Journal of Biomechanics 47 (2014) 3546–3554

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Patient-specific analysis of displacement forces acting on fenestrated stent grafts for endovascular aneurysm repair

Harkamaljot Kandail [a], Mohammad Hamady [b], Xiao Yun Xu [a,*]

[a] Department of Chemical Engineering, Imperial College London, South Kensington Campus, London, UK
[b] Department of Interventional Radiology, St Mary's Hospital, Imperial College Healthcare NHS Trust, London, UK

ARTICLE INFO

Article history:
Accepted 9 August 2014

Keywords:
Abdominal aortic aneurysm
F-EVAR
Displacement force

ABSTRACT

Treatment options for abdominal aortic aneurysm (AAA) include highly invasive open surgical repair or minimally invasive endovascular aneurysm repair (EVAR). Despite being minimally invasive, some patients are not suitable for EVAR due to hostile AAA morphology. Fenestrated-EVAR (F-EVAR) was introduced to address these limitations of standard EVAR, where AAA is treated using a Fenestrated Stent Graft (FSG). In order to assess durability of F-EVAR, displacement forces acting on FSGs were analysed in this study, based on patient-specific geometries reconstructed from computed tomography (CT) scans. The magnitude and direction of the resultant displacement forces acting on the FSG were numerically computed using computational fluid dynamics (CFD) with a rigid wall assumption. Although displacement force arises from blood pressure and friction due to blood flow, numerical simulations elucidated that net blood pressure is the dominant contributor to the overall displacement force; as a result, time dependence of the resultant displacement force followed pressure waveform very closely. The magnitude of peak displacement force varied from 1.9 N to 14.3 N with a median of 7.0 N. A strong positive correlation was found between inlet cross-sectional area (CSA), anterior/posterior (A/P) angle and the peak displacement force i.e. as inlet CSA or A/P angle increases, the magnitude of resultant displacement increases. This study manifests that while loads exerted by the pulsatile flow dictates the cyclic variation of the displacement force, its magnitude depends not only on blood pressure but also the FSG morphology, with the latter determining the direction of the displacement force.

1. Introduction

Abdominal aortic aneurysm (AAA) is an irreversible dilation of the abdominal aorta. If left untreated, AAA can rupture, leading to catastrophic internal bleeding which can be fatal. Treatment options include highly invasive open surgical repair or minimally invasive endovascular aneurysm repair (EVAR). In open surgical repair, an incision is made across the abdomen and the AAA is exposed. Once exposed, the diseased arterial wall is replaced with a polyester tube. In EVAR, a small incision is made to expose the femoral artery and a catheter is then guided fluoroscopically through the femoral artery to the AAA. Once a suitable landing zone is identified, endograft is released from the catheter and it forms an artificial conduit for the blood flow thus preventing the AAA from loads exerted by the pulsatile blood flow.

Although EVAR is minimally invasive and is associated with low mortality and low morbidity rates (Greenhalgh., 2004), some patients are unsuitable for this procedure due to complex AAA morphological features, such as short and/or angulated aortic neck. Fenestrated-endovascular aneurysm repair (F-EVAR) was introduced to overcome the limitations of standard EVAR. A customised endograft known as Fenestrated Stent Graft (FSG) has been used in F-EVAR. Depending on patient-specific AAA anatomy, FSG comprises of fenestrations in the main endograft which are aligned with ostia of the visceral arteries, a separate stent graft is then introduced via these fenestrations that protrude into renal arteries and/or superior mesenteric artery. These fenestrated vessels exclude the aneurysm sac from any blood flow and also provide the additional anchoring force which is required in order to ensure safe deployment of the endograft. One of the commercially available FSG which has recently been used in F-EVAR is the Anaconda™ FSG (Vascutek, Inchinnan, UK). This device shows promising early results but the question of durability remains unanswered, namely migration, i.e. movement of the proximal portion of the stent (Bungay et al., 2011). One of the primary factors responsible for migration is the resultant displacement force acting on the FSG. Although standard EVAR has been studied extensively in the past (Figueroa et al., 2010; Frauenfelder et al., 2006; Li and Kleinstreuer, 2005; Molony et al., 2010; Morris

* Corresponding author. Tel.: +44 20 7594 5588.
E-mail address: yun.xu@imperial.ac.uk (X.Y. Xu).

http://dx.doi.org/10.1016/j.jbiomech.2014.08.011
0021-9290/

et al., 2004), little has been done on FSGs (Avrahami et al., 2012; Sun and Chaichana, 2009, 2010). Sun and Chaichana (2009, 2010) examined blood flow patterns in the renal arteries of a patient after F-EVAR, but they did not evaluate the displacement force acting on the FSG. The displacement force experienced by FSG was analysed by Avrahami et al. (2012); however their analysis was based on a simplified FSG with ideal inflow and outlet boundary conditions. A more recent study carried out by Georgakarakos et al. (2014) discussed the displacement forces acting on idealised FSGs and the influence of misaligned renal fenestrations on these forces. The present study is the first patient-specific analysis of displacement forces acting on the Anaconda™ FSG. Quantification of the displacement force is important because its magnitude determines the risk of migration of the device, which in turn affects the long-term outcome of F-EVAR. A severely migrated FSG can lead to plethora of future complications such as occlusion of one of the visceral arterial branches or Type I endoleaks.

This study included six patients with different AAA morphologies. Pre- and post-operative geometries were reconstructed for each patient from the corresponding computed tomography (CT) images, and physiologically realistic inflow and outlet boundary conditions were incorporated. In addition to evaluation of the magnitude and direction of displacement forces, blood flow patterns in the pre- and post-operative AAAs were compared.

2. Methodology

2.1. Anatomical data

Pre- and post-operative contrast enhanced CT scans for six patients were obtained from St Mary's Hospital, London. All these patients were implanted with a commercially available Anaconda™ AAA stent graft (Vascutek, Inchinnan, UK). Out of these six patients, five were treated with FSGs and one with a standard non-fenestrated stent graft. The criteria on which these patients were selected are summarised in Table 1. Internal institutional review board approval was not required for this limited retrospective and anonymised study.

2.2. Reconstruction of patient-specific models

Transverse images acquired during CT scans were stored in DICOM (Digital Imaging and Communications in Medicine) format and processed using Mimics (Materialise, Leuven, Belgium). Fig. 1 shows the reconstructed pre- and post-operative geometries, which were stored in STL (Stereo lithography) format. Lateral as well as anterior/posterior (A/P) aortic neck angulations were measured following the approach adopted by Molony et al. (2010), as shown in Fig. 2(a) and (c). Anaconda™ FSG is made of woven polyester supported by nitinol wires which are clearly visible on contrast-enhanced CT scans. The inlet of each model was defined at the location where the stent graft fixation hooks first appear, as shown in Fig. 2(b).

2.3. Flow model and assumptions

The continuity and Cauchy momentum equations were used to describe the 3D incompressible pulsatile blood flow in the lumen and these equations are:

$$\textit{Continuity}: u_{i,i}=0 \tag{1}$$

$$\textit{Momentum}: \rho\left(\frac{\partial u_i}{\partial t}+u_j u_{i,j}\right)=-p_i+\tau_{ij,j} \tag{2}$$

$$\textit{Stress tensor}: \tau_{ij}=\eta\dot{\gamma}_{ij} \tag{3}$$

where u_i is the velocity vector, p is the pressure scalar, ρ is the fluid density (1060 kg/m^3), τ_{ij} is the shear stress tensor and $\dot{\gamma}_{ij}$ is the shear rate tensor. As patients were lying down when CT scans were taken, gravitational effects were ignored. The Quemada viscosity model was adopted to describe the non-Newtonian viscous behaviour of blood.

$$\textit{Apparent viscosity}: \eta=\frac{\mu_P}{[1-(1/2)(k_0+k_\infty(\sqrt{\dot{\gamma}/\dot{\gamma}_c})/(1+\sqrt{\dot{\gamma}/\dot{\gamma}_c})\Psi]^2} \tag{4}$$

where μ_p is the viscosity of blood plasma (1.2×10^{-3} Pa s), Ψ is haematocrit (0.45), $\dot{\gamma}/\dot{\gamma}_c$ is relative shear rate and k_0, k_∞ and $\dot{\gamma}_c$ are constants with values of 4.33, 2.07 and 1.88 s^{-1} (Neofytou, 2004). The flow was assumed to be laminar since the peak Reynolds number at maximum flow was found to be in the range of 1149–1700 for the six pre- and post-operative patient-specific models concerned.

Table 1
Criteria for selection of patients.

	Number of fenestrations	Specific aortic neck morphology
Patient 1	2	Straight aortic neck
Patient 2	3	Straight aortic neck
Patient 3	2	Severe aortic neck angulation
Patient 4	0	Angulated aortic neck
Patient 5	1	Angulated aortic neck
Patient 6	3	Angulated aortic neck

2.4. Boundary conditions

Physiologically realistic volumetric flow rate and pressure waveforms were adopted from the literature (Olufsen et al., 2000; Taylor et al., 1998) and described using Fourier series. It was assumed that 15% of the inlet flow entered each fenestrated vessel; the corresponding waveforms are shown in Fig. 3. Pulsatile velocity profiles, derived from the Womersley solution (Womersley, 1955) were imposed at the inlet as:

$$w(r,t)=\frac{2a_0}{\pi R^2}\left[1-\left(\frac{r}{R}\right)^2\right]+\Re\sum_{n=1}^{N}\left\{\frac{a_n\ \cos(n\omega t)+b_n\ \sin(n\omega t)}{\pi R^2}\times\left[\frac{1-J_0(\alpha_n(r/R)i^{3/2})/(J_0(\alpha_n i^{3/2}))}{1-2J_1(\alpha_n i^{3/2})/((\alpha_n i^{3/2})(J_0\alpha_n i^{3/2}))}\right]\right\} \tag{5}$$

where a_n and b_n are the Fourier series coefficients representing the inlet volumetric flow rate, $\omega=2\pi/T$ with T being the duration of the cardiac cycle, R is the inlet radius, J_0 and J_1 are Bessel functions of the first kind of order 0 and 1, respectively. The non-dimensional parameter $\alpha=R\sqrt{\omega/\nu}$ is called the *Womersley number* and it represents the ratio of transient forces, originating from the pulse wave, to the viscous force. v in α is the apparent kinematic viscosity of the blood. In addition to α, α_n is defined as $R\sqrt{(n\omega)/\nu}$. The Womersley velocity profiles were obtained using MATLAB (MathWorks, Natick, MA, USA) and a user function was written to couple these profiles with an external numerical flow solver. A representative pressure waveform was prescribed at the outlets in the iliac arteries, while flow waveforms given in Fig. 3 were imposed at the fenestrated vessel outlets. No slip conditions were specified at the wall, which were assumed to be rigid.

2.5. FSG dynamics

FSG experiences time dependent displacement force due to pressure and friction exerted by blood flow on its walls. Displacement forces acting on the FSG over a cardiac cycle are arguably the most important factor in determining the risk of device migration and future complications such as Type I endoleaks. This displacement force is a 3D force and can be calculated by taking an area integral of the net pressure and WSS over the entire wall of the FSG in x, y and z directions:

$$\textit{Displacement force}: F_{d,i}=\int_{A,i}pdA+\int_{A,i}\left(-\eta_w\frac{\partial u_t}{\partial n}\bigg|_{n=0}\right)dA \tag{6}$$

where $\int_{A,i}pdA$ is the pressure force and $\int_{A,i}(-\eta_w(\partial u_t/\partial n|_{n=0}))dA$ is the WSS force. The resultant displacement force (F_d) and direction cosines $(\theta_\alpha,\theta_\beta,\theta_\gamma)$ can then be calculated as:

$$\textit{Resultant displacement force}: |F_d|=\sqrt{(F_{d,x})^2+(F_{d,y})^2+(F_{d,z})^2} \tag{7}$$

$$\textit{Direction cosines}: \theta_\alpha=\cos^{-1}\left(\frac{F_{d,x}}{|F_d|}\right)\quad \theta_\beta=\cos^{-1}\left(\frac{F_{d,y}}{|F_d|}\right)\quad \theta_\gamma=\cos^{-1}\left(\frac{F_{d,z}}{|F_d|}\right) \tag{8}$$

Angles θ_α, θ_β and θ_γ are defined with respect to x, y and z axis, respectively.

3. Numerical approach

Due to the complex pre- and post-operative AAA morphologies, 3D unstructured meshes were generated which comprised a combination of tetrahedral and prism elements. A commercial mesh generator ANSYS ICEM CFD (ANSYS, Canonsburg, PA, USA) was used to generate meshes for patient-specific geometries. In order to ensure mesh quality, aspect ratio and expansion factor were kept less than 50 and 20, respectively while orthogonality angle was always larger than 20°. The governing flow equations were solved numerically using ANSYS CFX (ANSYS, Canonsburg, PA, USA), which uses finite volume method to discretise the governing partial differential equations into a set of linear algebraic equations and these linear equations are then solved using an algebraic multigrid method. The convergence criterion was