

Numerical Study of Haemodynamics Behaviour in Normal and Single Stenosed Renal Artery using Fluid – Structure Interaction

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ABSTRACT

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The haemodynamics of cardiovascular diseases such as atherosclerosis can be demonstrated using an advance technique of Fluid-Structure Interaction Simulation. This technique have the ability of studying the complex interaction of blood flow through elastic arteries using the computational aid. The aim of present study is to investigate the haemodynamics behaviour of blood flow in a 3D model of an idealistic abdominal aorta with renal branches with the normal and single stenosed cases under normal and high blood pressure condition. The 3D abdominal aorta model is generated using single slice technique base on Computed Tomography (CT) image. Numerical analysis is performed using FSI solver of two-way coupling in ANSYS-17. The blood flow is assumed to be homogeneous and Newtonian, incompressible, while linear elastic behaviour is assumed for artery wall. A transient analysis of three pulse cycle is conducted in order to investigate the hemodynamics parameters of flow velocity, pressure contour, Wall Shear Stress (WSS), arterial wall deformation and von-Mises stress at the bifurcation and critical zones for both cases under.

Keywords:

Haemodynamics, Renal Artery, Exercise and Resting condition, Atherosclerosis

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1. Introduction

Atherosclerosis is a serious complication that can be described as the leading cause of death worldwide. Atherosclerosis occurs due to the result of plaques formation of accumulation fatty substance inside the lumen which lead to the hardened of arteries. It is generally occurs nearly to bifurcations and curvatures of artery wall which resulting in obstruction of blood flow. The risk of atherosclerosis may increase due to the buildup of plaque on the arterial wall, thereby narrowing the

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arteries and hence, these cause the plaque clogged up the artery and disrupting the flow of blood around the human body. This type of disease occurs at all large and medium-sized arteries, including carotid manifested brain, coronary manifested heart, peripheral manifested leg, arms and lower body as well as renal arteries supply blood to kidney. Renal artery stenosis is widely known as the leading cause of kidney failure. The stenosis restricts the blood supply to the kidney which resulted in narrow size of the arteries due to less volume of oxygenated blood to kidney or may ultimately lead to the failure of kidney.

Recently, a lot of studies have demonstrated the blood flow behavior using computational method to explore the complex interactions of hemodynamics in cardiovascular disease [1]. The critical anatomical regions such as arterial bifurcation/ branching or curvature has higher potential of developing atherosclerosis due to the existing of complex flow depending to locations [2]. The emerging coupling technique of clinical imaging such as Magnetic Resonance Imaging (MRI), Computed Tomography (CT) and Ultrasound Doppler Imaging with numerical simulation such as Fluid-Structure Interaction (FSI) have been developed by researchers in recent years [3,4]. This helps researchers to understand detailed of blood flow behaviour especially the mechanism of stenosis development and progression in patients [5,6]. Hence, physiological parameters like flow velocity, pressure, wall shear stress (WSS), arterial wall deformation and von-Mises stress should assessed around the region of stenosis.

In the present work, FSI simulation was carried out to investigate the inter-relationship between normal case and stenosed case of renal artery during normal blood pressure (NBP) and high blood pressure (HBP) conditions. A 3D models of abdominal aorta with renal branches is constructed from a single mid-slice of CT image using MIMICS software and FSI simulation was accomplished using ANSYS 17. The simulation provides advantages for the medical expertise to understand blood flow's behavior in term of velocity, pressure, wall shear stress (WSS), arterial wall deformation and von-Mises stress due to aggravated of stenosis at the blood vessel.

2. Methodology

In the present study, blood flow is assumed to be Newtonian fluid, incompressible [7]. The dynamic flow is governed by the continuity and Navier-stokes equations [8], [9], respectively as

$$\nabla \cdot u = 0 \quad (1)$$

$$\rho \left(\frac{\partial u}{\partial t} + u \cdot \nabla u \right) = -\nabla p + \mu \nabla^2 u \quad (2)$$

Where p is the pressure, ρ is the density, u is the velocity and μ is the viscosity.

Modified momentum equation is adopted in addition to continuity equation as shown in Equation 3.

$$\frac{\partial}{\partial t} \int_{\Omega} \rho \partial \Omega + \int_S \rho (v - v_b) \cdot n \partial S = \int_S (\tau_{ij} i_j - P i_i) \cdot n \cdot \partial S + \int_{\Omega} b_i \partial \Omega \quad (3)$$

Where ρ is the density, u is the velocity vector, τ is the stress tensor, P is the pressure, u_b is the grid velocity, b_i is the body force at time, t .

Artery wall is assumed to be linearly elastic, isotropic, incompressible and homogeneous (Fung, 1984). Structural solution considering the transient behavior is described by the Equation 4.

$$[M]\{\ddot{U}\} + [C]\{\dot{U}\} + [K]\{U\} = \{F^a\} \quad (4)$$

Where M is the structural mass matrix, C is the structural damping matrix, K is structural stiffness matrix, F_a is the applied load vector and \ddot{U} , \dot{U} and U represents acceleration, velocity and displacement vector.

Two-way sequentially coupled transient FSI analysis is performed using FSI solver in ANSYS 17, a numerical simulation software. FSI solver solves fluid and solid domain separately using ANSYS Fluent and ANSYS Mechanical respectively. Pressure loads from initial ANSYS Fluent solution is transferred to the solid domain through FSI interface and later ANSYS structural domain is solved. Further details of FSI solver are described in detail as observed [10].

In the current study, 3D geometry of idealistic abdominal aorta with the renal branch is constructed from a single mid-slice of CT image and the configuration of the model is shown in Figure 1. Figure 2 (a) shows the 3D model of normal idealistic abdominal model with renal branches, while (b) shows the single stenosis in left artery.

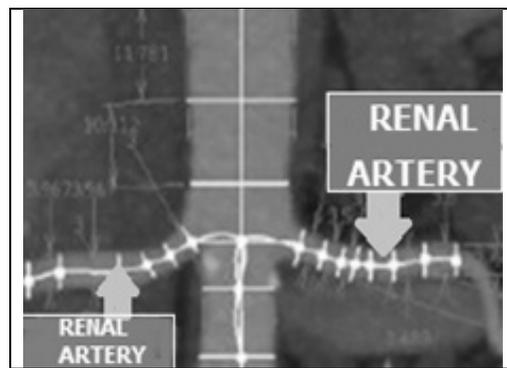


Fig. 1. CT data details Geometry from CT Slice

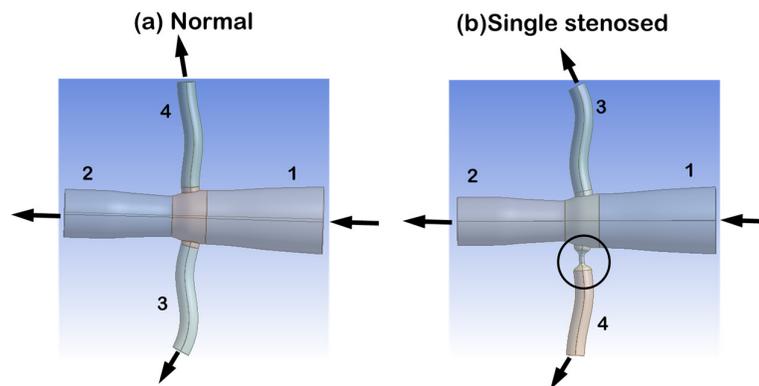


Fig. 2. Geometric description of various CFD models

3D fluid and solid models of abdominal aorta with renal branches are meshed with 85000 and 55000 hexahedral elements. Grid independency study is carried out under steady state condition at early systolic velocity. Flow variables such as velocity, WSS and pressure are monitored by maintaining the grid quality. At inlet and outlet of the fluid model, time varying pulsatile periodic velocity and pressure is applied respectively as shown in Figure 3 and Figure 4.

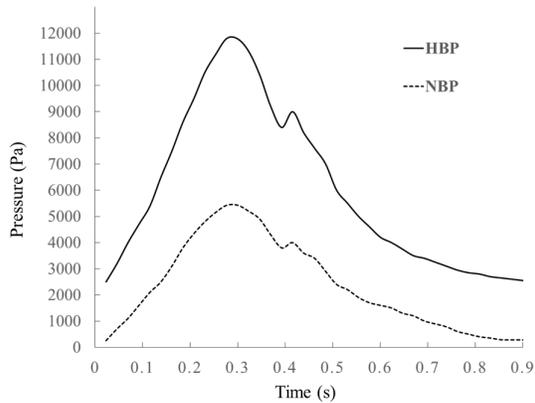


Fig. 3. Pulsatile pressure applied at outlet

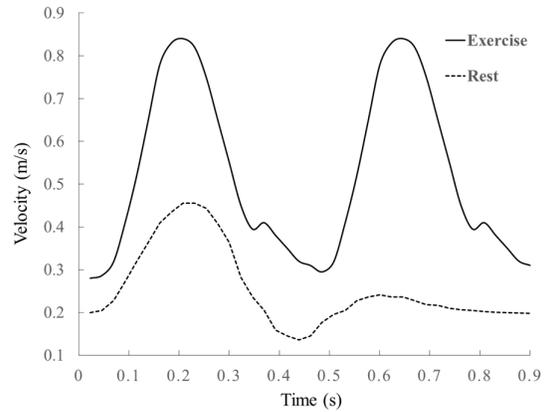


Fig. 4. Pulsatile velocity applied at inlet

3. Results

Numerical simulation of normal and single stenosed models with normal blood pressure (NBP) and high blood pressure (HBP) are carried out for three pulse cycle and results obtained in the last cycle is considered for the investigation. The haemodynamics parameters such as velocity, pressure, WSS, arterial wall deformation and von-Mises stress are studied at specific instants of pulse cycle like early systole (ES), peak systole (PS), early diastole (ED) and late diastole (LD). These parameters varies with time due to the pulsatility of the flow waveform and the maximum value generally occurs at the peak systole when the inflow is maximum.

Figure 5 shows the comparison of velocity contour and maximum velocity in normal and stenosed case during NBP and HBP. It describes the flow separation such that the flow divides into two streams with maximum velocity at the distal wall of the renal bifurcation and slower moving fluid on the proximal wall [11]. It is observed that there is no huge differences of velocity contour between the normal case and stenosed case except at the stenosed renal branches where the velocity of blood flow reduced due to the acute diameter of stenosed. Referring to the graph, Stenosed-NBP shows the highest value of maximum velocity follows by the Stenosed-HBP, Normal-NBP and Normal-HBP at each state.

Figure 6 demonstrates the pressure distribution during peak systole in NBP and HBP condition in normal and stenosed case. Upstream abdominal aorta performed higher pressure distribution in comparison with downstream side. Stenosed-HBP and Normal-HBP showed higher pressure build up in upstream side of abdominal aorta and at the renal bifurcation tip unlike Normal-NBP and Stenosed-NBP as in Figure 6. However, HBP case had larger pressure range over NBP condition as that depicted from outlet pressure profile.

At renal walls, WSS distribution is found to be low along the proximal wall and high WSS at distal renal wall [12]. It is also observed that distal wall in neighbourhood of bifurcation experience high WSS, whereas proximal walls nearby the flow separation region suffer from relatively low WSS as clearly in Figure 7. Maximum WSS observed at renal bifurcation in distal side is relatively more intense at stenosed case compared to normal case, respectively. Quantitatively, the maximum WSS value of stenosed case for NBP and HBP performed higher value than normal case for both condition, respectively.

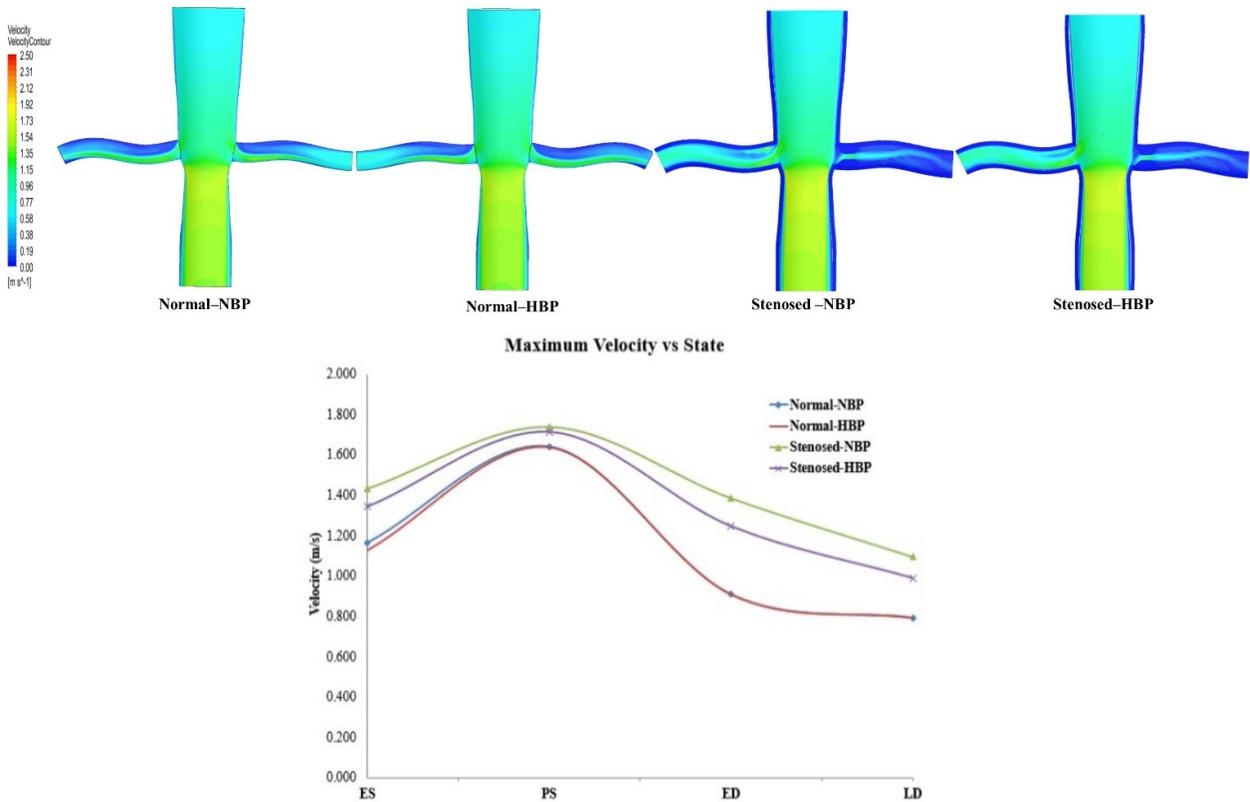


Fig. 5. Comparison of velocity contour and maximum velocity in normal and stenosed case during NBP and HBP

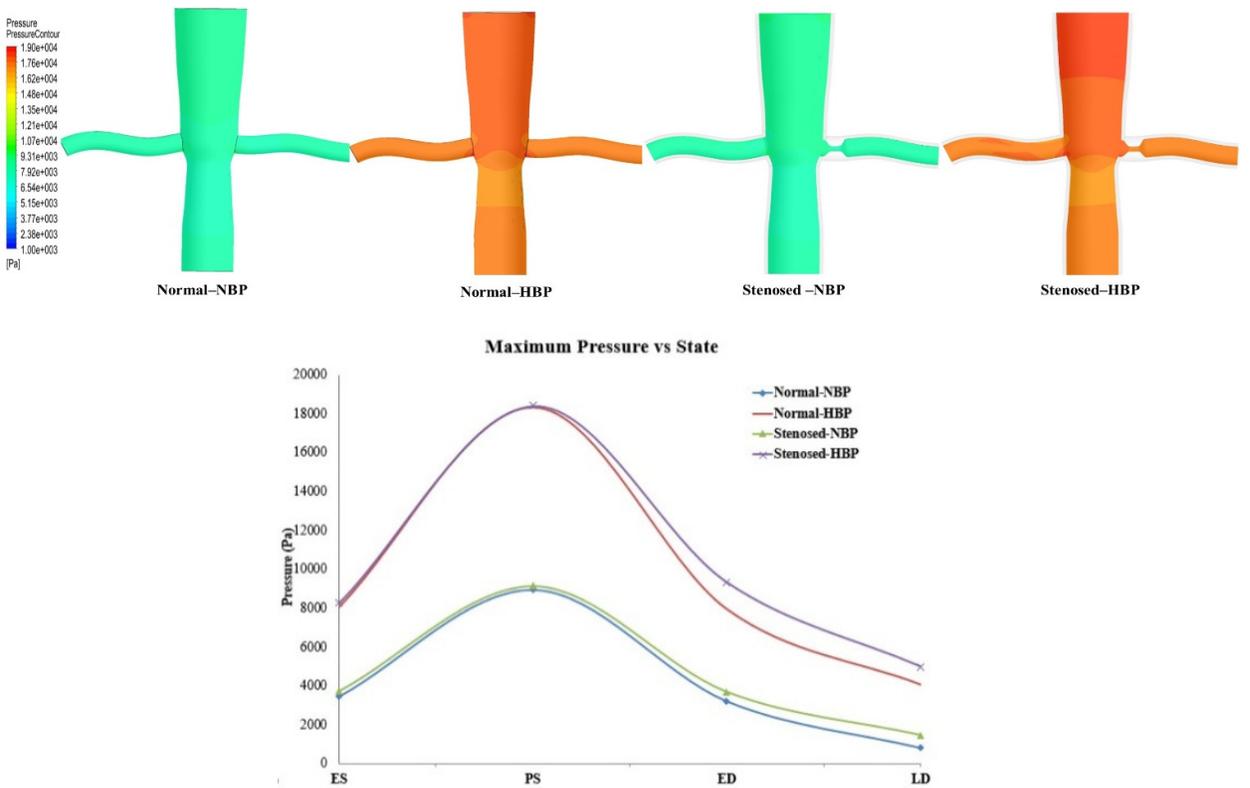


Fig. 6. Comparison of pressure contour in normal and stenosed case during NBP and HBP

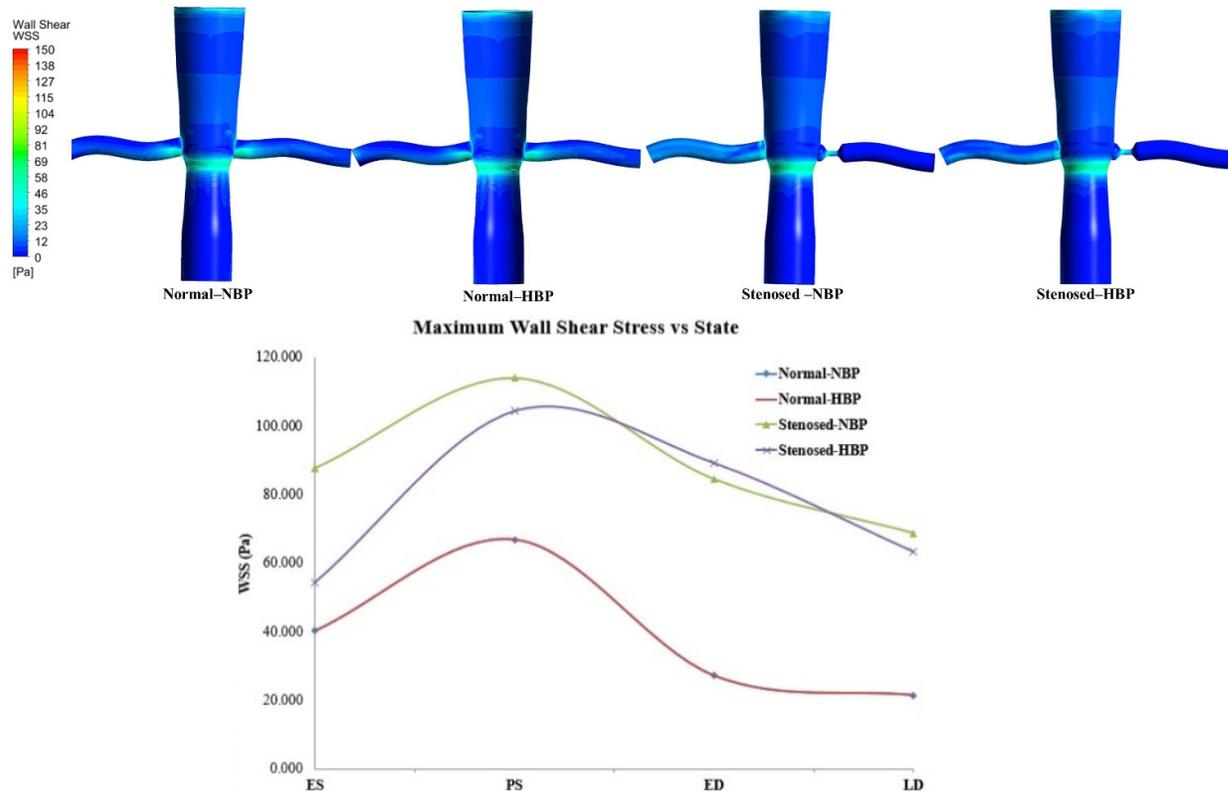


Fig. 7. Comparison of WSS in normal and stenosed case during NBP and HBP

The total arterial wall deformation behavior is compared during peak systole for normal and stenosed cases during NBP and HBP cases as in Figure 8. It is observed that upstream side of abdominal aorta deform significantly unlike downstream side. In NBP case, mild deformation is noticed at entry region of renal artery bifurcation on distal wall side and quite low at proximal wall side for normal and stenosed cases. However, in HBP case, wall deformation behavior extends to a larger length especially for stenosed case compared to normal case. The graph of maximum WSS shows that the Stenosed-HBP produced the highest value followed by Normal-HBP, Stenosed-NBP and Normal-NBP, respectively.

Figure 9 compares the von-Mises stress distribution during peak systole in NBP and HBP for normal and stenosed cases. In cardiac pulse cycle, stress distribution is found to be more visible during peak systole and flow deceleration, however during late diastole stress distribution reduces significantly. Upstream region of abdominal aorta is found to have slightly larger stress distribution unlike downstream side. Stress distribution is found to be negligible at the renal artery bifurcation in both branches of NBP condition for normal and stenosed case. However, during HBP condition, mild stress distribution is found to spread the entire renal branches, both in normal and stenosed cases.

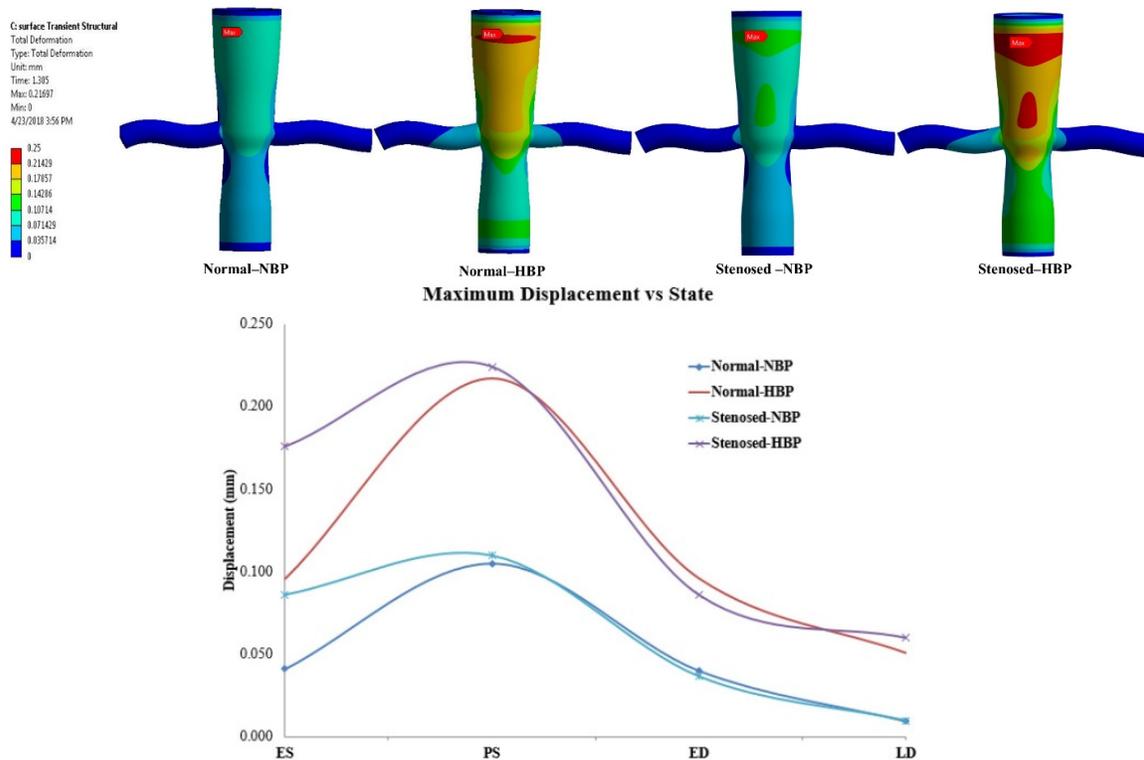


Fig. 8. Comparison of arterial wall deformation in normal and stenosed case during NBP and HBP

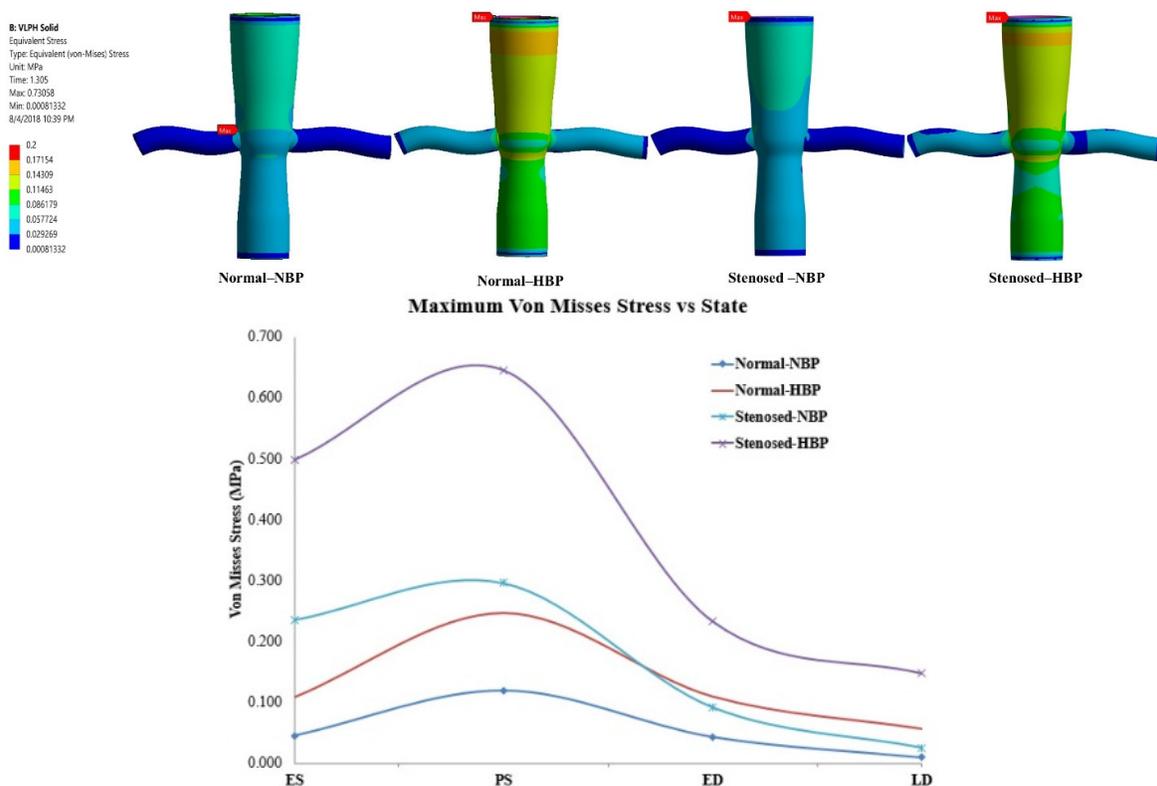


Fig. 9. Comparison of arterial von-Mises stress in normal and stenosed case during NBP and HBP

4. Conclusions

The haemodynamics behaviour of blood flow in a 3D model of an idealistic abdominal aorta with renal branches with the normal and single stenosed cases under NBP and HBP condition is carried out in this present study with the aid of two-way FSI simulation. Reduction of blood flow distribution is observed at stenosed renal branch compared to normal case due to the acute luminal diameter of the blood vessel. However, upstream region of abdominal aorta is found have slightly larger pressure distribution, arterial deformation and stress distribution compared to downstream side of abdominal aorta especially for Normal-HBP and Stenosed-HBP cases. Effect of HBP on renal flow is clearly observed in which Stenosed-HBP indicated the highest maximum value of velocity, pressure, WSS arterial wall deformation and von-Mises stress distribution compared to Stenosed-NBP, Normal-HBP and Normal-NBP. The present study is demonstrates the fundamental aspects of haemodynamics in idealized abdominal aorta with renal branches and can be further extended to patient specific cases.

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References

- [1] Taylor, Charles A., Thomas JR Hughes, and Christopher K. Zarins. "Finite element modeling of three-dimensional pulsatile flow in the abdominal aorta: relevance to atherosclerosis." *Annals of biomedical engineering* 26, no. 6 (1998): 975-987.
- [2] Fung, Y. "C, 1984." *Biodynamics: Circulation*.
- [3] Kagadis, George C., Eugene D. Skouras, George C. Bourantas, Christakis A. Paraskeva, Konstantinos Katsanos, Dimitris Karnabatidis, and George C. Nikiforidis. "Computational representation and hemodynamic characterization of in vivo acquired severe stenotic renal artery geometries using turbulence modeling." *Medical engineering & physics* 30, no. 5 (2008): 647-660.
- [4] Marshall, Ian, Shunzhi Zhao, Panorea Papathanasopoulou, Peter Hoskins, and X. Yun Xu. "MRI and CFD studies of pulsatile flow in healthy and stenosed carotid bifurcation models." *Journal of biomechanics* 37, no. 5 (2004): 679-687.
- [5] Suh, Ga-Young, Andrea S. Les, Adam S. Tenforde, Shawn C. Shadden, Ryan L. Spilker, Janice J. Yeung, Christopher P. Cheng, Robert J. Herfkens, Ronald L. Dalman, and Charles A. Taylor. "Quantification of particle residence time in abdominal aortic aneurysms using magnetic resonance imaging and computational fluid dynamics." *Annals of biomedical engineering* 39, no. 2 (2011): 864-883.
- [6] Suh, Ga-Young, Andrea S. Les, Adam S. Tenforde, Shawn C. Shadden, Ryan L. Spilker, Janice J. Yeung, Christopher P. Cheng, Robert J. Herfkens, Ronald L. Dalman, and Charles A. Taylor. "Hemodynamic changes quantified in abdominal aortic aneurysms with increasing exercise intensity using MR exercise imaging and image-based computational fluid dynamics." *Annals of biomedical engineering* 39, no. 8 (2011): 2186-2202.
- [7] Razavi, Seyed Esmail, Ramin Zanbouri, and Omid Arjmandi-Tash. "Simulation of blood flow coronary artery with consecutive stenosis and coronary-coronary bypass." *BioImpacts: BI* 1, no. 2 (2011): 99.
- [8] Soulis, Johannes V., George D. Giannoglou, Yiannis S. Chatzizisis, Kypriani V. Seralidou, George E. Parcharidis, and George E. Louridas. "Non-Newtonian models for molecular viscosity and wall shear stress in a 3D reconstructed human left coronary artery." *Medical engineering & physics* 30, no. 1 (2008): 9-19.
- [9] Ferziger, Joel H., and Milovan Perić. "Basic Concepts of Fluid Flow." In *Computational Methods for Fluid Dynamics*, pp. 1-20. Springer, Berlin, Heidelberg, 2002.
- [10] ANSYS Release 17.0 Documentation (2016), ANSYS Company, Pittsburgh, PA.
- [11] LIANG, Fuyou, Ryuhei YAMAGUCHI, and Hao LIU. "Fluid dynamics in normal and stenosed human renal arteries: an experimental and computational study." *Journal of Biomechanical Science and Engineering* 1, no. 1 (2006): 171-182.
- [12] Taylor, Charles A., Thomas JR Hughes, and Christopher K. Zarins. "Effect of exercise on hemodynamic conditions in the abdominal aorta." *Journal of vascular surgery* 29, no. 6 (1999): 1077-1089.