

Experimental corrosion and computational hemodynamic assessment of a 3D-printed biodegradable stent design

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ABSTRACT

Coronary artery disease (CAD) is a leading cardiovascular disorder caused by stenosis or occlusion of the coronary arteries. Stenting and bypass surgery are common treatment methods; however, conventional metal stents are prone to re-occlusion over time due to their permanent residence in the body. As a result, researchers have shown growing interest in biodegradable polymer stents. In this study, a novel stent and insert were 3D-printed with FDM and SLA using biocompatible PLA filament/resin. To assess corrosion resistance, the specimens were immersed in a simulated body fluid (SBF), and three corrosion tests were performed. Then, using field emission scanning electron (FESEM) images and 3D scanning, dimensional changes in the samples were extracted. The observations showed that in some places, the thickness of the samples increased by 0.1 mm compared to the original sample. Therefore, three samples with thickness increases of 0.1, 0.2, and 0.3 mm were designed and flow simulations were subsequently carried out. In simulations, the Carreau viscosity model and pulsatile inlet flow were used. It was found that with increasing stent thickness, the insert becomes progressively less effective at increasing wall shear stress, leading to a higher probability of restenosis.

1. Introduction

The heart functions as the central organ responsible for systemic blood circulation. Cardiovascular diseases, particularly coronary artery disease (CAD), remain a major contributor to mortality across the globe [1]. Researchers predict that by 2030, CAD will be responsible for 14.2 percent of all death people in the world. According to data provided by the American Heart Association, 34.4 percent of all deaths in the United States in 2006 were due to heart problems [2]. Sometimes, coronary arteries (CA) undergo stenosis (narrowing) as a result of plaque accumulation or other phenomena [1]. There are several methods for the treatment of CAD, such as drug therapy, stenting, and coronary artery bypass grafting (CABG) [2]. Researchers are developing bioabsorbable stents to address complications associated with conventional permanent stents, including both metal and drug-eluting types [3]. The Igaki-Tamai stent was the first bioabsorbable cardiac stent to be implanted in humans, marking a significant advancement in the field of cardiovascular intervention. This stent was made of poly(L-lactide). At standard physiological temperatures, the stent took 20 min to open

spontaneously. Furthermore, the stent completely degraded within three years [4]. Given the high importance of treating vascular occlusive diseases, the development of such advanced stents requires extensive research in several key areas. For example, many researchers have worked on fabricating various polymer stents using various methods such as 3D printing, while others have investigated the effect of stent properties on blood flow through a variety of simulations. A number of studies conducted in these areas were reviewed below.

Kabinejadian et al. [5] investigated velocity and wall shear stress in a new sequential anastomotic graft design under pulsatile flow conditions in vitro. The researchers used particle image velocimetry (PIV) to measure velocity and wall shear stress. The new design, which combined end-to-side and side-to-side anastomoses, outperformed end-to-end anastomoses. In another study, Kabinejadian et al. [6] examined a new spiral/helical graft design to improve the distal graft anastomoses by numerical simulation. They found that the presence of the spiral created a rotating flow. As a result of this flow pattern, wall shear stress (WSS) rose, and the region of stagnation points was diminished. Therefore, it may improve the lifespan of the graft. Soler et al. [7]

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investigated the optimization of the new helical bypass graft using computational fluid dynamics (CFD). This was a parametric study aimed at improving the performance of new artificial graft. Their research on four design parameters showed that the trailing edge is one of the most important parameters for creating rotating flow. Keshmiri et al. [8] worked on the numerical simulation of a new design of bypass graft. They conducted 270-degree trailing edge had better results than 90-degree because it could create strong rotational flow in the host vessel. Zhang et al. [9] evaluated the effect of the stents on intramural hematoma using fluid–solid interaction. The deformation, wall shear stress, and von mises stress, on the stent and vessel were investigated. Findings indicated that an increase in vessel diameter or a decrease in its length resulted in higher Von Mises stress.

Cheng and Zhang [10] employed the finite element method (FEM) for simulating re-occlusion of stents. The results showed that after placing a stent in the region of the vessel, the Von Mises stress in plaque increased significantly, especially in the region of stent and vessel contact. Martin et al. [11] analyzed balloon-expandable coronary stents with CFD. Some hemodynamic parameter results, such as Oscillatory shear index (OSI), and Time-averaged wall shear stress (TAWSS) were extracted. The researchers concluded that stent and vessel deformation significantly affected hemodynamic parameters.

Yeazel and Becker [12] investigated the use of shape-memory polymer stents with bioabsorbability using a 3D printer. The researchers found that the commonly used stents are permanent and are not biodegradable. Many industrial polymer stents use Polylactic Acid (PLA) and Polycaprolactone (PCL), because their behavior was proved in literature. In conclusion, the results indicated the researchers were able to create a patient-specific stent for each patient using a 3D printer. Cabrera et al. [13] combined computational tools with 3D printing to design polymer stents exhibiting mechanical performance similar to nitinol stents tested in animal models. Stents fabricated from Thermoplastic Copolyester Elastomer (TPC) underwent crush and crimping tests, which were validated against computational predictions. The study demonstrated that Fused Deposition Modeling (FDM) can serve as a practical method for producing prototype stents. Guerra and Ciurana [3] examined how 3D-printing parameters influence the absorption time of stents under laboratory conditions. They used PCL as a stent material. The results showed that the speed of destruction was affected by increasing the temperature and velocity of the flow rate during the print. Chang et al. [14] employed 3D printing combined with laser micro-machining to fabricate biodegradable stents with high precision. They used PLA for production prototype samples. After fabrication, the stent was subjected to compression and expansion to evaluate its radial force performance following deployment. A radial force of 0.079 N/mm was obtained for the fabricated stent, confirming that it satisfied the necessary specifications for a peripheral artery stent. Langi et al. [15] compared the microstructures and mechanical properties of stents produced with using additive manufacturing and commercial metal stent methods. The microstructures were examined through the field emission scanning electron microscopy (FESEM), and electron backscatter diffraction. Ultimately, the researchers found that stiffness and modulus of stents made with additives were higher than commercial stents. The stress–strain curve and yield strength were similar both type of stents. Amnieh et al. [16] investigated the biodegradation function of PLA stents under the effect of geometrical parameters, the risk of breakage, and separation of the parts. The researchers used a simulation to predict PLA degradation. The results showed that geometrical parameters such as the number of struts, curve radius, and design of stent cells were effective on behavior degradation. Okereke et al. [17] developed biodegradable 3D printing coronary stents. They designed 10 new stents and manufactured them with the FDM method. The results showed that stents with a Race-track cell shape were suitable for angioplasty.

Recent investigations on biodegradable stents have emphasized the importance of material design, structural customization, surface modification, and degradation control in improving implant performance.

Studies on functionalized biodegradable stents, polymeric customizable stent designs, surface-modified magnesium-based stents, and novel biodegradable metallic alloys indicate that corrosion behavior, mechanical stability, geometry, and biocompatibility must be carefully considered in the design of temporary stent systems [18–21].

Hol et al. [22] investigated stent corrosion and metal release in vessels exposed to mechanical pressure and electrochemical conditions. Restenosis frequently occurs and is typically managed by placing a new stent inside the existing one or by overlapping stents to cover a larger vessel area. Stents were exposed to pulsatile radial pressure of up to 8% strain at 1 Hz and to electrochemical stress. The results indicate that mechanical strain combined with a weak electric potential causes corrosion and significant stent fractures, particularly in overlapping stents. Hassanpor et al. [23] investigated the in vitro corrosion behavior of PLA–Mg composites for potential use in cardiovascular stents. Their study emphasized the role of magnesium volume fraction as a key factor governing the absorption rate in the design of patient-specific cardiovascular stents fabricated from PLA–Mg composites. The prototype stent and composite samples were placed in simulated body fluid (SBF) for corrosion testing. Finally, they concluded that, for stent manufacturing, the magnesium content should not exceed 3 percent to maintain suitable corrosion properties. Shen et al. [24] developed and analyzed a numerical corrosion-fatigue model for biodegradable magnesium alloy stents. Results demonstrated that the model reliably predicted degradation rate, radial stiffness loss, and critical stent locations. It can also be extended to alloys like Fe and Zn to optimize degradation and improve fatigue resistance. Wang et al. [25] studied the corrosion of MgZnCa and AZ31 absorbable stents under vascular-like flow. They found that flow velocity, shear stress, and hydrodynamics strongly influence corrosion behavior. Hosseini et al. [26] investigated the corrosion of a fatigue standard sample in a 1X SBF solution. The material used was PLA. After 330 days of testing, they found that they had gained approximately 15% of their weight, which was attributed to water absorption. Baran et al. [27] investigated the effect of surface roughness on the corrosion of the samples. The material used was PLA. They ultimately found that the rougher the sample surface, the greater the increase in water absorption, leading to a higher degree of polymer degradation.

Previous studies show that stenting and bypass surgery are important treatments for coronary artery disease[2]; therefore, many researchers worked on various designs of stents using different materials and product methods. These stents are permanent in the body and it found that it was not good for the body [3,4], so the researches directed to biodegradable stent polymers that can be manufactured with 3D printing [16–22] and some simulation researches [5–15] and tests, such as corrosion tests [23–30], were performed.

The innovation of this study lies in investigating and simulating the effect of corrosion on a newly designed polymer stent with an internal insert, as well as its impact on blood flow in a bypass vessel. Stent and insert were fabricated with 3D printer. Unlike previous studies that focused only on fluid flow through intact stents or on stent corrosion separately, the present work is the first to combine both aspects. By comparing the results before and after corrosion, the study highlights how corrosion alters blood flow behavior. In the following sections, the research method and results of this study are presented.

2. Method

In a previous work, a novel geometry for the stent and insert in bypass configuration was first designed and evaluated by computational fluid dynamics (CFD) simulation by the authors [29]. CFD was used to analyze fluid flow behavior virtually. Physical samples were fabricated using FDM with PLA filament and Stereolithography (SLA) with PLA resin. The samples underwent corrosion testing in SBF. After corrosion testing, geometric changes were extracted by 3D scanning and FESEM imaging and the corroded geometry was reconstructed. CFD simulations were then repeated using the corroded geometry.

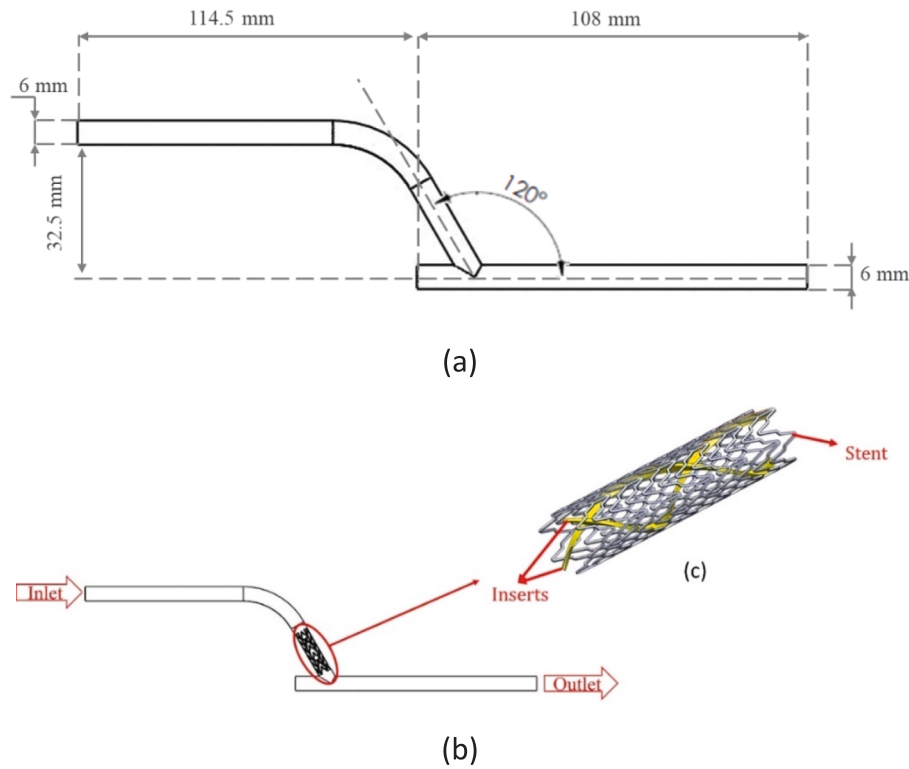


Fig. 1. (a) The dimension of the bypass vessel (b) the placement of the stent within the bypass vessel and (c) the new design of stent and insert.

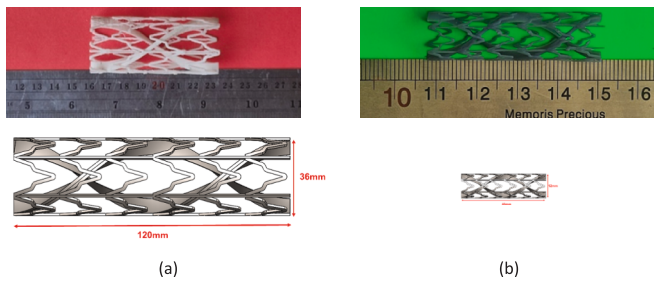


Fig. 2. Printed samples included scales, (a) FDM method with PLA filament and (b) SLA method with PLA resin.

Table 1

The nominal concentrations of different body simulator solutions [26,31].

Ion	Blood plasma (mM)	SBF 1X (mM)	SBF 10X (mM)
Na ⁺	142.0	142.0	1420
K ⁺	5.0	5.0	50
Mg ²⁺	1.5	1.5	15
Ca ²⁺	2.5	2.5	25
Cl ⁻	103.0	103.0	1030
HCO ³⁻	27.0	27.0	270
HPO ₄ ²⁻	1.0	1.0	10
SO ₄ ²⁻	0.5	0.5	5

2.1. Geometry

Fig. 1(a) presents the stent location within the bypass vessel, Fig. 1(b) illustrates the vessel configuration, and Fig. 1(c) displays the newly designed stent and insert. The stent has a diameter of 6 mm and strut thickness of 0.2 mm. The insert has a triangular cross-section with two sides measuring 1.13 mm and the third side measuring 1 mm, and it wraps one full turn along the entire length of the stent. A novel bypass

geometry for the stent and insert was adopted from our previous work [29]. This configuration is particularly effective for generating swirling flow, which increases wall shear stress and reduces sediment deposition at the bypass junction, where shear is otherwise low. As shown in Fig. 1(b), the stent is positioned within the bypass vessel at the proper junction with the main artery, ensuring realistic placement and alignment for the simulations. The stent is placed inside the bypass vessel with the same diameter of the vessel without oversizing. The bypass geometry and stent insert were generated using SolidWorks 2019, following the parametric specifications described Fig. 1(a).

2.2. 3D printing and corrosion test

Stents and inserts were 3D-printed using two methods: FDM with 1.75-mm YOUSU PLA filament and SLA with eSUN PLA Pro Resin. In general, PLA is a biocompatible material that is used in stent manufacturing [16]. The stent and insert were designed in SolidWorks 2019. Then, they were printed using the FDM method; however, due to the limitations of this printing method, the samples were printed much larger than their actual size (approximately 5 times larger). Then, the SLA method was used, as it could print a smaller size than the FDM method (approximately twice the actual size). Fig. 2 presents representative FDM- and SLA-fabricated specimens along with their corresponding CAD geometry views. The CAD views provide the main dimensional information and serve as a consistent geometric reference to support visual comparison between the printed specimens.

After fabrication of the stent and inserts, three corrosion tests were performed. In the first test, the FDM-made sample was placed in SBF 1X. In the second test, another FDM-made sample was placed in SBF 10X. In the last test, the SLA-made sample was placed in SBF 10X. In all three tests, the samples were initially placed in the SBF for 28 days. During the test, they were removed from the solution and weighed to record the changes in the sample weights [30]. For a more detailed study, the first and third tests were examined over a longer period (the first corrosion test lasted 328 days, and the third corrosion test lasted 105 days).

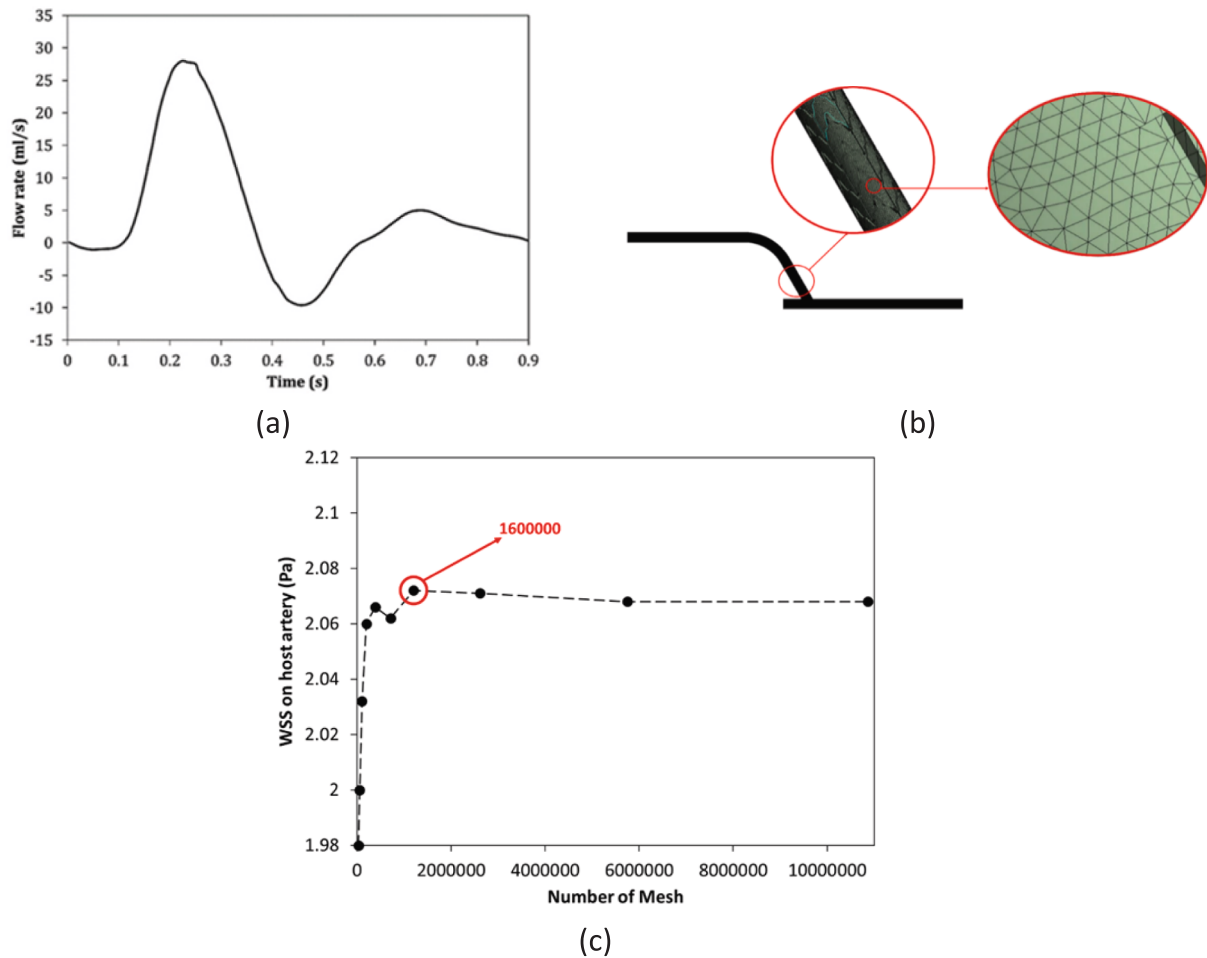


Fig. 3. (a) Inlet velocity waveform [6], (b) The mesh around the stent and insert, and (c) The mesh independency diagram for WSS on host artery.

Table 1 gives the nominal concentrations of different body simulator solutions. After the corrosion test of the samples was completed, FESEM was used to examine the microstructure. Finally, the geometry of the stent and insert was extracted with the help of a 3D scanning.

2.3. Simulation

CFD simulation provides velocity, pressure, and stress data within the fluid domain, helping identify regions at risk of plaque formation and development [32]. The assumptions used to perform the simulation are as follows:

- The vessel is rigid; so, no diameter changes,
- The flow is incompressible,
- The non-Newtonian fluid is considered,
- The flow is unsteady in a cardiac cycle,
- A fully developed flow is assumed at the vessel inlet.

Eqs. (1)–(4) are the mass conservation and Navier-Stokes equations, governing the laminar flow in Cartesian coordinates [29]. The Reynolds number is less than 1770 in all cases.

$$\frac{\partial \rho}{\partial t} + \nabla \cdot (\rho V) = 0 \quad (1)$$

$$\rho \left(\frac{\partial u}{\partial t} + u \frac{\partial u}{\partial x} + v \frac{\partial u}{\partial y} + w \frac{\partial u}{\partial z} \right) = -\frac{\partial p}{\partial x} + \mu \left(\frac{\partial^2 \tau_{xx}}{\partial x^2} + \frac{\partial^2 \tau_{yx}}{\partial y^2} + \frac{\partial^2 \tau_{zx}}{\partial z^2} \right) + \rho g_x \quad (2)$$

$$\rho \left(\frac{\partial v}{\partial t} + u \frac{\partial v}{\partial x} + v \frac{\partial v}{\partial y} + w \frac{\partial v}{\partial z} \right) = -\frac{\partial p}{\partial y} + \mu \left(\frac{\partial^2 \tau_{xy}}{\partial x^2} + \frac{\partial^2 \tau_{yy}}{\partial y^2} + \frac{\partial^2 \tau_{zy}}{\partial z^2} \right) + \rho g_y \quad (3)$$

$$\rho \left(\frac{\partial w}{\partial t} + u \frac{\partial w}{\partial x} + v \frac{\partial w}{\partial y} + w \frac{\partial w}{\partial z} \right) = -\frac{\partial p}{\partial z} + \mu \left(\frac{\partial^2 \tau_{xz}}{\partial x^2} + \frac{\partial^2 \tau_{yz}}{\partial y^2} + \frac{\partial^2 \tau_{zz}}{\partial z^2} \right) + \rho g_z \quad (4)$$

Cardiac motion has a minimal hemodynamic effect on the coronary arteries, so coronary artery compliance has a negligible impact on flow [32]. Simulations showed that rigid and compliant models differ only slightly in hemodynamics. Therefore, most models assume rigid, fixed coronary artery walls, particularly in cases of disease or stenting [32]. A growing body of evidence links atherosclerosis and intimal thickening to local hemodynamic measures. The geometry was designed in the SolidWorks 2019 software. The fluid used was a fluid with blood parameters, and the density was 1050 kg/m³. The fluid employed in the study was non-Newtonian, with viscosity defined by the Carreau model. Shear stress is related in Eq. (5), while Eqs. (6)–(8) define strain rate, shear rate, and Carreau viscosity, respectively [33].

$$\tau = \mu D \quad (5)$$

$$D = \frac{1}{2} [\text{grad} V + (\text{grad} V)^T] \quad (6)$$

$$\dot{\gamma} = \sqrt{2 \text{tr}(D^2)} \quad (7)$$

$$\mu = \mu_{\infty} + (\mu_0 - \mu_{\infty}) [(1 + (\dot{\gamma})^2)]^{\frac{n-1}{2}} \quad (8)$$

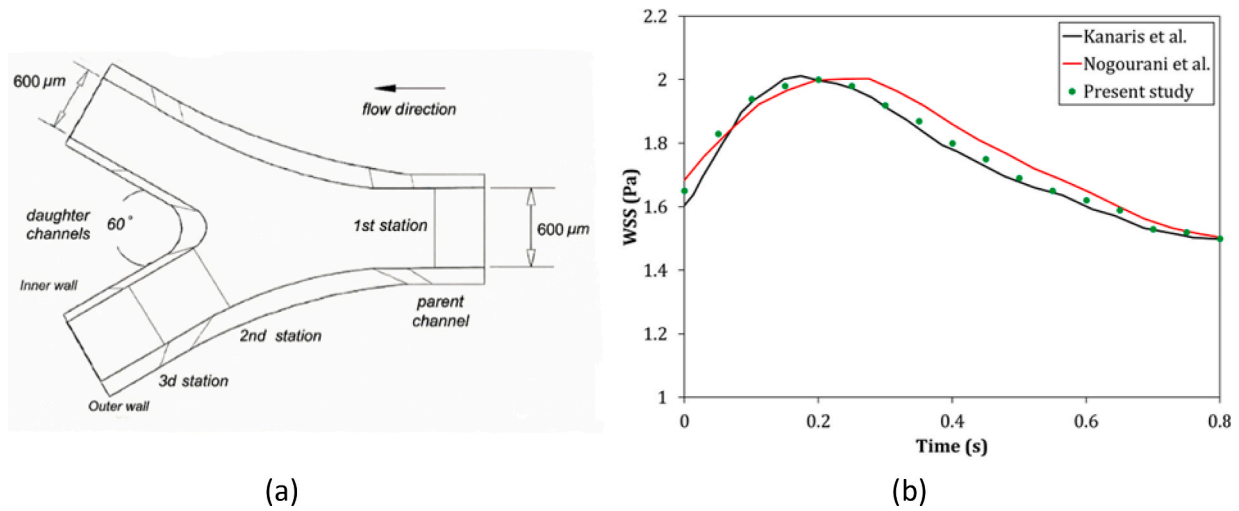


Fig. 4. (a)Vessel used in validation simulation [36], and (b) Results of bifurcated vessel wall shear stress at different times of the cardiac cycle [36,37].

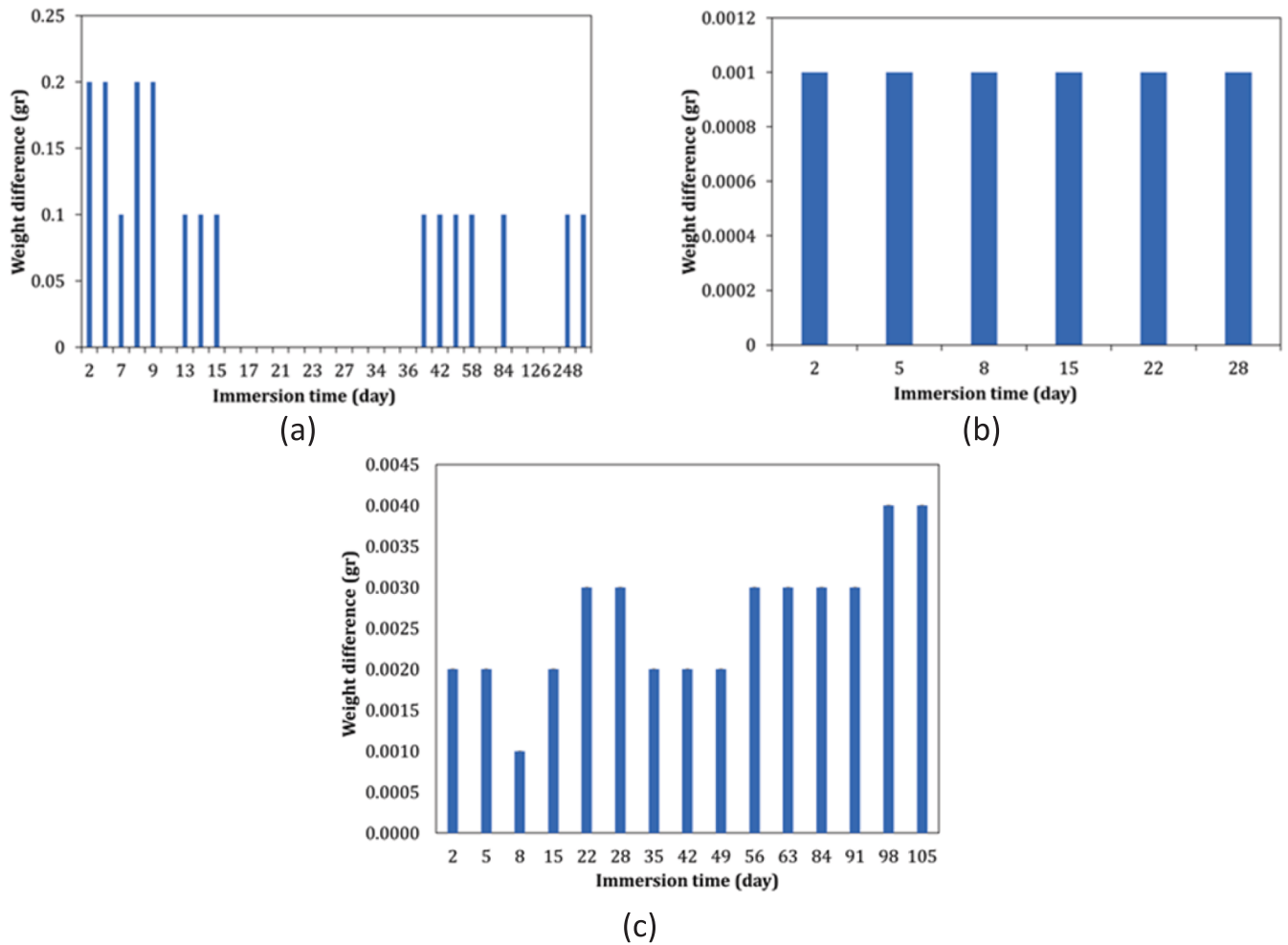


Fig. 5. The samples weighing results (a) PLA filament in SBF 1X (b) PLA filament in SBF 10X (c) PLA resin in SBF 10X.

μ_0 and μ_∞ are the zero and infinite shear rates with a value of 0.022 $\frac{\text{kg}}{\text{ms}}$ and 0.0022 $\frac{\text{kg}}{\text{ms}}$, Γ and n are time constant and power-law index, whose values are 0.11 s and 0.392, respectively [34]. Boundary conditions included a fully developed pulsatile flow at the graft inlet and a traction-free specification at the artery outlet. A rigid-wall assumption with no-

slip conditions was adopted for all surfaces. Data were extracted at the completion of the fourth cycle, with the inlet velocity waveform depicted in Fig. 3(a).

Another key parameter, the oscillatory shear index (OSI), is given in Eq. (9) [35].

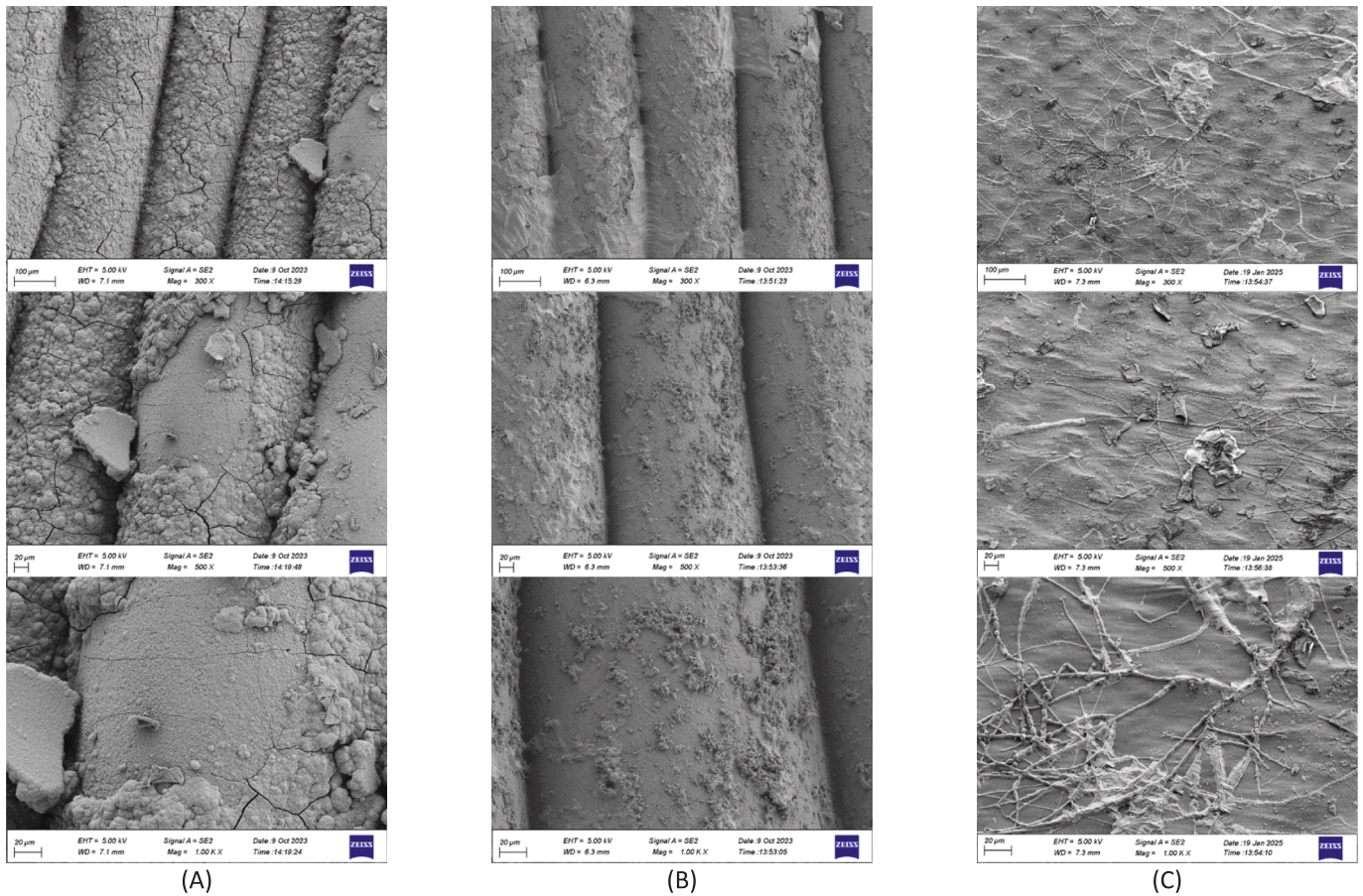


Fig. 6. The FESEM images on the insert of different sizes (A) PLA filament in 1X solution (B) PLA filament in 10X solution (C) PLA resin in 10X solution.

$$OSI = \frac{1}{2} \left(1 - \frac{\left| \int_0^T \tau_w dt \right|}{\int_0^T |\tau_w| dt} \right) \quad (9)$$

The computational framework was implemented in Ansys Fluent 2020, utilizing the SIMPLE algorithm for pressure–velocity coupling. Spatial discretization employed least-squares (gradient), second-order (pressure), and second-order upwind (momentum) schemes, while temporal discretization followed a second-order implicit formulation. Numerical stability was maintained by enforcing the Courant–Friedrichs–Lewy (CFL) condition, which dictates that fluid displacement per time step must not exceed one cell length. The corresponding definition of the CFL number is provided in Eq. (10).

$$CFL = \frac{u \Delta t}{\Delta x} \quad (10)$$

where, Δt is the time step, u is the local flow velocity, and Δx is the smallest cell size.

The mesh configuration applied in the simulations is presented in Fig. 3(b). As demonstrated in Fig. 3(c), the mesh independence study confirmed that a 1.6-million-cell mesh achieved sufficient accuracy based on WSS values, and this number of cells was employed in all subsequent analyses. The difference in wall shear stress between the 1,600,000-element mesh and the 10,800,000-element mesh was less than 0.5%. Therefore, considering the negligible change in the key hemodynamic result and the significantly higher computational cost of the finer mesh, the 1,600,000-element mesh was selected for the final simulations.

3. Validation

Fig. 4(a) shows the geometry of a bifurcated vessel (so-called Y-shaped) which was studied by Kanaris et al. [36] and Nogourani et al. [37]. The considered blood as non-Newtonian fluid using the Hemato-crit model with a density of 1050 kg/m^3 . This model relies on the volume of red blood cells in the total blood volume, which incorporates Eq. (12) for shear stress, 13 for blood viscosity at high shear rates, 14 for yield stress, and 15 for viscosity [38].

$$\sqrt{\tau} = \sqrt{\tau_y} + \sqrt{n_N \dot{\gamma}} \quad (12)$$

$$n_N = n_p [1 + 0.025 H_t + 7.35 \times 10^{-4} H_t^2] \quad (13)$$

$$\tau_y = A(H_t - H_c)^3 \quad (14)$$

$$\mu = \frac{\tau}{\dot{\gamma}} \quad (15)$$

where, τ is the WSS, τ_y is the yield stress (the minimum amount of shear stress that must be applied to the blood to start moving), n_N is the blood viscosity for high shear rates, n_p is the blood plasma viscosity, $\dot{\gamma}$ is the shear rate, H_t is the percentage of hematocrit, H_c is the critical hematocrit, in a range of 4 and 8, and A is a constant between 0.6×10^{-7} and $1.2 \times 10^{-7} \text{ Pa}$.

The present numerical method was performed for this geometry and the obtained results is compared with the two others in Fig. 4(b).

4. Results and discussion

Three corrosion tests were performed under simulated physiological

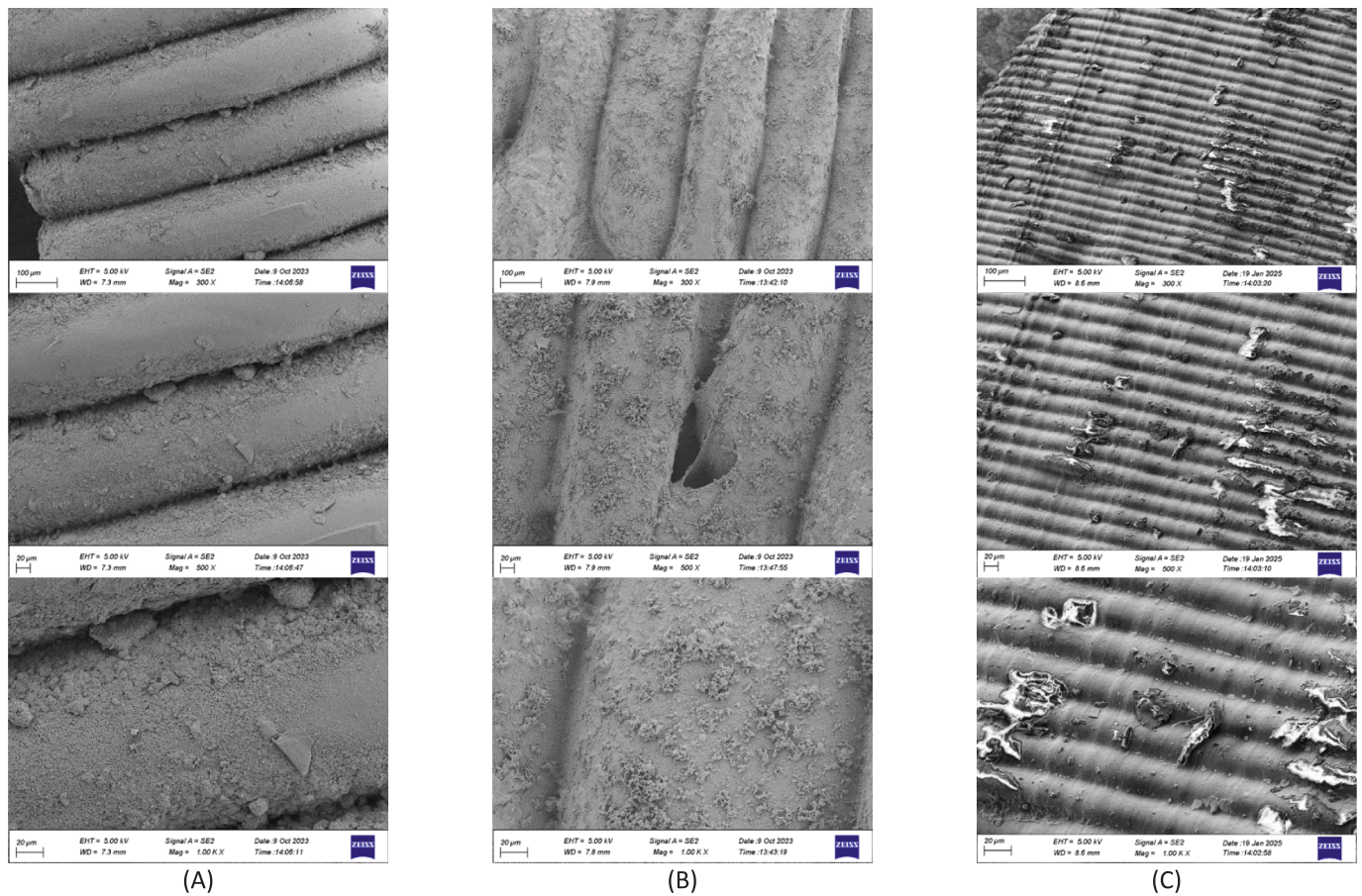


Fig. 7. The FESEM images on the strut of different sizes (A) PLA filament in 1X solution (B) PLA filament in 10X solution (C) PLA resin in 10X solution.



Fig. 8. 3D scanning image.

conditions. The term “corrosion” is used to describe the overall chemical or physical changes observed in PLA samples in simulated body fluid, including water absorption, and surface deposition. In Test A, PLA specimens produced by FDM at approximately five times the original stent and insert size were immersed in $1 \times$ SBF for 328 days. In Test B, FDM-printed PLA specimens were exposed to $10 \times$ SBF for 28 days. In Test C, SLA-fabricated PLA-resin specimens at twice the original size

were immersed in $10 \times$ SBF for 105 days. During corrosion testing, specimens were periodically removed, dried, and weighed. At the conclusion of each test, FESEM imaging was performed.

Fig. 5 compares the results of weighing the samples. It shows that the mass-change profiles of PLA stents/inserts in SBF depend strongly on the printing route and exposure time. In Test A (FDM, SBF $1 \times$), the specimen exhibited a pronounced early transient, the weight difference reached 0.20 by days 2–5, toward 0 by days 10–16, so the mean weight difference was 0.0706 g, before the 28th day (-9.5×10^{-3} g per day). Over extended immersion to 328 days weight difference remained largely within 0.0–0.1 g, with a slow late rise to 0.10 by days 248–328, indicating that after immersion in SBF solution the rate of further change was very low at the available resolution. In Test B (FDM, SBF $10 \times$), weight difference was essentially constant at 0.001 through 28 days, suggesting that any SBF concentrations from $1 \times$ to $10 \times$, affected FDM mass change over four weeks was below detection under the current setup. In contrast, Test C (SLA resin, SBF $10 \times$) showed a small but measurable monotonic increase from 0.002 (day 2) to 0.004 (day 105), mean weight difference before 28 days was 0.00217 g ($+1.9 \times 10^{-5}$ g per day), consistent with a slow, cumulative process. Collectively, the data indicate that FDM parts require an initial stabilization period (2 weeks) after which mass change slows markedly, while SLA parts in $10 \times$ SBF undergo slow, continuing change over months. Although the specimens were printed at approximately two and five times the actual stent size, the dimensions were scaled uniformly in all directions, preserving geometric proportions. As shown in Fig. 5, the mass-change profiles illustrate the effect of printing route and exposure time. These scaled results can serve as a relative degradation analysis for real-size stents, providing insight into expected degradation trends, even though absolute kinetics may differ slightly in smaller geometries.

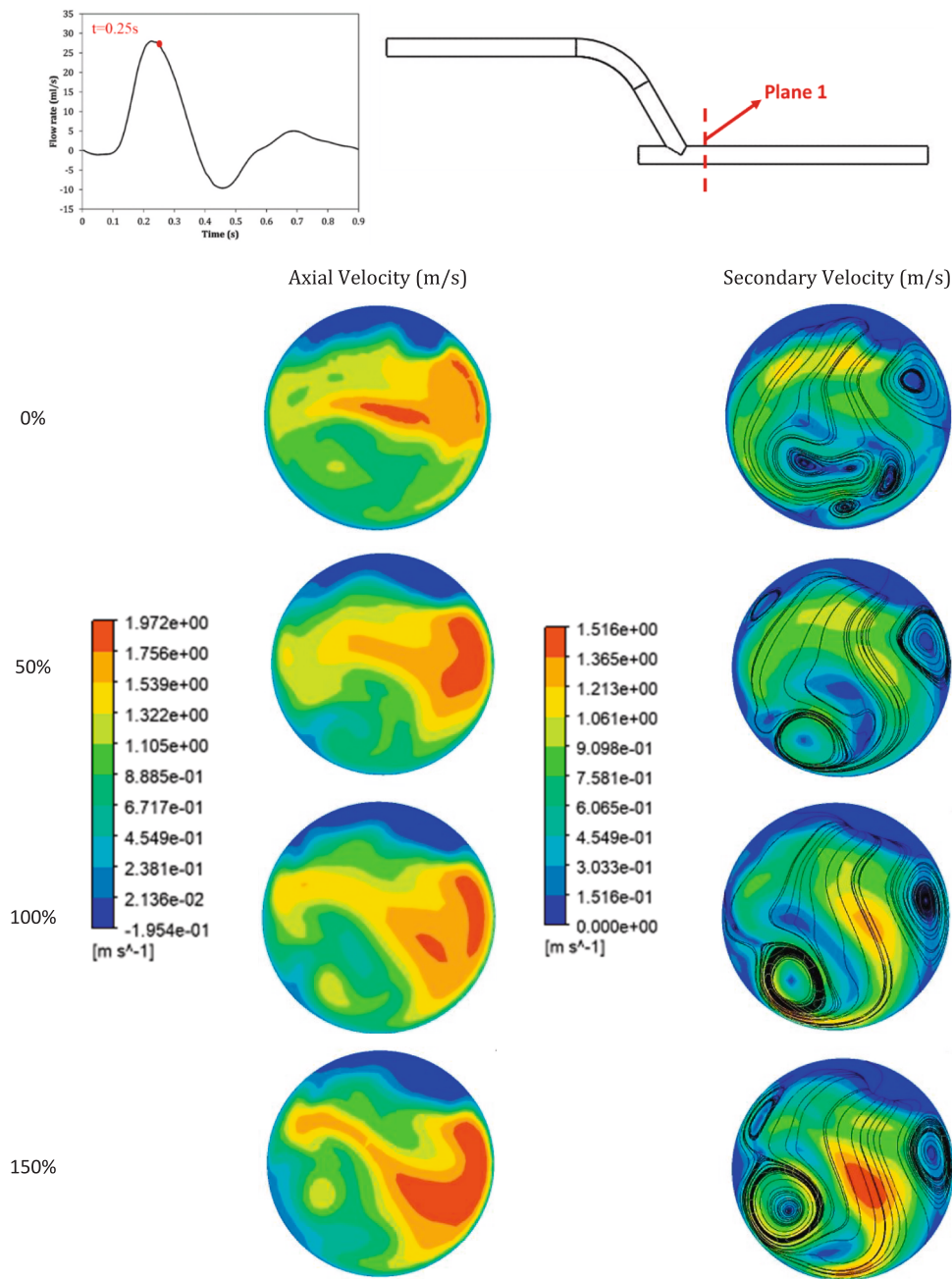


Fig. 9. Comparison of stent thickness increase at 1 mm after anastomosis at $t = 0.25$ s.

Figs. 8 and 9 show FESEM images taken from the surface of the samples, these figures show more sediment was deposited on the samples printed using the FDM method than on the samples printed using the SLA method. This is because, in the FDM method, the larger sample size, greater surface roughness, and the presence of more pores result in more sediment being absorbed onto the samples, which is consistent with the results of Baran et al. [27] and Singh et al. [38]. It was also found that in all three tests, the samples gained weight, which was attributed to sediment formation and water absorption. This result is consistent with the findings of other researchers [26,39,40].

According to Figs. 6 and 7, SLA and FDM differ mainly in how their layers are formed and bonded. FDM lays down molten filament roads that fuse as they cool, leaving bead boundaries and micro-voids between layers; this creates visible layer lines, higher surface roughness, and stronger anisotropy (the Z-direction is weakest) [41–43]. SLA, by contrast, photocures very thin layers that chemically crosslink to the

previous layer, yielding finer layer heights, smoother surfaces, and generally lower porosity with less severe anisotropy [44]. Consistent with prior reports on FDM-printed PLA in SBF and on SLA photopolymers in physiological media [45–47], the results show distinct time courses. FDM exhibits an early transient in weight change. This is attributable to its layer-level microstructure. SLA shows a slower, steadier change in SBF consistent with diffusion-controlled water uptake and/or surface deposition. Together, these results indicate that the printing route and the microstructure it imposes govern the kinetics of mass change under physiological conditions.

The Test C specimen was selected for corrosion simulations because it most closely matched the target geometry and exhibited the highest fabrication quality. Pre- and post-exposure 3D scanning was used to quantify geometric changes (Fig. 8). The combined corrosion and scanning results showed local thickness increases up to 0.10 mm (approximately 47% relative to the initial thickness) after 105 days. The

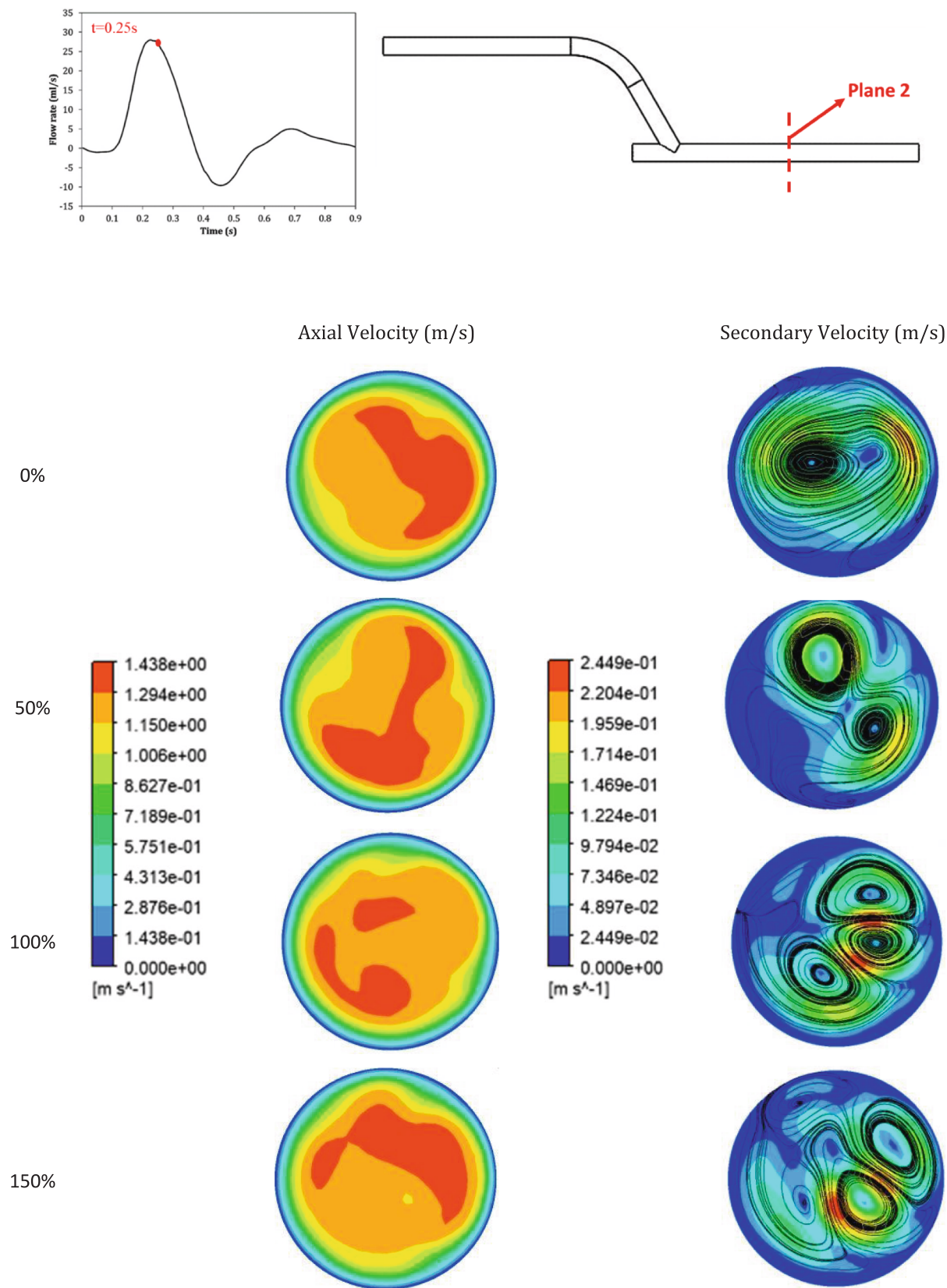


Fig. 10. Comparison of stent thickness increase at 50 mm after anastomosis at $t = 0.25\ s$.

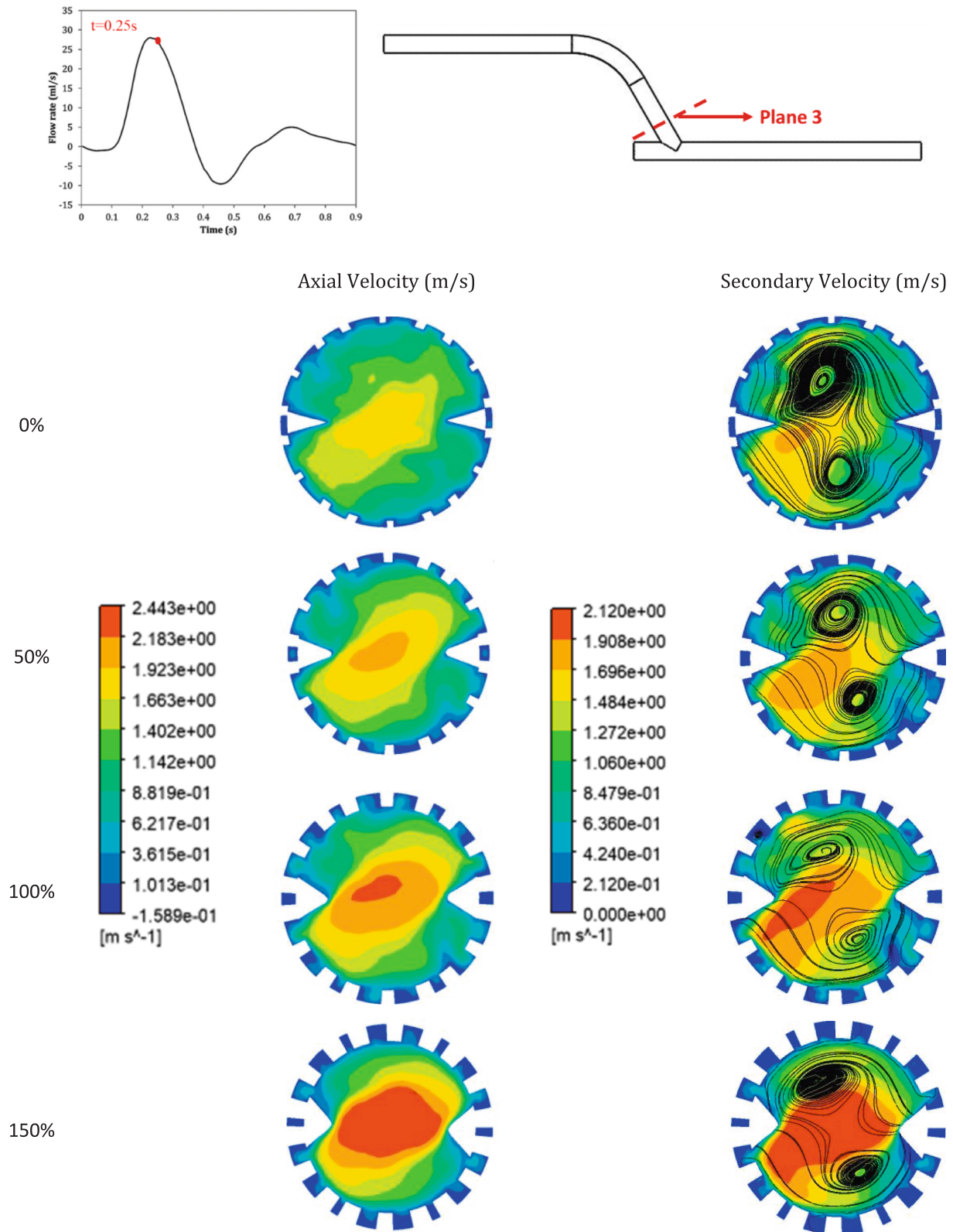


Fig. 11. The comparison of stent thickness increase in the middle of the stent and insert at 0.25 s.

corrosion experiments were essential to reveal the true behavior of the stent, including possible weight loss, structural changes, or mass increase due to sediment deposition, which informed the geometry used for computational simulations. To evaluate the effect of thickening, we generated parametric models by uniformly increasing the initial

thickness by 50%, 100%, and 150% along the full length of the stent and insert. These models were simulated in Fluent, and the outputs were compared with those from the baseline (pre-corrosion) geometry.

Fig. 9 compares the axial and secondary velocities 1 mm after anastomosis in the systole region (time 0.25 s). It suggests that with

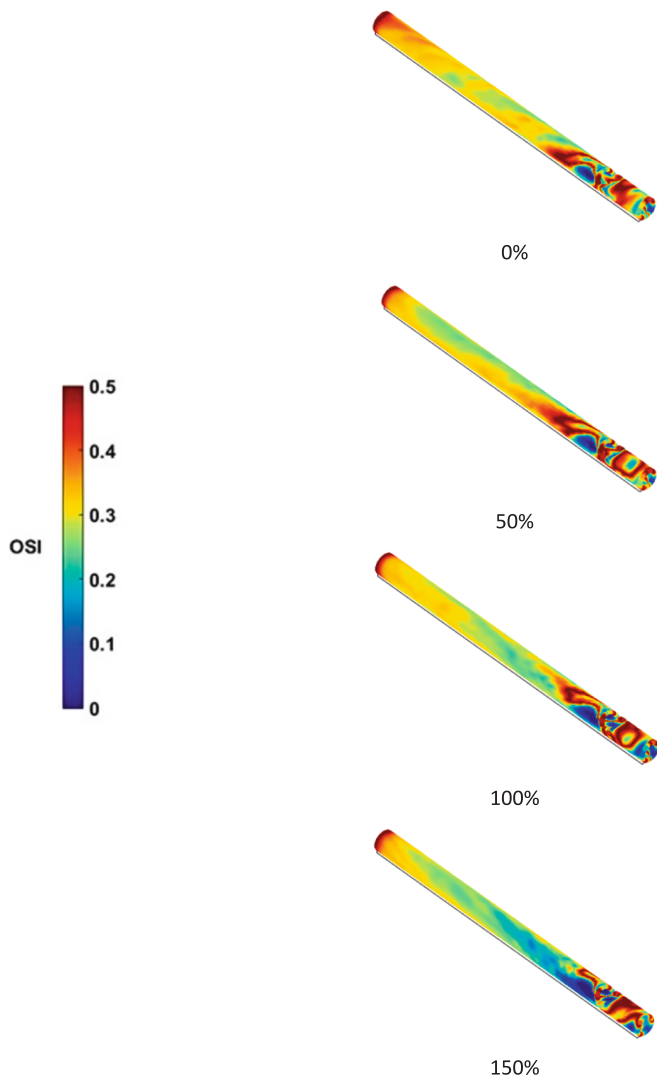


Fig. 12. The comparing the effect of increasing thickness on OSI.

increasing thickness, the axial velocity increases in the right half-plane, but the velocity decreases in the left half-plane, especially near the walls. It is also clear from the secondary velocity images that as the thickness increases, the maximum secondary velocity shifts from the wall side to the center, resulting in stagnant areas that increase the risk of reocclusion [6,8].

Fig. 10 compares the axial and secondary velocities at a distance of 50 mm after anastomosis at $t = 0.25$ s. A notable point regarding the axial velocity is that as the thickness increases, the area with the maximum velocity decreases, and its location shifts from near the wall to the center of the plate. In the secondary velocity, it is clear that as the thickness increases, the number of vortices on one side of the wall increases, but the area with zero velocity on the other side increases compared to the pre-corrosion state. The reason for these changes in the axial and secondary velocities is that with increasing thickness, the surface of the insert is covered by deposits. As a result, the effect of their presence decreases, causing less swirling flow, which also leads to an increase in low-velocity areas. This is consistent with the results of [7,29,48].

Fig. 11 examines and compares the axial and secondary velocities at the center of the stent-insert at $t = 0.25$ s. As the figure illustrates, the axial velocity at the center of the plate increases with thickness. The most important reason for this is the reduction in the cross-sectional area of the flow, which is consistent with reference [29]. For the secondary

velocity, there are two symmetrical vortices in all four cases. However, as the thickness increases, the strength of the vortices decreases. The main reason is the reduction in the spiral's length due to increased thickness and deposition, which lessens the spiral's impact.

Fig. 12 compares the effect of increasing thickness on OSI results. Considering the OSI relationship, it can be said that the results of this figure are for a complete cycle, allowing for more accurate decisions to be made. According to the OSI results, as the thickness increases, the flow rate along the tube becomes more uniform ($0 < \text{OSI} < 0.3$) and the amount of flow non-uniformity decreases ($0.3 < \text{OSI} < 0.5$) [29]. The most significant reason for the decrease in shear stress with increasing thickness is the reduction in the effect of the insert. Overall, the size and prominence of the strut disrupt the near-wall flow field. Consequently, shear stress near the wall decreases at these points, which are linked to neointima and thrombosis. This pattern was observed in the 3D simulations of Ladisa et al. [49], Mejia et al. [50], Martin and Boyle [51], Beier et al. [28], and Hsiao et al. [52]. Laboratory and animal studies, including the evaluation by Kolandaivelu et al. [53], also confirm this finding.

In the case of helical stents, previous studies have also indicated that they are suitable when the twist reaches the wall and is maintained along the vessel length. In this situation, transverse mixing is improved, wall shear stress is increased, and the oscillatory index (OSI) in areas prone to deposition is not unduly reduced. In addition, review and preclinical/clinical data suggest that twisting flows can increase wall shear stress and limit low-flow areas. Any factor that shortens the helix length or confines the twist to the central core (for example, increasing the effective thickness of the legs) weakens the protective effect. As a result, OSI maps tend toward more unidirectionality and less mixing [54,55].

5. Conclusion

In the present study, the corrosion of a new stent and insert design in a body-simulating solution, as well as its effect on blood flow, were investigated. Initially, three corrosion tests were conducted on stent and insert samples printed with two FDM and SLA methods. Based on geometric fidelity and print quality, one specimen (Test C) was selected for flow simulations. The dimensions of the new sample were extracted using FESEM images and 3D scanning. It was found that in some areas, the thickness was increased by a maximum of 0.1 mm. Parametric models with uniform thickness increases of 50%, 100%, and 150% relative to the initial geometry were generated to assess the impact of thickening, and CFD (Ansys Fluent) was conducted using a vein-bypass configuration with the stent and insert placed within the bypass. The key findings are summarized below.

- The corrosion tests showed that the samples exhibited weight gain instead of weight loss, likely due to SBF absorption and sediment deposition. The weight increase percentages recorded in the first, second, and third tests were 7.2%, 1.0%, and 1.5%, respectively.
- More surface sediment formed on FDM samples than on SLA samples, which is linked to larger size, higher roughness, and greater porosity of FDM parts.
- The bead/void of the FDM architecture promotes rapid early fluid ingress and leaching, whereas SLA's crosslinked, smoother, lower-porosity layers produce a slower, steadier change in SBF.
- FDM samples require an initial stabilization (2 weeks) before quasi-steady behavior, while SLA parts in $10 \times$ SBF undergo slow, continuing change over months.
- FDM- $1 \times$ changes rapidly (-9.5×10^{-3} g/day), whereas SLA- $10 \times$ changes more slowly ($+1.9 \times 10^{-5}$ g/day). This suggests that the printing route has a greater impact on mass-change kinetics than the difference between the $1 \times$ and $10 \times$ SBF conditions.
- In the corrosion simulation, it was found that increasing the thickness reduces the effect of the insert; the peak of the axial velocity

tends towards the center. As a result, the velocity near the walls has dropped, and the secondary velocity maximum has shifted from the side of the wall towards the center.

- OSI results in corrosion simulation showed that as the thickness increased, the flow became more uniform along the host vessel, resulting in a decrease in the amount of wall shear stress along the vessel.

While this study provides insights into the corrosion behavior and hemodynamic effects of the stent and insert design, several limitations should be noted. The experiments were conducted on enlarged-scale, in vitro specimens, and the CFD models assumed rigid vessel walls. Future studies could extend this work by investigating patient-specific geometries, incorporating fluid–structure interactions, and performing long-term in vivo validation to better capture physiological conditions and predict clinical performance.

CRedit authorship contribution statement

Bahram Vaziri: Writing – original draft, Software, Investigation, Formal analysis, Data curation. **Saadat Zirak:** Writing – review & editing, Validation, Supervision, Investigation, Funding acquisition. **Mohammad Azadi:** Writing – review & editing, Validation, Supervision, Investigation, Funding acquisition, Conceptualization. **Amir Keshmiri:** Writing – review & editing, Supervision, Investigation, Funding acquisition, Conceptualization. **Nima Shokri:** Writing – review & editing, Supervision, Investigation.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at <https://doi.org/10.1016/j.matdes.2026.116357>.

Data availability

Data will be made available on request.

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