introduces a temporal shift in the vessel wall motion, with the expansion phase being shortened relative to the unstented configuration. Both the amplitude increase and the phase shift are most pronounced at the lowest Womersley number. The ensemble-averaged velocity profiles measured at $x = 0.5 \cdot D_0$ show that the stent exerts a moderate influence on the bulk flow field, with slightly higher maximum velocities in the stented case at the phase angles of maximum and minimum inner diameter. The results provide new quantitative insight into the coupling between vessel compliance, stent mechanics, and pulsatile haemodynamics, with direct implications for the understanding of stent-related complications such as restenosis and thrombosis in large compliant vessels.

1. Introduction

Cardiovascular diseases, such as atherosclerosis, aneurysms, and occlusive vascular pathologies, remain one of the leading causes of death worldwide Xu et al. (2021). In recent decades, endovascular interventions such as stent implantation have become a standard therapeutic approach to

address this challenge. These interventions aim to restore vessel patency, reconstruct the parent artery, and prevent disease progression. However, despite the clinical success of this approach, stented vessels are frequently associated with complications such as restenosis, thrombosis, or delayed occlusion. These complications are strongly linked to altered local haemodynamics and vessel–stent interactions Williamson et al. (2022). Thus, improving the design of the device and hence the long-term clinical outcome requires a detailed understanding of the flow field in stented arteries. Previous experimental and numerical studies have demonstrated that stent implantation can substantially modify wall-shear stress (WSS), oscillatory shear index (OSI), and flow patterns, particularly in the vicinity of stent struts and at the transition between compliant vessels and comparatively stiff devices Alkhalili et al. (2018). However, the degree of physiological realism in existing investigations varies considerably, particularly with respect to vessel compliance, pulsatility of the flow, fluid–structure interaction effects, and stent design. While the influence of individual parameters, such as pulsatile flow or stent geometry has been investigated in isolation, the combined effect on compliant vessels remains insufficiently understood.

Studies investigating the effect of stent implantation on the flow field in compliant vessels using spatially resolved measurements such as particle-image velocimetry (PIV) are scarce. Charonko et al. (2009) employed time-resolved PIV in compliant coronary artery phantoms at two Reynolds numbers $Re = [150, 300]$ and demonstrated that different commercial stent designs result in significant variations in WSS and OSI under pulsatile flow conditions. The study identified an inverse correlation between the WSS and OSI within the stent, i.e., while the WSS decreases, the OSI increases. These results highlight the importance of stent geometry in determining the near-wall flow environment and its effects on the endothelial cells. Similarly, Dennis et al. (2015) investigated the steady flow field at three Reynolds numbers using tomographic PIV in an elastic aneurysm model, quantifying the haemodynamic impact of two flow diverters. The study showed that both devices removed the large baseline vortex inside the aneurysm dome at all flow rates, reducing peak aneurysmal velocities by around 90% by blocking the flow into the neck. Consequently, the inflow shifted from the distal neck to the mid- or proximal neck. The transition between the compliant native vessel and the comparatively stiff stent or stent graft represents a critical region for haemodynamic disturbances. Yazdi et al. (2021) investigated this transition in a compliant iliac artery model during pulsatile flow for a peak Reynolds number of $Re = 958$ using standard two-dimensional two-component (2D/2C) PIV. The study identified the formation of low-velocity regions, flow recirculation, and fluctuating shear stresses close to the wall and distal to the employed stent graft, which can be attributed to a compliance mismatch between the graft and the parent artery. This may consequently result in intimal thickening, limb occlusion, and graft obstruction.

Despite the clinical relevance of stenting in large arteries, experimental in-vitro investigations that combine non-intrusive flow measurements, elastic vessel models, physiological flow conditions, and different stent design parameters and resulting mechanics are sparse. Thus, the need to identify and quantify the haemodynamic disturbances at the transition between the stented and na-

tive vessel, where compliance mismatch is known to promote restenosis and intimal hyperplasia, has not yet been met. This study therefore presents an interdisciplinary approach of combining state-of-the-art braided stents and non-intrusive (2D/2C) particle-image velocimetry (PIV) measurements inside an elastic straight vessel using refractive-index matching. This study aims to accurately capture the pulsatile inflow and examine the flow field and fluid–structure interaction (FSI) in both stented and unstented vessels to achieve a more comprehensive understanding of the underlying interaction mechanisms.

2. Experimental Setup and data evaluation

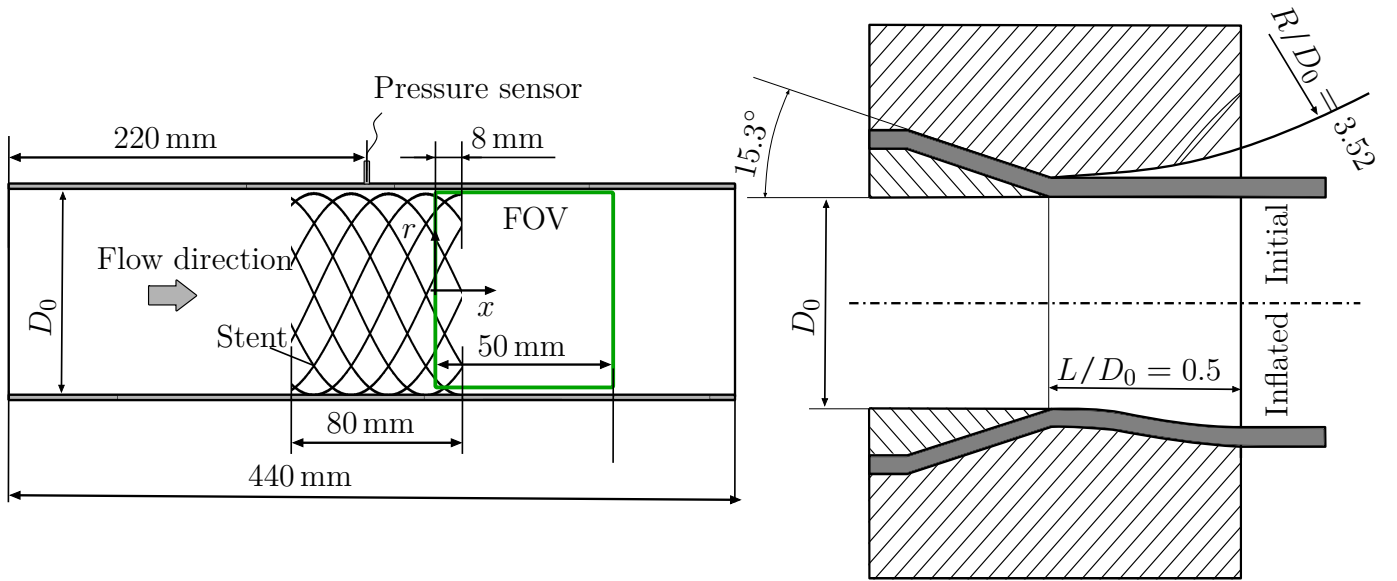
All of the experiments described in this paper were conducted in an open-loop setup, as illustrated in Figure 2. The individual components of the experimental setup, i.e., vessel properties, fluid properties, boundary conditions, the pump system, and the measurement system, are discussed in detail in the following sections.

2.1. Silicone model

The silicone model is manufactured from RTV-615, a two-component polydimethylsiloxan (PDMS), using a centrifugal casting process inside a precision steel tube with an inner diameter of $D_0 = 24$ mm, following the approach introduced by Burgmann et al. (2009) and Pielhop et al. (2015). The wall thickness of the vessel after curing depends on the volume of silicone inserted into the steel tube. In order to match the material properties of the arterial walls, see Learoyd & Taylor (1966), the volume of silicone is chosen so that wall thickness of the phantom is constant at 1 mm. Consequently, the elastic vessel has an inner diameter of $D_0 = 22$ mm. During the mixing of the two PDMS components, a small amount of Rhodamine B dissolved in methanol is added to the silicone, resulting in a clearly visible fluorescence of the vessel. This fluorescence enables the position of the vessel wall to be determined by analyzing the grayscale distribution in the measurement images. Before casting, the steel tube is polished with a release agent to simplify the demoulding process and is visually checked for dust contamination. The tube has an 8 mm transverse hole in its center to create a negative mould for a small pressure sensor. The vessel is then cast in two consecutive steps at controlled temperatures and fixed rotations per minute (rpm). The rotational speed is measured using a digital tachometer, set to 1500 rpm, and controlled by a programmable power supply. First, the temperature is set below 15 °C to evenly distribute the silicone in the casting tube for four hours. Next, the temperature inside the insulating box is increased to 65 ± 2 °C and held constant for four additional hours to fully cure the silicone.

After curing, the vessel is cut to a length of 400 mm. During the experiments, the vessel is stretched by 10 %, resulting in a final length of approximately $L = 440$ mm. The geometry of the straight

vessel under experimental conditions, the position of the stent, the field of view (FOV), and the coordinate system are shown in Figure 1 a). The origin of the axial–radial (x – r) coordinate system is located approximately 8 mm ($0.37 \cdot D_0$) upstream of the distal end of the stent. Figure 1 b) shows the vessel mount during experiments, for an initial and inflated load, as introduced by Pielhop et al. (2015). Special adapters are designed to connect the flexible vessel to rigid pipes up- and downstream of the vessel to minimize diameter changes in the vicinity of the mounts. Their longitudinal curvature possesses a radius of $R = 3.52 \cdot D_0$ to compensate for inflation and thus buckling of the compliant vessel.



a) Stent position, FOV, and coordinate system, b) Vessel mount during initial and inflated load

Figure 1. Sketch of the elastic vessel, the positioning of the stent and the mounting of the vessel during initial and inflated load, adapted from Pielhop et al. (2015).

2.2. Flow facility

The refractive index of the silicone model is matched by a water/glycerine mixture with a refractive index of $n = 1.406$. This mixture allows optical access without noticeable distortion. The mixture consists of 56.77 wt% glycerine and 43.23 wt% distilled water, resulting in a density of $\rho_{WG} = 1.139 \text{ kg m}^{-3}$ and a dynamic viscosity of $\eta_{WG} = 0.0069 \text{ Pa s}$ for a temperature of $T = 30^\circ\text{C}$, which is held constant throughout all experiments, using six active temperature controllers and heating cables throughout the open test loop. The elastic vessel is then placed inside a closed compartment that is filled with the same transparent refractive-index matched fluid that is used as the working fluid. The container serves two purposes.

On the one hand, the combination of the refractive-index matched fluid in the container and inside

the vessel, the flat transparent cover, and the flat side walls provides an undistorted view of the flow inside the artery model. On the other hand, two reservoirs serve as a mechanism to adapt the transmural pressure p_{trans} , i.e., the pressure difference between the inside and the outside of the flow phantom: one is connected to the compartment (reservoir 1), and the other is connected to the inner flow (overflow reservoir 2). By positioning these reservoirs at different heights (h_1 and $h_2 > h_1$), as indicated in Fig. 2), the transmural pressure difference can be adjusted according to $p_{trans} = \rho_{WG} \cdot g \cdot (h_2 - h_1)$, with g as the gravitational acceleration. In accordance with previous investigations of flow in other geometries (Pielhop et al., 2015; Dörner et al., 2021), the transmural pressure is set to $p_{trans} = 12\,467\text{ Pa}$. This ensures that the pressure difference remains positive throughout the entire excitation cycle.

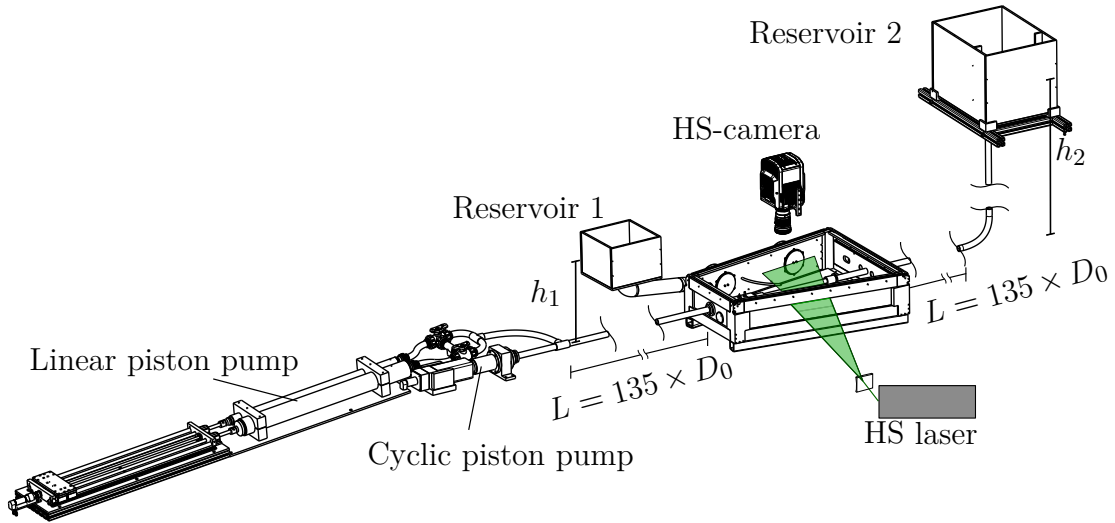


Figure 2. Isometrical view of the experimental PIV setup.

The vessel is connected to rigid straight pipes with $L/D_0 = 138$ (L : length, D_0 : inner diameter of the straight vessel) upstream and downstream of the measurement tank to ensure a fully developed velocity profile. As shown in Figure 3, the time-dependent pulsatile volume flux $\dot{V}_{sum}(\phi)$ is generated by the superposition of two piston pumps. The first pump performs a sinusoidal motion, while the linear piston offsets the negative volume flux with a constant positive flux. The phase angle $\phi = 0^\circ$ is defined at the zero crossing of the piston displacement

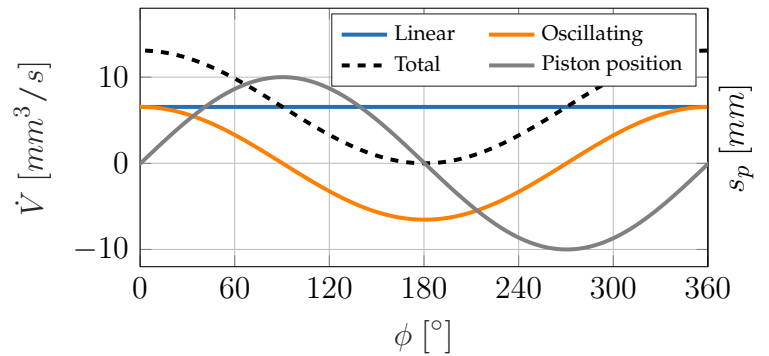


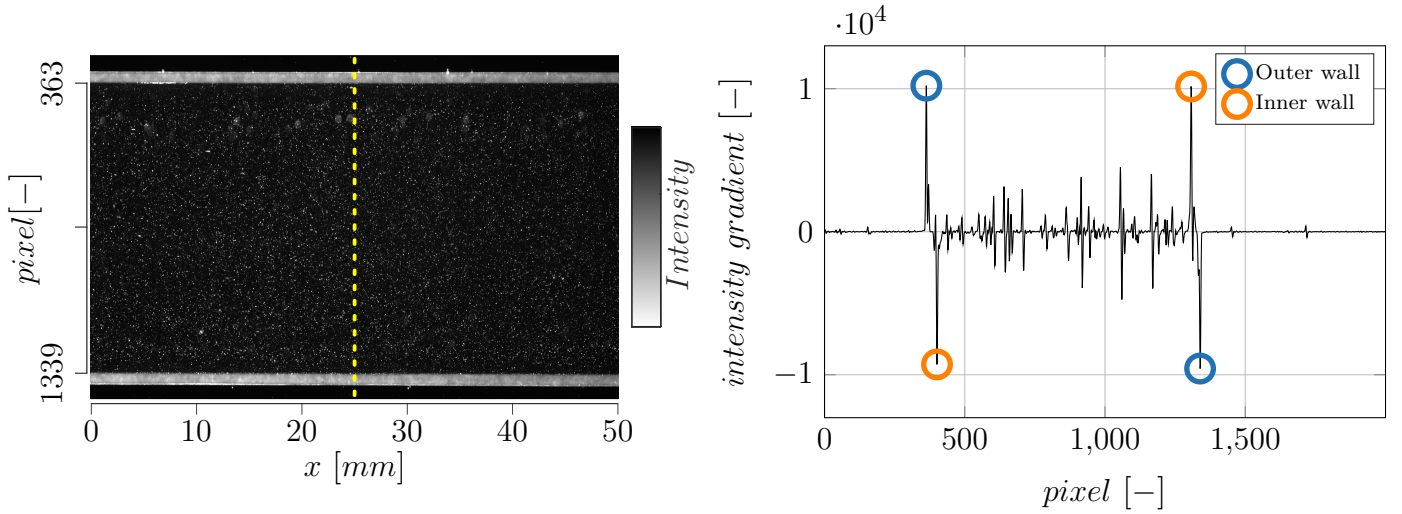
Figure 3. Volume flux for the oscillating piston $\dot{V}_{osc}(\phi)$, linear piston $\dot{V}_{linear}(\phi)$, the sum $\dot{V}_{sum}(\phi)$, and piston position $s_p(\phi)$ for an exemplary pulsatile waveform.

ment $s_p(\phi) = 0$ during the expansion stroke, with one complete pulsatile cycle corresponding to $\phi = 360^\circ$, as illustrated in Figure 3 for an exemplary waveform. The pumps are driven by high-precision motors with active position feedback. Teflon rings seal the pistons to prevent stick-slip and ensure a tight but smooth seal.

2.3. Measurement technique and data evaluation

The measurement setup consists of a particle-image velocimetry (PIV) system and linear sensors for measuring the stroke of the piston. In order to synchronize the stroke of the cyclic piston, i.e., the phase angle of the pulsatile cycle, with the PIV results, the piston position is measured using a magnetostrictive linear sensor that is mounted on the piston rod. The PIV system comprises one PCO dimax HS4 high-speed camera equipped with a Tamron 180 mm macro lens. As illustrated in Fig. 2, the camera is positioned perpendicular to the measurement plane inside the silicone model, at a distance of approximately $\delta_{cam} = 400$ mm from the measurement plane. This provides a spatial resolution of 37 px mm^{-1} . The PIV images are recorded at a constant frequency of 200 Hz. A Quantronix Darwin-Duo 100 mJ Nd:YLF laser with a wavelength of $\lambda = 527$ nm equipped with a light sheet optic is used to illuminate the measurement plane, as indicated in Figure 2. The FOV is positioned directly downstream of the distal stent edge, as illustrated in Figure 1 a), targeting the transition zone between the stented and native vessel, which is particularly prone to atherosclerotic plaque formation and calcification due to compliance mismatch-induced flow disturbances.

To determine the instantaneous position of the vessel wall, an automated detection routine is used to identify the wall location directly from the intensity distribution of the particle images, see Figure 4. Due to the addition of Rhodamine B, the vessel wall appears as a distinct intensity transition in the image. Therefore, the intensity gradient in the wall-normal direction, i.e., along the y -direction, is calculated. For each x -position, the expected upper and lower wall locations are identified within a predefined search window, and the local inner wall positions are then used to calculate the local inner vessel diameter based on the mapping of the calibration. The inner diameter D_i is subsequently obtained by spatially averaging over all x -positions within the field of view.



(a) Single particle image colorcoded by the grayscale intensity, with the dashed yellow line indicating the position of the extracted intensity gradient. (b) Intensity gradient along a single pixel in wall-normal direction. Red and green circles indicate the outer and inner wall positions extracted from the gradient distribution.

Figure 4. Workflow of the automated wall detection procedure. a) shows the raw particle image and b) shows the extract of the vertical (wall-normal) gradient of the particle image.

2.4. Stent

This study uses a commercial self-expanding nitinol stent (sinus-XL) with a diameter of $d_{stent} = 24$ mm and a length of $l_{stent} = 800$ mm. The sinus-XL is a self-expanding vascular stent indicated for implantation in large vessels such as the aorta and vena cava. It is used for the treatment of vascular stenoses and vena cava obstructions, including those associated with external tumor compression.

2.5. Flow parameters

To characterize the dynamic properties of the test rig and assess whether the presence of the stent alters the dynamic response of the coupled fluid–structure system, a step response excitation is applied and the resulting pressure decay is analyzed for both the unstented and stented configuration. Figure 5 shows the dynamic response of the system, by plotting the resulting pressure signal relative to the actuation represented by the piston velocity $v_n = v(t)/v_{\max}$ and the resulting normalized static pressure signal $p_n = p(t)/p_{\max(s)}$, where $v_{\max}$ and $p_{\max}$ denote the maximum values of the piston velocity and static pressure at the first peak of the step response excitation for two configurations, i.e., the unstented elastic vessel and a stented elastic vessel. The distinct damped oscillatory behaviour indicated by the pressure data reveals the dynamic properties, i.e., the eigenfrequency of the setup. The eigenfrequency of the test rig is a key characteristic of the coupled fluid–structure system because it governs the elastic vessel’s dynamic response to oscil-

latory excitation. In pulsatile flow experiments, the interaction between the fluid inertia and the structural compliance of the vessel leads to resonance phenomena that significantly impact pressure, flow rate, and vessel deformation. The eigenfrequency f_0 for both test rig configurations is determined from the decay of the static pressure following step response excitation. The system behaves as a damped oscillator and the eigenfrequency is obtained from the damped oscillation frequency f_d and the damping coefficient δ as

$$f_0 = \sqrt{f_d^2 + \frac{\delta^2}{4\pi^2}}, \text{ with } f_d = \frac{1}{T_d}, \text{ and } T_d = t_{i+1} - t_i \quad (1)$$

where f_d denotes the damped frequency that is defined by the period of successive peaks of the decay signal. The damping coefficient $\delta = \Lambda \cdot f_d$ is determined by the logarithmic decrement $\Lambda = \ln\left(\frac{y_i}{y_{i+1}}\right)$ using the successive amplitudes of the pressure decay curve y_i and y_{i+1} . The unstented setup exhibited a damped frequency of $f_d = 0.567$ Hz with a damping coefficient of $\delta = 0.018$, whereas the stented setup showed a damped frequency of $f_d = 0.571$ Hz and a damping coefficient of $\delta = 0.012$. Although the pressure decay curves differ in shape, the oscillatory response remains sufficiently distinct to determine the damped frequency reliably. The resulting damped frequencies of both configurations are nearly identical, indicating that the presence of the stent does not substantially alter the dominant eigenfrequency of the coupled fluid–structure system. However, the stented configuration exhibits a reduced damping coefficient of approximately 33 %, consistent with the increased structural stiffness introduced by the stent limiting the energy dissipation capacity of the vessel wall.

To evaluate how close the excitation is to resonance, the normalized frequency $f_n = \frac{f}{f_0}$ for each case is shown in table 1, where f denotes the excitation frequency of the piston-driven flow.

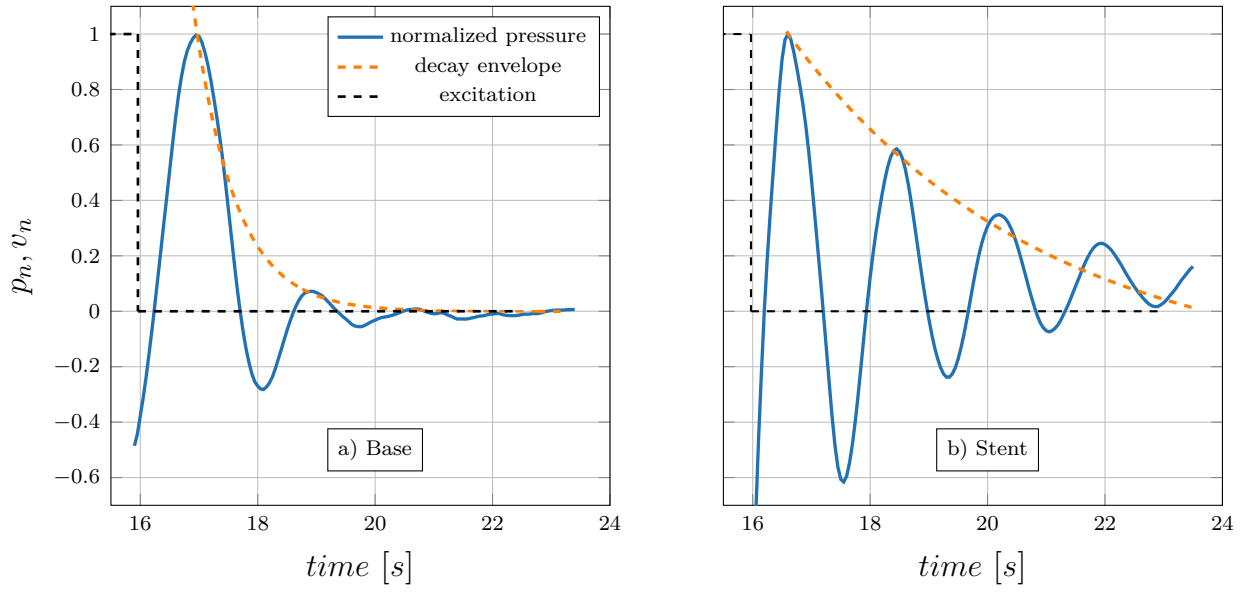


Figure 5. Normalized static pressure signal p_n and piston velocity v_n for the unstented and stented elastic vessel following step response excitation, used to determine the eigenfrequency of the system.

The oscillating Reynolds number is defined as

$$Re_D = \frac{\rho_{WG,30^\circ\text{C}} \cdot u_B \cdot D_0}{\eta_{WG,30^\circ\text{C}}} \quad (2)$$

and is calculated using the bulk velocity at maximum volume flux u_B , the inlet diameter of the straight vessel D_0 , and the density and dynamic viscosity of the refractive-index matched fluid at 30°C , i.e., ρ_{WG} and η_{WG} . The Womersley number reads

$$Wo_D = (D_0/2) \cdot \sqrt{\frac{\omega \cdot \rho_{WG}}{\eta_{WG}}}, \quad (3)$$

where ω denotes the angular frequency of the driving signal. The flow parameters of the conducted experiments, namely the Reynolds number Re_D , the Womersley number Wo_D , the excitation frequency f , the normalized frequency f_n , the bulk velocity at maximum volume flux u_B , and the number of cycles per ensemble N_i , for the baseline and the stent respectively, are summarized in Table 1.

Table 1. Overview of the investigated cases, Reynolds and Womersley numbers with the corresponding piston excitation frequencies f , the normalized frequency f_n , the bulk velocity at maximum volume flux u_B , and the number of cycles per ensemble N_i for the baseline and stented case.

Case	Re_D	Wo_D	f	f_n	u_B	N_{base}	N_{stent}
[-]	[-]	[-]	[Hz]	[-]	[m s ⁻¹]	[-]	[-]
A1	1000	13	0.98	1.74	0.22	7	7
A2	1000	15	1.31	2.31	0.22	16	16
A3	1000	17	1.68	2.97	0.22	19	19
B1	1500	13	0.98	1.74	0.33	13	13
B2	1500	15	1.31	2.31	0.33	14	14
B3	1500	17	1.68	2.97	0.33	19	19

3. Results

The following section presents the structural and haemodynamic response of the elastic vessel for both the baseline and the stented configuration. First, the influence of the stent on the phase-resolved inner vessel diameter is analyzed. Subsequently, the corresponding ensemble-averaged velocity profiles are evaluated at characteristic phase angles of the pulsatile cycle.

3.1. Structural response of the vessel

Figure 6 shows the inner diameter D_i as a function of the phase angle ϕ for all six investigated cases. Herein, blue lines correspond to $Re_D = 1000$ and orange lines represent the measurements at $Re_D = 1500$, respectively. While solid lines indicate the base configuration, dashed lines correspond to the stented configuration. The wall dilatation exhibits a sinusoidal oscillation around the initial diameter of $D_i = 23.4$ mm and $D_i = 23.6$ mm for the baseline and stented configuration at rest, respectively. The small difference of 0.2 mm between both configurations is attributed to the moderate oversizing of the stent ($\approx 2.6\%$ with respect to the initial diameter of the unstented vessel). The stent-induced increase in peak dilatation grows with Reynolds number, reaching approximately 0.52% and 1.44% for $Re_D = 1000$ and $Re_D = 1500$, respectively. Furthermore, as expected, a clear trend is observed whereby the difference in dilatation between the baseline and stented vessel is most pronounced at the lowest Womersley number $Wo_D = 13$ and diminishes with increasing Wo_D . This trend is consistently present for both Reynolds numbers investigated.

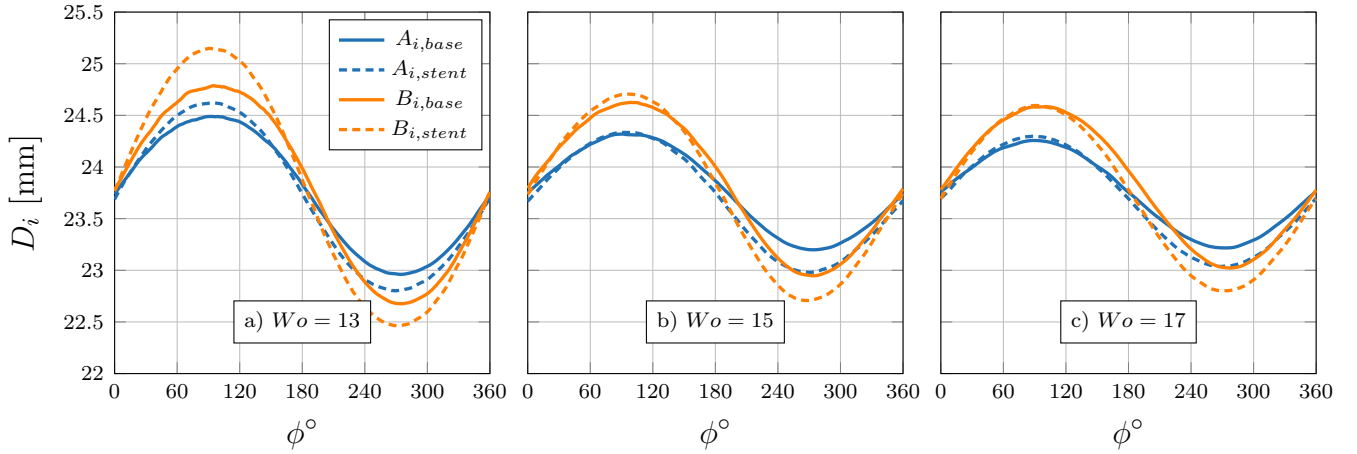


Figure 6. Inner diameter D_i of the baseline and the stented vessel as a function of the phase angle for the six cases.

Figure 7 shows the difference in inner diameter ΔD_i between the baseline and stented vessel as a function of phase angle, highlighting the influence of the stent on the vessel's structural response. ΔD_i exhibits a sinusoidal variation over the pulsatile cycle, with the zero crossing occurring well before $\phi = 180^\circ$ for all investigated cases, which is primarily attributable to a negative time-averaged offset of ΔD_i , indicating that the stent shifts the dynamic equilibrium towards a smaller mean inner diameter relative to the baseline. The amplitude of ΔD_i is larger in both the expansion and relaxation phases, with the maximum differences occurring in the vicinity of the phase angles of maximum and minimum inner diameter, consistent with a stent-induced force variation that is proportional to the local pressure difference. For a linear spring-damper system, one would expect the stent to introduce a constant additional stiffness, resulting in symmetric oscillations around a new equilibrium diameter. However, since the stent exerts an outward radial force only for $D_i < d_{stent} = 24$ mm, it loses contact with the vessel wall once D_i exceeds d_{stent} , introducing a geometric nonlinearity into the system. As the measured inner diameter exceeds d_{stent} during the expansion phase, the stent intermittently loses contact with the vessel wall, which may contribute to the observed asymmetry between the expansion and relaxation phases, most prominently at the lowest Womersley number $Wo_D = 13$ where the dilatation amplitude is largest.

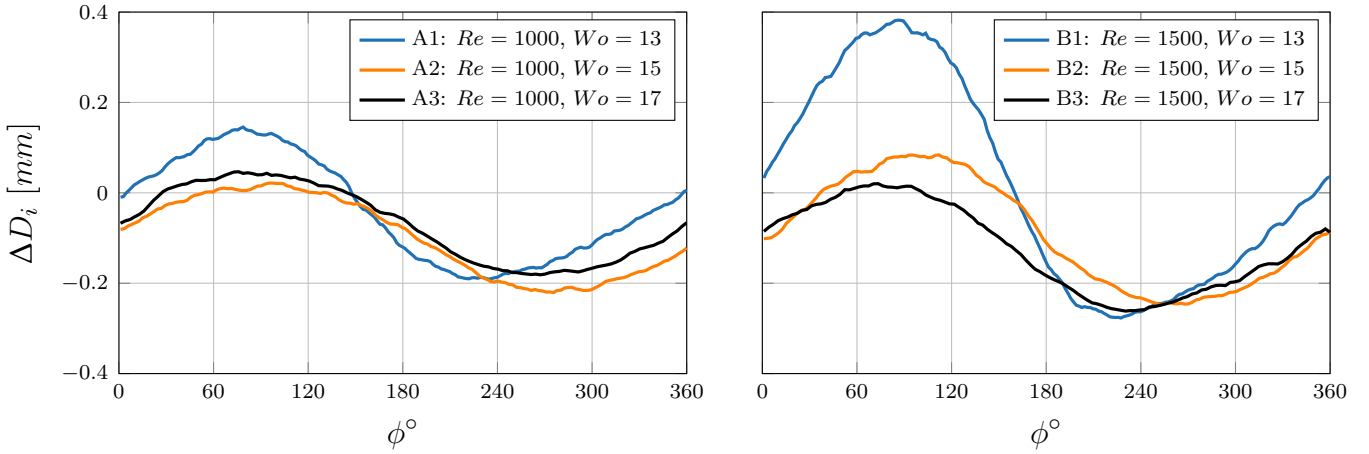


Figure 7. Difference of the inner diameter ΔD_i of the baseline and the stented vessel as a function of the phase angle for the six cases.

3.2. Flow field

Figures 8-10 show the ensemble-averaged velocity profiles $u(r/D_i)$ at $x = 0.5 \cdot D_0 = 11$ mm for the baseline and stented configurations at three characteristic phase angles corresponding to maximum ($\phi(D_{\max})$), nominal ($\phi(D_{\text{avg}})$), and minimum ($\phi(D_{\min})$) inner diameter and the difference Δu between the baseline and the stented configuration. As the observed differences between the baseline and stented configurations are qualitatively consistent across all investigated cases and more pronounced at the higher Reynolds number, the velocity profiles for $Re_D = 1000$ are presented in Appendix A. To assess the statistical reliability of the observed differences, the pulse-to-pulse standard deviation of the velocity was evaluated for all cases, yielding typical values of $0.003\text{--}0.009$ m/s ($\approx 1\text{--}3\%$ of u_∞). The stent-induced velocity differences at $Re_D = 1500$ are of comparable or greater magnitude than this variability, suggesting that the observed differences are statistically meaningful, particularly at the peak and deceleration phases. Overall, the differences between the baseline and stented velocity profiles remain small across all investigated cases, reflecting the moderate influence of the stent on the bulk flow field under the present conditions. Notably, the velocity profiles retain a well-developed parabolic shape of comparable amplitude at all three characteristic phase angles, with no significant velocity reduction observed between them. This behavior can be attributed to the phase shift between the flow excitation and the structural response of the vessel wall, such that the selected phase angles do not correspond to fundamentally different flow states but rather to different stages of the coupled fluid–structure response. Nevertheless, a consistent trend is identified that the stented case exhibits slightly elevated velocities relative to the baseline at $\phi(D_{\max})$, converging near $\phi(D_{\text{avg}})$, and returning to higher values at $\phi(D_{\min})$. This behavior is most pronounced for case B1 ($Re_D = 1500$, $Wo_D = 13$), which also exhibits the largest vessel dilatation among the investigated cases, suggesting a coupling between the structural response amplitude and the resulting flow field modification.

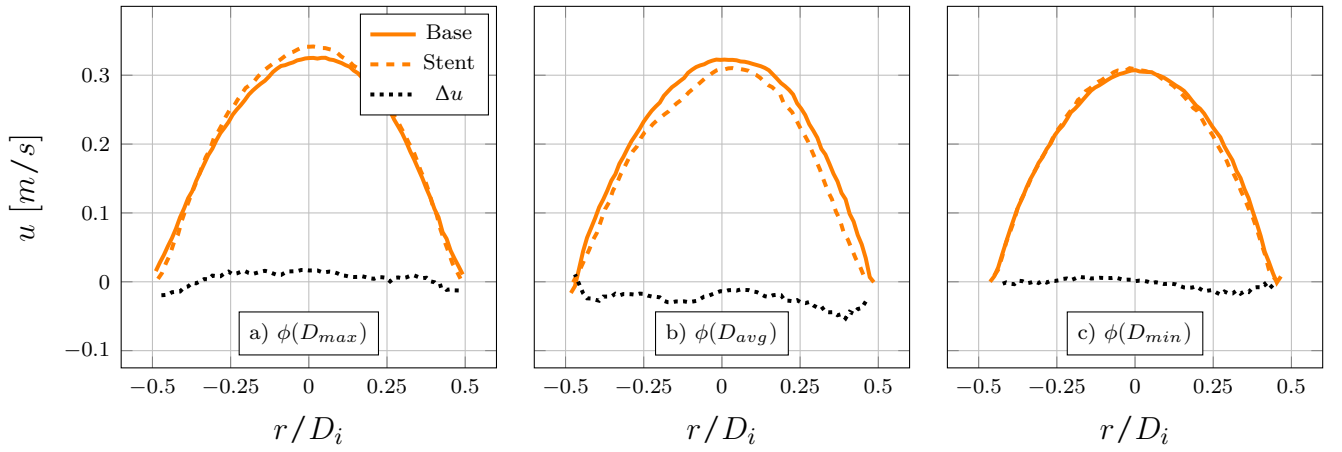


Figure 8. Ensemble-averaged velocity profiles $u(r/D_i)$ for the baseline and stented vessel at $Re_D = 1500$, $Wo_D = 13$, shown at phase angles of maximum ($\phi(D_{\max})$), mean ($\phi(D_{\text{avg}})$), and minimum ($\phi(D_{\min})$) inner diameter. The difference Δu between both configurations is indicated in a blue line.

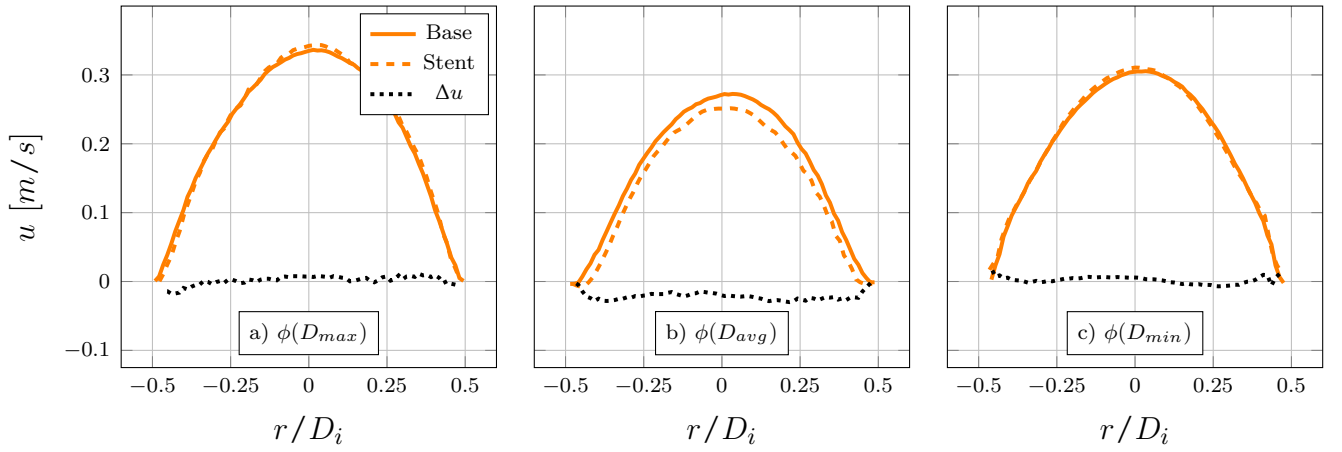


Figure 9. Ensemble-averaged velocity profiles $u(r/D_i)$ for the baseline and stented vessel at $Re_D = 1500$, $Wo_D = 15$, shown at phase angles of maximum ($\phi(D_{\max})$), mean ($\phi(D_{\text{avg}})$), and minimum ($\phi(D_{\min})$) inner diameter. The difference Δu between both configurations is indicated in a blue line.

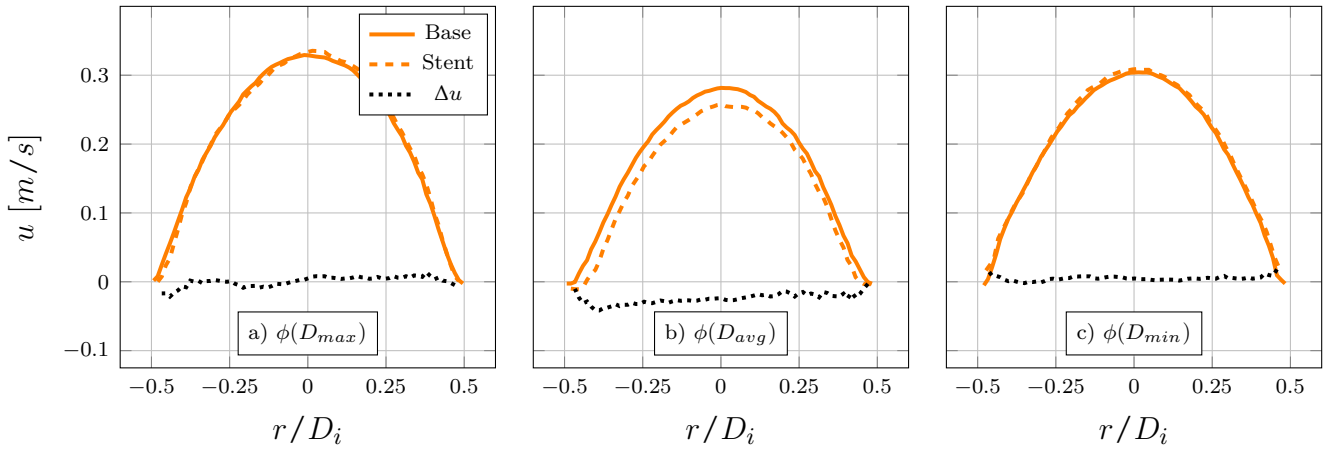


Figure 10. Ensemble-averaged velocity profiles $u(r/D_i)$ for the baseline and stented vessel at $Re_D = 1500$, $Wo_D = 17$, shown at phase angles of maximum ($\phi(D_{\max})$), mean ($\phi(D_{\text{avg}})$), and minimum ($\phi(D_{\min})$) inner diameter. The difference Δu between both configurations is indicated in a black dotted line.

4. Conclusion and Outlook

This study presents an experimental investigation of the fluid–structure interaction in a stented elastic straight vessel under pulsatile flow conditions, combining two-dimensional two-component particle-image velocimetry with refractive-index matching and an automated wall detection routine. Six cases were investigated, covering Reynolds numbers of $Re_D = 1000$ and $Re_D = 1500$ and Womersley numbers of $Wo_D = 13, 15$, and 17 , with excitation frequencies spanning the eigenfrequency of the coupled fluid–structure system.

The structural response of the vessel demonstrates that stent implantation increases the amplitude of pulsatile wall dilatation during both the expansion and relaxation phases of the pulsatile cycle, while leaving the mean inner diameter largely unaffected. The stent induces three coupled effects on the vessel wall dynamics, i.e., a temporal asymmetry between the expansion and relaxation phases, larger amplitudes of ΔD_i in both phases, and a negative time-averaged value of ΔD_i , indicating that the stented vessel exhibits a smaller mean inner diameter than the baseline under pulsatile flow conditions, despite being slightly larger at rest. Both the amplitude increase and the temporal asymmetry diminish with increasing Womersley number and are consistently observed for both Reynolds numbers investigated, reflecting the frequency-dependent nature of the stent–vessel interaction.

The ensemble-averaged velocity profiles reveal that the stent exerts a moderate influence on the bulk flow field under the present conditions. A consistent trend is identified across all cases, with the stented configuration exhibiting slightly elevated velocities at the phase angles of maximum and minimum inner diameter, converging towards the baseline near the mean diameter. The coupling between the structural response amplitude and the resulting flow field modification is most

pronounced for case B1 ($Re_D = 1500$, $Wo_D = 13$), which exhibits the largest vessel dilatation among all investigated cases.

The findings provide quantitative experimental evidence of the interaction between vessel compliance, stent mechanics, and pulsatile haemodynamics in a controlled in-vitro setting. Future work will focus on extending the investigation to curved vessel geometries and additional stent designs, as well as spatially resolved wall shear stress measurements to further characterize the near-wall flow environment in stented compliant vessels.

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Nomenclature

D_0	Inner diameter of the steel tube / vessel inlet [mm]
D_i	Local inner diameter of the vessel [mm]
ΔD_i	Inner diameter difference, baseline vs. stented vessel [mm]
d_{stent}	Stent diameter [mm]
δ	Damping coefficient [-]
δ_{cam}	Camera distance to measurement plane [mm]
η_{WG}	Dynamic viscosity of the water/glycerine mixture [Pa s]
f	Piston excitation frequency [Hz]
f_0	Natural frequency of the system [Hz]
f_d	Damped oscillation frequency [Hz]
f_n	Normalized frequency [-]
g	Gravitational acceleration [m s^{-2}]
h_1, h_2	Reservoir heights [m]
Λ	Logarithmic decrement [-]
λ	Laser wavelength [nm]
L	Vessel length [mm]
l_{stent}	Stent length [mm]
n	Refractive index [-]
N_{base}	Number of cycles per ensemble, baseline [-]
N_{stent}	Number of cycles per ensemble, stented [-]

ω	Angular frequency of the driving signal [rad s^{-1}]
ϕ	Phase angle [$^{\circ}$]
p_n	Normalized static pressure [-]
p_{trans}	Transmural pressure [Pa]
r	Radial coordinate [mm]
R	Adapter curvature radius [mm]
Re_D	Reynolds number [-]
ρ_{WG}	Density of the water/glycerine mixture [kg m^{-3}]
s_p	Piston displacement [mm]
T	Fluid temperature [$^{\circ}\text{C}$]
T_d	Period of successive pressure decay peaks [s]
t	Time [s]
u	Axial velocity [m s^{-1}]
u_B	Bulk velocity at maximum volume flux [m s^{-1}]
Δu	Velocity difference, baseline vs. stented profile [m s^{-1}]
v_n	Normalized piston velocity [-]
$\dot{V}_{osc}$	Volume flux of the oscillating piston [$\text{m}^3 \text{s}^{-1}$]
$\dot{V}_{linear}$	Volume flux of the linear piston [$\text{m}^3 \text{s}^{-1}$]
$\dot{V}_{sum}$	Total volume flux [$\text{m}^3 \text{s}^{-1}$]
Wo_D	Womersley number [-]
x	Axial coordinate [mm]
y_i, y_{i+1}	Successive amplitudes of the pressure decay curve [-]

References

- Alkhalili, K., Hannallah, J., Cobb, M., Chalouhi, N., Philips, J. L., Echeverria, A. B., ... Gonzalez, L. F. (2018). The effect of stents in cerebral aneurysms: A review. *Asian journal of neurosurgery*, 13(2), 201–211. doi: 10.4103/1793-5482.175639
- Burgmann, S., Große, S., Schröder, W., Roggenkamp, J., Jansen, S., Gräf, F., & Büsen, M. (2009). A refractive index-matched facility for fluid–structure interaction studies of pulsatile and oscillating flow in elastic vessels of adjustable compliance. *Experiments in Fluids*, 47(4-5). doi: 10.1007/s00348-009-0681-y
- Charonko, J., Karri, S., Schmieg, J., Prabhu, S., & Vlachos, P. (2009). In vitro, time-resolved piv comparison of the effect of stent design on wall shear stress. *Annals of biomedical engineering*, 37(7), 1310–1321. doi: 10.1007/s10439-009-9697-y

- Dennis, K. D., Rossman, T. L., Kallmes, D. F., & Dragomir-Daescu, D. (2015). Intra-aneurysmal flow rates are reduced by two flow diverters: an experiment using tomographic particle image velocimetry in an aneurysm model. *Journal of neurointerventional surgery*, 7(12), 937–942. doi: 10.1136/neurintsurg-2014-011323
- Dörner, P., Schröder, W., & Klaas, M. (2021). Experimental quantification of oscillating flow in finite-length straight elastic vessels for newtonian and non-newtonian fluids. *European Journal of Mechanics - B/Fluids*, 87, 180–195. doi: 10.1016/j.euromechflu.2021.02.001
- Learoyd, B. M., & Taylor, M. G. (1966). Alterations with age in the viscoelastic properties of human arterial walls. *Circ. Res.*, 18(3), 278-292. doi: 10.1161/01.RES.18.3.278
- Pielhop, K., Klaas, M., & Schröder, W. (2015). Experimental analysis of the fluid–structure interaction in finite-length straight elastic vessels. *European Journal of Mechanics - B/Fluids*, 50, 71–88. doi: 10.1016/j.euromechflu.2014.11.001
- Williamson, P. N., Docherty, P. D., Yazdi, S. G., Khanafer, A., Kabaliuk, N., Jermy, M. C., & Geoghegan, P. H. (2022). Review of the development of hemodynamic modeling techniques to capture flow behavior in arteries affected by aneurysm, atherosclerosis, and stenting. *Journal of biomechanical engineering*, 144(4). doi: 10.1115/1.4053082
- Xu, J., Murphy, S. L., Kochanek, K. D., Bastian, B., & Arias, E. (2021). Deaths: Final data for 2019. *National Center for Health Statistics*(Vol. 70, No. 8).
- Yazdi, S. G., Docherty, P. D., Khanafer, A., Jermy, M. C., Kabaliuk, N., Geoghegan, P. H., & Williamson, P. (2021). In-vitro particle image velocimetry assessment of the endovascular haemodynamic features distal of stent-grafts that are associated with development of limb occlusion. *Journal of the Royal Society of New Zealand*, 51(2), 361–374. doi: 10.1080/03036758.2020.1826988

A. Additional velocity profiles for $Re_D = 1000$

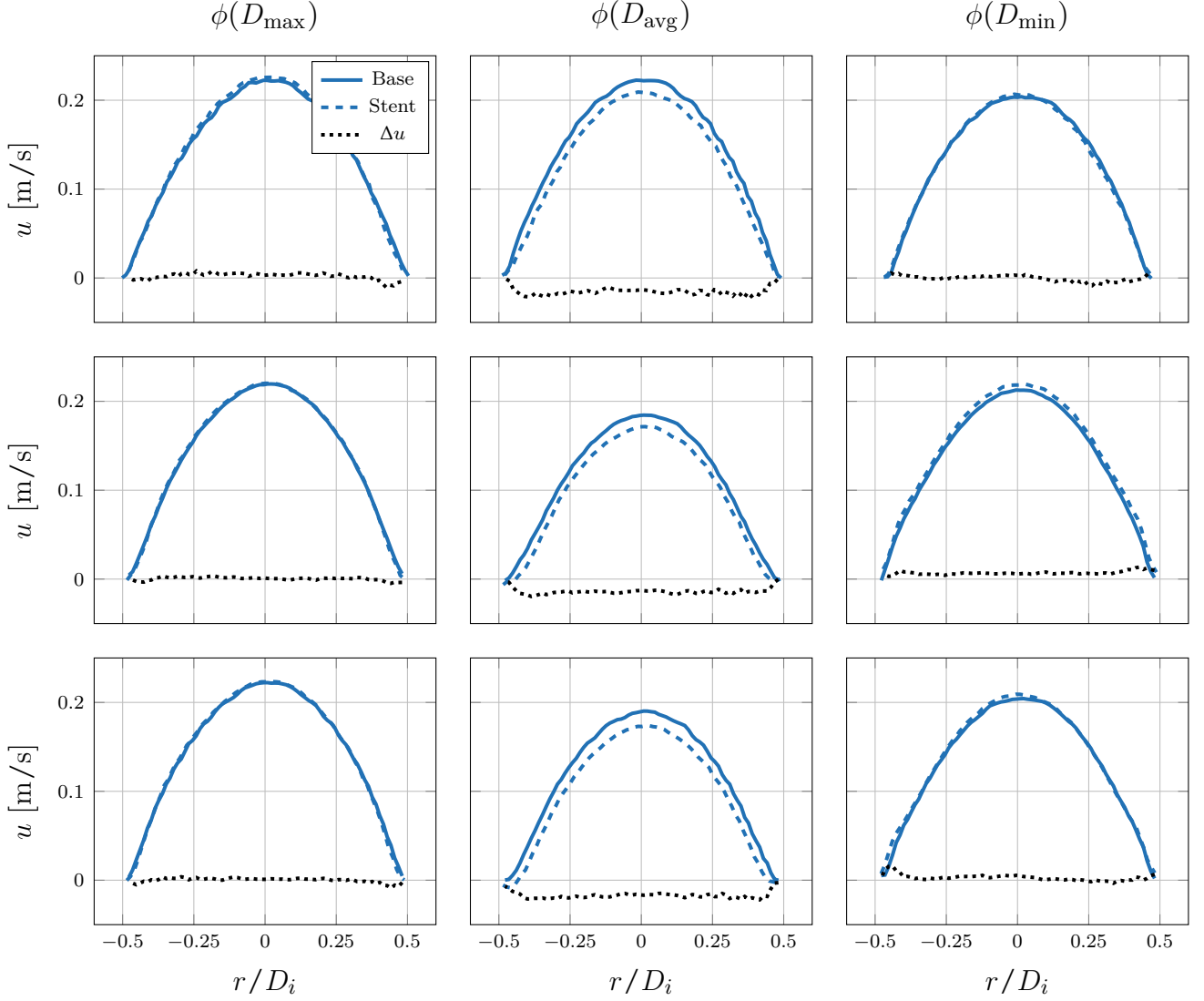


Figure 11. Ensemble-averaged velocity profiles $u(r/D_i)$ for the baseline and stented vessel at $Re_D = 1000$, shown at phase angles of maximum ($\phi(D_{\max})$), mean ($\phi(D_{\text{avg}})$), and minimum ($\phi(D_{\min})$) inner diameter for $Wo_D = 13$ (top), $Wo_D = 15$ (middle), and $Wo_D = 17$ (bottom). The difference Δu between both configurations is indicated by the black dotted line.