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A 3D flow simulation study for optimizing the geometry of artificial end-to-side vascular anastomoses using biopreparates

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Author Contributions

Conceptualization and methodology, P.K., L.A.F., and N.N.; surgical operations, P.K., L.A.F.; investigation and data analysis, L.A.F., P.K., B.G., L.K-P., and N.N.; writing—original draft, L.A.F., P.K.; writing—review and editing, L.A.F and N.N.; supervision and funding acquisition, N.N.

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1. ABSTRACT

Introduction: End-to-side anastomosis is widely used in organ transplantation, arteriovenous fistula creation, and vascular reconstruction. Its geometry critically influences flow conditions, wall shear stress, and endothelial function, thereby affecting vascular remodeling, regeneration, and long-term anastomotic patency. This study aimed to evaluate the hemodynamic effects of different anastomotic angles using 3D simulations.

Materials and methods: The sciatic artery was dissected in its entire length from fifty chicken tight biopreparations, from which anastomoses at angles of 15°, 30°, 45°, 60° and 90° were prepared using microsurgical techniques (2 vessels/anastomosis, n=5/group). The anastomoses were performed with 12 sutures (8/0 monofilament polypropylene suture), with 16 sutures required only in the 15° group. After preparation, a 22G cannula was secured with ligatures to the acceptor vessel, and the vessels were filled with physiological saline solution. Then, casts were made using a special two-component plastic. The tissue was dissected from the casts, and after 3D scanning, software-based 3D flow simulation measurements were performed (pressure, flow velocity, wall shear stress, vorticity, Reynolds number).

Results: The digitisation and flow simulation of the plastic casts made it possible to compare surgically created arterial anastomoses. The individual flow parameters showed dynamically different changes along the vessel segments. Vessel wall shear stress was too low for anastomoses at an angle of 15°, while it was high for vessel connections made at angles above 45°. The Reynolds number normalised at the shortest distance from the anastomosis opening for anastomoses at angles of 30° and 45°.

Summary: 3D flow simulation is suitable for determining the optimal angle of different types of vascular anastomoses during training. Vascular connections at different angles altered the flow conditions to varying degrees. From a flow perspective, the 30° and 45° angles proved to be the most optimal among the cases examined.

2. INTRODUCTION

Over the past century, the development of vascular anastomosis has undergone remarkable progress, fundamentally shaping the field of modern surgery. Early procedures relied exclusively on vessel ligation until Nikolai V. Eck performed the first vascular anastomosis in 1877, creating a side-to-side portocaval shunt in a dog (Rickard et al., 2014). Subsequent milestones included Jassinowsky's successful carotid artery anastomoses in animals (Jassinowsky et al., 1891), Murphy's human application of the invagination technique (Murphy et al., 1897), and Alexis Carrel's triangular method (Carrel, 1902), refined in collaboration with Guthrie, which provided more favourable outcomes by incorporating intimal transposition (Rickard et al., 2014). These pioneering techniques laid the foundation for microvascular surgery, later revolutionized by Jacobson and Suarez in 1960 with the first successful microvascular anastomosis on small-diameter dog veins, followed by Komatsu and Tamai's groundbreaking finger replantations in 1965 (Rickard et al., 2014).

Today, numerous techniques are available for creating anastomoses, ranging from manual suturing to mechanical stapling and sutureless methods (Seidenberg et al., 1963). Based on vessel orientation, anastomoses are commonly classified as end-to-end, side-to-side, or end-to-side, the latter being of particular clinical relevance in procedures requiring redistribution of blood flow. Ensuring the long-term patency of these anastomoses, however, depends not only on surgical precision but also on the hemodynamic environment created within the junction. Blood flow dynamics, specifically the transition between laminar and turbulent flow, play a crucial role in vascular integrity. Laminar flow, characterized by concentric fluid layers moving in parallel according to the Hagen–Poiseuille law, supports endothelial health, while turbulent flow, described by the Reynolds number, introduces chaotic shear forces that can predispose to

vascular dysfunction (Fonyó, 2011; Baskurt, 2007; Caballero and Lain, 2015; Németh, 2020; Pisano, 2021). Importantly, the flow profile and resulting wall shear stress distribution directly influence endothelial behaviour, thereby affecting long-term anastomotic outcomes (Gribbin et al., 1976, Zhou, 2014; Chistiakov, 2017).

Building upon these biomechanical principles, we hypothesized that the insertion angle of an end-to-side anastomosis acts as a primary determinant of the local hemodynamic microenvironment. Specifically, we postulated that altering the anastomotic angle would induce distinct, quantifiable shifts in critical fluid dynamic parameters—including wall shear stress, vorticity, and the Reynolds number. Furthermore, we hypothesized that systematically mapping these flow variations would allow us to identify a specific angular range that minimizes pathological shear forces, avoids stagnant flow zones, and provides a physiologically optimal profile for endothelial preservation.

Crucially, while most existing computational studies rely on idealized, mathematically perfect vessel geometries, clinical reality is inherently more complex (Németh, 2022; Wongchadukul, 2024). The final architecture of an anastomosis is significantly altered by human and mechanical variables, including the operator's manual technique, tissue traction and deformation, individual suture placement, and the physical characteristics of the suture material itself. We hypothesized that capturing these real-world imperfections—by performing simulations on authentic three-dimensional casts of manually constructed anastomoses on biological tissues—would provide a fundamentally more accurate and clinically translational assessment of the optimal anastomotic angle than conventional theoretical models.

Therefore, the aim of the present study was to comprehensively analyze the flow conditions within end-to-side anastomoses constructed at varying angles (15°, 30°, 45°, 60°, and 90°) using advanced three-dimensional computational simulations based on these authentic microsurgical casts. By bridging the gap between surgically executed models and precise hemodynamic analysis, we sought to determine the optimal geometric configuration that maximizes hemodynamic stability, thereby offering evidence-based guidance to improve long-term graft patency and clinical success.

3. MATERIALS AND METHODS

3.1. Biological model and tools used

The experiments on the non-living biopreparates were registered and approved by the National Food Chain Safety Office (registration Nr. HB/15-ELB/20733-3/2021). The tests were performed on biopreparates, which in our case was the sciatic artery of commercially available chicken upper thighs. The biomodels were obtained freshly (2-3 hours prior to the procedures) from the same location each time. We created 5 anastomoses at different angles (15°, 30°, 45°, 60° and 90° angles, using 10-10 vessels per group (one donor and one acceptor per anastomosis), with a diameter of $2.5 \text{ mm} \pm 0.5 \text{ mm}$ and a length of $6 \text{ cm} \pm 2 \text{ cm}$. We used microvascular clips with a millimetre scale to determine the diameters. To achieve the correct angle, we used a 3D-printed angle gauge designed by us (printed by Zoltán Máté Csák, material: Polylactic acid, 3DPower, Warszawa, Poland), which was also suitable for determining the correct angle at the inlet of the recipient vessel segment during preparation and for stabilising the completed, already joined vessel segments at the required angle. For the anastomoses, we used microsurgical hand instruments and atraumatic 8/0 monofilament polypropylene suture material with serosa needles (Prolene, Ethicon, Raritan, New Jersey, USA). The procedures were performed with the utmost precision using the ALPHA ENT-4A (Elektro Optika, Érd, Hungary) microscope available. To create realistic, tangible models, we used a special two-component plastic (ME3D-Graft, Hungary) and a 22G blue cannula (BBraun Hungary, Budapest, Hungary) to fill the vessels. For digitisation, we used a dental scanner (CARAT dental scanner, Kulzer, Japan) for digitisation, and a CFD program (Ansys 20 R1 background,

Ansys, Inc, Canonsburg, PA, USA) for the simulations, in collaboration with a company specialising in such tasks (ME3D-Graft, Hungary).

3.2. Preparation of anastomoses

To prepare the anastomosis, we first dissected the sciatic arteries from the biomodels using a thorough and atraumatic microsurgical technique. Following skin and muscle incision, the fat and connective tissue surrounding the arteries were removed, exposing the main arterial trunk and the branches opening from it, which were ligated at their origin (8/0, Prolene, Ethicon, Raritan, New Jersey, USA) and then completely removed them from the main trunk. In the case of the donor vessel segments, the vessels were removed from the chicken thighs after complete transection both proximally and distally, on which we then created the appropriate angle inlet using a 3D-printed protractor.

For easier and more accurate handling, the acceptor vessel segment was left in its



Figure 1 3D-printed angle gauge and a sciatic artery bioprepate

anatomical position in the chicken thigh until the end of the procedure. An arteriotomy of the appropriate size was made in the middle of the recipient artery, free from side branches, corresponding to the proximal opening of the above-mentioned donor vessel. Prior to the anastomosis, we flushed the vessels with physiological saline solution (using the 22G blue cannula) to remove any thrombi or debris remaining in the vessels. We placed an insulating plastic strip under the recipient vessel segment at the anastomosis site and then placed the approximator on top of it.

We then proceeded to perform the actual end-to-side anastomosis, which was carried out in all cases with 12 simple knotted sutures, except for the 15° angle, as a lower angle resulted in a larger entry opening. We chose simple knot stitches because other stitching techniques would result in significant deformities (indentation, loss of vessel wall elasticity) at the anastomosis level, which would act as distorting factors in the examination of the target parameters set for the research. As a first step, we placed the two main retaining sutures, then made another temporary retaining suture halfway along the front wall, and then placed the remaining 2 or 3 sutures between the already placed sutures, depending on the angle. We performed the same procedure on the rear wall as on the front wall (Figure 2).

We cannulated the proximal end of the combined vessel segment, through which we performed thorough irrigation to remove air from the vessels and check for any deficiencies in the suture line.

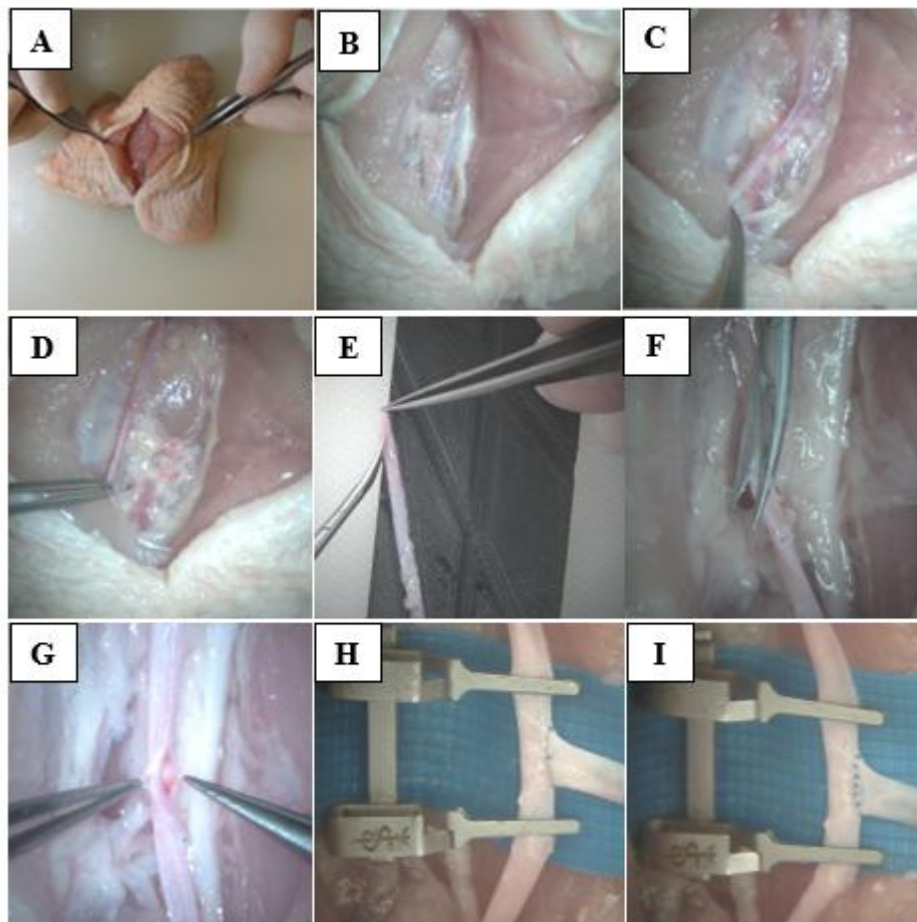


Figure 2 Steps in preparing the anastomosis

A: skin-muscle incision B: preparation of the sciatic artery C-D: locating, closing, and transecting the side branches E: forming the appropriate angle F-G: arteriotomy of the recipient vessel H: placement of holding sutures I: suture line I: Completion of suture line (equally on front and back walls)

3.3. Three-dimensional model

We mixed the two-component plastic material, drew it into a 5 ml syringe, and injected it into the vessel model through the cannula previously inserted. Once the plastic had passed through the entire vessel segment, its distal ends were ligated and the appropriate pressure was created in the vessels. During the 5-7 minute setting time of the two-component plastic, the fused vessel segment was fixed in the desired shape using the 3D printed gauge. After the model had solidified, the entire vessel wall was dissected using a microsurgical, atraumatic method, resulting in the three-dimensional structures shown in Figure 3, from which virtual 3D models were created using the scanner and the Ansys 20 R1 background software (Figure 4).

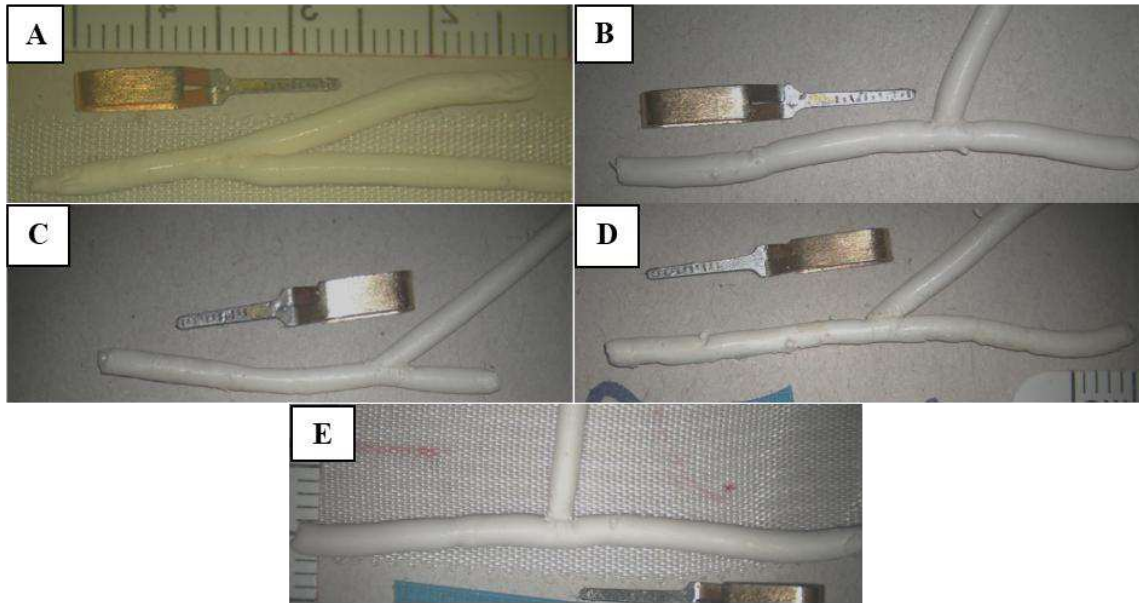


Figure 3 Representative two-component plastic-filled vessels

A: 15° B: 30° C: 45° D: 60° E: 90°

To compare certain angles, we examined the most significant parameters, such as flow velocity, pressure profile, wall shear stress, Reynolds number, and vorticity and turbulence in the vessels. To examine the individual values, we determined all measurement points projected onto the wall in a given plane on the digital coin models. At a distance of 5 mm from the connection, we recorded the temporal medians (under the entire pulse curve) of the spatial

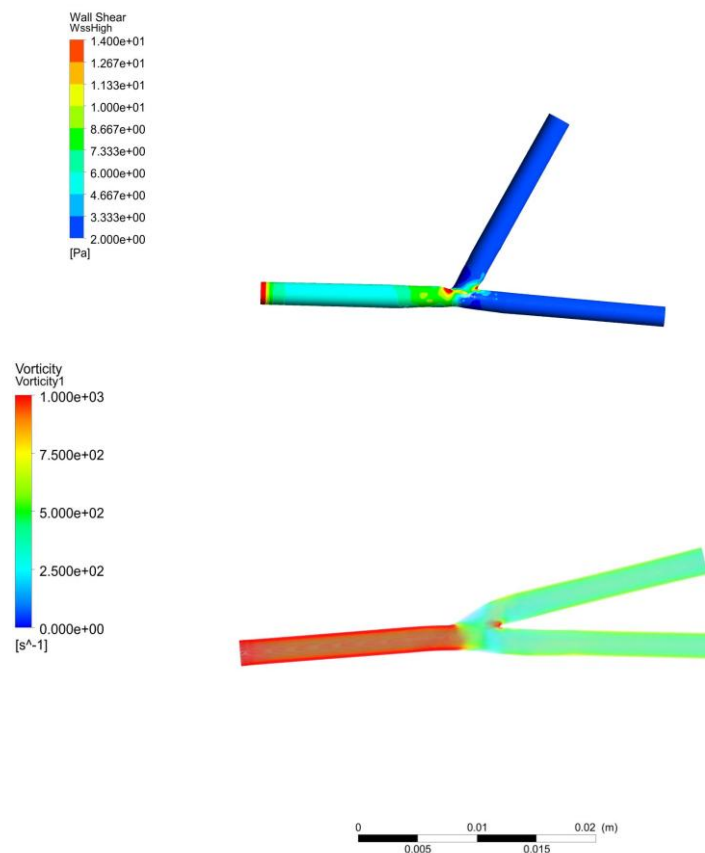


Figure 4 Representative 3D simulation images (wall shear and vorticity)

values of the aforementioned indicators, with the exception of flow velocity, in the proximal, distal and side branch sections. The change in flow velocity at the end points of the combined vessel segment was examined. We determined not only the location but also the time of recording each value, which we marked as the maximum of the simulated pulse curve, so that the measurements could be performed at the most intense flow possible (Figure 5).

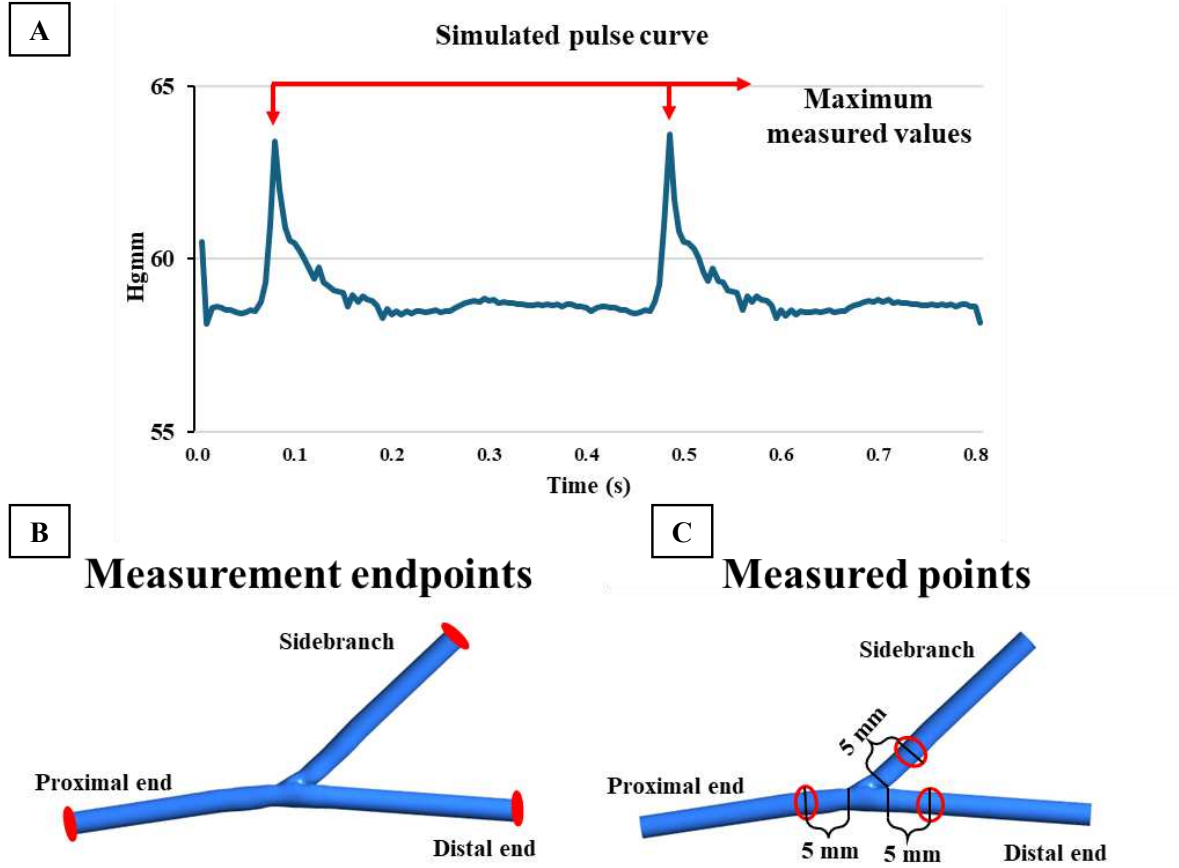


Figure 5 A: Simulated pulse curve image; B: Measurement endpoints for flow velocity C: Measurement points near the anastomosis (pressure profile, wall shear stress, Reynolds number, vorticity)

3.4. Statistical analysis

The values are presented as mean $\pm$ standard deviation (S.D.). The results showed a normal distribution, and we used paired T-tests and one-way ANOVA tests for statistical analysis, depending on the data to be compared (Whitley and Ball, 2002; Kim, 2015). SPSS Statistics version 22.0 and Microsoft Excel 2016 were used for the calculations. The significance level was set at $p < 0.05$ in all cases.

4. RESULTS

4.1. Pressure profile

The main driving force of blood flow in the vascular system is the pressure difference between the central and peripheral vascular segments, so the first value determined in the study was the pressure profile. Looking at the results shown in the graph, we can see that the initial proximal (P) pressure in the vascular segments following the anastomosis of the artery shows a decrease in pressure in both the distal and side branch segments due to an increase in the total vessel diameter and a decrease in resistance. The pressure values between the individual angles do not show significant differences in the examined vessel segments, but it can be seen that in all three examined segments, the highest pressure values were obtained with the 90° anastomosis, while the lowest values were obtained with the 15° construction (Figure 6).

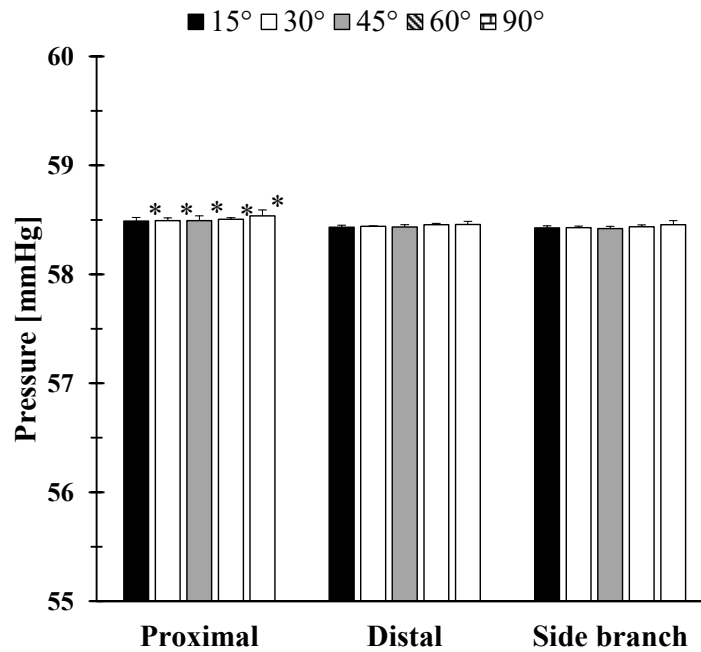


Figure 6
Pressure profile at the determined points of the examined vessel
mean±SD; *p<0.05 vs. distal and vs. side branch

4.2. Flow velocity

The flow velocity in each section of the vessel was determined by measuring the values at the maximum of the simulated pulse curve during the tests and examining their distribution, thus obtaining an accurate picture of the velocity of the blood flowing through each section of the vessel.

The maximum velocity, average velocity and velocities measured at identical points on each side branch were determined for all angles examined. The recorded data clearly show that, in accordance with the simulation mechanism, the velocity measured at the proximal end before the anastomosis was identical at different angles due to the negligible differences between the input simulation values and the measured values. However, the values measured in the sections following the anastomosis (distal and side branches) play a more significant role in determining the differences between the individual angles. In both branches, we can see that as the angle of the anastomosis decreased (below 60°), the flow velocity also decreased (Figure 7).

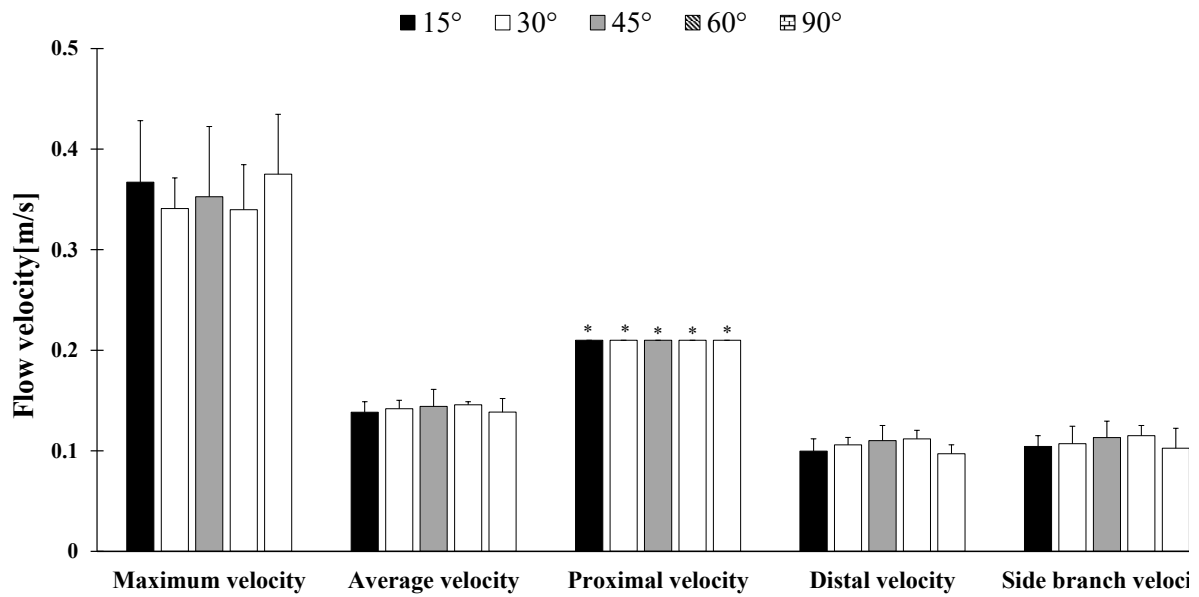


Figure 7

Flow velocity values at the peak of the simulated pulse waveform
 average±SD; *<0.05 vs. distal, vs. side branch, vs. average and maximum velocity

4.3. Wall shear stress

The results obtained show that the highest values were obtained in all cases for the model prepared at 90°, while the lowest results were obtained for the distal and lateral sections for structures at angles of 15° and 45°. In the case of the proximal vessel section, the lowest shear stress was also observed at 45° (Figure 8).

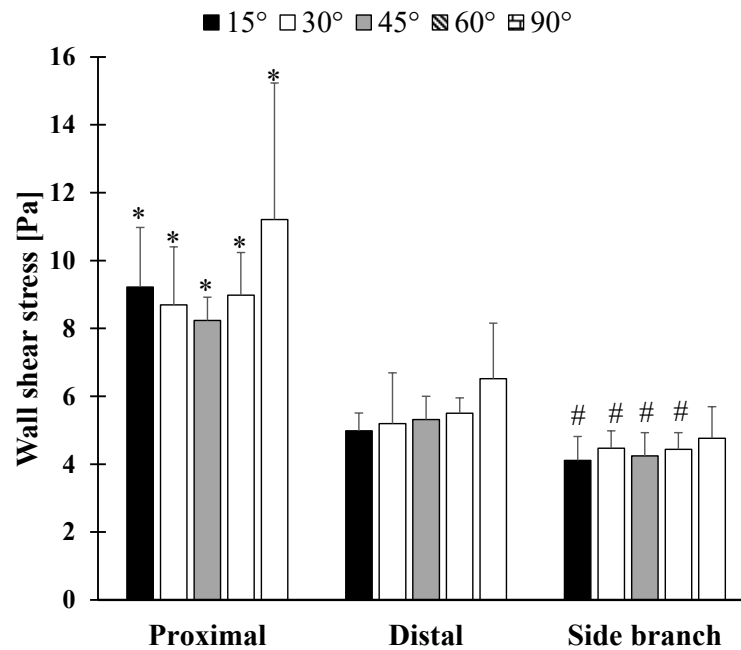


Figure 8

Wall shear stress at the determined points of the examined vessel
 mean±SD; *p<0.05 vs. distal and side branch; #p<0.05 vs. 90° distal

4.4. Reynolds number

The Reynolds number is a dimensionless value, the magnitude of which allows us to infer the degree of turbulence in the flow. The higher the value, the greater the likelihood of turbulence occurring. The values obtained from the analysis of the data show that the lowest values were obtained at 30° and 45° distally from the anastomosis and at 15° and 30° in the case of the side branch (Figure 9).

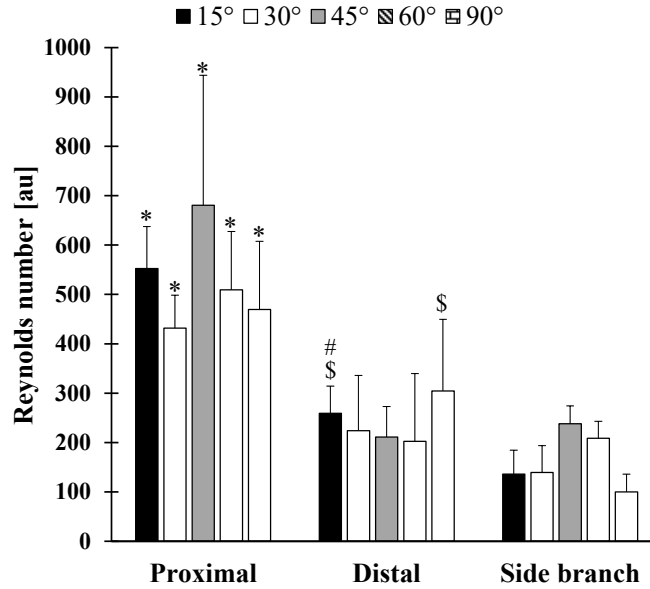


Figure 9

Reynolds number values at the determined points of the examined vessel

mean±SD; *p<0.05 vs. distal and side branch; #p<0.05 vs. 90° distal; \$ p<0.05 vs. 90° side branch

4.5. Vorticity

We quantified the vorticity, or vortex formation, in the combined vessel segments. Based on the values obtained during the evaluation, it is evident that the 90° model produced the greatest vortex formation at all points examined in the combined vessel segment, thus allowing the flowing blood and its shaped elements to exert a direct force on the vessel wall and all its layers. Considering the side branch and the distal end of the artery, we can see a similar pattern, i.e. lower vorticity values were measured at 15°, 30° and 45° (Figure 10).

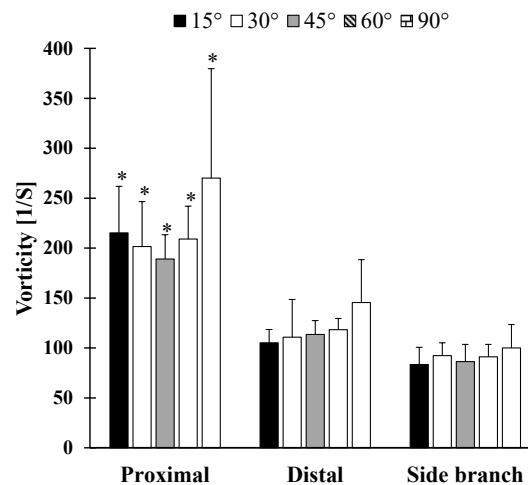


Figure 10 Vorticity at the determined points of the examined vessel

mean±SD; *p<0.05 vs. distal and vs. side branch

5. DISCUSSION

While the evolution of vascular anastomosis techniques has historically paralleled advancements in both surgery and microscopy, the hemodynamic evaluation of these procedures often relies on computational models that assume idealized conditions. Such models typically feature uniform vessel cross-sections and perfectly executed sutures—circumstances that rarely reflect clinical reality. Although previous investigations have compared various vessel-joining techniques, the present study addresses a more focused and clinically critical question: what is the optimal angle for an end-to-side anastomosis to achieve favorable hemodynamic conditions? This determination is of substantial practical importance, particularly for procedures such as arteriovenous fistula creation for hemodialysis and solid organ transplantation, where optimal flow characteristics are paramount for long-term patency and overall clinical success (Hughes et al., 1995; 1996; Loth et al., 2008; Roy-Chaudhury et al., 2006).

Given these considerations, the geometry of end-to-side anastomoses is pivotal. The junction angle directly dictates flow patterns, wall shear stress (WSS), and ultimately, graft survival. Because the fundamental laws governing fluid mechanics and circulatory dynamics remain constant, variations in hemodynamic outcomes arise predominantly from operator-dependent factors during surgical construction. To investigate these effects while accounting for the natural variability of surgical technique and vessel wall elasticity, we manually constructed anastomoses on real biological specimens. Because the origin angle of the graft stem is the primary geometric determinant in end-to-side configurations, establishing a highly accurate geometric baseline was essential. To this end, we utilized a custom-made angular measuring device paired with a three-dimensional (3D) printed acrylonitrile butadiene styrene (ABS) vascular scaffold, enabling the precise preparation of both the anastomotic angle and the proximal graft segment. The utilization of such 3D-printed customized templates represents a critical advancement in standardizing *ex vivo* models for accurate computational fluid dynamics (CFD) validation (Lashkarinia 2018; Chinchilla-Mendez 2022; Yu, 2024).

This individualized, manual preparation of the vessels yielded a physiologically relevant experimental model that more closely approximates *in vivo* conditions than conventional, idealized simulations. By utilizing chicken sciatic nerve biopreparations alongside 3D plastic models, our experiments facilitated flow simulation testing under highly reproducible conditions with minimal variability. Crucially, the resulting data—derived from precise 3D casts of the manually constructed specimens—encapsulate not only the hemodynamic consequences of anastomotic geometry but also the inherent human factors of surgical practice. To our knowledge, this methodological synthesis—combining precisely controlled geometry, manual anastomosis construction on real biological specimens, and subsequent computational analysis—represents a unique tier of experimental evidence without direct precedent in the current literature (Németh, 2022; Wongchadakul, 2024), expanding upon the growing paradigm of patient-specific and image-based hemodynamic modeling (Taylor & Figueroa, 2009).

Our subsequent computational analyses provided high-fidelity data elucidating the impact of geometric variations on flow characteristics. The observed differences confirm that the anastomotic angle exerts a profound effect on key hemodynamic parameters, including velocity distribution, Reynolds number, vorticity, and WSS. Collectively, these parameters have well-established downstream consequences on endothelial integrity and vascular function, underscoring the critical need for geometric optimization in the surgical construction of anastomoses (Katsoudas, 2026).

Our findings are consistent with earlier studies by Hughes and colleagues, who demonstrated that variations in Reynolds number and velocity profiles can alter the transition between laminar and turbulent flow, with implications for vortex formation and local shear forces (Hughes et al., 1995; 1996). Similarly, the work of Sarpkaya et al. (1992) underscores

the difficulty in pinpointing the exact threshold for turbulence, due to the stochastic nature of fluid particle motion. In agreement with these reports, our study demonstrated that turbulence and disordered flow increase with certain anastomotic angles, predisposing to eddy formation. Such vortices, particularly under low-flow conditions, contribute to prolonged platelet contact, endothelial activation, and the release of procoagulant and growth factors, which can drive intimal thickening and thrombus formation (Burkel et al., 1988; Henry et al., 1996).

The role of wall shear stress (WSS) warrants particular emphasis. WSS is defined as the tangential force exerted on the endothelial surface by the friction of blood flow and its components (Jacek et al., 2003; Malek et al., 1999). This parameter is one of the most critical determinants of vascular health, as it directly influences endothelial structure and function. In the present study, differences in WSS values provided meaningful information for comparing anastomosis angles. Importantly, WSS is shaped by multiple factors, including blood flow velocity, viscosity, pressure, and vessel wall mechanics (Papaioannou et al., 2005). By using numerical simulation, we were able to calculate these values with high precision, thereby ensuring reliable evaluation of the flow conditions created by different geometries.

Different researches indicate that both excessively high and excessively low WSS can be detrimental. High WSS promotes endothelial hyperplasia and adaptive remodeling (Ojha et al., 1993; Sottiurai et al., 1989; Sottiurai et al., 1983, Purcell 1997), while low WSS favours platelet adhesion, coagulation factor accumulation, and thrombus formation due to prolonged residence time of blood elements near the vessel wall (Henry et al., 1996; Szabó et al 2022; Malek et al., 1999). This duality underscores the importance of identifying an optimal shear range. Advanced mechanobiological studies have confirmed that these low and oscillatory shear zones activate specific pro-inflammatory transcription factors at the molecular level, aggressively driving early intimal hyperplasia (Souilhol et al., 2020; Szabó et al, 2022). In our study, models with a 15–30° angle demonstrated favourable vorticity profiles, while 30° in the distal host vessel and 45° in the side branch produced advantageous WSS distributions, suggesting these geometries may best balance hemodynamic stability and vascular health.

In addition to hemodynamic conditions and flow characteristics, the rheological properties of blood fundamentally determine the distribution of wall shear stress (WSS) developing on the endothelial surface. Changes in blood viscosity, as well as in the aggregation or deformability of red blood cells, can directly alter the mechanical stimuli acting on the endothelium, thereby significantly influencing its functional state (Németh, 2022). Treating blood as a Newtonian fluid is a widely applied simplification in hemodynamic modeling; however, numerous studies have demonstrated the critical importance of its non-Newtonian characteristics regarding flow conditions, wall shear stress, viscosity, and the development of their clinical implications (Ballyk et al., 1994; Liepsch et al., 1984). The non-Newtonian behavior of blood is dependent on both hematocrit and shear rate. At lower flow velocities, blood viscosity increases (thixotropic effect) due to the viscosity-elevating impact of red blood cell aggregation. Conversely, at higher shear rates, the cells disaggregate and elongate in the direction of flow due to their deformability and elastic properties (Baskurt, 2007; Németh, 2020).

Beyond geometric considerations, systemic factors such as hypertension also play a significant role in vascular outcomes. As described by Rizzoni et al. (1998) and Kumar et al. (2017), chronic pressure elevation impairs endothelial function, increases vessel wall susceptibility to degenerative changes, and negatively impacts tissue regeneration. Since anastomosis creation inherently disrupts vascular continuity, such stressors may further exacerbate adverse remodeling processes (Chiu et al., 2011). Thus, identifying geometries that minimize hemodynamic stress may help mitigate these risks in clinical practice.

As noted in previous computational fluid dynamics (CFD) research, laboratory studies provide highly controlled and precise measurements but can only approximate the complexity

of *in vivo* conditions (Hughes et al., 1996). Therefore, further research using animal models and, eventually, clinical studies will be necessary to validate the applicability of the geometries identified here. While the use of scanned plastic-filled different angled anastomoses and numerical CFD simulations ensured highly accurate geometry and flow analyses, the *in vitro* environment inevitably lacks several physiological factors that influence vascular healing and remodeling *in vivo*. Specifically, the absence of pulsatile pressure fluctuations generated by a living heart, the lack of active vasomotor responses, and the missing contribution of blood cellular elements (platelets, leukocytes, endothelial progenitor cells) limit the biological realism of the models (Ku, 1997). In particular, endothelial regeneration, inflammatory signaling, and thrombogenicity could not be evaluated under these conditions. Furthermore, standard rigid-wall CFD simulations inherently omit fluid-structure interaction (the dynamic expansion and recoil of the vessel wall during the cardiac cycle) which modern research identifies as a significant modifier of instantaneous shear stress profiles (Arzani & Shadden, 2018).

Our experiments utilized the chicken arteria ischiadica model, chosen for its excellent structural similarity to human vasculature and technical availability. To place the results in the proper context, it is also necessary to take into account the differences in the physiological parameters of the avian model. The average heart rate in chickens is higher ($\approx 286/\text{min}$ vs. $\approx 90/\text{min}$ in humans), accompanied by elevated systolic and diastolic pressures (142/117.8 mmHg vs. 120/80 mmHg). As a result, the mean arterial pressure (MAP) is higher (126 mmHg vs. 93 mmHg), alongside a lower central venous pressure (4.4 mmHg vs. 10 mmHg). In addition, the body temperature of chickens (41 °C) could potentially cause discrepancies in the assessment of enzyme activity and vascular wall elasticity. Rheologically, whole blood viscosity (WBV) and plasma viscosity (PV) in the avian model are inherently lower (WBV: 2.141 mPa·s, PV: 0.944 mPa·s) than in humans (WBV: 5.996 mPa·s, PV: 1.24 mPa·s), reflecting a lower baseline hematocrit (27.6% vs. 40%) (Maxwell 1992; Windberger 2003). These specific physiological characteristics create a consistent and specific hemodynamic environment for the assessment of flow resistance, shear stress on the vessel wall, and vortex formation. However, since our study used only fully developed vascular preparations, did not involve regenerative processes, and involved scanning and CFD analysis of vessels that had been harvested, sutured and then filled with plastic, these differences are likely to have only a minor impact on the applicability of our results to humans. (Padmanaban 2021; Maxwell 1992; Windberger 2003).

To ensure high standardization in our experimental setup, we utilized specific methodological approaches. The application of simple interrupted sutures was deliberately chosen to minimize geometric distortion at the anastomosis site, providing a clear and highly reproducible baseline geometry for analysis, even if it represents a specific subset of broader clinical microsurgical techniques. Biomechanical evaluations underscore that the choice of suturing technique and the resulting compliance mismatch at the suture line significantly perturb local hemodynamics (Desai, 2011). Furthermore, utilizing an injected two-component polymer was essential for achieving highly precise 3D reconstructions. This technique effectively isolates the purely geometric and structural effects of the anastomosis angle on flow dynamics, separating them from the variables of natural vessel wall elasticity.

In summary, our biomodel system provides a robust and reproducible platform for investigating angle-dependent hemodynamic changes, which are also influenced by the stitches and knots. By factoring in the specific physiological and rheological parameters of the model, the structural insights gained from this standardized setup offer valuable, translatable data for clinical microvascular practice. Furthermore, our study reinforces the critical importance of anastomotic geometry in shaping flow dynamics. The results suggest that end-to-side anastomoses constructed at 30° and 45° angles may provide the most favourable hemodynamic

conditions, effectively balancing vorticity and wall shear stress (WSS) to reduce the risk of thrombosis and pathological remodeling.

6. CLINICAL IMPLICATIONS

From a clinical perspective, the optimization of anastomotic geometry holds considerable promise in improving outcomes for patients undergoing vascular reconstructive procedures. In arteriovenous fistula surgery, for example, ensuring stable flow patterns and minimizing turbulence may prolong fistula patency, reducing the frequency of reinterventions. Similarly, in transplant surgery, where the success of the graft is highly dependent on adequate blood supply, selecting an anastomotic angle that ensures optimal wall shear stress distribution may help prevent early thrombotic events and long-term vascular complications. Our results indicate that 30°–45° constructions provide the most stable hemodynamic conditions, which may serve as a practical guideline for surgeons. These results apply to the settings we used with the arterial parameters, which correspond to the direction of flow toward the anastomosed branch, where we measured the angle between the two distal branches; this should be taken into account in clinical applications. Future translation of these findings into clinical practice could contribute to the refinement of microsurgical techniques and the improvement of patient outcomes.

7. LIMITATION

Although the applied methodology allowed us to investigate angle-dependent hemodynamic alterations with a high degree of reproducibility and precision, some limitations should be acknowledged when interpreting the results.

Although computational fluid dynamics provided highly detailed information on velocity fields, pressure profiles, and wall shear stress distribution, these results inherently depend on the assumptions and boundary conditions set during the simulations. Parameters such as viscosity, inlet velocity, and pressure profiles were based on average values, which may not capture the full variability of physiological blood flow *in vivo*.

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10. DISCLOSURE

The authors report no proprietary or commercial interest in any product or concept discussed in this article.

11. DATA AVAILABILITY STATEMENT

The datasets used and/or analysed during the current study available from the corresponding author on reasonable request.

12. REFERENCES

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