Comparison of Blood Flow in Branched and Fenestrated Stent-grafts for Endovascular Repair of Abdominal Aortic Aneurysms

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ABSTRACT

Purpose

Reliability of antegrade and retrograde stent-grafts, collectively known as branched stent-grafts (BSG), has not been appropriately analysed. This computational study assesses the haemodynamic outcomes of BSG for different anatomical variations.

Methods

Ideal models of BSG and fenestrated stent-grafts (FSG) were constructed with different visceral take-off angles (TOA) and lateral aortic neck angle. TOA was defined as the angle between the centrelines of the main stent-graft and side branch, with 90° representing normal alignment while 30° and 120° representing angulated side branches. Computational simulations were performed by solving the conservation equations governing the blood flow under physiologically realistic conditions.

Results

Largest renal flow recirculation zones (FRZ) were observed in FSG (at TOA of 30°) and the smallest FRZ was also found in FSG (at TOA of 120°). For straight neck stent-grafts with TOA of 90° , mean flow in each renal artery was 0.54, 0.46 and 0.62 L/min in antegrade BSG, retrograde BSG and FSG, respectively. For angulated stent-grafts, the corresponding values were 0.53, 0.48 and 0.63 L/min. All straight neck stent-grafts experienced equal cycle-averaged displacement forces of 1.25, 1.69 and 1.95 N at TOA of 30° , 90° and 120° , respectively. Angulated stent-grafts experienced equal cycle-averaged displacement force of 3.6 N.

23 **Conclusions**

24 Blood flow rate in renal arteries depends on the configuration of stent-graft with FSG giving
25 maximum and retrograde BSG resulting in minimum renal flow. Nevertheless, the difference
26 is small, up to 0.09 L/min. Displacement forces exerted on stent-grafts is very sensitive to
27 lateral neck angle, but not on the configuration of stent-graft.

28 **Key words:** Antegrade stent-grafts, retrograde stent-grafts, fenestrated stent-grafts,
29 haemodynamics, renal artery blood flow, displacement force, Computational fluid dynamics.

INTRODUCTION

Since its introduction in 1991 by Dr Juan Parodi¹, endovascular aneurysm repair (EVAR) has been continuously evolving throughout. Although EVAR is minimally invasive and is associated with lower mortality rates as compared to conventional open surgical repair (OSR), certain groups of patients remain unsuitable for EVAR due to unfavourable abdominal aortic aneurysm (AAA) morphology such as short neck, conical and/or angulated aortic neck, calcified aortic wall and atheroma depositions. Fenestrated and branched stent-grafts were introduced to overcome some known anatomical limitations of standard EVAR.

Fenestrated stent-graft (FSG) is a customised stent graft in which fenestration(s) in the main endograft are aligned with the ostia of the visceral artery and a covered stent is then deployed via this fenestration that protrudes into the visceral vessel²⁻⁶. The same procedure is then repeated for the number of desired visceral arteries. These fenestrated vessels provide a seal that excludes any blood flow from entering the aneurysm sac whilst providing an additional anchoring force in order to safely deploy the endograft. Off the shelf branched stent-grafts (BSG) can be seen as successors to FSG, this is because the length of seal between the main endograft and visceral vessel branch is much longer as compared to FSG, where the whole sealing zone consists of the metal ring surrounding the fenestration only⁵. BSG also make it easier for aortic interventionalists to cannulate the visceral arteries as the inner docking lumen in the main endograft provides a room for additional manoeuvring.

Schematics of FSG and BSG are shown in Figure 1. Although BSG are conceptually more appealing than FSG with respect to branch stability and ease of implantation, there are a number of critical questions which must be answered beforehand in order to ensure durable device design. Some of these key questions are: (a) the haemodynamic impact of antegrade and retrograde blood flow on the functionality of BSG⁷, and (b) the consequences of

misaligned visceral stent-grafts⁴ on the durability of both FSG and BSG. Answering these questions will help to assess the reliability of both stent-grafts for EVAR before treatment, which may aid surgeons in predicting the likelihood of future complications such as migration and endoleaks as well as intimal hyperplasia and occlusion. The objective of this study is to answer the two aforementioned questions by simulating blood flow in ideal models of FSG and BSG under physiological conditions using computational fluid dynamics (CFD). Both antegrade and retrograde BSG are examined, and FSG with the same renal outlet configuration are also analysed to allow direct comparison of the results. Key results that are scrutinised in this study include: (a) recirculation zones in renal branches, (b) outflow in renal and iliac arteries and (c) resultant displacement force experienced by the stent-grafts to assess the risk of device migration.

METHODS

Performing CFD simulations is a systematic and multi-step procedure which requires construction of a 3D geometric model for the fluid domain of interest, specifying assumptions made in the governing equations and physiologically relevant boundary conditions, and choosing an appropriate numerical scheme to solve the set of coupled equations. Idealised 3D models of stent-grafts were constructed using commercially available software SolidWorks (Dassault Systemes, Velizy, France). In order to account for misaligned visceral stent-grafts, three different take-off angles (TOA) between the main stent-graft body and the visceral stent-graft centreline axis were considered i.e. 90°, representing normal alignment, 30° and 120° representing misaligned renal outlets of BSG and FSG. The effect of aortic neck angulation on the performance of BSG and FSG was evaluated by building additional models with TOA of 90° and lateral aortic neck angulation of 60°. Schematics of all the studied stent-grafts together with definitions of visceral TOA and lateral aortic neck angle are further

explained in Figure 1 and Figure 2. For all the analysed BSG and FSG, inlet diameter to renal diameter ratio was set at 4^{4,7}; inlet diameter to iliac diameter ratio was set at 1.54 while the iliac bifurcation angle was fixed at 30°. For antegrade and retrograde BSG, the length of each visceral cuff, present inside the main endograft, was 8 mm and 15 mm, respectively. Surface areas of antegrade BSG, retrograde BSG and FSG were 15,817 mm², 16,336 mm² and 15,252 mm², respectively. A cylindrical segment of 20 mm in length was added to the inlet while visceral (renal) arteries were artificially extended by 30 mm in order to ensure that the resolved flow field was not influenced by the location of inlet and outlet boundaries.

Conservation equations were used to describe the laminar pulsatile blood flow in the lumen, these equations are based on the principle that for a given system, mass, momentum and energy are conserved. Since blood flow in a human body can be assumed to be incompressible (constant density) and isothermal (constant temperature), energy conservation equation was not needed and the resulting governing equations consisted of continuity and Cauchy momentum equations⁶. As blood is a suspension of erythrocytes, leucocytes, thrombocytes and various other proteins in plasma, it was assumed to be a non-Newtonian fluid described by the Quemada viscosity model. Physiologically realistic volumetric flow rate waveform was extracted from the literature⁸ and was imposed at the inlet along with Womersley velocity profiles⁹. Corresponding pressure waveforms were prescribed at the outlets and these pressure waveforms were obtained by coupling the outlet of each terminal vessel with a 3-Element Windkessel Model (3-EWM). The parameters of the 3-EWM, i.e. proximal resistance R_1 , distal resistance R_2 and compliance C , were obtained using Nelder-Mead Simplex algorithm. No slip boundary conditions were specified at the stent-graft walls, which were assumed to be non-distensible. In order to delineate the effect of geometry on haemodynamics in stent-grafts, the same set of boundary conditions were applied to all computational models concerned in this study. The inflow waveform expressed in terms of

instantaneous Reynolds number along with the schematic of the employed computational model is shown in **Figure 3**.

The 3D stent-graft models were discretised into tetrahedral and prism elements using ANSYS ICEM CFD (ANSYS, Canonsburg, PA, USA). The governing equations were solved numerically using ANSYS CFX (ANSYS, Canonsburg, PA, USA). In order to achieve well-converged solutions, convergence criterion based on root mean square residual was set to be 1×10^{-6} . A uniform time-step of 0.001 s was used and all simulations were performed for three cardiac cycles in order to achieve periodicity⁶. Grid independence tests were carried out and the results were declared grid independent when velocity fields and displacement forces did not alter by $\pm 2\%$ between two successive meshes. For the studied geometries a minimum of 350,000 elements were required to achieve grid independence. The number of elements adopted in the final analysis ranged from 1.5 million to 1.8 million.

Flow recirculation zones (FRZ) are the regions of local flow reversal. FRZ in the renal branches were quantified by measuring the distance between the flow separation and re-attachment points (Figure 4a), following the method proposed by Kenwright *et al.*¹⁰. It is important to identify and compare FRZ in renal branches because these regions are associated with low wall shear stress (WSS), and are therefore prone to increased risk of thrombus formation^{11, 12, 13}. In addition, all stent grafts experience time-dependent displacement forces which are generated due to blood pressure and friction exerted by blood flow on stent-graft walls¹⁴. Large displacement forces are related to future complications such as endograft migration and type I endoleaks, therefore it is critical to quantify these forces. Displacement forces were calculated by integrating the traction vectors, pressure and wall shear stress (WSS), over the entire surface of the BSGs and FSG in coronal, sagittal and transverse directions.

127 **RESULTS**

128 In order to evaluate the performance of all stent-grafts for every visceral TOA and lateral
129 aortic neck angle, results are presented and discussed with regard to the following:
130 recirculation zones in the renal arteries; comparison of renal and iliac flow rates between
131 BSG and FSG and finally, displacement forces experienced by BSG and FSG.

132 **Recirculation Zones in the Renal Arteries**

133 The formation of flow recirculation zone (FRZ) is one of the most important flow phenomena
134 in the renal branches. This is caused by a sudden change in flow direction at the renal
135 bifurcation, and the size and location of FRZ are strongly dependent on local geometry.
136 Instantaneous time point of maximum deceleration ($T_b = 0.3$ s) was chosen to compare FRZ
137 in the renal arteries for all stent-grafts because this is the time point when the largest FRZ
138 were observed. As summarised in Figure 4b, the largest FRZ were found in antegrade BSG
139 and FSG at TOA of 30° , while the smallest FRZ was observed in FSG at TOA of 120° . At
140 TOA of 30° the length of FRZ in antegrade BSG and FSG was significantly larger than that
141 of retrograde BSG, because there were two FRZ in each of the antegrade BSG and FSG and
142 the reported length was the sum of the two lengths. Of all the geometrical variations of stent-
143 grafts studied here, the minimum standard deviation in length of FRZ was found in retrograde
144 BSG (0.8 mm) while the maximum was found in FSG (6.6 mm). Locations of the FRZ can be
145 seen from the time-averaged wall shear stress (TAWSS) contours in Figures 5-8, which also
146 demonstrate large spatial variations of wall shear stress in the renal branches.

147 **Swirling Patterns in Main Endograft Body for Stent-grafts with Straight and Angulated** 148 **Aortic Neck**

For stent-grafts with a straight aortic neck, four symmetric swirling zones were found in the main endograft body, distal of renal ostia, as shown in **Figure 9a**. However in case of stent-grafts with lateral aortic neck angle of 60° , only two swirling zones were found in the main endograft body, distal of renal ostia, as shown in **Figure 9b**.

Comparison of Renal and Iliac Flow Rates between BSG and FSG with Straight Aortic Neck

Based on our numerical simulation results, there was an equal flow division between the left and right renal arteries in all stent-grafts with a straight aortic neck, and the same was found for the left and right iliac arteries owing to geometric symmetry. For all the simulated scenarios, renal flow waveform was characteristically different from the iliac flow waveform such that in renal arteries the instantaneous flow rate was antegrade throughout the cardiac cycle with relatively high diastolic flow as compared to flow in the iliac arteries. As shown in **Figure 10**, the maximum renal flow was observed for FSG and minimum flow for retrograde BSG for all TOA examined. In antegrade BSG, renal flow was as high as in FSG during systolic acceleration but soon after peak systole, renal flow dropped down to about the same level as in retrograde BSG. **Table 1** gives a summary of the cycle-averaged renal flow for all the stent-grafts and TOA concerned in this study.

Iliac flow waveforms for all the studied stent-grafts at every TOA are also included in **Figure 10**. The time dependence of iliac flow waveform was very similar to inlet flow waveform. In contrast to renal flow waveform, during early systole, the maximum flow was observed for retrograde BSG and after peak systole, especially at late diastole, blood flow in the iliac arteries settled at almost the same value for all the stent-grafts. Peak systolic flow rate for iliac arteries was four times higher than peak systolic renal flow in FSG. However, just like inlet flow rate waveform, there was flow reversal in the iliac arteries during early diastole and

average late diastolic flow was almost zero. Cycle-averaged mean flow in the iliac arteries for all the stent-grafts at every visceral TOA is also summarised in [Table 1](#).

Comparison of Renal and Iliac Flow Rates between BSG and FSG with Angulated Aortic Neck

With an angulated aortic neck, the BSG and FSG models became asymmetric, thus the flow division between the two renal branches was no longer equal, especially during early systole when the flow was accelerating (Figure 11). The results show that peak systolic flow through outlet 1 was higher in retrograde BSG but lower in FSG and antegrade BSG. Cycle-averaged renal and iliac flow rates for all stent-grafts with an angulated neck are summarised in Table 2. Iliac flow waveforms for stent-grafts with an angulated aortic neck followed the same trend as that of stent-grafts with a straight aortic neck.

Displacement Forces Experienced by Straight and Angulated Aortic Neck BSG and FSG

Time dependence of displacement forces acting on all the stent-grafts follows the pressure waveform very closely, as shown in [Figure 12a](#) for the straight aortic neck models with a TOA of 30°. The peak displacement force for all the stent-grafts was observed at $t = 0.3$ s, corresponding to the time point of peak aortic pressure. In stent-grafts with a straight aortic neck, it is evident from [Table 3](#) that the magnitude of the displacement forces depends very strongly on the TOA and lateral aortic neck angle but not on the type of stent-graft. Mean (cycle-averaged) displacement force of 1.95 N was observed for BSG and FSG at TOA of 120° while mean displacement force of 1.24 N was observed for these stent-grafts at 30° TOA. At TOA of 90°, an intermediate mean displacement force of 1.69 N was observed for all types of stent-grafts. Displacement force is a 3D vector with components in the coronal, sagittal and transverse planes. However, as the constructed stent-grafts with straight aortic

neck were planar and symmetric, all the displacement forces were acting vertically downwards in the coronal plane (Figure 12b). The cycle-averaged angle between x -axis and displacement force was 90° .

The time trend for displacement forces observed for stent-grafts with different TOA also hold true for stent-grafts with a lateral aortic angle. That is, not only the displacement force followed the aortic pressure waveform very closely but also the same displacement force was recorded for antegrade and retrograde BSG as well as FSG with an aortic neck angle of 60° , both in terms of time dependence and magnitude. The peak displacement force for all the stent-grafts with an aortic neck was found to be 5.27 N while the minimum and mean displacement force was 2.69 N and 3.65 N, respectively. With respect to maximum mean displacement for stent-graft with a straight aortic neck at TOA of 120° , there was an increase in displacement force by 86.22%. As angulated stent-grafts were planar but asymmetric, displacement force was no longer acting vertically downwards (Figure 12c) and the cycle-averaged angle between x -axis and displacement force was 6.25° .

DISCUSSION

FSG are customised on patient-specific basis to ensure that the fenestrations in the main endograft body are aligned properly with the visceral arteries. These fenestrations then house the supplemental stent-grafts that protrude into visceral arteries, so that the aneurysm sac is separated from blood flow while the visceral (renal) stent-grafts provide additional anchoring force to prevent device migration. Due to the presence of these fenestrations, tailor-made FSG require several weeks in the manufacturing process, rendering their use expensive and unsuitable for urgent cases. FSGs are also susceptible to intra- and inter-observer discrepancies^{15, 16} which can lead to potentially misaligned visceral stent-grafts thus affecting their patency which can lead to branch occlusion. One can therefore argue that the unique

advantage of FSG is also their biggest drawback. In order to overcome these limitations of FSG, BSG namely antegrade and retrograde stent-grafts were introduced¹⁷. BSG have a fixed stent-graft which is parallel and anastomosed to the proximal part of the main endograft. These fixed stent-grafts can be oriented in either antegrade or retrograde fashion and house the docking point for visceral stent-grafts thus eliminating the need for custom-made fenestrations and make them adaptable to a wide variety of hostile anatomies. The overall objective of this study was to utilise CFD simulations to evaluate the haemodynamic consequences of antegrade and retrograde BSG as compared to FSG at various visceral take-off angles and lateral aortic neck angles.

Persistent FRZ were identified in renal arteries (outlets 1 and 2) of most stent-grafts and the largest recirculation zones were observed at the time point of maximum flow deceleration, T_b ($t = 0.3$ s). Flow in FRZ is disturbed, resulting in low wall shear stress (as shown in Figures 5-8) which has been associated with the development of atherosclerotic plaques. The presence of permanent FRZ may also favour thrombus formation, leading to partial or complete occlusion of the visceral branch and sometimes distal embolization. Therefore, large and permanent FRZ should be avoided as much as possible. For stent-grafts with a straight aortic neck, as summarised in **Figure 4**, two features can be observed: (1) FRZ were much larger in antegrade and retrograde BSG than in FSG except for TOA of 30° ; (2) the size of renal FRZ in BSG depends very strongly on the renal TOA as the largest recirculation zones were found in BSG at TOA of 30° . The presence of FRZ caused large spatial variations in TAWSS near the renal junction, as shown in Figures 5-7. For stent-grafts with an angulated aortic neck however, FRZ in the renal arteries varied differently from those found in stent-grafts with a straight aortic neck. With aortic neck angulation of 60° , the flow was directed towards one side of the endograft due to inertia i.e. towards outlet 2 (right renal artery), thus FRZ in the right renal branches were not mirror images of the ones found in the

left, as reflected in the TAWSS contours in [Figure 8](#). Lateral neck angulation also has a strong effect on the flow swirling patterns in the main endograft body distal to renal ostia, as shown in [Figure 9](#). The experimental study carried out by Ha and Lee¹⁸ demonstrated the beneficial effects of pulsatile swirling flow, namely reducing the oscillatory shear index and the propagation length of jet flow.

Volumetric flow rate waveforms in renal arteries are characteristically distinct from iliac arteries due to relatively low resistance of the distal vascular beds of the kidneys¹⁹ thus avoiding any backflow throughout a cardiac cycle. As shown in [Figure 10](#) and Table 1, retrograde BSG had the lowest renal flow rate at all TOA examined, while the highest renal flow was achieved in FSG. A complete opposite trend was observed in iliac arteries with the highest flow in retrograde BSG and lowest in FSG. This is dictated by the conservation of mass, so that stent-graft inflow must be balanced by the sum of outflow through the renal and iliac arteries. For antegrade and retrograde BSG with TOA of 90°, outflow into each renal branch was 0.08 L/min higher in antegrade than retrograde BSG. Computational simulations carried out by Sutalo *et al.*⁷ on a simplified model of BSG with TOA of 90° also found a similar trend. They reported that outflow into renal branches of antegrade BSG was 0.07 L/min higher than that of retrograde BSG for a 200-mm conduit, and the difference was reduced to 0.03 L/min for a 40-mm conduit. Although our results are in qualitative agreement with those of Sutalo *et al.*, some quantitative differences exist, which can be attributed to the different stent-graft model geometry and flow conditions employed in the two studies. Sutalo *et al.*⁷ examined a single branch conduit without including the iliac bifurcation, whereas our model incorporated two parallel branch conduits and the iliac bifurcation. Moreover, we applied a realistic abdominal aortic flow waveform based on the *in vivo* study of Fraser *et al.*⁸

Regarding the effect of TOA on renal flow, Table 1 shows clearly that antegrade BSG is less sensitive to TOA than retrograde BSG and FSG. When renal TOA reduced from 90° to 30°,

271 mean renal flow reduced by 0.04 L/min for retrograde BSG and 0.03 L/min for FSG. As
272 TOA increased from 90° to 120°, no change in renal flow was found in retrograde BSG,
273 while mean flow rate decreased by 0.02 L/min in FSG. These results indicate that renal flow
274 in retrograde BSG is sensitive to TOA, and an acute TOA (e.g. 30°) tends to reduce renal
275 flow; however, the quantitative effect of TOA on mean renal flow is relatively minor.

276 For stent-grafts with lateral neck angulation, the flow division between renal outlets 1 and 2
277 was not equal (Figure 11). The asymmetry in flow between the left and right renal arteries
278 was amplified at peak systole. This is because blood flowing at a high velocity tends to keep
279 flowing in the same direction. Due to sharp angulation, the blood close to the inner wall does
280 not have enough centripetal force to flow along the same path and centrifugal force becomes
281 predominant. As a result, flow tends to favour the branch which requires less change in flow
282 direction. By comparing Tables 1 and 2, it is clear that an angulated aortic neck caused a
283 slight increase in renal flow. Compared with straight aortic neck stent-grafts, total mean
284 outflow into the renal arteries was 0.04 L/min higher in retrograde BSG with an angulated
285 aortic neck, and 0.01 L/min higher in FSG. For antegrade BSG, there was a slight reduction
286 of 0.01 L/min in each renal artery. This can be explained on the fact that aortic neck
287 angulation gives rise to secondary motion in the main stent-graft body (as shown in Figure 8)
288 and this secondary motion helps divert blood to renal branches in retrograde BSG and FSG
289 thus yielding slightly improved renal flow. However, just like stent-grafts with a straight
290 aortic neck the maximum renal flow was observed in angulated FSG, followed by antegrade
291 BSG while minimum flow rate was observed in retrograde BSG.

292 With respect to displacement forces acting on the stent-grafts, it can be seen from Table 3
293 that the magnitude of cycle-averaged displacement force is dependent more on TOA rather
294 than the orientation of the stent-graft. Linear regression analysis between TOA and
295 displacement forces yielded an R^2 value of 0.998 for BSG and FSG, suggesting that

displacement force increased almost linearly with TOA. Nevertheless, the increase in the magnitude of displacement force with TOA was relatively small compared to the change brought on by aortic neck angulation. When TOA increased from 30° to 90°, the cycle-averaged displacement force increased from 1.25 N and 1.69 N, but when neck angle changed from straight to 60° for a TOA of 90°, the mean displacement force was more than doubled from 1.69 N to 3.6 N. Therefore, the results suggest that the displacement force experienced by stent-grafts is more sensitive to lateral aortic neck angle than TOA. This is consistent with the finding of Georgakarakos *et al.*⁴ who reported that with increasing TOA, the value of displacement force did not vary much from 5.55 N. It is interesting to note that Georgakarakos *et al.* reported a higher value of displacement force than those found in this study. This difference is mainly attributed to the different dimensions of the stent-grafts and their non-planarity; the stent-graft models examined in the present study were planar whereas those adopted by Georgakarakos *et al.* was non-planar. As shown in patient-specific study carried out by Kandail *et al.*⁶, the magnitude of displacement force depends very strongly on non-planar anterior-posterior angle, and increasing anterior-posterior angle can significant increase the magnitude of displacement force.

CONCLUSIONS

Our computational simulations clearly show that all the stent-grafts examined provide adequate perfusion to the visceral (renal) arteries even when the visceral stent-grafts are misaligned. Renal flow rate is higher with antegrade BSG than retrograde BSG, although FSG offers the highest renal flow. However, misaligned renal stent-grafts give rise to recirculation zones in the renal arteries which may become potential sites for thrombus formation or stenosis. The size of the renal recirculation zone depends strongly on the angle between BSG and the renal artery. Due to comparable dimensions of antegrade, retrograde

320 and fenestrated stent-grafts, the displacement forces acting on the stent-grafts are independent
321 of their type but dependent on the degree of misalignment and lateral aortic neck angle, with
322 the latter being a more important determinant of the displacement force.

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371 TABLES

372 **Table 1:** Cycle-averaged flow rate in the renal and iliac arteries for all stent-grafts with a
 373 straight aortic neck, at different TOA.

Mean flow rate (L/min)						
TOA	Renal artery outlet			Iliac artery outlet		
	30°	90°	120°	30°	90°	120°
Antegrade BSG	0.53	0.54	0.53	0.51	0.50	0.51
Retrograde BSG	0.42	0.46	0.46	0.62	0.58	0.58
FSG	0.59	0.62	0.60	0.46	0.42	0.45

374

375 **Table 2:** Cycle-averaged flow rate in the renal and iliac arteries for all stent-grafts with a
 376 lateral aortic neck angulation of 60° and TOA of 90°.

	Mean flow rate (L/min)			
	Renal arteries		Iliac arteries	
	Outlet 1	Outlet 2	Outlet 3	Outlet 4
Antegrade BSG	0.53	0.53	0.49	0.53
Retrograde BSG	0.49	0.47	0.53	0.60
FSG	0.61	0.64	0.40	0.43

377

378 **Table 3:** Cycle-averaged displacement forces acting on all stent-grafts with a straight aortic
379 neck (at different TOA) and all stent-grafts with angulated aortic neck (LNA stands for lateral
380 aortic neck angle of 60°).

TOA	Displacement Force (N)								
	Antegrade BSG			Retrograde BSG			FSG		
	Mean	Min	Max	Mean	Min	Max	Mean	Min	Max
30°	1.25	0.83	1.83	1.26	0.85	1.83	1.24	0.83	1.82
90°	1.69	1.16	2.47	1.69	1.17	2.47	1.69	1.15	2.47
120°	1.95	1.35	2.83	1.95	1.36	2.83	1.95	1.35	2.84
LNA	3.65	2.69	5.27	3.65	2.70	5.26	3.64	2.69	5.26

381

382 **FIGURE CAPTIONS**

383 **Figure 1:** Schematics of the studied branched stent-grafts (BSG) and fenestrated stent-graft
384 (FSG). Antegrade BSG are shown in column 1, Retrograde BSG in column 2 and FSG in
385 column 3. Outlets 1 and 2 are labelled in antegrade BSG with TOA of 90° and the
386 terminology is consistent for all the other stent-grafts.

387 **Figure 2:** Schematics illustrating the definition of take-off angle (TOA) and lateral aortic
388 neck angle (planar) used in this study.

389 **Figure 3:** Schematic of the computational model adopted in the study along with the inflow
390 waveform, shown here in terms of instantaneous Reynolds number.

391 **Figure 4:** (a) Velocity vectors highlighting the separation point and reattachment point which
392 define the flow recirculation zone (FRZ) found in renal branches, (b) Bar chart showing the
393 lengths of FRZ (mm) in all the stent-grafts examined. For antegrade BSG and FSG at TOA of
394 30° , the length shown here corresponds to the sum of the two FRZ found in each case.

395 **Figure 5:** Time-averaged wall shear stress (TAWSS) contours for stent-grafts with straight
396 aortic neck and TOA of 90° . Locations of FRZ are marked by arrows which also correspond
397 to regions of low TAWSS.

398 **Figure 6:** Time-averaged wall shear stress (TAWSS) contours for stent-grafts with straight
399 aortic neck and TOA of 30° . Locations of FRZ are marked by arrows which also correspond
400 to regions of low TAWSS.

401 **Figure 7:** Time-averaged wall shear stress (TAWSS) contours for stent-grafts with straight
402 aortic neck and TOA of 120° . Locations of FRZ are marked by arrows which also correspond
403 to regions of low TAWSS.

Figure 8: Time-averaged wall shear stress (TAWSS) contours for stent-grafts with lateral aortic neck angle of 60° and TOA of 90° . Locations of FRZ are marked by arrows which also correspond to regions of low TAWSS.

Figure 9: Swirling flow patterns in main endograft body of (a) stent-graft with straight aortic neck and (b) stent-grafts with lateral aortic neck angle of 60° , at instantaneous time point of $T_b = 0.3$ s, which corresponds to maximum flow deceleration. Only retrograde BSG are shown as similar patterns were observed for all the other stent-grafts.

Figure 10: Renal (left) and iliac (right) flow rate waveforms over one cardiac cycle for all the stent-grafts with a straight aortic neck angle and (a) TOA of 90° , (b) TOA of 30° and (c) TOA of 120° .

Figure 11: Renal (left) and iliac (right) flow rate waveforms over one cardiac cycle for stent-grafts with a lateral aortic neck angle of 60° and TOA of 90° , where (a) flow rate waveforms in antegrade BSG, (b) flow rate waveforms in retrograde BSG and (c) flow rate waveforms in FSG.

Figure 12: (a) Time variation and magnitude of the displacement forces acting on all the stent-grafts with a straight aortic neck at TOA of 30° . Time dependence of displacement forces follows cardiac pressure waveform very closely and this trend holds true for all the other stent-grafts. The magnitude of displacement force changes with TOA and aortic neck angle, (b) Direction vector (red arrow) for the resultant displacement force for all stent-grafts with a straight aortic neck and (c) Direction vector (red arrow) for the resultant displacement force for all stent-grafts with an angulated aortic neck.

	Antegrade BSG	Retrograde BSG	FSG
<p>TOA = 90°</p> <p>Straight neck</p>
<p>TOA = 30°</p> <p>Straight neck</p>
<p>TOA = 120°</p> <p>Straight neck</p>
<p>TOA = 90°</p> <p>Angulated neck</p>

Figure 1.

Definition of take-off angle (TOA)

Definition of lateral aortic neck angle

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430 Figure 2.

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432 Figure 3.

Figure 4

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Figure 11

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458 Figure 12