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Computational fluid dynamics of a novel perfusion strategy using direct perfusion of a left carotid-subclavian bypass during hybrid thoracic aortic repair

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Glossary of Abbreviations

CFD = computational fluid dynamics
CPB = cardiopulmonary bypass
FET = frozen elephant trunk
RAA = right axillary artery
LCA = left carotid artery
LNH = localized normalized helicity
LSA = left subclavian artery
LVA = left vertebral artery
NACSA = National Adult Cardiac Surgery Audit
NICOR = National Institute for Cardiovascular Outcomes Research
WSS = wall shear stress
WSSG = wall shear stress spatial gradient

Abstract

Objective: We aimed to computationally evaluate the effects of direct cerebral perfusion strategy through a left carotid-subclavian bypass on hemodynamics in a patient-specific thoracic aorta model.

Methods: Between July 2016 and March 2019, eleven consecutive patients underwent single-stage frozen elephant trunk operation using the left carotid-subclavian bypass with a side graft anastomosis and a right axillary cannulation for systemic and brain perfusion. A multiscale model realized coupling 3D computational fluid dynamics was developed and validated with in vivo data. A model comparison with direct antegrade cannulation of all epiaortic vessels was performed. Wall shear stress, wall shear stress spatial gradient, and localized normalized helicity were selected as hemodynamic indicators. Four cerebral perfusion flows were tested (6 to 15 ml/kg/min).

Results: Direct cerebral perfusion of the left-subclavian bypass resulted in higher flow rates with augmented speeds in all epiaortic vessels in comparison with traditional perfusion model. At the level of left vertebral artery, a speed of 22.5 *vs* 21 ml/min and mean velocity of 3.07 cm/s *vs* 2.93 cm/s were registered, respectively. With a cerebral perfusion flow of 15 ml/kg, lower left vertebral artery wall shear stress (1.596 vs 2.030 N/m^2) and wall shear stress gradient (1445 vs 5882 N/m^3) were observed. A less disturbed flow considering the localized normalized helicity was documented. Similar results persisted at different cerebral perfusion flows. No patients experienced neurological/spinal cord damages.

Conclusions: The direct perfusion of a left-carotid bypass proved to be cerebroprotective, resulting in a more physiological and stable anterior and posterior cerebral perfusion.

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Central message

Additional direct perfusion of a left-subclavian bypass may provide a more physiological and stable cerebral perfusion during aortic arch and descending thoracic aortic repairs.

Perspective statement

The present image-based computational fluid dynamics (CFD) analysis with in vivo validation demonstrated that additional direct perfusion of a left-subclavian bypass offers a more physiological and stable cerebral perfusion than conventional perfusion methods. This strategy could offer significant clinical advantages during aortic arch and descending thoracic surgeries.

Introduction

The frozen elephant trunk (FET) represents a simplified treatment for complex diseases of the thoracic aorta, and has rapidly gained in popularity for its clinical and surgical advantages.[1,3] FET allows for single-stage therapy in case of multilevel aortic diseases, favours the expansion of true lumen in type A acute dissections, and also offers a potential landing zone for subsequent transfemoral endovascular aortic repairs.[1,2] However, in this context the optimal cerebral protection strategy remains controversial, and procedure-related FET complications are not remote.[1-4] Neurological and spinal cord complications occur in 2.5% to 21% of treated patients.[3,4]

To mitigate the risk of perioperative neurological complications during FET procedures, we have developed a modified cerebral perfusion strategy to both preserve the anterior and posterior cerebral circulation by the simultaneous perfusion of right the axillary artery (RAA) and a left carotid-subclavian bypass. In the present study through an image-based computational fluid dynamics (CFD) analysis, we aimed to assess the fluid dynamics and vascular biomechanical properties of this novel perfusion strategy, and to better understand the relationship between antegrade cerebral perfusion and pathophysiology of neurological complications during aortic arch surgery.

MATERIAL AND METHODS

Study Population

Between July 2016 and March 2018, eleven consecutive patients underwent ‘single-stage’ operation with the Thoraflex hybrid stent graft (Vascutek, Terumo, Inchinnan, UK) at University Hospital of Leicester, Glenfield Aortic Centre (United Kingdom) for the repair of complex thoracic aortic diseases involving the aortic arch and the proximal descending aorta using our novel cerebral perfusion strategy. All patient data were prospectively collected in the National Institute for Cardiovascular Outcomes Research (NICOR) of the National Adult Cardiac Surgery Audit (NACSA) registry.[5]

Operative Technique

Total intravenous anesthesia was routinely administered. Both radial and left femoral arteries were cannulated to monitor the perfusion pressures to the brain and the lower body part, especially during circulatory arrest and selective antegrade cerebral perfusion. In the same operating session and before sternotomy, a left carotid-subclavian bypass was created in all patients through a standard left supraclavicular incision, by using an 8-mm Dacron graft (Vascutek Terumo, Renfrewshire, Scotland). An additional 8-mm Dacron graft was then anastomosed ("T" configuration) to the same left carotid-subclavian bypass. This constituted the first perfusion line for the institution of the cardiopulmonary bypass (CPB). A second arterial cannulation site was also created through an 8-mm Dacron graft anastomosed to the RAA. After the median sternotomy, innominate artery, the left common carotid (LCA) and subclavian arteries (LSA) were mobilized and taped. After systemic heparinization, CPB was instituted through the two above mentioned perfusion lines. The venous drainage was achieved by cannulation of the right atrium. The left ventricle was vented through the right superior pulmonary vein. Myocardial protection was achieved with antegrade and retrograde administration of intermittent cold blood cardioplegia of Harefield Hospital Formulation (IVEX Pharmaceuticals Ltd, Larne, Northern Ireland, UK). The ascending aortic aneurysm/dissection was then excised, and circulatory arrest was established at a target nasopharyngeal temperature of 23-25°C. The innominate artery and LCA were then clamped and disconnected from the native aortic arch, while the LSA was permanently occluded at its origin. Therefore, cerebral perfusion was never interrupted, being maintained through the RAA and the left carotid-subclavian bypass perfusion lines. In all patients, near-infrared spectroscopy (INVOS cerebral oximeter; Somanetics Corporation, Troy, MI, USA) was utilized to guide the cerebral perfusion, and target radial and femoral pressures were maintained at 50-70 and 20-30 mmHg, respectively (perfusate flow: 8-10 ml/kg). The arch was then opened longitudinally, generally between the LCA and LSA origins. The Thoraflex hybrid graft was bent slightly to conform to the curvature of the descending thoracic aorta and deployed under direct vision, without using any guidewires. The hybrid device was selected according to the anatomic characteristics of the aortic arch/descending aorta and type of lesion. Generally, a 15 cm stent was deployed in chronic atherosclerotic aneurysm, while a 10 cm stent graft in acute aortic syndromes. After the distal aortic arch reconstruction, the systemic perfusion through the side-branch of the hybrid graft was re-initiated. Subsequently, the proximal aortic repair was accomplished, and the cross-clamp removed. The re-implantation of the innominate artery and the LCA to the graft branches was then performed in beating heart, and the initial perfusion lines into the RAA and the left subclavian bypass were subsequently excluded. Cerebro-spinal fluid drainage was never adopted.

Computational Modeling of the Aorta

An image-based model of a 52-year-old male patient who undergone FET with the above-mentioned cerebral perfusion configuration was created, using the postoperative computed tomography angiography of the entire thoracic aorta. The images of the brachiocephalic trunk, common carotid arteries, subclavian and vertebral arteries were all retained and used to reconstruct the aortic geometry in order to perform the CFD model. The 3D anatomical model was constructed by imaging segmentation and 3D reconstruction processes using the commercial application software for 3D open source medical images Itk-Snap 3.0 (http://www.itksnap.org). Since the segmentation software provides a stereolithographic file (stl format), the 3D aorta surface was subjected to the reverse engineering process to obtain a better surface quality, smoothing the aorta and the supra-aortic vessels. Finally, the surface model was converted in a 3D solid model. The aorta model included the aortic arch, the epiaortic vessels and the two vertebral arteries. Then, the parameters used to calculate the image of the function and the weights related to the different types of speed driving the evolution of the segmentation were selected. Segmentation produced a 3D surface that was imported into RHINOCEROS v.4.0 software (Robert McNeel & Associates, Seattle, WA, USA). The surface was blunted, and the volume rebuilt through the sweep command. The two following perfusion configurations were then compared, and the corresponding hemodynamic changes calculated, evaluating the blood flow in the aorta and supra-aortic vessels (Figure 1). Configuration 1 represented a traditional cerebral perfusion strategy using direct cannulation of the epiaortic vessels including the LSA and was used for comparison.[6-8] Configuration 2 (our novel perfusion configuration) consisted in the simultaneous cerebral and systemic perfusion through two 8-mm Dacron grafts anastomosed to the RAA and to the left carotid-subclavian bypass (Figure 1).

Mathematical Model and Hemodynamic Indicators

Two interrelated mathematical models were adopted: one for the blood and one for the vessel wall. The blood flow was considered Newtonian, an accepted assumption for flows in large vessels as the aorta with density of the blood and viscosity equal to 1060 kg/m^3 and 0.0035 Pa·s, respectively.[9-13] The blood motion was modeled as laminar, and the Navier-Stokes equations for incompressible fluids were used:

$$\nabla \cdot \mathrm{u} = 0, \tag{1}$$

$$\rho(\delta \mathrm{u}/\delta t) + \rho(\mathrm{u} \cdot \nabla)\mathrm{u} = \nabla \cdot [-p\mathrm{I} + \mu(\nabla \mathrm{u} + (\nabla \mathrm{u})^{\mathrm{T}})] + \mathrm{F}, \tag{2}$$

where u is the fluid velocity vector, p the static pressure, μ the dynamic viscosity, ρ the density of blood, I the identity matrix, and F the volume force field.[9] The term F was neglected in the computational study and the effect of gravity was ignored, because the surgical procedures were conducted with patients in supine position.[12,13]

In order to compare the two types of perfusion configurations and to investigate the blood flow, several hemodynamic indicators were selected, including the Wall Shear Stress (WSS), the Wall Shear Stress Spatial Gradient (WSSG) and the Localized Normalized Helicity (LNH). WSS represents the stress induced by the flow in the fluid layer near the wall of the vessel. It is expressed in units of force per unit area and is defined as:

$$WSS = \sqrt{(\tau_x)^2 + (\tau_y)^2 + (\tau_z)^2} \tag{3}$$

where τ is the viscous stress in x, y, and z directions.[14-16] Values less than 1 N/m^2 correlate with atherogenesis and plaque progression, while values greater than 3 N/m^2 correlate with the development of endothelial lesions, including plaque rupture and debris dislodgement.[14-16] WSSG is the spatial derivative of WSS along the flow direction with respect to the streamwise distance, and a marker of endothelial cell tension, and values greater than 2200 N/m^3 are implicated in the derangements and lesions of endothelium due to spatially changing hemodynamic factors.[16,17] It is defined as follow:

$$\text{WSSG} = \frac{1}{T}\sqrt{\left(\frac{d\tau x}{dx}\right)^2 + \left(\frac{d\tau y}{dy}\right)^2 + \left(\frac{d\tau z}{dz}\right)^2} \qquad (4)$$

where T is the duration of one cardiac cycle, δ/δx, δ/δy, and δ/δz are the partial derivates with respect to *x*, *y*, and *z* coordinates.[16] Finally, LNH was calculated to measure the helical structures of the aortic blood flow along the cardiac cycle:

$$LNH(s;t) = \frac{V(s;t) * \omega(s;t)}{|V(s;t)||\omega(s;t)|} = cos\varphi(s;t) \qquad (5)$$

where V is the velocity vector, s the position, t the time, and ω the vorticity vector.[18] LNH variates between −1 and +1, representing the local value of the *cosine* of the angle between the velocity and the vorticity vectors.[18] Positive values indicate regions where the flow rotation is right-handed, negative values reflect left-handed rotation, while symmetrical flow occurs when LNH is equal to zero.[18]

Computational Fluid Dynamic Analysis

For the fluid-dynamic analysis, identical continuous flows were applied as for the inlet and outlet boundaries for each supra-aortic vessel, and for both configurations (Figure 2). The inlet boundary was set at the tip of the cannula, while the outlet was set in the output boundaries of supra-aortic vessels. At the inlet level, four different constant flows were tested for both configurations, mimicking different perfusion flow regimens. These values were derived from the European Association for Cardiothoracic Surgery (EACTS) survey on neuroprotection in aortic arch surgery.[7] Perfusate flow was reported to be fairly consistent across 400 European centers in the average of 10-15 ml/kg/min.[8] For a patient weighing 70 kg, we tested: 1) a total flow of 420 ml/min corresponding to 6 ml/kg/min (case A), 2) a total flow of 560 ml/min corresponding to 8 ml/kg/min (case B), a total flow of 700 ml/min corresponding to 10 ml/kg/min (case C) and, finally, a total flow of 1050 ml/min corresponding to 15 ml/kg/min (case D). For configuration 1, the total flow was divided by the three inlet cannulas, whereas for configuration 2 by two cannulas. The outputs were set equal to 60 mmHg at the level of all supra-aortic vessels, corresponding to the usual cerebral perfusate pressure.[9] Aortic walls and perfusion cannulas were assumed to be rigid and impermeable, and a no-slip condition ($v_{\text{wall}} = 0$) was adopted. For the numerical simulation, the post-process and the visualization of numerical results, a finite-element-based commercial software package was used (COMSOL 4.3a, Inc., Stockholm, Sweden). A fine mesh consisting of tetrahedral elements was then generated. The blood flow was investigated in terms of velocity streamlines, pressure and shear stress indices. A GMRES (Generalized Minimal Residual) algorithm for solving a non-symmetrical linear system of equations was used.[19]

RESULTS

Patient Population

The 11 patients had a mean age of 63.9 ± 11.0 years (range, 47 to 79 years), and underwent replacement of the aortic arch and repair of the descending aorta using the Thoraflex hybrid prosthesis, using our technique as the only cerebral perfusion strategy. Baselines, operative and postoperative characteristics of enrolled patients are summarized in Table E1 in the Appendix. Briefly, treated aortic lesions included chronic atherosclerotic or dissecting aneurysms (n=8) and acute aortic syndromes (n=3). No patient had previous cardiac, thoracic and abdominal aortic surgery. No history of cerebrovascular accidents was documented. Cumulative CPB time was 219.4 ± 39.8 minutes, and lower body circulatory arrest time was 22.7 ± 13.5 minutes. Concomitant cardiac procedures were performed in 7 (64%) cases. None of the approached patients experienced temporary/permanent neurological or spinal cord injuries.

Computational Modeling results

Figure 3A and 3B show the trend of the velocities with a maximum perfusion flow of 1050 ml/min (15 ml/kg). Configuration 1 resulted in lower flow rates with reduced speeds in the vertebral arteries, LCA and LSA compared to configuration 2. The calculated flow in the left vertebral artery (LVA) for configuration 1 was 21 ml/min with a mean velocity of 2.93 cm/s, while configuration 2 demonstrated higher values equal to 22.5 ml/min with a mean velocity of 3.07 cm/s. In configuration 1, the presence of vortices at the level of vertebral arteries was also more frequently observed than in configuration 2 (Figures 3C and D). The percentage of variations in term of reduction of flow and mean velocity in the vertebral arteries was also calculated. The values of three different perfusate flows (cases A, B and C) were compared with those of a perfusate flow of 1050 ml/min (case D). Table 1 reports the variations in terms of reduction of flow and mean velocity in the vertebral arteries for both configuration models. Again, configuration 1 resulted in less stability during cannula flow variations, resulting in higher percentages of variation at vertebral level.

Figure 4 shows the results of the WSS calculation in both configuration models. Configuration 1 demonstrated higher values at cannula level as a consequence of a greater flow. With a perfusate flow of 1050 ml/min, a WSS of 2.030 N/m^2 was observed at the level of LVA in the configuration 1, while a lower WSS (1.596 N/m^2) was encountered in the configuration 2. Higher WSS values are associated with an increased risk of endothelial damage and disruption. Similarly, in configuration 1, the WSSG calculated as the maximum value along the vertebral artery surface resulted to be equal to 5882 N/m^3. In configuration 2, the WSSG value was lower and equal to 1445 N/m^3. Results obtained for cases A, B, C were finally compared with those of case D, and percentages of variations were then calculated (Table 2). Configuration 1 brought a considerable WSS reduction (up to 80%) with respect to the maximum flow value. Finally, considering the LNH at the vertebral level, configuration 1 resulted in a more disturbed flow than configuration 2 (Figure 5).

Discussion

Several methods for cerebral perfusion have been adopted to address aortic arch and descending aortic disease repairs, including metabolic suppression with anaesthetic agents, antegrade cerebral perfusion (ACP), hypothermic circulatory arrest, and retrograde cerebral perfusion.[6-8,20] In a recent survey reporting currents trends in cannulation and neuroprotection during aortic arch surgery in Europe, bilateral and unilateral ACPs resulted as the most frequent utilized methods, accounting for 53% and

38% of strategies in the acute setting, and for 65% and 33% in chronic aortic conditions, respectively.[8] Although all of these operative strategies have been directed at reducing operative mortality and morbidity, the occurrence of temporary and permanent neurologic deficits remain high.[1-4] Even in elective proximal arch surgeries, a 6% rate of paraplegia is encountered, highlighting the need for further measures to reduce this devastating complication, especially in case of more time demanding extensive aortic repairs.[21] In this context, additional perfusion of LSA seems to be beneficial, particularly in critical vascular conditions such as concomitant carotid dissections, acute occlusion of the right vertebral artery, dominant left vertebral artery, or inadequate intracranial arterial communications.[22-27] As a matter of fact, studies reporting on outcomes after thoracic endovascular aortic repair (TEVAR) with overstenting of the LSA, have demonstrated an increased risk of left-hemispheric stroke, and permanent paraplegia.[28-30] In addition, the perfusion of the left vertebral artery (LVA) through the LSA is of utmost importance in presence of posterior anomalies of the Willis circle (type IA and type IIA variations).[27]

Moriyama et al.[23,24] firstly introduced the selective perfusion of the LSA during the repair of the descending thoracic and thoracoabdominal aortic aneurysms under deep hypothermia. Avoiding retrograde perfusion, they did not encounter any brain injury.[24] Kurisu et al.[25] described the use of bilateral cerebral perfusion through cannulation of both axillary arteries in 12 patients undergoing aortic arch surgery. Although this was a preliminary and limited series, the authors did not observe any temporary and permanent neurologic deficits nor paraplegia.[25] Xydas et al.[26] similarly did not reported any neurological or paraplegia complication during aortic arch reconstructions with the routine use of a carotid-subclavian arterial bypass.

In order to minimize cerebral ischemia and the risk of inadequate cerebral perfusion, preserving both anterior and posterior cerebral circulation, we recently introduced the use of a perioperative left carotid-subclavian bypass as CPB arterial inflow, warranting the simultaneous perfusion of the LCA, LSA and LVA. In the present study, we were able to demonstrate through an image-based CFD analysis that our modified cerebral perfusion configuration resulted in a more physiological and stable cerebral blood perfusion. Our excellent neurological outcomes also corroborate the image-based CFD data, although we are conscious that this is a preliminary and limited patient series.

Our cerebral configuration presents several advantages. First, the simultaneous perfusion of the LCA, LSA, and LVA maintain the complete blood circulation in the anterior and posterior cerebral circle, avoiding the risk of an inadequante perfusion in presence of undected Willis anomalies. This is of importance in the acute setting, when an accurate and immediate intracranial arterial imaging is not feasible for the impeding risk of an aortic rupture. Second, a direct LSA anastomosis through the sternotomy is often challenging, requiring unduly prolonged cerebral ischemia time and poor LSA visualization with the risk of uncontrolled bleeding, especially in case of a dissected and fragile LSA.[31] Third, the distal FET anastomosis can be easily performed in zone 0 and 1 with excellent visualization, minimizing the cerebral ischemia time, and favouring a better and direct haemostatic control. Fourth, the avoidance of direct cannulation of the supra-aortic vessels greatly reduced the risks of cerebrovascular accidents resulting from air embolism or dislodgement of atherosclerotic debris.[31,32] In acute aortic dissection involving the aortic arch vessels, the direct carotid cannulation could potentially lead to the direct damage of the arterial intima wall, resulting in bleeding complications and serious malperfusion.[31] The latter can also be encountered with an improper insertion of the perfusion carotid cannulas.[33] Finally, our technique is fully compatible with other aortic arch and descending repairs or with aortic root reconstructions.

The configuration of our cerebral perfusion strategy is pathophysiologically justified by clinical and experimental evidences, suggesting that spinal cord perfusion does not principally depend by a single branch artery of the descending thoracic aorta, the so-called artery of Adamkiewicz.[34,35] It has been demonstrated that spinal cord perfusion is supported by an extensive integrated collateral arterial network, including the segmental and epidural arteries, and the anterior spinal artery. All these vessels are interconnected with the subclavian arteries cranially, and the hypogastric arteries distally.[35] The result is an extensive collateral compensatory flow to spinal cord even when some collaterals are irreparably compromomised or in case of an anatomically incomplete circle of Willis.[22,35] In rats, the bilateral direct ACP alone resulted in perfusion of only 30% of the spinal cord through to the retrograde flow of the vertebrobasilar system. The additional perfusion of the subclavian arteries alone resulted in greater spinal cord perfusion (up to 40%).[22] The simultaneous bilateral ACP with at least one of the subclavian arteries was demonstrated to provide a much better perfusion to both the spinal cord and the brain.[23] This evidence is consonant with our image-based CFD analysis that demonstrated a more physiological and stable cerebral blood perfusion when the carotid-subclavian bypass is used as direct arterial inflow for cerebral perfusion.

Certainly, the present study is limited by its non-randomized and observational nature, other than the limited patient population. In addition, we used idealized boundary conditions for investigating the impact of our cerebral perfusion configuration. Possible bias originating by the cardiac function of the patients as well as concomitant cardiac diseases, and hypertension were not considered in our calculations. Therefore, there may be a discrepancy between the individually measured data and calculated data.

In conclusion, the additional direct perfusion of a left carotid-subclavian bypass provides a more physiological and stable cerebral perfusion, warranting an adequate and complete anterior and posterior cerebral circulation. This technique may decrease the risk of neurological and spinal cord complications associated with aortic arch and descending aortic repairs, especially in case of undetected vascular compromises such as a dominant left vertebral artery, carotid artery disease or inadequate intracranial arterial communication.

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Figure legends

CENTRAL PICTURE. Comparison of a standard and a novel cerebral perfusion strategy configuration including the additional direct perfusion of a left-subclavian bypass (upper panels). Wall Shear Stress distribution recorded in all supra-aortic vessels during cerebral perfusion shows a more physiological and stable cerebral perfusion in case of the novel perfusion system (lower panel).

FIGURE 1. Patient specific aortic model with geometrical reconstruction, including the supra-aortic vessels and the aortic arch. Both configuration models are represented. Arterial inflows are indicated by *triangles*. *Asterisks* identify the 8-mm Dacron grafts anastomosed to the RAA (configurations 1 and 2) and the left carotid-subclavian bypass (configuration 2). The left-carotid subclavian bypass is underlined by a *blue circle*. (*LCA* = left carotid artery; *LSA* = left subclavian artery; *LVA* = left vertebral artery; *RAA* = right axillary artery; *RCA* = right carotid artery; *RVA* = right vertebral artery)

FIGURE 2. Geometrical reconstruction with related inlet and outlet boundaries for each supra-aortic vessel and for both configurations.

FIGURE 3. Velocity stream lines (m/s) recorded in all supra-aortic vessels for the cerebral perfusion model obtained with configuration 1 (A,C) and with configuration 2 (B,D). Panel C and D also show details of velocity stream lines at the level of left vertebral arteries. Colours denote velocity values, from smallest (blue) to highest (red).

FIGURE 4. Wall Shear Stress (WSS) distribution recorded in all supra-aortic vessels during cerebral perfusion for configurations 1 and 2. Colours denote WSS values, from smallest (blue) to highest (red).

FIGURE 5. Localized Normalized Helicity (LNH) distribution recorded in all supra-aortic vessels during cerebral perfusion for configurations 1 and 2. Colours denote LNH values, from negative (blue) to positive (red).

Tables

TABLE 1. Percentage variations of flow and velocity in configurations 1 and. Different perfusate flows are compared with a maximum perfusate flow of 1050 ml/min (case D)

Perfusate flow [ml/min]	**Configuration 1**				**Configuration 2**			
	Flow		**Velocity mean**		**Flow**		**Velocity mean**	
	[ml/min]	**[%]**	**[cm/s]**	**[%]**	**[ml/min]**	**[%]**	**[cm/s]**	**[%]**
Case A (420 *ml/min*) *vs* case D	6.9	67%	1.0	67%	7.9	65%	1.2	65%
Case B (560 *ml/min*) *vs* case D	10.9	48%	1.5	48%	13.5	40%	1.9	40%
Case C (700 *ml/min*) *vs* case D	14.1	33%	2.0	33%	16.2	28%	2.2	28%

TABLE 2. Percentage variations of WSS and WSSG in configurations 1 and 2. Different perfusate flows are compared with a maximum perfusate flow of 1050 ml/min (case D)

Perfusate flow [ml/min]	**Configuration 1**				**Configuration 2**			
	WSS		**GWSS**		**WSS**		**GWSS**	
	[N/m^2]	**[%]**	**[N/m^3]**	**[%]**	**[N/m^2]**	**[%]**	**[N/m^3]**	**[%]**
Case A (420 *ml/min*) *vs* case D	0.4	80%	411	93%	0.7	61%	463	68%
Case B (560 *ml/min*) *vs* case D	0.7	64%	1235	79%	0.9	47%	607	58%
Case C (700 *ml/min*) *vs* case D	1.0	51%	3529	60%	1.1	33%	766	47%

WSS, wall shear stress; *WSSG*, wall shear stress gradient.

Appendix

TABLE E1. Baseline characteristics of the patient population

Variable	**Total (N = 11)**
Demographic	
Age (y)	63.9 (47-79)
Sex (male)	7 (63.6)
Body mass index (kg/m^2)	27.3 (22.5-35-7)
Cardiac status	
Emergent/urgent	6 (54.5)
NYHA IV/III	2 (18.2)
Prior myocardial infarction	1 (9)
Prior percutaneous coronary intervention	1 (9)
Coronary artery disease	3 (27.3)
Left ventricular ejection fraction	53.5 (40-70)
Pulmonary hypertension (> 35 mmHg)	2 (18.2)
Comorbidities	
Hypertension	9 (81.8)
COPD	1 (9)
Peripheral vascular disease	7 (63.6)
Creatinine	89.4 (50-122)
Aortic pathology	
Aortic aneurysm	8 (72.7)
Type A Acute Aortic dissection	3 (27.3)
Operative details	
CPB time (min)	219.8 (152-298)
Cross clamp time (min)	106.9 (80-177)
Lower body circulatory arrest	22.7 (21-35)
Concomitant surgical procedures	7 (63.6)
CABG	2 (18.2)
Valve surgery	5 (45.5)
Aortic root surgery	1 (9)
Thoraflex hybrid **prosthesis** size	
26/28 x 15 cm	2
28/20 x 15 cm	1
30/32 x 10 cm	1
30/36 x 15 cm	2
30/38 x 10 cm	1
30/38 x 15 cm	1
30/40 x 10 cm	1
32/40 x 15 cm	2

Table E1. Continued

Outcomes	
Hospital mortality	1 (9)
CRRT	2 (18.2)
Re-exploration for bleeding	1 (9)
Stroke	0
Spinal cord ischemia	0

Data presented as mean and min/max values for continuous variables and n (%) for categoric variables.
CABG, coronary artery bypass grafting: *COPD*, chronic obstructive pulmonary disease; *CPB*, cardiopulmonary bypass; *CRRT*, continuous renal replacement therapy; *NYHA*, New Your Heart Association (class).