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Impact of Aortic Grafts on Arterial Pressure: A Computational Fluid Dynamics Study

O. Vardoulis [a,*], E. Coppens [b], B. Martin [a], P. Reymond [a], P. Tozzi [c], N. Stergiopulos [a]

[a] Laboratory of Hemodynamics and Cardiovascular Technology, Ecole Polytechnique Fédérale de Lausanne, Lausanne, Switzerland
[b] Faculty of Engineering, Division of Biomedical Technology, Katholieke Universiteit Leuven, Leuven, Belgium
[c] Cardiovascular Surgery Department, Centre Hospitalier Universitaire Vaudois, CHUV, Lausanne, Switzerland

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Nonlinear graft compliance;
Wave reflection;
Characteristic impedance

Abstract *Objective:* Vascular prostheses currently used in vascular surgery do not have the same mechanical properties as human arteries. This computational study analyses the mechanisms by which grafts, placed in the ascending aorta (proximal) and descending aorta (distal), affect arterial blood pressure.
Methods: A one-dimensional cardiovascular model was developed and adapted to include the graft geometry with *in vitro* measured mechanical properties. Pressure at the aortic root and haemodynamic parameters were computed and compared for a control, proximal and distal graft case.
Results: In comparison to the control case, the proximal graft increased characteristic impedance by 58% versus only 1% change for the distal graft. The proximal and distal graft increased pulse pressure by 21% and 10%, respectively.
Conclusions: The mechanisms underlying pulse pressure increase are different for proximal and distal grafts. For the proximal graft, the primary reason for pulse pressure rise is augmentation of the forward wave, resulting from characteristic impedance increase. For the distal graft, the pulse pressure rise is associated with augmented wave reflections resulting from compliance mismatch. Overall, the proximal aortic graft resulted in greater haemodynamic alterations than the distal graft. Thus, it is likely that patients who receive ascending aorta grafts are more prone to systolic hypertension and therefore deserve closer blood pressure monitoring.

* Corresponding author. O. Vardoulis, EPFL/STI/IBI/LHTC, BM 5128, Station 17, CH-1015 Lausanne, Switzerland. Tel.: +41 21 6938342; fax: +41 21 6939635.
E-mail address: orestis.vardoulis@epfl.ch (O. Vardoulis).

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Dacron grafts are currently used to replace segments of thoracic and abdominal aorta with excellent long-term results and few structural changes since their first clinical application in 1951.[1–3] The mechanical properties of woven Dacron grafts differ significantly from human aorta and aortic graft implantation changes the aortic geometry and mechanical properties, thus affecting haemodynamics. Tremblay et al.[4] demonstrated that woven Dacron graft material is up to 18 times stiffer than dilated ascending aorta, which is 1.3 times stiffer than healthy aorta under physiologic stretch. Significant differences in the material stiffness and anisotropy could affect the aortic compliance and generate changes in arterial blood pressure that eventually may affect the late outcome of aortic replacement surgery.

Ascending aorta properties have an important impact on haemodynamic parameters. Proximal aortic compliance and dimensions define the characteristic impedance of the arterial tree, and the volumetric compliance of the ascending aorta is about half of the total arterial compliance.[5] Insertion of Dacron prosthesis in the descending aorta has also been shown to cause an increase in characteristic impedance along with a decrease in total arterial compliance.[6,7] *In vivo* studies have revealed that the introduction of an aortic graft augments systolic and pulse pressure (PP), alters waveforms and increases ventricular afterload.[5,8,9] It was also found that aortic arch repair can cause shorter inflection time, leading to an increased systolic pressure in comparison to an age-matched control.[9] In combination, these haemodynamic alterations may contribute to subsequent cardiovascular complications such as hypertension, myocardial infarction and coronary heart disease.

Much remains to be understood about the global cardiovascular impact of synthetic aortic grafts with respect to hypertension, wave reflections and characteristic impedance. Human studies have examined aortic grafts in terms of biological aspects and failure modes such as leaks and graft migration.[10–13] Research has also been conducted to develop grafts with mechanical properties similar to the native tissue by a tissue engineering and novel biomaterials approach.[14,15] However, it remains unclear how graft implantation location and graft mechanical properties affect systolic arterial pressure and PP. The following computational study was designed to assess the impact of the graft implantation location and compliance mismatch on the aortic haemodynamics using a dedicated fluid dynamics model developed in our laboratory. Graft impact was assessed in terms of pressure indices, wave reflection and aortic characteristic impedance at the proximal aorta.

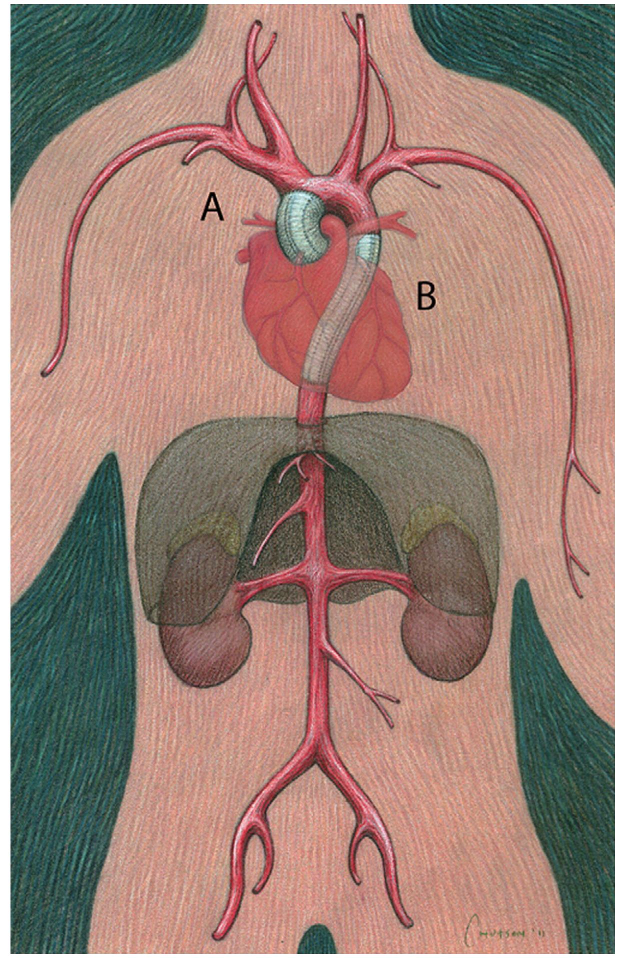

Figure 1 Schematic representation of the grafted arterial tree with the proximal graft (A) and distal graft (B) implantation location indicated (graft diameter and properties in Table 1). The proximal graft represents replacement of the aorta from the aortic valve to the brachiocephalic artery bifurcation. The distal graft represents replacement of the descending thoracic aorta.

Methods

1-D computational model

A one-dimensional (1-D) cardiovascular tree model was implemented to simulate the impact of (1) a proximal aortic graft located at the aortic root (Fig. 1(A)) and (2) a distal aortic graft located at the descending thoracic aorta (Fig. 1(B)). The simulations were compared with a control case without any graft present. The arteries were treated as straight, long, tapered cylindrical segments. The governing equations for the model are obtained by integrating the continuity and longitudinal momentum equations of the Navier–Stokes equations to obtain their 1-D form. Full details on the formulation of the 1-D model are provided by Reymond et al.[16] The model incorporated all systemic arteries greater than 2 mm in diameter, including a complete circle of Willis and distal cerebrovasculature. The arterial behaviour was considered to be nonlinear and viscoelastic using the methodology of Holenstein et al.,[17] based on the published data of Bergel.[18] Left-ventricular function was simulated with the varying elastance model described by Sagawa[19] that allows flexibility when changes occur either in cardiac parameters (e.g., heart rate, contractility, filling, etc.) or in arterial parameters (e.g., resistance, compliance, etc.). Distal vessels were terminated with three-element Windkessel models and intimal shear was modelled using the Witzig–Womersley theory. Forward wave reflections at the arterial bifurcations were minimised by adapting the characteristic impedance of downstream branches. Pressure and flow throughout the systemic tree were obtained by solving the governing equations of the model with appropriate boundary conditions, using an implicit finite difference scheme.

Dacron graft geometry and properties

This study focussed on woven Dacron grafts used in open aneurysm repair. The 1-D model was adapted to include the graft geometry and properties. Graft diameters and lengths were based on the average values of the aortic sections being replaced. Studies have shown native aortic tissue distensiblity to be from 1×10^{-3} to 5.5×10^{-3} $mmHg^{-1}$.[16] In addition, aortic grafts have been tested to have a range of distensibility from 0.3×10^{-3} to 1×10^{-3} $mmHg^{-1}$, depending on the material type and graft size. Table 1 summarises the geometric and mechanical characteristics of the simulated grafts.

The Dacron graft material is a condensation polymer produced from ethylene glycol and terephthalic acid and was considered to be nonlinearly elastic.[2] The graft compliance, C_{graft}, was determined by *in vitro* pressure–diameter tests following the methodology of Cengiz et al.[20] Based on the *in vitro* tests, the pressure–diameter relationship for a 17-mm diameter Dacron graft is provided:

$$D = a + b \cdot \left(1 - \exp\left(-\frac{P^c}{d}\right)\right) \tag{1}$$

where P is the pressure (mmHg), D is the diameter (mm), $a = 3.71$, $b = 16.33$, $c = 0.12$ and $d = 0.77$.[20–22] Distensibility is the normalised area compliance and can be related to the diameter compliance by

$$\text{Distensibility} = \frac{dA}{dP} \cdot \frac{1}{A} = 2 \cdot \frac{dD}{dP} \cdot \frac{1}{D} \tag{2}$$

Fig. 2 shows the relation between graft distensibility and pressure, compared with native aortic distensibility provided by Langewouters.[23] Compliance of the graft segments was calculated with Eq. (3), taking into account the cross-sectional area of the graft at a reference pressure of 100 mmHg. Graft viscoelasticity was considered to be zero.

$$C_{graft} = \text{Area}_{100\ mmHg} \times \text{Distensibility} \tag{3}$$

Haemodynamic parameter calculation

The impact of the graft was assessed in terms of eight parameters: (1) peak systolic pressure (P_{syst}), (2) PP, (3) PP of the forward pressure wave (PP^f), (4) PP of the backward pressure wave (PP^b), (5) reflection coefficient (Γ^*), (6) characteristic impedance (Z_c), (7) total systemic compliance (C_{syst}) and (8) pulse wave velocity (PWV). These parameters were calculated for the proximal graft, distal graft and control case, while having identical vascular geometry and properties, except for the graft location. Parameters 1 through 6 were calculated at the sinotubular junction of the proximal aorta. PWV was calculated based on the travel time of the pressure wave foot between the ascending and distal abdominal aorta (iliac bifurcation). Total systemic compliance was calculated by adding the compliance of all tree segments with the compliance of the terminal Windkessels. Characteristic impedance was calculated by averaging the input impedance modulus (Z_{in}), between the fourth and fifteenth harmonic using[24]

$$Z_{in} = \left|\frac{F(P(t))}{F(Q(t))}\right| \tag{4}$$

$$Z_c = \overline{Z}_{in} \tag{5}$$

The forward and backward pressure waves, $P^f(t)$ and $P^b(t)$, were determined by

$$P^f = \frac{P + Z_c \cdot Q}{2} \tag{6}$$

$$P^b = \frac{P - Z_c \cdot Q}{2} \tag{7}$$

The reflection coefficient was calculated with Eq. (8) as the ratio of the backward to the PP^f. $\Gamma*$ provides a simplified measure of reflection in the aorta due to the impedance mismatch between the graft and native tissue:

$$\Gamma^* = \frac{PP^b}{PP^f} \tag{8}$$

Table 1 Geometric and mechanical properties for the proximal and distal graft.

	Length (mm)	Diameter (mm)	Volume (mm^3)	Distensibility ($mmHg^{-1}$)
Proximal	90	30	6.36×10^4	0.5×10^{-3}
Distal	160	20	4.95×10^4	0.5×10^{-3}

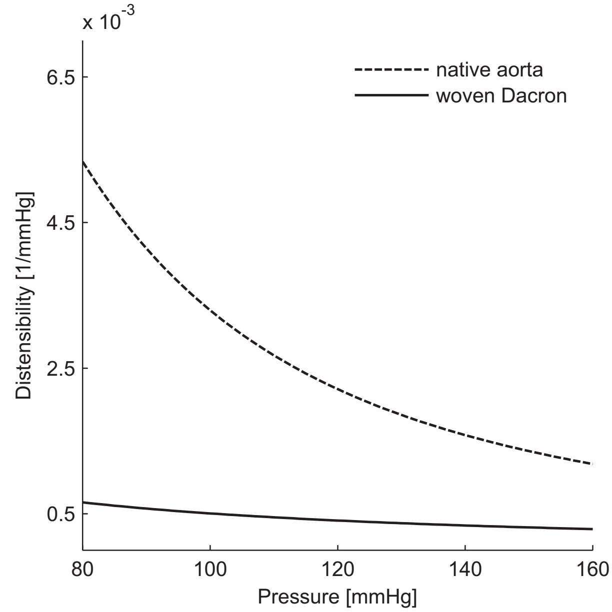

Figure 2 Distensibility plotted as a function of pressure for the a) thoracic and abdominal aorta as determined by Langewouters et al.[23] (dashed line) and b) woven Dacron aortic graft as formulated in Eq. (2) (solid line).

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Total peripheral arterial resistance was modified for each simulation so that diastolic pressure for the three cases was the same (80.6 mmHg). To achieve this, the peripheral resistance for the proximal and distal graft was increased by a factor of 1.08 in comparison to the control case.

Results

Table 2 summarises results for the control, proximal and distal graft. Aortic P_{syst} and PP increased for both graft cases. PP for the control, proximal and distal graft was 28.75, 34.76 and 31.78 mmHg, respectively. PP^f increased by 33% with a proximal graft and by 7% with a distal graft. PP^b increased for the proximal graft by 12%, whereas for the distal graft PP^b increased by 14%. Fig. 3 shows the reflection analysis results calculated for the proximal graft (left column) and distal graft (right column).

Characteristic impedance for the proximal graft was 58% greater than the control case, while for the distal case there was only a 1% decrease (Fig. 4). Total systemic compliance for the control, proximal and distal graft was 1.58, 1.25 and 1.38 ml $mmHg^{-1}$.

The reflection coefficient was altered in both graft cases in comparison to the control. For the proximal aorta replacement, the reflection coefficient decreased by 16%. On the contrary, the reflection coefficient increased by 10% for the distal aortic graft. The foot-to-foot PWV was more influenced by a distal graft than a proximal graft. PWV for the control, proximal and distal graft was 4.74, 5.16 and 5.95 m s^{-1}, respectively.

Discussion

The closer the Dacron graft is placed to the heart, the greater the impact on systolic arterial pressure and PP

The proximal graft increased aortic P_{syst} and PP to a greater degree than a distal aortic graft. The percent change of PP for the proximal graft was twice as high as the percent change of PP for the distal graft. As PP is inversely proportional to total systemic compliance, a greater increase in PP is expected to coincide with proximal aorta replacement because the proximal aorta is a major contributor to total arterial compliance and replacement of the proximal aorta leads to a larger decrease in compliance (Table 2).[5]

Table 2 Summary of results for the control, proximal and distal graft.

	Control Case	Proximal graft	Distal graft
P_{syst} [mmHg]	109.17	115.37 (+6%)	112.46 (+3%)
PP [mmHg]	28.75	34.76 (+21%)	31.78 (+10%)
PP^f [mmHg]	19.27	25.68 (+33%)	20.61 (+7%)
PP^b [mmHg]	10.70	11.95 (+12%)	12.16 (+14%)
Γ^*	0.55	0.46 (−16%)	0.60 (+9%)
Z_c [mmHg ml^{-1} s^{-1}]	0.0305	0.0483 (+58%)	0.0300 (−1%)
C_{syst} [ml/mmHg]	1.58	1.25 (−20%)	1.38 (−13%)
PWV [m/s]	4.74	5.16 (+9%)	5.94 (+25%)

Results for systolic pressure (P_{syst}), pulse pressure (PP), pulse pressure of the forward (PP^f) and backward (PP^b) moving wave, reflection coefficient (Γ^*), characteristic impedance (Z_c), total systemic compliance (C_{syst}) and pulse wave velocity (PWV).

Two mechanisms of pressure augmentation

In our simulations, a proximal graft resulted in a major increase of the proximal aortic characteristic impedance (58%), which, in turn, led to an amplification of the forward pressure wave. In consequence, the PP increase in the proximal graft case was primarily caused by forward wave augmentation (Fig. 3). Inversely, for the distal graft the characteristic impedance remained unaffected. In the latter case, the PP increase was imparted by the amplified backward pressure waves raised by compliance mismatch between the descending thoracic aorta and the graft (Fig. 3). Furthermore, the aorto-iliac PWV increase after the distal graft implantation was greater than the corresponding value in the proximal graft case. This difference in aorto-iliac PWV can be attributed to the fact that distal graft's length is nearly two times greater than the proximal graft.

The computational results compare favourably to *in vivo* and *in vitro* measurements in the literature

Our results had a number of similarities with *in vivo* and *in vitro* measurements conducted on aortic repairs. As has been reported in the work of Swillens et al.,[25] the reflection coefficient changed after implantation of a geometrically different, less compliant graft. For the simulated descending aorta replacement, the reflection coefficient increased by 9% whereas in the case of the proximal graft, the increased stiffness, along with the impedance mismatch, resulted in a decreased wave reflection coefficient. Dobson et al.[13] reported an increase of 64% in the aortic characteristic impedance after endografting the thoracic aorta in dogs. In agreement to these results, the simulations of the current study showed that a proximal graft resulted in an increase of 58% in the aortic characteristic impedance. In addition, the calculated compliance loss and pressure increase after graft implantation is consistent with the experimental findings of Ioannou et al.[5] and is greater in the case of a proximal graft because most of the arterial compliance resides in the proximal aorta. Lantelme et al.[6] reported an increase of 1 m s^{-1} in carotid femoral PWV after a 47 days' follow-up in a group of 50 patients with aortic graft replacement. The 1-D model results for aorto-iliac PWV showed an increase of 0.42 and 1.2 m s^{-1} for the proximal and distal graft, respectively (Table 2).

Utility of the 1-D model for graft design optimisation

These findings suggest that improvements are needed in prosthetic material design to mimic better native tissue. Still, the experimentation required for the initial design

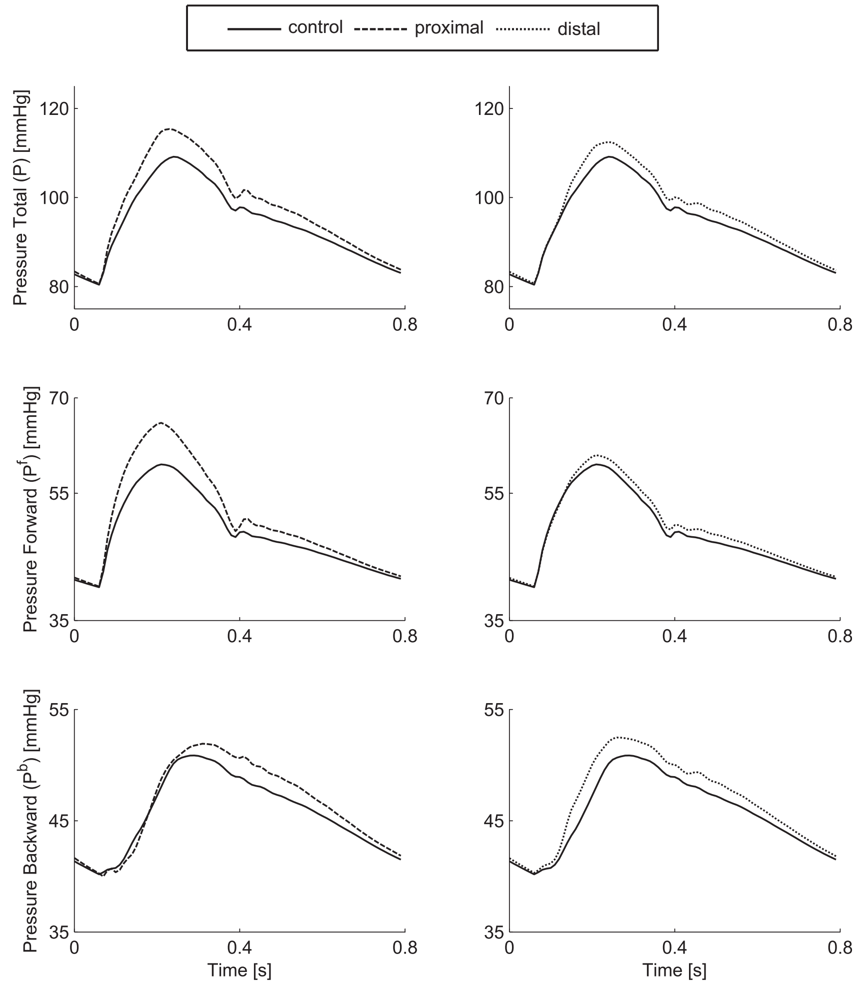

Figure 3 Comparison of the pressure waves for the proximal aortic replacement (dashed lines), distal aortic replacement (dotted lines) and the control case (solid lines). The first row depicts total pressure waves. The second and third row depict forward and backward pressure waves respectively.

phase is not practical in the clinical environment and *in vitro* testing can be costly. The presented model can be used as a low-cost, basic analysis tool to assess *in silico* graft designs in terms of geometry and materials. Comparison between the values of the graft and aortic compliance reveals that the technological distance to cover towards a fully compliant proximal graft is within 1 order of magnitude. Graft implantation location should also be taken into consideration, as aortic properties vary along the aorta and between patients. The surgeon's choice of graft material, diameter and position could play an important role with respect to compliance mismatch and haemodynamic impact.

Limitations

This study included the basic graft parameters that influence haemodynamics by taking into account graft diameter, length and nonlinear compliance. Still, there are parameters not included in our model, such as post-stitching graft diameter reduction and post-surgical graft dilation, which could also have important influence on aortic hypertension and graft failure and should be addressed in future work. The 1-D formulation of the phenomenon has some limitations, particularly when compared with 3-D methods of computational fluid dynamics. In specific, it does not account for complex shear stress distributions and secondary flow structures, such as swirls, which might influence blood pressure. The used woven Dacron graft compliance was calculated based on a different diameter graft than that simulated. Experimental data of compliance for a range of graft dimensions could further refine the results of pressure increase. Furthermore, the tissue surrounding the graft will affect the Dacron mechanical behaviour. In addition, the simulated control case represented an arterial tree with the mechanical and haemodynamic properties of a healthy person. Ideally, the 1-D model should be adapted to reflect the arterial geometry and elastic properties for an aged or diseased person. Note, the results of this study cannot be extended directly to endoprosthesis techniques, such as AEVAR/TEVAR.[26] The endovascular grafts are made of composite materials with different mechanical properties from Dacron. In addition, after endovascular repair of the aorta, the diseased artery stays in place resulting in

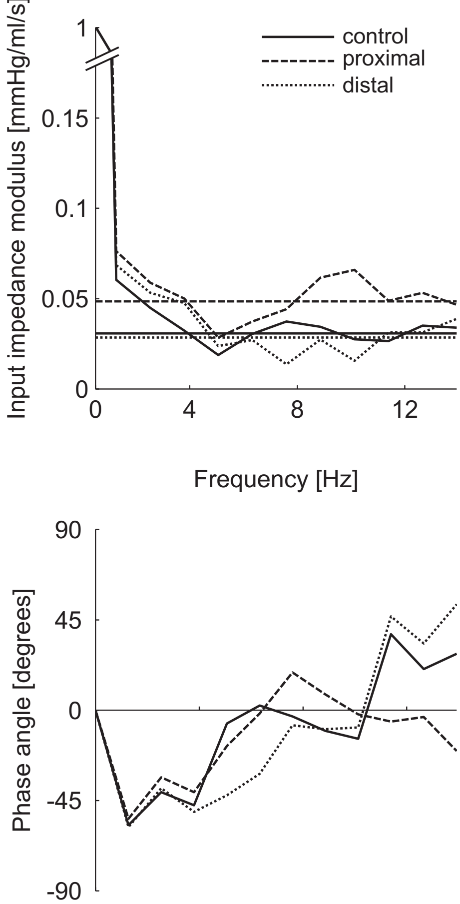

Figure 4 Input impedance modulus and phase calculated over a range of frequencies for the control case (solid lines), proximal graft (dashed lines), and distal graft (dotted lines). Horizontal lines depict values of characteristic impedance for each case.

a modification of the graft mechanical properties. Modelling of the complex interaction between endografts and aortic tissue along with branching—debranching techniques was outside the scope of this study.

Conclusions

The haemodynamic impact of a proximal and a distal aortic graft was modelled using an adapted 1-D cardiovascular model that included the graft geometry. The results showed that the rise in PP, after aortic replacement, is critically affected by the position of the graft and the level of compliance mismatch. The proximal graft increased proximal aortic PP to a greater extent, and this is achieved primarily through forward wave augmentation. The distal graft led to a smaller increase in PP and this was mainly due to increased wave reflections. To confirm the results and further strengthen our conclusions, these findings need to be compared with *in vivo* measurements of pressure and flow in patients receiving aortic grafts to confirm our theoretical results and further strengthen our conclusions. If confirmed, the results would support that ascending aortic graft recipients are potentially at a greater risk for systolic hypertension and therefore deserve closer blood pressure monitoring until more physiologically similar grafts are available.

Conflict of Interest

None.

Funding

None.

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