

Hemodynamic Consequences of Renal Artery Ostium Positioning After Inner-Branch Endografting for Juxtarenal Aortic Aneurysms

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Background: To investigate the hemodynamic consequences of renal artery ostium positioning following endovascular repair of juxtarenal abdominal aortic aneurysms (JAAAs) using the novel WeFlow-JAAA inner-branch system.

Methods: Patient-specific computational fluid dynamics analysis was performed on 3 postoperative JAAA cases treated with the WeFlow-JAAA endograft. For each case, 4 models systematically simulated progressive distal migration of the bilateral renal ostia, ranging from a proximal position near the celiac artery to their native anatomical locations. Renal perfusion rates, aortic flow patterns, and established wall shear stress (WSS)-derived metrics (time-averaged wall shear stress [TAWSS], oscillatory shear index [OSI], relative residence time [RRT]) were quantitatively evaluated.

Results: Distal renal ostium positioning consistently yielded higher renal perfusion rates compared to proximal placements. However, this improved perfusion was associated with the development of potentially adverse hemodynamic conditions (characterized by low TAWSS and high OSI/RRT) on the perirenal aortic wall adjacent to the ostia. Conversely, proximal placements, while mitigating these adverse perirenal WSS patterns, compromised renal perfusion and generated pronounced flow disturbances in the infrarenal aorta distal to the termination of the inner-branch parallel segments.

Conclusion: The preliminary findings suggest that renal ostium positioning after WeFlow-JAAA implantation may critically influence postoperative hemodynamics. The simulations indicate maximizing renal perfusion via distal placement may potentially expose the perirenal aortic wall to less favorable long-term WSS conditions. These results underscore the need for careful and potentially patient-specific consideration of device placement to appropriately balance perfusion and aortic wall remodeling risks in juxtarenal endovascular repair.

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INTRODUCTION

Juxtarenal abdominal aortic aneurysms (JAAAs), constituting approximately 16% of infrarenal aneurysms, presents a significant therapeutic challenge due to aneurysmal dilatation extending near the renal arteries, typically resulting in an inadequate proximal sealing neck (<10 mm) that precludes standard endovascular aneurysm repair (EVAR).^{1,2} Consequently, complex EVAR utilizing custom-manufactured fenestrated or branched endografts (F/BEVAR) represents the current mainstay treatment. These devices facilitate renal artery preservation through precisely positioned fenestrations or branches designed to align with native ostia, accommodating bridging stents to maintain perfusion while excluding the aneurysm.^{3–7}

A major limitation of custom F/BEVAR devices, including physician-modified endografts, is the requisite manufacturing or modification lead time, often exceeding several weeks, rendering them unsuitable for urgent or emergent clinical scenarios.⁸ This limitation has driven the development of “off-the-shelf” (OTS) endovascular solutions for JAAAs.^{9,10} The WeFlow-JAAA system exemplifies such an innovation, featuring a novel hybrid design incorporating standardized inner branches for renal artery perfusion alongside modified fenestrations for other visceral vessels.¹¹ The system comprises proximal and distal main bodies (the former containing the renal inner branches), iliac limbs, and dedicated bridging stent grafts (Fig. 1A).¹¹ Preliminary clinical evaluations suggest favorable technical success and short-term outcomes, positioning this technology as a potentially valuable addition to the JAAAs treatment armamentarium.¹²

Despite meticulous planning aimed at minimizing alterations to renal inflow pathways, post-procedural renal function impairment remains a significant morbidity associated with these complex repairs,^{5,13} highlighting the sensitivity of renal outcomes to procedural and device-related factors. For example, a relatively high incidence of type I or III endoleak following FEVAR can induce acute and chronic renal dysfunction.¹⁴ In contrast, BEVAR is associated with a higher rate of renal artery occlusion, with 1 study reporting an incidence of 9.6% at 2 years postoperation versus 2.3% for FEVAR.¹⁵ While different endovascular techniques affect the renal arteries through varying mechanisms, preserving renal function post-EVAR remains a persistent challenge across all modalities. OTS inner-branch devices like the WeFlow-JAAA are no exception.

Although this design may mitigate certain failure modes associated with fenestration/bridging stent misalignment,^{16,17} the inner-branch design introduces a distinct and unavoidable anatomical alteration. It inherently alters the effective entry position and inflow angle of the renal arteries relative to the aortic lumen, creating a segment where the renal bridging stent runs parallel within the main endograft body. This mandatory geometric repositioning has hemodynamic ramifications. Specifically, its impact on local flow dynamics is currently uncharacterized. This includes the flow at the ostium, within the parallel segment, and in the aorta distal to the branch termination. Understanding these device-specific hemodynamic alterations is crucial, as they represent a critical knowledge gap concerning the potential long-term effects on both renal perfusion and aortic wall pathobiology.

Therefore, this exploratory study aimed to quantify the hemodynamic consequences arising specifically from varying renal artery ostium positioning secondary to WeFlow-JAAA inner-branch endografting. Utilizing patient-specific computational fluid dynamics (CFD) models derived from 3 clinical cases treated with the WeFlow-JAAA system, we systematically investigated the impact of progressive distal migration of the renal artery inner-branch ostia on both renal perfusion metrics and aortic hemodynamic parameters. By elucidating these mechanistic links between surgically dictated ostial position and resultant hemodynamics, this work seeks to provide critical insights into the potential long-term renal and aortic implications associated with this novel inner-branch design, thereby contributing foundational knowledge relevant to the optimization of JEVAR strategies employing such technology.

MATERIAL AND METHODS

Image-Based Model Reconstruction

Postoperative computed tomography angiography (CTA) data were obtained from 3 male patients (aged 65, 71, and 66 years, respectively) who underwent endovascular repair for JAAAs using the WeFlow-JAAA inner-branch endograft system. For each patient, three-dimensional intraluminal geometries of the aorta, renal arteries (including the stented inner-branch segments), celiac artery (CA), and superior mesenteric artery (SMA) were reconstructed from the CTA datasets using Mimics

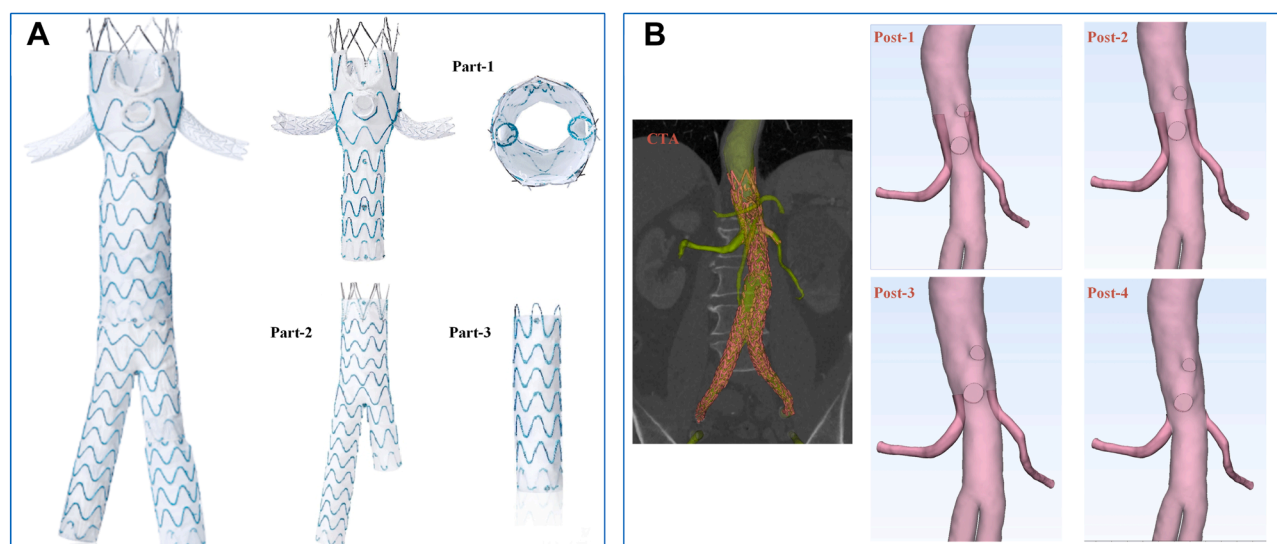


Fig. 1. Characteristics of the WeFlow-JAAA stent and optimization of postoperative model reconstruction. **(A)** Composition features of WeFlow-JAAA stent system. **(B)** Schematic diagram of postoperative model reconstruction and structural optimization.

software (Version 19.0; Materialize, Plymouth, MI, USA).

The initial reconstruction directly represented the as-implanted postoperative anatomy, where the functional ostia of the renal artery inner branches were observed to be located relatively proximally, near the axial level of the native CA origin. This patient-specific geometry served as the baseline configuration for subsequent analysis and was designated as the post-1 model (Fig. 1B). To systematically investigate the hemodynamic impact of renal inner-branch ostial positioning, 3 additional hypothetical geometric models were computationally generated for each patient, simulating progressive caudal migration of the bilateral renal branch origins along the aortic axis relative to the post-1 configuration. The specific axial positions were defined based on key anatomical landmarks identified from preoperative imaging. Post-2 model is the renal inner-branch ostia were repositioned to an axial level located midway between the origins of the native CA and SMA. Post-3 model is the renal inner-branch ostia were relocated to the axial level corresponding to the native SMA origin. Post-4 model is the renal inner-branch ostia were positioned at the axial level corresponding to their preoperative anatomical (native) locations.

Crucially, all models (post-1 – post-4) maintained the fundamental architecture of the WeFlow-JAAA system, including the presence of the parallel inner-branch segment connecting the repositioned ostium to the renal artery proper. The

geometric representations of these 4 configurations for the 3 patient cases are illustrated in Figures 1B and 2. This setup allowed for a controlled comparison of hemodynamic characteristics specifically attributable to variations in the axial take-off position of the inner renal branches.

Calculation and Boundary Conditions

Numerical simulations of pulsatile blood flow were performed for all 4 geometric configurations for each patient case using the finite element method implemented in COMSOL Multiphysics® (Version 5.5, COMSOL AB, Stockholm, Sweden). The computational domain encompassed the intra-luminal blood volume. Finite element meshes primarily composed of tetrahedral elements were generated for the fluid domain. To accurately resolve near-wall velocity gradients crucial for wall shear stress (WSS) calculations, 5 layers of prism elements were constructed in the boundary layer region adjacent to the vessel wall interface. A mesh independence study was rigorously performed for a representative model by systematically refining the mesh density until cycle-averaged flow rates in major branches and area-averaged WSS in key regions varied by less than 3% between consecutive refinement levels, ensuring that the simulation results were independent of mesh resolution. The selected mesh density was then applied consistently across all models.

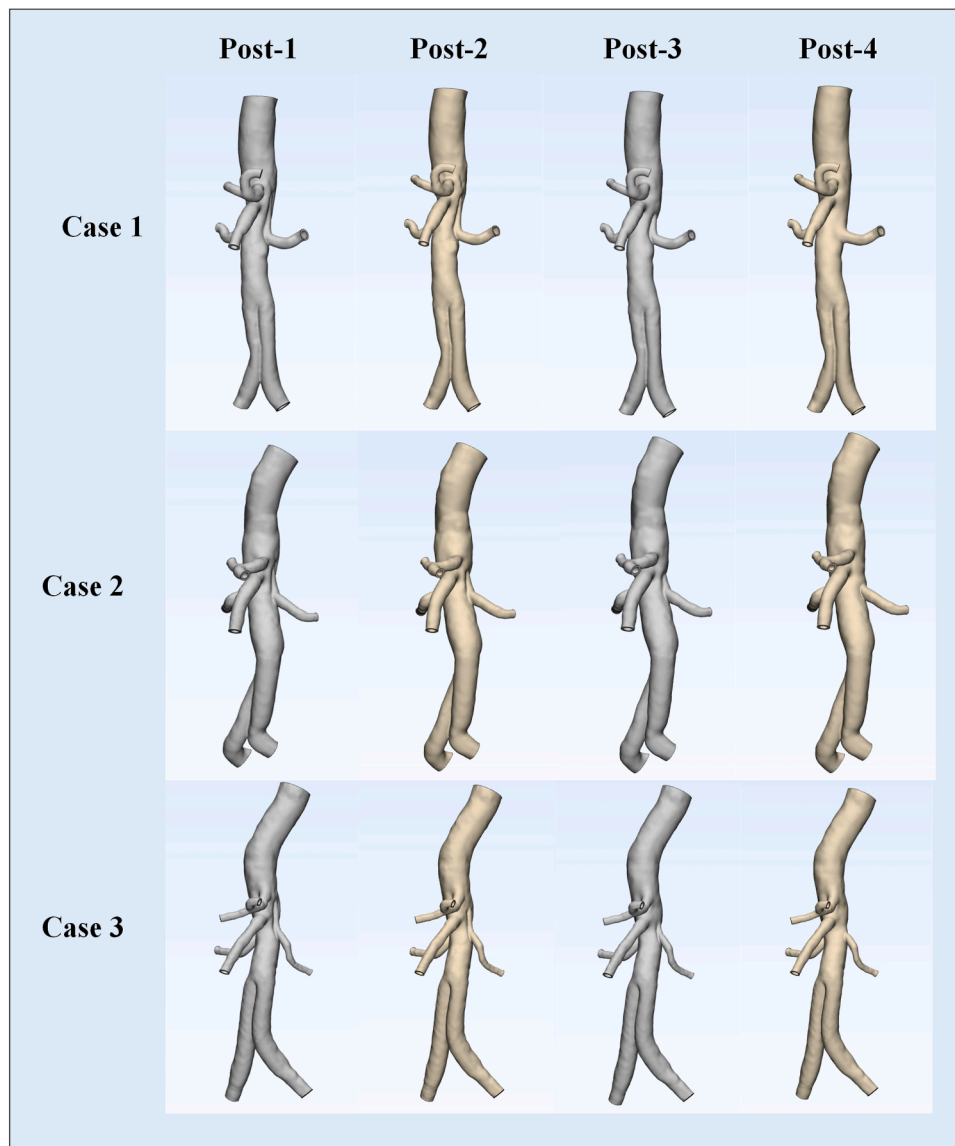


Fig. 2. The postoperative reconstructions and optimized models of the 3 cases.

Blood flow was modeled as an incompressible, homogeneous Newtonian fluid, governed by the Navier-Stokes equations. Standard physiological values were assigned for fluid properties: density (ρ) = 1,060 kg/m³ and dynamic viscosity (μ) = 0.0035 Pa s. A critical assumption employed in this fluid dynamics simulation was that the vessel walls were rigid and nondeformable.

Time-varying, physiologically representative boundary conditions were implemented, adapted from established literature data¹⁸ representing typical resting hemodynamics, acknowledging that these were not derived from patient-specific measurements.

A pulsatile volumetric flow rate waveform (Fig. 3) was prescribed uniformly across the proximal aortic inlet surface of each model. Pulsatile pressure waveforms (Fig. 3) were applied at all distal outlets to account for downstream peripheral vascular resistance. Consistent with physiological branching patterns and impedance characteristics, identical pressure waveforms were assigned to the CA and SMA outlets. Likewise, uniform pulsatile pressure boundary conditions were imposed on both renal artery outlets. The iliac artery outlets were also subjected to matching pressure waveforms. A key aspect of the simulation strategy was the application of the identical set of inlet flow and

outlet pressure boundary conditions across all 4 geometric models for each individual patient case. This approach was adopted to specifically isolate and compare the hemodynamic alterations attributable solely to the geometric variations in renal inner-branch ostial positioning.

Related Indicators

Postprocessing analysis focused on quantifying the hemodynamic environment within key regions susceptible to alteration by the inner-branch endograft configuration. Specifically, analysis included the luminal surface of the reconstructed renal arteries and critical areas on the aortic wall, namely the perirenal aortic surface adjacent to the inner-branch ostia and the infrarenal aortic segment distal to the termination point of the parallel inner branches.

Several key hemodynamic parameters derived from the WSS vector field were computed to characterize the local flow environment. WSS, the tangential force exerted by flowing blood on the endothelial surface, is a critical regulator of vascular pathobiology.

Time-averaged wall shear stress (TAWSS): Represents the mean magnitude of WSS over the cardiac cycle (T), calculated as:

$$TAWSS = \frac{1}{T} \int_0^T |\mathbf{WSS}(s, t)| dt \quad (1)$$

where $\mathbf{WSS}(s, t)$ is the instantaneous WSS vector at position “s” on the wall surface at time “t.”

Oscillatory shear index (OSI): Quantifies the degree of directional change in the WSS vector throughout the cardiac cycle, ranging from 0 to 0.5. It is calculated as:

$$OSI = \frac{1}{2} \left[1 - \frac{\left| \int_0^T \mathbf{WSS}(s, t) dt \right|}{\int_0^T |\mathbf{WSS}(s, t)| dt} \right] \quad (2)$$

High OSI indicates disturbed flow patterns often associated with endothelial dysfunction and atherosclerotic plaque development.

Relative residence time (RRT): An inverse indicator of TAWSS that incorporates the influence of flow oscillation, reflecting the time near-wall particles are likely to reside in a specific region. It is calculated as:

$$RRT = \frac{1}{(1 - 2 \times OSI) \times TAWSS} \quad (3)$$

High RRT signifies areas of flow stagnation or recirculation, potentially promoting thrombosis and inflammatory cell adhesion.

These WSS-derived metrics are frequently evaluated against established thresholds linked to heightened vascular risk. While detailed spatial analysis is presented in the results, regions potentially exhibiting low TAWSS (<0.4 Pa), high OSI (>0.25), and/or high RRT (>5 1/Pa) are often considered hemodynamically unfavorable based on previous investigations.^{19–21} Additionally, to assess the direct impact on target vessel perfusion, the cycle-averaged volumetric flow rate through each renal artery outlet was quantified from the simulation results. All parameters were computed using the postprocessing modules within COMSOL Multiphysics.

RESULTS

Flow Pattern

Streamline patterns, visualized at $t = 0.3$ s representing the approximate average peak systolic phase across outlets, reveal distinct hemodynamic characteristics associated with renal ostium positioning (Fig. 4). In the post-1, post-2, and post-3 models, which feature embedded renal artery segments running parallel to the main aortic lumen, several common features were observed across all 3 patient cases (1): Elevated flow velocities within the parallel-running renal artery segments. (2) Significant flow disruption, characterized by complex secondary flow patterns and apparent recirculation zones, located on the anterior aspect of the infrarenal aorta immediately distal to the termination point of these parallel segments. (3) Marked streamline deviation in the suprarenal aorta preceding the renal artery inlets, indicative of flow division. In striking contrast, the post-4 configuration, simulating ostia positioning closer to physiological norms without substantial parallel segments, exhibited comparatively ordered flow. In this model, streamline deflection was primarily localized near the branch origins (CA, SMA, and renal), and the infrarenal aorta displayed substantially reduced flow complexity and mixing compared to the other 3 models.

The Flow Ratio of Bilateral Renal Arteries

The influence of renal artery ostium positioning on renal perfusion (right renal artery, RRA; left renal artery, LRA) was evaluated by normalizing the calculated flow rates in the post-2, post-3, and post-4 models relative to the baseline post-1 configuration for each respective artery and case (Fig. 5). A consistent finding across all 3 cases was that the

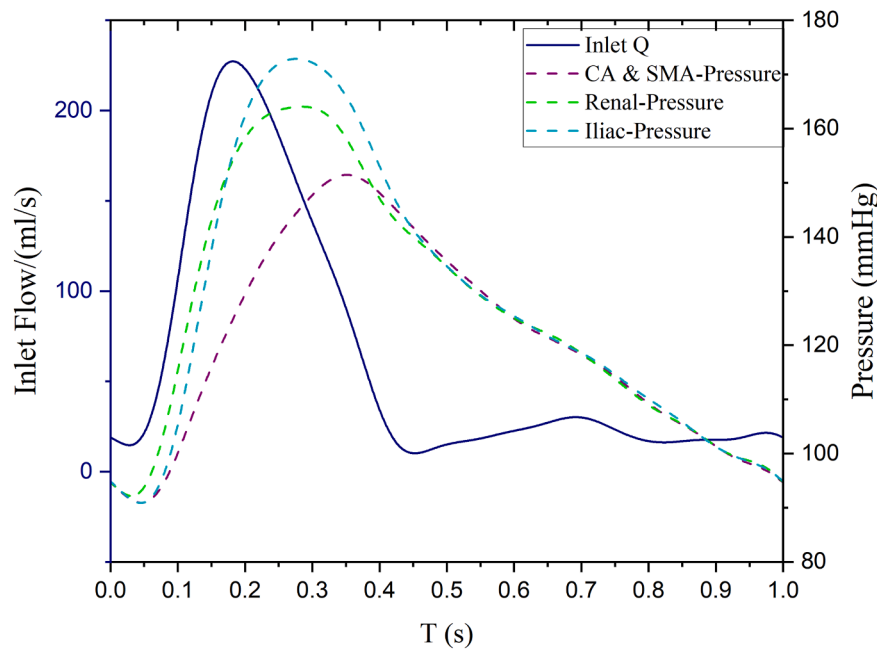


Fig. 3. Boundary conditions applied to the model: flow-based inlet and pressure-based outlet.

post-4 model, simulating ostia positioned closest to their physiological locations, yielded the highest normalized perfusion rates for both RRA and LRA, reaching values up to approximately 1.5-fold (RRA, case 3) and 2.7-fold (LRA, case 1) increase over baseline. Conversely, the post-1 configuration inherently represented the lowest (baseline) perfusion. Analysis of the intermediate configurations revealed that the post-2 model generally resulted in higher normalized perfusion than the post-3 model, although this trend exhibited some variability between cases and arteries (e.g., RRA in case 3 showed slightly higher perfusion in post-3 than post-2).

Distribution Characteristics of WSS Characteristic Values

To elucidate the impact of renal artery ostial positioning on the local hemodynamic environment, WSS-derived metrics (TAWSS, OSI, and RRT) were performed across the 4 simulated configurations (post-1 – post-4) for each patient case.

Time-Averaged Wall Shear Stress. The positioning of renal artery ostia exerted a significant influence on both the spatial distribution and total area of low TAWSS regions (TAWSS <0.4 Pa) (Fig. 6). Spatial maps (Fig. 6A) indicate that distal ostial migration leads to an expansion of low TAWSS areas on the adjacent dorsal aortic wall, coupled with a

corresponding reduction in the infrarenal aorta distal to the parallel segments. Quantitative analysis of the total low TAWSS area, normalized relative to post-1 (Fig. 6B), numerically reflects these competing spatial changes. In cases 1 and 3, the normalized low TAWSS area demonstrated a generally monotonic increase with distal ostial migration, reaching approximately 3.5-fold and 1.8-fold of the respective baseline areas in the post-4 configuration. In contrast, case 2 exhibited nonmonotonic behavior, reaching a minimum normalized area at the post-2 configuration, primarily because the reduction in the infrarenal low TAWSS area surpassed the expansion near the renal ostia for that specific model. To quantify the intercase variability and provide a more robust statistical basis, we have incorporated pooled descriptive statistics (mean $\pm$ SD) for the normalized areas. The mean curve (Fig. 6B) shows a clear monotonic increase in the average area, while the significant increase in the standard deviation at distal positions.

Oscillatory Shear Index. Analysis of the OSI revealed significant alterations in regions indicative of disturbed, oscillatory flow (high OSI) influenced by renal inner-branch ostial positioning (Fig. 7). Spatial distribution maps (Fig. 7A) consistently demonstrated that progressive distal migration of the ostia tended to foster the emergence or expansion of high OSI zones, particularly on the dorsal aortic wall adjacent to the renal inlets.

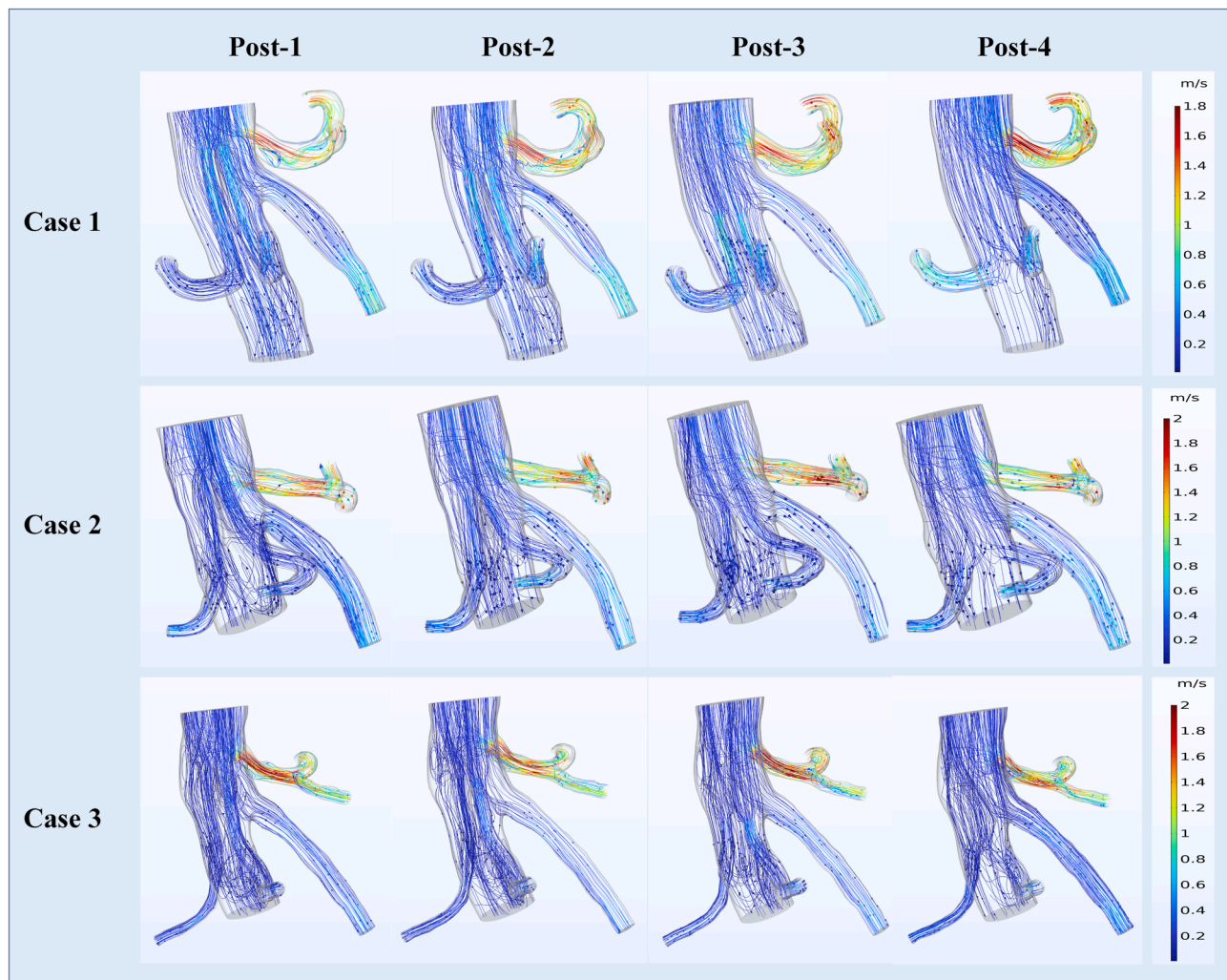


Fig. 4. Aortic flow patterns visualized via streamlines for all patient cases (cases 1–3) and simulated ostial positions (post-1 – post-4). Images depict flow conditions at $t = 0.3$ s (approximate average peak systole).

Concurrently, this distal shift was generally associated with a reduction in high OSI regions within the infrarenal aortic segment that were prominent in the more proximal ostial configurations.

However, in contrast to the trends observed for TAWSS, the quantitative assessment of the total surface area exhibiting high OSI (normalized relative to the post-1 baseline for each case, Fig. 7B) demonstrated notably complex, nonmonotonic, and highly case-specific responses. Specifically, case 1 showed a pronounced peak in normalized area at the post-2 configuration, potentially associated with a transient ventral high OSI region near the ostia, followed by a sharp decline for post-3 and post-4. Case 2 displayed relatively modest fluctuations, predominantly remaining at or below the baseline level

across configurations. In contrast, case 3 exhibited a nadir at post-2 before a notable increase exceeding baseline at post-3, coinciding with prominent dorsal high OSI patches near the inlets, followed by a decrease toward baseline at post-4. The mean curves for OSI show complex variations rather than a clear trend, and their standard deviations are relatively smaller at certain positions than TAWSS.

Relative Residence Time. Analysis of RRT focused on identifying regions potentially prone to prolonged near-wall particle interaction and flow stagnation. Qualitatively, the spatial distribution maps (Fig. 8A) closely resembled those observed for high OSI. Progressive distal ostial migration generally coincided with an expansion of high RRT zones, predominantly located on the dorsal aortic wall

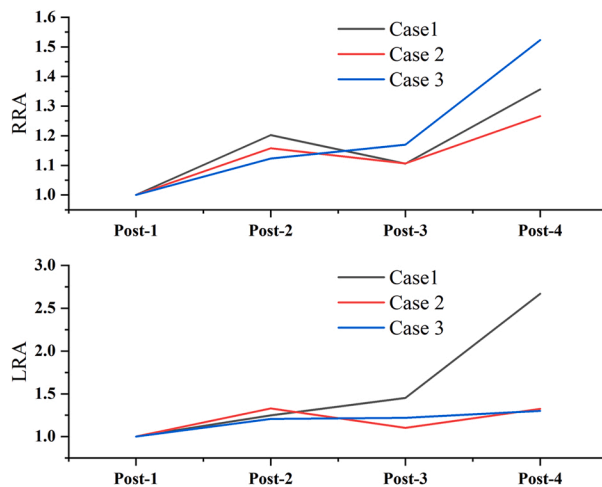


Fig. 5. Blood perfusion variation characteristics of bilateral renal arteries. RRA, right renal artery; LRA, left renal artery

adjacent to the renal inlets. Concurrently, a reduction of high RRT areas was typically observed within the infrarenal aortic segment compared to the more proximal ostial configurations.

However, the quantitative analysis of the total surface area exhibiting high RRT (normalized relative to the post-1 baseline for each case, Fig. 8B) revealed complex, case-dependent, and nonmonotonic variations. While sharing some general characteristics with the OSI trends (Section 3.3.2), these RRT quantitative patterns displayed distinct features. Case 1 showed a peak normalized area at post-2 followed by a decline below baseline. Case 2 exhibited a minimum at post-2 before partly recovering but remaining below baseline. Case 3 showed a minimum at post-2 followed by a peak above baseline at post-3. Notably, the mean curve for RRT follows a similar nonmonotonic pattern of variation as that for OSI.

DISCUSSION

This exploratory study investigated the hemodynamic consequences of varying renal artery ostium positions following endovascular repair of JAAAs repair using the WeFlow-JAAA inner-branch endovascular system. By employing patient-specific CFD, our preliminary findings suggest that a critical interplay may exist between the surgically dictated ostial placement, subsequent renal perfusion, and the complex flow patterns established within the supra-renal and infrarenal aorta. These results help to offer valuable insights into the device-specific

hemodynamics and contribute a quantitative basis to potentially inform the optimization of JEVAR strategies, particularly those employing inner branch or parallel graft designs.

The inherent anatomical complexity of the juxtarenal aorta, involving closely spaced major visceral branches (CA, SMA, and renal), predisposes this region to intricate baseline hemodynamics influenced by geometry, pulsatility, and flow division phenomena like the Coanda effect.^{22–24} While our post-4 model aimed to approximate a physiological branching pattern, even this configuration exhibited notable streamline deflection, underscoring the baseline complexity that endovascular repair must accommodate (Fig. 4). The WeFlow-JAAA device, with its prefabricated inner renal branches, introduces specific structural alterations including potential pathway lengthening and the creation of a parallel segment within the main aortic lumen.^{12,25} This parallel segment inherently alters local flow dynamics, creating zones of acceleration followed by expansion.²⁶

Our simulations illuminate the direct consequences of these structural variations. A key finding was the progressively enhanced renal perfusion observed with more distal ostial positioning, culminating in the highest rates in the post-4 configuration (Fig. 5). This likely reflects a less tortuous and hydrodynamically more efficient pathway compared to the longer, embedded proximal designs (post-1 demonstrating the lowest perfusion), appear to increase flow path resistance, thereby compromising renal blood flow. These are highly consistent with patient-specific computational simulations by Tran et al.²⁷, which demonstrated that branched endografts (bEVAR), characterized by longer pathways and complex ‘U-shaped’ geometric topologies, inherently generate higher flow resistance and a subsequent reduction in perfusion than fenestrated designs. However, this perfusion benefit with distal placement is counterbalanced by distinct alterations in aortic flow patterns. While the post-4 model exhibited relatively organized infrarenal flow, the more proximal configurations (post-1, post-2, and post-3) generated significant flow disturbances distal to the parallel segment termination (Fig. 4). The extent of this distal turbulence appeared inversely related to the parallel segment length, suggesting a complex relationship between segment length, expansion effects, and potential inter-branch flow interference.

From a vasculopathic perspective, the interpretation of WSS-related metrics reveals further complexity and potential trade-offs. In the proximal models (post-1, post-2, and post-3), flow

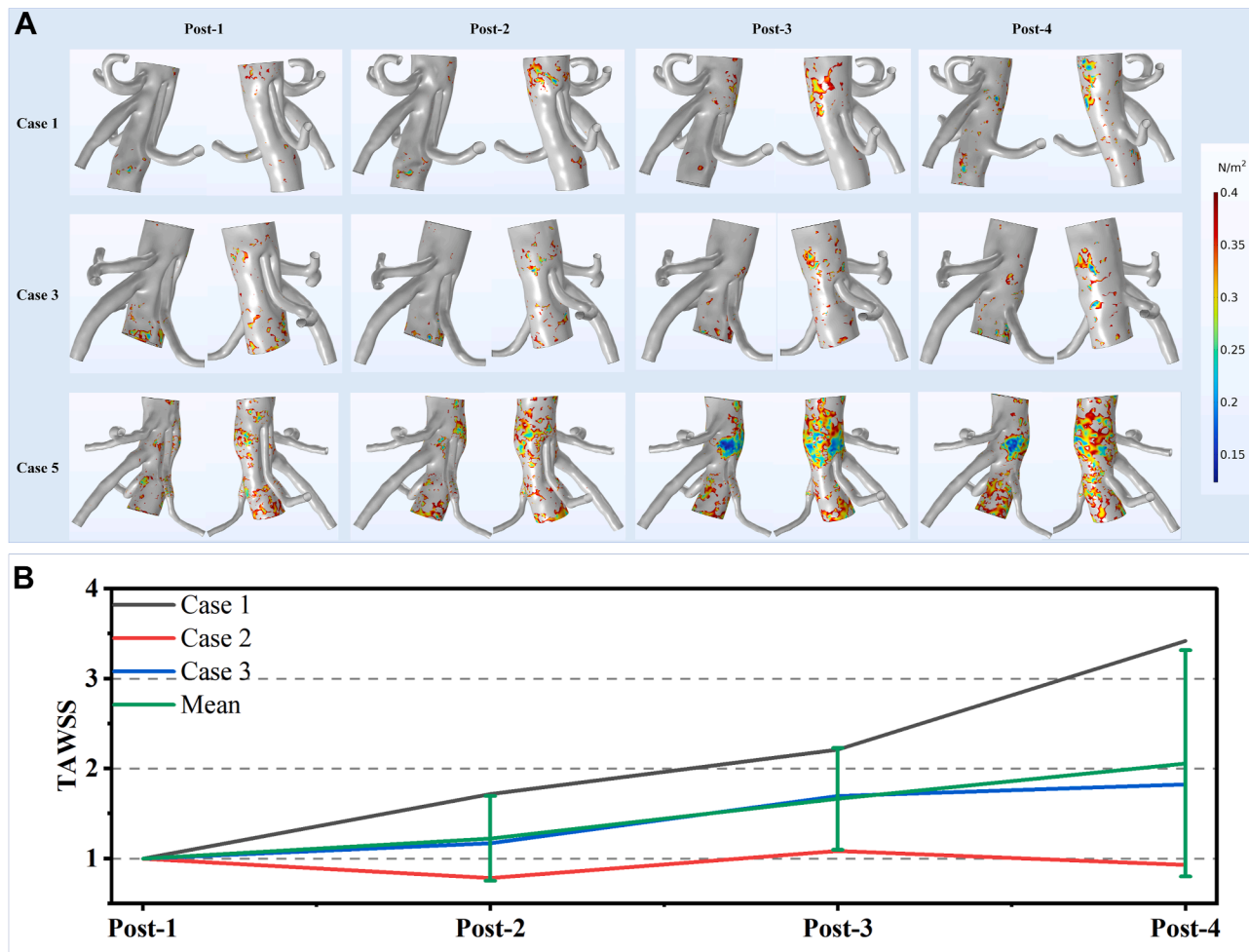


Fig. 6. Characteristics of spatial distribution and area variations in regions where TAWSS < 0.4 Pa. **(A)** Distribution characteristics of regions with TAWSS < 0.4 Pa.

(B) Normalized area variation characteristics of TAWSS < 0.4 Pa regions. TAWSS, time-averaged wall shear stress.

acceleration within the parallel segments might lead to relatively smaller areas of low TAWSS (<0.4 Pa) within those specific segments. However, the situation differs in the post-4 model, it exhibited the largest overall low TAWSS area despite its superior renal perfusion. These areas were concentrated on the aortic wall near the more ‘physiological’ renal artery ostia. This suggests that while parallel grafts altering the primary take-off angle and impingement patterns might locally elevate WSS.

Furthermore, regions of high OSI (>0.25) and high RRT (>5 1/Pa), often considered indicative of flow oscillation/stagnation and correlated with turbulence and vascular pathology,^{19,20,28} showed distinct spatial distributions largely mirroring each other (Figs. 7A, 8A). High OSI/RRT zones were

prominent near the ostia (especially the dorsal region of the abdominal aorta during proximal placement) and within the turbulent infrarenal regions generated by the parallel graft expansion effect in proximal models. Importantly, the quantitative total area analysis for high OSI and RRT (Figs. 7B, 8B) revealed nonmonotonic and case-dependent responses to ostial positioning, often differing from the TAWSS trends (Fig. 6B). This underscores that optimizing 1 WSS parameter may not optimize others, and the net effect on proatherogenic potential is complex. The variability between cases, despite similar positional changes, suggests that subtle differences in local aortic geometry (e.g., curvature, branch angles beyond just axial position) likely play a significant role in modulating these complex flow patterns, meriting further investigation.

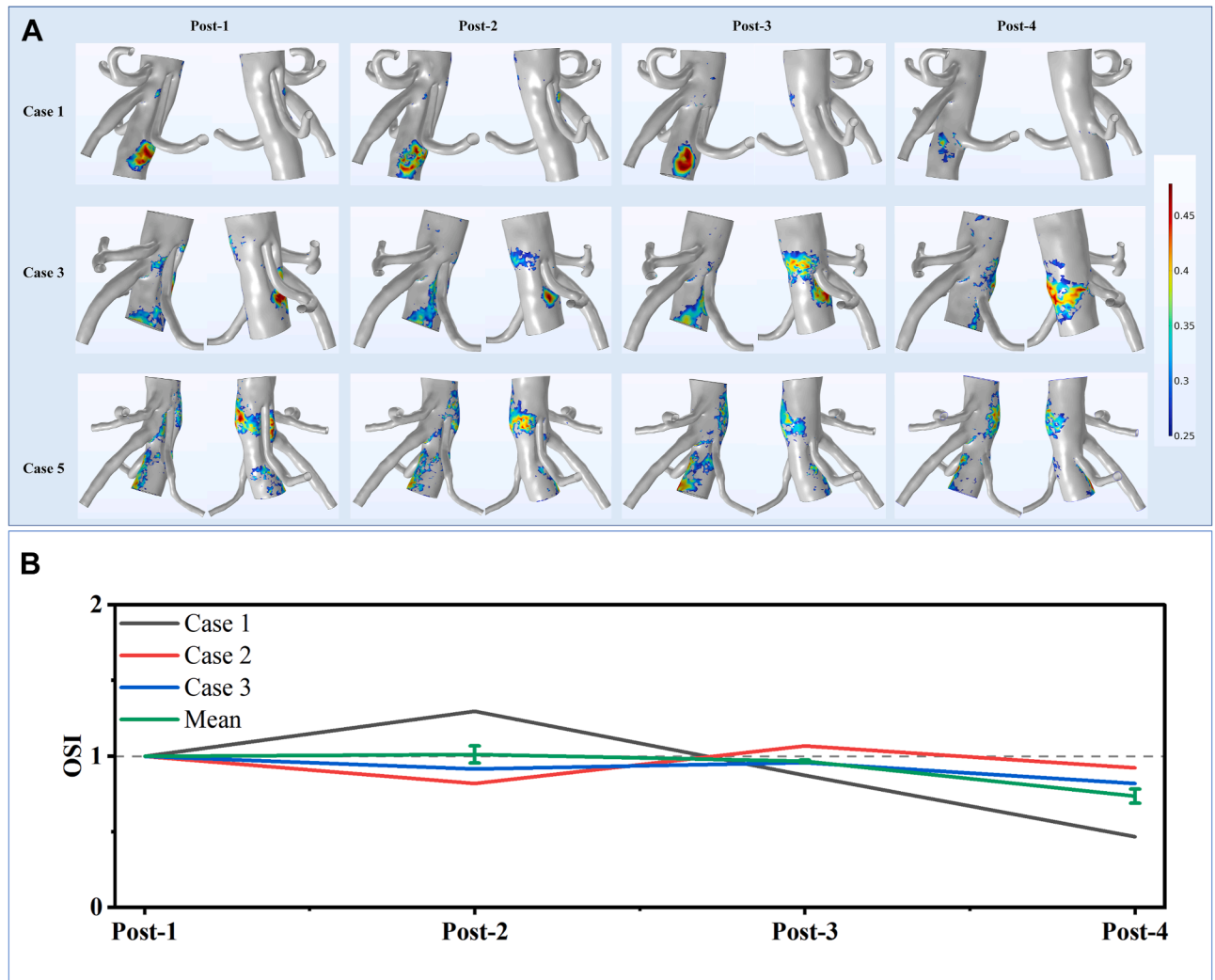


Fig. 7. Characteristics of spatial distribution and area variations in regions where OSI > 0.25. **(A)** Distribution characteristics of regions with OSI > 0.25. **(B)** Normalized

area variation characteristics of OSI > 0.25 regions. OSI, oscillatory shear index.

The clinical implication of these findings is that selecting the renal ostium position during WeFlow-JAAA deployment involves a nuanced weighing of competing risks. Prioritizing maximal renal perfusion might favor a more distal placement, but this could potentially expose the perirenal aortic wall to less favorable long-term WSS conditions. Conversely, a more proximal placement might create a more favorable WSS environment near the ostia but at the expense of reduced renal flow and the induction of significant infrarenal flow disturbances, whose long-term consequences (e.g., impact on distal sealing zones or limb patency) are unknown.

While a recent multicenter study²⁹ reported excellent 30-day technical success (99.1%) and branch patency (98.2%) for the WeFlow-JAAA

system, our findings are complementary to, rather than contradictory with, these clinical outcomes. The marginal reduction in renal perfusion and less favorable WSS conditions identified in our models are likely insufficient to trigger acute occlusion in the short term, which explains the high early clinical patency observed. However, the localized high-risk zones of OSI and RRT identified in our simulations reveal potential subclinical thrombotic risks that may emerge over time. Furthermore, the clinical necessity for frequent intraoperative axial and rotational adjustments to accommodate patient-specific anatomy²⁹ underscores the practical relevance of our work. By elucidating the hemodynamic consequences of these positional adjustments, our findings provide a predictive biomechanical framework

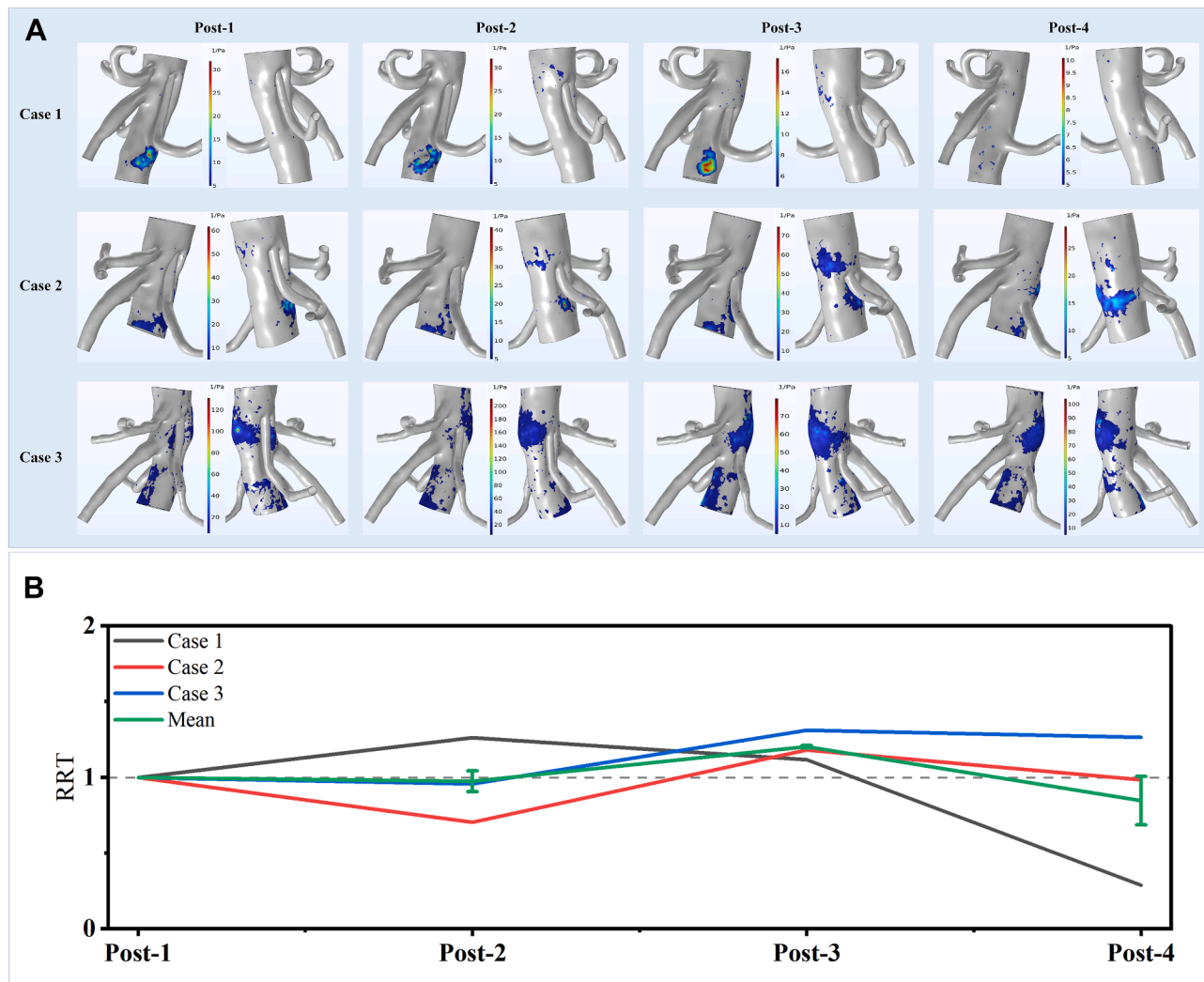


Fig. 8. Characteristics of spatial distribution and area variations in regions where $RRT > 5 \text{ 1/Pa}$. **(A)** Distribution characteristics of regions with $RRT > 5 \text{ 1/Pa}$.

(B) Normalized area variation characteristics of $RRT > 5 \text{ 1/Pa}$ regions. RRT, relative residence time.

for evaluating long-term device durability. Ultimately, the optimal strategy may therefore not be universal but rather patient-specific, requiring consideration of factors like baseline renal function, aortic morphology, cardiovascular risk factors, and life expectancy.

Limitations

While this study provides valuable insights, its interpretation is subject to certain limitations that warrant careful consideration. The most significant constraint is the small sample size ($N = 3$). This inherently limits the generalizability of our findings, precludes robust statistical analysis, and means the observed trends,

particularly the nonmonotonic WSS responses, must be considered preliminary. Future research must involve large-scale patient cohorts to definitively confirm these preliminary observations and comprehensively evaluate interpatient variability. A second key limitation is the absence of direct experimental validation for our CFD model. While the model was built using established methods, the lack of in vitro or in vivo data means the quantitative accuracy of the hemodynamic parameters, such as WSS and renal perfusion rates, could not be definitively confirmed. This restricts our ability to fully adhere to established validation and verification guidelines and should be considered when interpreting the quantitative hemodynamic characteristics presented.

Methodologically, the assumption of rigid vessel walls omits the damping effects of arterial compliance and potential fluid-structure interactions, which could influence wave propagation, flow separation dynamics, and the precise magnitude and distribution of WSS, especially during the pulsatile cycle. Similarly, modeling blood as a Newtonian fluid neglects its non-Newtonian behavior at physiological extremes of shear rate, potentially influencing viscosity estimations in specific flow regions (e.g., recirculation zones) and thus affecting calculated WSS values. Finally, the application of identical outlet pressure boundary conditions across all models simplifies the complex downstream vascular impedance. Actually, patient-specific or adaptive outlet conditions responding to proximal geometric changes could subtly influence flow distribution and renal perfusion results differently than predicted here. These modeling assumptions, common in CFD studies for tractability, should be considered when interpreting the quantitative hemodynamic characteristics presented.

CONCLUSION

This computational study suggests a crucial design consideration in JEVAR involving inner branches like the WeFlow-JAAA system: the positioning of the renal ostium appears to create a significant hemodynamic trade-off. Our findings indicate that maximizing renal perfusion, generally favored by more distal ostial placement, may paradoxically shift potentially adverse WSS conditions (low TAWSS, high OSI/RRT) toward the perirenal aortic segment. The complex, nonlinear responses of OSI and RRT further suggest that a simple strategy of maximizing distal placement might not yield the universally optimal hemodynamic profile. These findings, though derived from a limited number of cases, provide a mechanistic framework for understanding potential long-term outcomes associated with different WeFlow-JAAA placements. They tend to suggest that achieving an “optimal” hemodynamic state balancing renal health and aortic wall stability likely requires patient-specific planning rather than a one-size-fits-all approach. This may necessitate a nuanced consideration of an intermediate position to balance both perfusion deficits and adverse flow patterns inherent to current inner-branch designs.

CREDIT AUTHORSHIP CONTRIBUTION STATEMENT

Fen Li: Writing — original draft. **Weiham Lu:** Data curation. **Wenhui Xue:** Data curation. **Zhuoxin**

Deng: Data curation. **Yawei Zhao:** Data curation. **Lingfeng Chen:** Writing — review & editing. **Hui Song:** Conceptualization. **Hongpeng Zhang:** Supervision. **Haiping Tian:** Writing — review & editing.

DATA AVAILABILITY

The authors declare that the data supporting the findings of this study are available within the paper. Raw data files in other formats are available from the corresponding author upon reasonable request.

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