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REVIEW

Geometrical Factors Influencing the Hemodynamic Behavior of the AAA Stent Grafts: Essentials for the Clinician

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Abstract Endovascular aneurysm repair (EVAR) is considered to be the treatment of choice for abdominal aortic aneurysms (AAA). Despite the initial technical success, EVAR is amenable to early and late complications, among which the migration of the endograft (EG) with subsequent proximal endoleak (Type Ia) leads to re-pressurization of the AAA sac, exposure to excessive wall stress, and, hence, to potential rupture. This article discusses the influence that certain geometrical factors, such as neck angulation, iliac bifurcation, EG curvature, neck-to-iliac diameter, and length ratios, as well as iliac limbs configuration can exert on the hemodynamic behavior of the EGs. The information provided could help both clinicians and EG manufacturers towards further development and improvement of EG designs and better operational planning.

Keywords Endovascular graft · Migration · Displacement forces · Geometry · Abdominal aortic aneurysm · Biomechanics

Introduction

Endovascular aneurysm repair (EVAR) is considered to be the preferred treatment modality for abdominal aortic aneurysms (AAA) for over a decade. EVAR aims at the positioning of an endovascular graft (EVG) within the AAA sac using an over-the-wire technique either via surgical exposure of both common femoral arteries or totally percutaneously. Reduced perioperative morbidity and mortality comprise the major advantage of this minimally invasive technique compared with conventional open repair [1, 2]. However, despite the initial technical success and early discharge of the patient, this technique is amenable to early and late complications, such as EVG migration, endoleaks, and material failure, which may lead to reexposure of the AAA wall to pulsatile blood pressure and excessive wall stress and, thus, to potential rupture.

This short review discusses the geometrical factors that affect the hemodynamic behavior of the EVGs, providing both clinicians and EVG manufacturers with useful information for better planning of endovascular techniques and developing of EVG designs. In order to facilitate the understanding of the importance of geometric parameters,

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experimental data from our research group are displayed along with their potential clinical applications.

Geometrical Factors Affecting EVG Migration

According to the EUROSTAR study, 3.5 % of patients undergoing EVAR suffer from graft migration comprising a ≥5-mm movement of the stent-graft from its initial deployment site [3, 4]. Displacement of the EVG can cause loss of proper sealing at its proximal or distal landing zone (i.e., the aortic neck and the iliac sites, respectively), leading to Type Ia and Ib endoleaks, respectively and restoring the systemic pressure loading on the AAA sac, thus regenerating the risk of AAA rupture despite the preceded endovascular therapy. Many researchers have investigated the major determinants of EVG migration, depicting an association between the aortic-EVG geometric features and the displacement forces acting on the latter.

Modern computational analytical/numerical methods can estimate the displacement forces, the magnitude of which is strongly influenced by the diameter and angulation of the EVG neck or the angulation of iliac bifurcation [5–10]. The computational sequence requires fundamentally three distinct work steps: (i) geometry reconstruction of the study model from medical images, (ii) biomechanical simulation (Finite-Element or Fluid–Structure Interaction computation), and (iii) interpretation of biomechanical properties [8]. Molony et al. [11] used fluid–structure interaction (FSI) simulation for a group of patient-specific 3D reconstructed EVGs to show that the antero-posterior neck angulation and the large inlet-to-outlet area ratios were the greatest determinants of the magnitude of drag forces. Furthermore, the curvature of the iliac limbs can also create additional sideways forces that may predispose to displacement of the iliac limbs and peripheral endoleaks (Type Ib) [8]. Interestingly, the greatest part of the displacement forces seems to act on the EVG bifurcation site compared to the proximal (neck) and distal (iliac) counterparts [10].

It should be delineated that endovascular treatment of AAA is not an instant fixation of an EVG in the aortic lumen, but rather an ongoing process of conformational changes in the aortic endograft during the post-EVAR shrinkage process of the sac, changing its geometry and, thus, putting increased axial strain and altering the distribution stress patterns on the different components of an EVG [12–15]. Therefore, the importance of studying the aortic and EVGs geometry is not limited solely to preoperative proper size planning of the endovascular device but also may have predictive role on the hemodynamic behavior and the resultant adverse effects related to the EVG; furthermore, the information drawn from such studies may effectively modify the design of newer devices to improve their accommodation on challenging AAA anatomies, thus improving their hemodynamic performance. The significance of each geometric parameter is summarized in Table 1.

Table 1 The influence of geometric factors of endografts on displacement forces

	Total forces	Forces on iliacs	Forces on bifurcation
Inlet (neck) diameter	+++	++	++
Neck angulation	++*	+	+
Endograft curvature	++	++	++
Iliac limbs bifurcation angle	+	+	+
Inlet to outlet diameter ratio	+++	++	++
Main body-limbs length ratio	+	+++	++

Evaluation of the relative influence of the geometric factors on the magnitude of displacement forces, from the lowest (+) to the highest (+++) influence, as displayed in the manuscript, based on data from the literature

* Neck angulation >60°

Diameter of the Neck

The significant influence of the inlet-to-outlet (i.e., neck-to-iliac limbs, $d1/d2$) diameter ratio has been recently studied by our study group and is depicted in Fig. 1. We estimated the maximum displacement forces over a cardiac cycle on two reconstructed EG models in the standard (BifG, Fig. 1A) or the cross-limbs (BalG, Fig. 1B) fashion using FSI with the validated ANSYS software (ANSYS version 12.1; Ansys Inc.). The calculations took place for two different inlet diameters, i.e., 36 and 24 mm, corresponding to the maximum and minimum inlet diameters of the commercially available aortic endografts respectively, with $d1/d2$ ranging from 1.5 to 3.0. The increase in $d1/d2$ caused a constant increase of the maximum total displacement force ranging from 2.6 to 14 and 1.2 to 7.1 N for $d1$ of 36 and 24 mm, respectively (Fig. 2A, B).

Clinical Relevance

The aforementioned findings come in accordance with previous studies, underscoring the role of inlet diameter as a major determinant of the hemodynamic behavior of an EVG, because their effect is coupled with an exponential raise of the displacement forces acting on the endograft. Compared with other geometrical parameters, such as the AAA neck angle, the endograft curvature, or angulation of the limb bifurcation, the neck diameter causes the most profound effect on the magnitude of displacement forces [12, 16]. Interestingly, these seem to confirm clinical observations regarding AVG migration and loss of adequate sealing, because proximal EVG fixation failure seems to be determined to a greater degree by an aortic

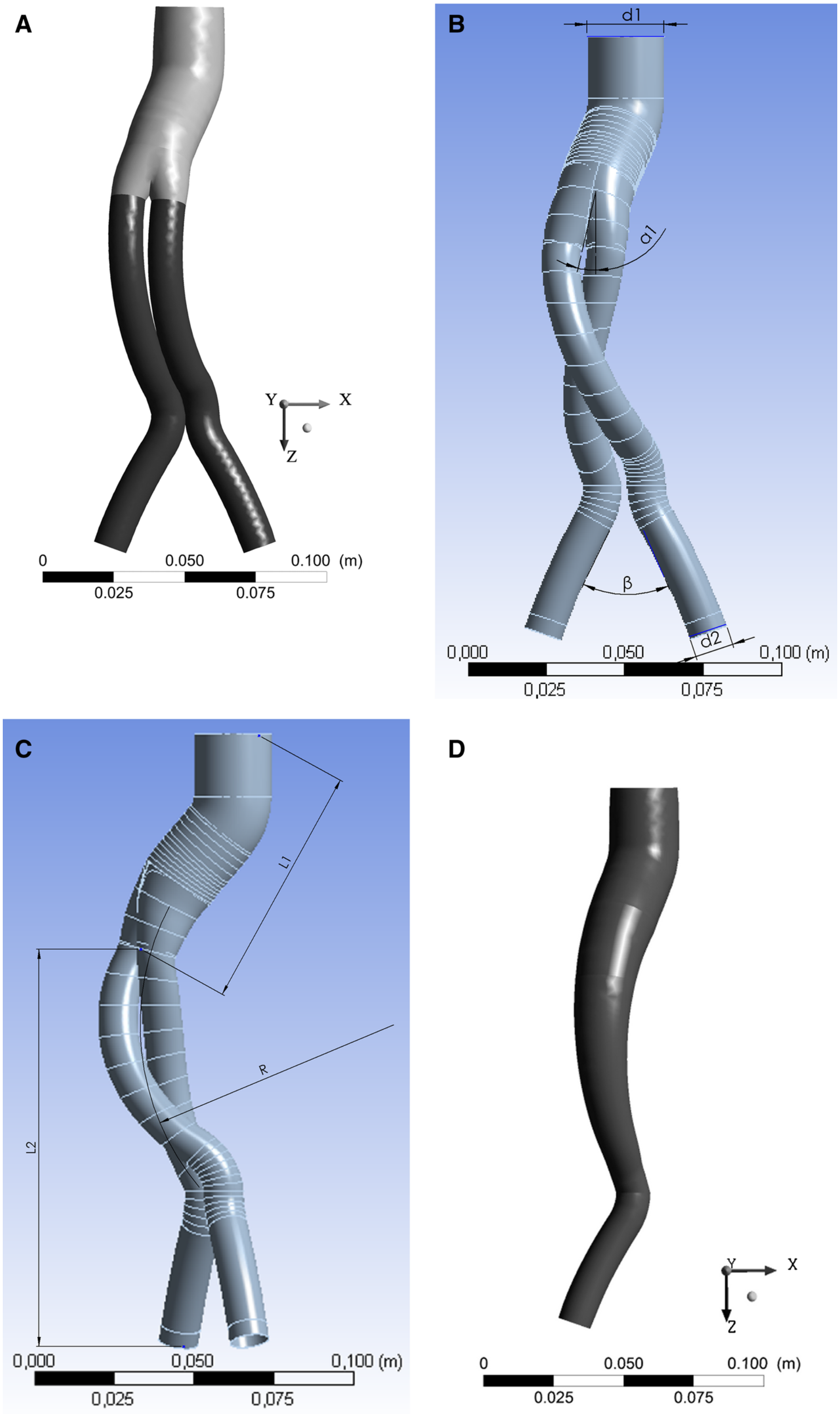

Fig. 1 Reconstructed models used to computational fluid dynamics. **A** Customary bifurcated model (BifG). **B**, **C** Cross-limbs model (BalG). **D** Aortouniiliac model (UniG). The a_1, β, $d1$, and $d2$ in **B** represent the lateral neck angulation, the angle of endograft limb bifurcation, and the inlet and outlet diameter, respectively. The main body length up to the flow divider and the iliac limb length are depicted as $L1$ and $L2$, respectively (**C**)

neck dilatation exceeding EVG oversizing (i.e., increase in *d*1) rather than simple migration distance, according to Litwinski et al. [17]. The importance of the (in)sufficient AAA neck length also is questioned by Hager et al. [18], who compared clinical performance of EVG with supra- and infrarenal fixation in short but straight proximal necks and reported equal and reported freedom intervals from early and late type 1a endoleaks.

Taking into consideration that a cut-off value of AAA neck diameter >28 mm is considered to represent a high

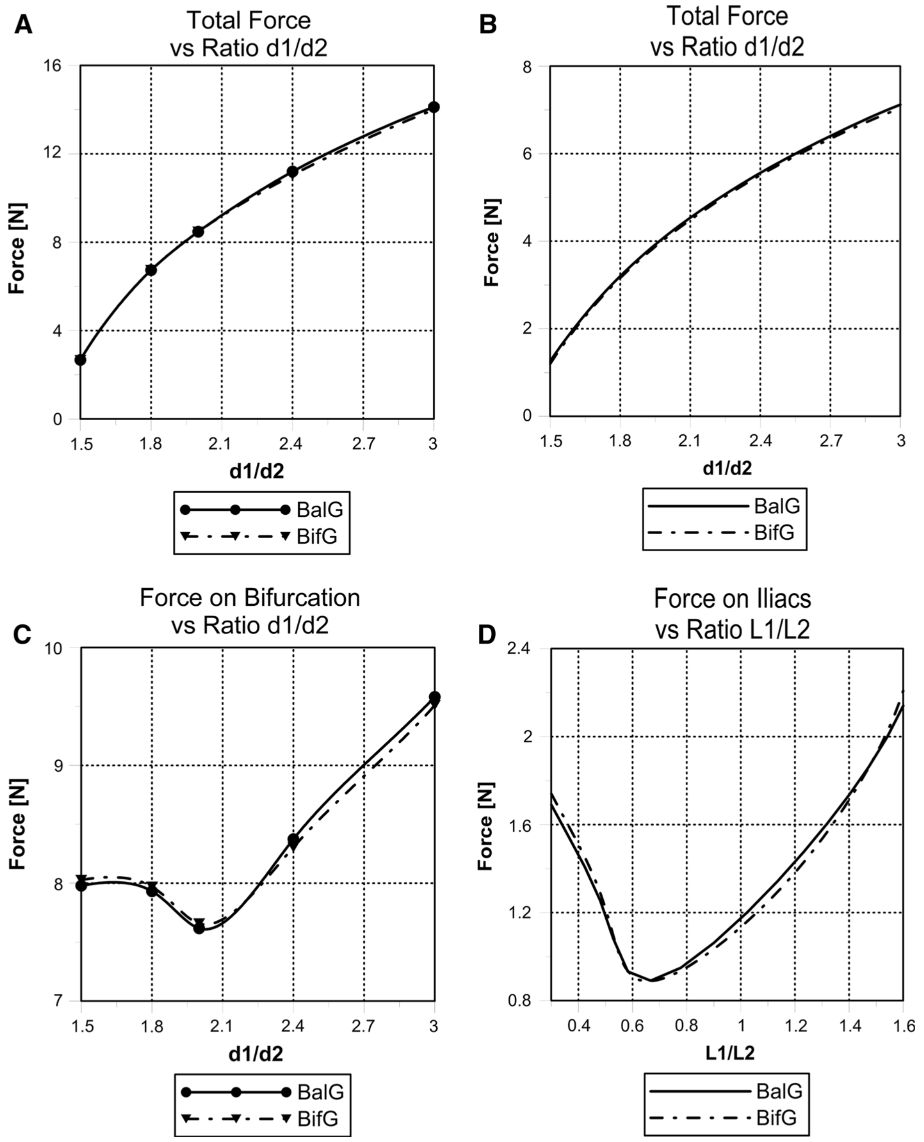

Fig. 2 Increase in inlet-to-outlet diameter ratio of an EVG is associated with an increase of the magnitude of total displacement forces (**A**, **B**). Ratio >2 predisposes to higher mechanical loading at the bifurcation site (**C**), whereas high or low main body-to-iliac length ratios enhance the instability of iliac limbs, creating greater forces exerted at this segment (**D**)

risk for EVAR according to the Society for Vascular Surgery/American Association for Vascular Surgery (SVS/AAVS) [19], many researchers conducted studies comparing the clinical performance of EVGs in the treatment of AAA with large and smaller neck diameters. Jim et al. [20] reported a higher rate of major adverse effects within the first year and higher migration rates at 5 years when treating AAAs with large (>28 mm) versus smaller (<28 mm) using the Talent device (a bimodular device with suprarenal fixation), whereas Stanley et al. [21] reported a migration incidence of 4.2 % in a series of 238 AAA treated with the Zenith device, identifying a neck diameter >28 mm ($p = 0.0024$) as the sole determinant of this complication. Generally, an oversizing of the EVG central fixation segment by 10–25 % is suggested to ensure an adequate radial force and proper sealing in AAA necks of >28 mm, providing an acceptable low migration rate and incidence of proximal endoleak (Type Ia) [22]. On the other hand, the continuous radial force exerted by self-expanding stent-grafts has been associated with progressive dilatation of the aortic neck postoperatively, predisposing to generation of higher displacement forces to migration (i.e., increase of $d1/d2$) and loss of proximal sealing [23–26]. Therefore, newer EVG aiming at proximal sealing with alternative modes, such as polymer-filled sealing rings merit greater attention [27–29].

Role of Diameter of Iliac Vessels

While clinical studies focus mainly on the geometrical characteristics of the AAA and endograft neck diameter, a computational evaluation of the influence of inlet diameter

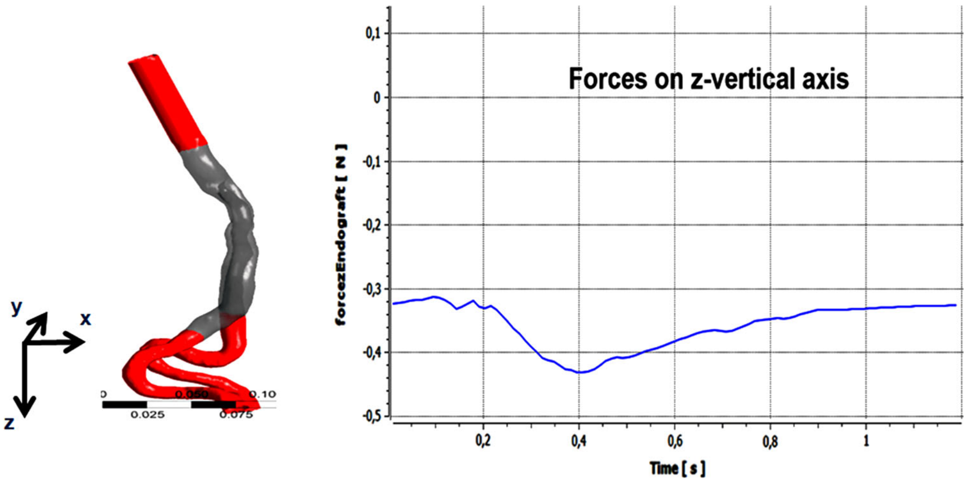

Fig. 3 Computational reconstruction of an endograft model (*grey color*) used to treat an AAA with iliacs (*red color*) of excessive tortuosity. Because the vertical *z*-axis is headed caudally, the estimated negative values of displacement forces (shown in diagram) corresponding to tendency for upward instability can be attributed to excessive iliac tortuosity

on the hemodynamic performance of an EVG unveils an important role for the iliac (outlet) diameters, as well. The computational estimation of the displacement forces acting on the bifurcation of an EVG with large (36 mm) neck diameter exemplifies this (Fig. 2C). As can be seen, an inlet-to-outlet diameter ratio of 3 (i.e., iliac arteries of 12 mm) renders a higher bifurcation force compared with an AAA of the same inlet diameter but with ectatic iliacs of 18 mm (ratio of 2). In such instances, adapting an accommodation mode of a main body actually sitting on the aortic bifurcation, such as the Powerlink XL EVG system [30] or of a long mainbody available with the Cook Zenith AAA EVG [31] renders the AAA-EVG conjugation theoretically more stable.

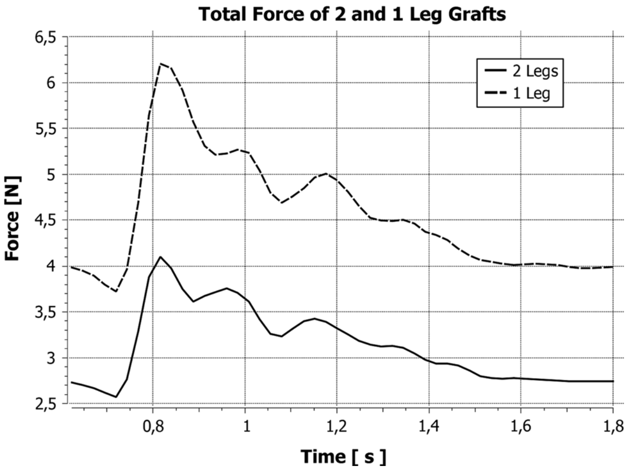

Fig. 4 Aortouniiliac endograft configuration (*dotted line*) shows predisposition for higher displacement forces compared to the bifurcated accommodation (*solid line*)

Investigating the Role of EVG Limb Length

The iliac diameter can affect the displacement forces and the stability of the peripheral fixation with respect to endoleaks Type Ib. A low bifurcation, as determined by the presence of long EVG main body accommodating at the aortoiliac bifurcation and/or short iliac limbs, renders the EG less prone to proximal migration [32]. Indeed, Benharash et al. [33] and Heikkinen et al. [34] underscored the migration-preventive role of long iliac fixation, especially in cases of suboptimal or inadequate proximal fixation. This finding also was supported by Waasdorp et al. [35], who suggested that the shorter the proximal fixation, the longer the iliac fixation has to be to prevent migration. In other words, an EVG of long proximal and short iliac fixation could bear the same migration risk compared to an EG of short proximal and long iliac fixation. The aforementioned indicate that the hemodynamic effect of the relative limb length with respect to the EVG main body should not be considered negligible. While the relative lengths influence the magnitude of the displacement forces at different EVG parts, the stabilization provided by the iliac fixation lengths contributes to resistance to migration.

Our computational example suggests that a high ratio of main body-to-iliac limb length (i.e., either a long main body or short limbs, both coinciding with a low bifurcation) favors hemodynamically EVGs with low bifurcation, such as the AFX® stent-graft (Endologix Inc., Irvine, CA, USA), the COOK Zenith EVG (available in five different lengths), or the Treovance® (Bolton Medical Inc., Sunrise, FL, USA) [36] but may theoretically attenuate the iliac limb stability (Fig. 2D). Indeed, while our model shows the beneficial role of an iliac-limb length twice as long the main-body, it seems that higher or lower length ratios beyond that point lead to increased forces along the iliac limbs, thereby reflecting a higher predisposition for the development of Type III endoleaks, due to modular

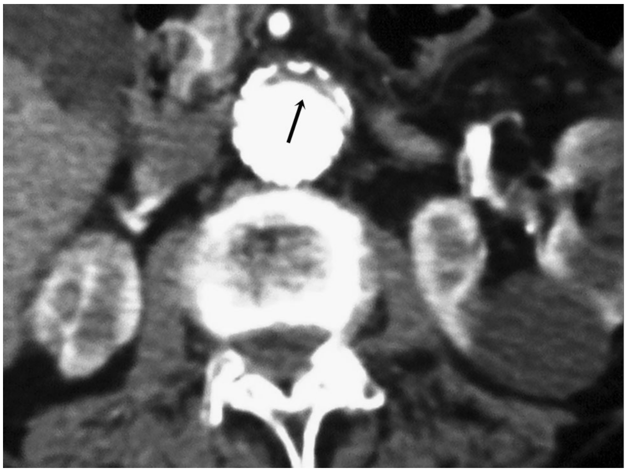

Fig. 5 Thrombus apposition (*arrow*) detected at the anterior surface of the main body of an endograft used for treatment of an AAA

disconnection. To counteract this problem when the patient's individualized geometry leads to an endograft configuration as previously described, certain mechanisms have been evolved to enhance limb's stability, such as the unique Lock-stent mechanism of five rounded bards for fixation within stent modules [37].

Influence of Iliac Configuration

Unfavourable iliac geometry comprises one of the commonest reasons that render AAA unsuitable for EVAR, with extreme iliac tortuosity accounting for 10 % of the exclusion criteria [38]. While iliac angulation and tortuosity have been implicated in endograft limb kinking and thrombosis [39, 40], it is worth mentioning that excessive iliac tortuosity (Fig. 3) may generate hemodynamic forces with a cephalad direction leading to upward migration of the endograft. Although EVG upward migration constitutes a very rare entity with only few cases reported in the immediate peri- or postoperative period, attributed either to surgical manipulating errors or material failure [41–43], there seems that iliac geometry may predispose to such events in the long-term, as demonstrated above. In routine practice, this tendency is counteracted by various proximal fixation mechanisms (hooks, pins, and barbs) available in newer generation endografts [44, 45].

Furthermore, one also should bear in mind that the upward movement of an EVG associated with angulation/tortuosity of the device's iliac EVG limbs either following the native iliac anatomy or secondarily caused by aneurysm sac shrinkage can lead to disconnection of the EVG's components and consequent endoleak Type III [46]. This complication occurs usually in the third to the sixth postoperative year [47].

Comparison of the Bifurcated EVG with the "Crossed-Limbs" and the Aortouniiliac Configuration

While the customary bifurcated EVG configuration lies in the center of both clinical and computational studies, clinicians may occasionally come along challenging anatomies or clinical circumstances where the successful sealing of the AAA is accomplished by adapting the cross-limb accommodation of the EVG or using an aortouniiliac configuration [48, 49]. A comparison of the displacement forces between the cross-limbs and customary bifurcated EVG configurations was recently performed [16, 50–52]. It has been shown recently that these configurations sustain similar displacement forces, irrespective of any variability in the EVG curvature, in the angulation or in the relative lengths and diameters of the proximal and distal EVG segments (Fig. 2A–D). Moreover, the similar oscillatory shear index pattern (expressing the shear stress vector deflection from blood flow's predominant direction during a cardiac cycle), expressed in both endograft configurations, suggest that thrombosis may occur similarly between the two configurations [52]. Indeed, the only-to-date clinical study comparing the clinical performance between the two configurations in terms 12 and 36 months freedom from of migrations, any type of endoleak and need for reintervention and limb thrombosis yielded similar clinical outcomes, with no statistical significance [53].

The use of aortomonoiliac EVG facilitates the management of AAA in cases of narrow terminal aorta, tortuous, kinked, small, calcified, or occluded contralateral iliac artery, emergent treatment of ruptured AAAs or treatment of endoleaks of previously implanted endoprostheses [45, 54, 55]. Therefore, limited data exist to compare directly the performance and hemodynamic profile of aortomonoiliac (Fig. 1D) versus bifurcated EVG (Fig. 1A). Our laboratory results suggest higher displacement forces over the entire EVG or, specifically, at the iliac area (Fig. 4) for the aortouniiliac configuration, predisposing to higher migration rates. Therefore, this mode of AAA treatment should be reserved only in selected cases with the aforementioned indications, rather than as an alternative to bifurcated endografts.

Thrombus Formation in Aortic Endografts

A frequent observation considering the endovascular repair of an AAA is the deposition of thrombus inside the stent-graft lumen (Fig. 5). Several studies have reported intragraft mural thrombus formation starting from the first month after the operation up to nearly 5 months [56, 57]. Mestres et al. [56] estimated the postoperative progression of intragraft thrombus using CT angiography and concluded that the presence of thrombus in the native aorta and the presence of the aortouniiliac configuration were independent predictive factors for the progression of EVG mural thrombus. Additional evidence was provided by Wegener et al. [57], who noted that more than one-fifth of patients developed intraluminal deposits of thrombotic material identifying; however, no potential risk factors associated with thrombus formation. Noteworthy, there was evidence that some of the thrombotic depositions resolved completely without any specific therapy and interestingly enough in these patients no episode of thromboembolism was noticed. Consistent with previous reports, intragraft thrombus deposition after postimplantation of endograft in EVAR patients was observed by Wu et al. [58]. who further reported that the incidence of intraprosthetic thrombosis increased in endografts with longer mainbody and in those with larger mainbody diameter compared with the iliac graft diameter (i.e., $d1/d2$), whereas no correlation was found between the preoperative presence of thrombus or the postoperative antiplatelet/anticoagulant treatment and the deposition of thrombus in the stent-graft. Finally, investigating the flow patterns in a bifurcated stent-graft deployed in a AAA model, Chong et al. reported that the geometry of the arterial vessel and the configuration of the stentgraft could have an impact on the formation of thrombus, with aortic neck angulation, iliac tortuosity, and configuration of the endograft identified as important parameters for the deposition of intra-stent thrombus [8, 59, 60].

Flow Patterns in AAA

Local geometric factors play a role in the determination of velocity values and flow patterns (recirculating zones, flow separation, skewed flow) [60, 61]. Chong et al. [60] described the flow patterns in several regions of a bifurcated, nonplanar stented-AAA model under pulsatile blood flow throughout the cardiac cycle and found a region of flow separation and recirculation at the anterior wall of the proximal stent, which increases with increasing angle and is most predominant during the diastolic phase. As the anteroposterior neck angle increases, the flow patterns present greater asymmetry with flow separation and recirculation zones at the posterior region of the graft main trunk, while most of the flow diverting towards the anterior wall. These phenomena occur mainly in the late deceleration and early systolic phase. Additionally, the flow patterns in the two EVG iliac limbs present quite similar, with skewing of the flow during the presystolic acceleration/peak systolic phase and with subsequent flow separation at the outer wall surface of the EVG limbs. Frauenfelder et al. [62] demonstrated a reduction of turbulence after placement of a stent-graft, with equal blood flow volume through both the stented iliac arteries coupled with a reduction of wall pressure and wall shear stress. They also concluded that high shear stress values develop at the junction site between the stump (iliac gate) of the main body and the contralateral limb (docking area), predisposing to type III endoleak, as also supported by Juchems et al. [63] and Kramer et al. [15].

Clilical Application of Hemodynamic Principles: from Theory to Practice

Computational simulation provides useful data to understand basic pathophysiological aspects and delineate the behavior of AAA and EVG. The demonstration presented above show that the combination of geometrical aspects provides more information to predict the postimplantation EVG performance rather than single geometric features. The advantages and disadvantages associated with certain conformational features of various EVGs show that there is no ideal EVG that serves better than others the purpose of AAA endovascular treatment; rather, each AAA possesses a unique anatomy in which some EVGs accommodate better than others, so that no single EVG pattern emerges as the best [64].

Accordingly, identification of certain AAA geometrical challenges, such as iliac tortuosity and severe angulation or large diameter of the neck, led to development of enhanced suprarenal fixation modes, supported iliac legs and modified stent designs. A recent study based on Finite Element Analysis estimated the stresses on different designs of iliac limb stents, showing differences between them, with spiral and circular stents providing greater flexibility and lower stress values than Z-stents [65]. This comes in accordance with the hybrid concept of combining components (main body, limb extensions) from different endografts, not only in emergent but also in selective clinical setting, treating challenging anatomies with components of different mechanical properties [65–67].

Finally, computational data showed that the displacement forces acting on an EVG are directed sideways rather than downward [11, 13, 14]. These were confirmed by clinical studies that identified a frequent late occurrence of postimplantation EVG sideway movements (27–35 % Rafii et al. [68] and Waasdorp et al. [69], respectively) [68–71].

Although interesting as computational simulations may seem, it should be stressed out that reproducibility and comparison between results of different studies should be cautiously approached, because these models carry certain limitations associated with pressure and flow parameters (boundary conditions) applied to these models; therefore, detailed information about preconditions and model assumptions should always be provided. Admittedly, future clinical studies are needed to validate clinically of computational results and expand further the practical applications of the latter.

In conclusion, this article discusses the influence that certain geometrical factors can exert on the hemodynamic behavior of the EVGs. No EVG design emerges as the best; rather, every AAA has a unique anatomy served better by some EVG than others and vice versa. The information derived seems to be in accordance with clinical observations and comprises a useful adjunct for both clinicians and manufacturers to further development and improvement of EVG designs and better operational planning.

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Conflict of interest Efstratios Georgakarakos, Christos Argyriou, Nikolaos Schoretsanitis, Chris V. Ioannou, Nikolaos Kontopodis, Robert Morgan declare that they have no conflict of interest. Dimitrios Tsetis; Dr. Tsetis gives lectures for Trivascular outside the scope of this article.

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