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# VASCULAR RECONSTRUCTION MODELLING OF LUMEN-ADAPTED ARTERIES WITH STIFFENED GRAFTS

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**Key words:** Shape Optimization, Genetic Algorithms, Biomedical Engineering, ANN Applications.

**Abstract.** Optimization of prosthetic graft configuration with regard to blood dynamics is the major target of this research. Hemodynamic simulations of idealized arterial bypass systems are acquired using a finite element arterial blood simulator exhibiting hemodynamic flow changes due to compliance differences of a stiff graft and an elastic arterial wall. An artificial neural network simulating hemodynamic specific conditions is developed in order to reduce computational time. Optimal graft configurations are searched using a multi-objective genetic algorithm. An optimal set of solutions are presented and analyzed.

## 1 INTRODUCTION

Vascular grafts are special tubes that serve as replacements for damaged blood vessels. When suitable autologous veins are unavailable, prosthetic graft materials are used for peripheral arterial revascularisations. Research studies over the last three decades have established that hemodynamic interactions with the vascular surface as well as surgical injury are inciting mechanisms capable of eliciting distal anastomotic intimal hyperplasia (IH) and ultimate bypass graft failure. Compliance and calibre mismatch between native vessel and graft contributes towards poor long term patency [1].

The ideal vascular bypass graft would replicate the mechanical properties of native artery perfectly. Research study of dynamic arterial wall properties of large arteries such as the carotid and femoral arteries is becoming more common. Using non-invasive techniques the maximum and minimum arterial diameters and the intima-media thickness (IMT) at the point of maximum diameter and minimum diameter have been determined over the cardiac cycle.

The pressure wave following the ejection of blood by the heart is gradually conveyed to the periphery. Close to the heart the wave velocity is of the order of 5 m/s and gradually increases towards the periphery [2]. The arterial diameter and intima-media thickness values can be used together with the blood pressure measurements to calculate several standard arterial stiffness indices [3]. Nevertheless, in vitro and in vivo experiments have demonstrated that the diameter-pressure relationship exhibits an exponential characteristic [4], which can be accounted for by replacing the pulse pressure in the distensibility coefficient by the logarithm of the ratio of systolic and diastolic pressure [5].

Computational approaches have been used simulating blood flow through idealized bypass models [6, 7, 8]. They exhibit particular patterns characterized by the presence of recirculation zones and secondary flows in certain regions. Pulsatile simulations of artery-graft systems show elevated and negative wall shear stresses at the toe, heel and hood regions of the anastomosis. The region where the wall shear stress (WSS) is negative corresponds to recirculation regions within the artery. Regions of reverse flow are usually associated with local deposit of particles, which results in a blockage of the artery. Therefore, potential artery-graft anastomosis design improvements that reduce the amount of wall shear stress and recirculation zones may have to be performed in order to increase the clinical success of vascular bypass grafts.

In this project a developed multi-objective genetic algorithm [6] is considered in order to reach optimal graft geometries for idealized arterial bypass systems of fully occluded host arteries. Genetic algorithms require a large number of computer simulations. So, an artificial neural network (ANN) is developed to efficiently calculate specific outputs associated with blood flow for predefined graft geometries. Input and target data have been acquired using a modified version of a finite element (FE) arterial blood simulator previously developed and tested considering fully unsteady incompressible Navier-Stokes equations and a three-dimensional geometry [9, 10].

## 2 MULTI-OBJECTIVE GENETIC ALGORITHM

Geometry plays the key role in determining the nature of hemodynamic patterns. This investigation will address rigid sinusoidal grafts with walls drawn by sine curves. In order to understand the dependence of the bypass blood flow on arterial wall variability, finite element model simulations were performed using a modified version of a previously developed code [9, 10]. Figure 1 presents the deformable model that includes both the proximal and distal bypass sections in order to analyze the flow development along the entire bypass.

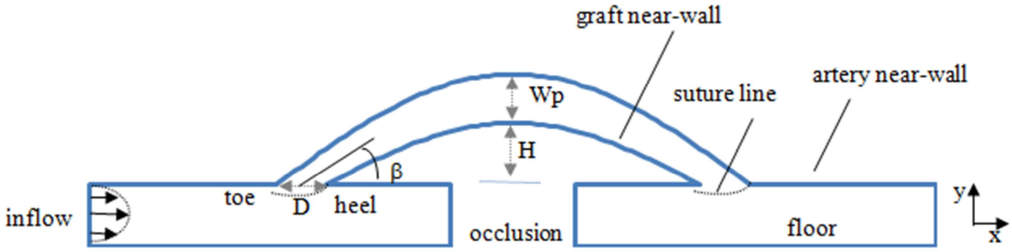


Figure 1: Anastomotic configuration and nomenclature of the graft/artery geometry

For modeling purpose the simplified arterial graft prosthesis is a tubular vessel disposed around a longitudinal axis as described in the bypass model given in Figure 1. Each design vector $\boldsymbol{b}$ has four geometric components $\boldsymbol{b} = (H, \beta, Wp, D)$ as displayed in Figure 1: the distance from the near wall of the graft to the near wall of the artery $H$, the junction angle $\beta$, the width of the prosthesis at its longitudinal symmetric line $Wp$ and the suture line dimension $D$. The host artery is assumed to be a fully stenosed conduit, simulated using two cylindrical tubes of 6 millimeter diameter, the proximal host artery before the obstruction and the distal host artery after obstruction. The graft is symmetric and meets the host artery with a side-to-end proximal anastomosis and an end-to-side distal anastomosis. As usually adopted by most previous investigations, vessels are assumed to be impermeable tubes.

A general multi-objective optimization seeks to optimize the components of a vector-valued objective function. For the shape optimization application presented here, the genetic algorithm (GA) begins by randomly setting an initial population of possible individuals, where individuals represent different graft geometries. The successive populations maintain the same number of individuals as it evolves throughout successive generations. Each individual is referred as a chromosome containing design variable values referred as genes of the chromosome over which genetic operators are applied. Operators such as selection/crossover, mutation and elimination supported by an elitist strategy are considered to ensure that fitness of the forthcoming generations is always improved [6, 9].

For the study presented here two functional values qualifying and quantifying the graft local hemodynamics are considered. Multiple hemodynamic factors capable of eliciting a hyperplastic response at the cellular level indicate the potential significance of platelet-wall interactions coinciding with regions of low WSS in the development of intima hyperplasia (IH). The first functional to be considered is:

$$f_1(\boldsymbol{b}) = \sum_1^{N_\Gamma} \|WSS\|^2 \tag{1}$$

where WSS are the smoothed wall shear stress values at the floor of the distal artery-graft junction. Sites of significant particle interactions with the vascular surface have been identified by functions associated to long near-wall residence times capturing a significant IH formation. A second component is associated with the fluid velocity distribution at the cross-section of the distal graft-artery junction:

$$f_2(\boldsymbol{b}) = \sum_{\Omega^*_{g-a}(\boldsymbol{b})} {\mathbf{v_x}}^2 \tag{2}$$

For the same inlet velocity profile, values of this function $f_2(\boldsymbol{b})$ will be larger for disturbed longitudinal velocity distributions along the distal graft-artery junction and smaller for smooth and parabolic distributions.

In the project reported here, an artificial neural network (ANN) is developed to efficiently simulate blood flow for specific graft geometries. A set of randomly generated 500 input vectors $\boldsymbol{b} = (H, \beta, Wp, D)$ and target vectors $\boldsymbol{f} = (f_1(\boldsymbol{b}), f_2(\boldsymbol{b}))$ has been collected using the FE code within the design space given as follows

$$\begin{gathered} 10 \leq H \leq 30\,mm \\ 0.15 \leq \beta \leq 0.785\,rad \\ 6 \leq Wp \leq 10\,mm \\ 6 \leq D \leq 10\,mm \end{gathered} \tag{3}$$

The ANN analysis was performed using MATLAB with the Neural Network Application Toolbox (The MathWorks Inc., MA, USA). The multilayer feed-forward neural network of the software is well suited for function fitting problems. Since the initial weights and biases are randomly set more than one trial was done. The regression analysis was considered using the R value as an indication of the relationship between the outputs and targets. For this example, the training data indicates a good fit. The validation and test results showed R values greater than 0.99. Once the ANN has demonstrated acceptable pattern recognition skills, its creation has been achieved being ready for use.

## 3 EFFECT OF LUMEN-ADAPTED ARTERY

The influence of wall variability on blood flow behavior has been studied and published either in vascular replicas as well as in real vascular anatomies. By their elastic nature, major conduit arteries should be able to store blood volume temporarily during systole and release it during diastole. This reduces the systolic blood pressure required for the flow of a given volume quantity and gradually suppresses the pulsatile flow pattern. The repetitive stretching of the wall (strains of up to 10 per cent) may cause fragmentation of the elastic fibers in the wall, modifying wall elasticity. To maintain wall stress the elastic arteries respond with a diameter increase in combination with an increase of arterial wall thickness. In recent years various methods have been developed to assess and monitor the above interaction. Most of these methods are based on ultrasound techniques because of its wide availability and its non-invasive and non-traumatic nature. Presently these techniques enable the assessment of wall thickness, diastolic diameter, distension waveform, i.e. the time-dependent change in diameter, the relative pulsatile increase in diameter, and pulse wave velocity, for elastic and muscular arteries in humans. For the carotid artery, comparison between the computed and experimentally measured wall movement during a cardiac cycle showed that the pressure waveform plays the main role in driving the wall movement while the pressure gradient resulting from the flow only has a secondary influence [11].

A basic problem for the assessment and evaluation of mechanical parameters is to acquire the local time dependent blood pressure. The artery responds to the pulsatile change in transmural pressure during the cardiac cycle with a pulsatile change in cross-sectional area. The transient wall motion can be divided into three phases: rapid dilation, rapid partial contraction, and slow contraction. The minimum artery diameter occurs during the low-pressure end-diastolic cardiac phase and the maximum artery diameter during the peak systolic phase. Evaluation of arterial vessel diameter variability along the cardiac cycle has been addressed by several authors. Using computerized edge detection-sequential multi-frame image processing Selzer et al. presented a typical plot of the continuous measurement of carotid arterial diameter over 2 cardiac cycles [3].

In this work, the material of the graft wall is assumed to be rigid and the material of the artery wall is assumed to be incrementally linear elastic. For the 6mm diameter artery, the basic geometrical parameters considered here are the initial (end-diastolic) diameter, the change in diameter and the wall thickness. As expected, intima-media thickness (IMT) varies according to the arterial dimension. When the arterial diameter is at its minimum, IMT is at its thickest point and when the arterial diameter is at its maximum, IMT is at its thinnest point. Using Selzer et al. measurements [3] polynomial curves were fitted defining both the arterial diameter and IMT over the cardiac cycle. The shape of the pressure waveform is highly site-dependent and is modified not only by proximal and distal bifurcations, curvatures and artery tapering, but also by the site-dependent mechanical characteristics of the artery itself. In the numerical simulation, a flow waveform boundary is specified at the model inlet. A pulsatile velocity curve along the cardiac cycle has been adapted by Carneiro [12] from Taylor and Draney [13].

In order to realize the necessity to introduce the arterial wall elastic behavior, FEM simulations were performed considering the same geometric bypass defined by parameters $\boldsymbol{b} = (H = 18.67mm, \beta = 0.782rad, Wp = 6.11mm, D = 7.40mm)$ at the end of the diastolic phase. In the computations, the non-Newtonian behavior of blood will follow the Casson model as used in the software validation by Sousa et al. [9]. This model shows both yield stress and shear-thinning non-Newtonian viscosity, broadly used to describe the shear thinning behavior of blood [14]. Comparison between the Newtonian and non-Newtonian fluid models has demonstrated that the velocity profile of the non-Newtonian fluid is somewhat flattened, due to its shear-thinning behavior.

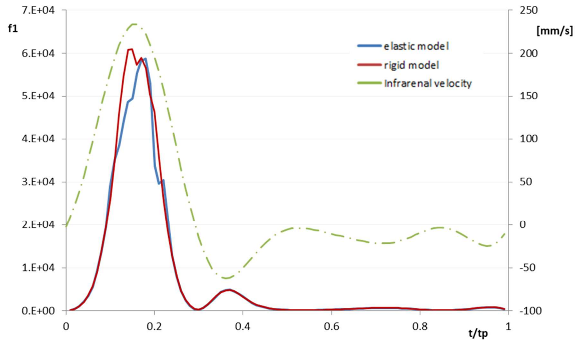


Figure 2: Magnitude variability of the functional associated to wall shear stress for elastic and rigid artery model along the cardiac cycle

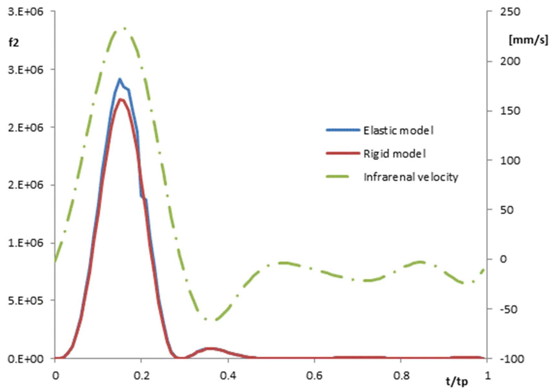


Figure 3: Magnitude variability of the functional associated to velocity distribution at the distal graft-artery junction for elastic and rigid artery model along the cardiac cycle

The introduction of arterial diameter variability along the pulsatile blood velocity curve induces 0.4mm diameter variability so the simulated diameter artery goes from 6mm at the end-diastolic phase up to 6.4mm at the systolic peak. Simulated values for the objective functions $f_1(\boldsymbol{b})$ and $f_2(\boldsymbol{b})$ considering different artery conditions are given bellow. Figure 2 presents a comparison of the magnitude variability of functional $f_1(\boldsymbol{b})$ associated to wall shear stress between the elastic and the rigid artery model together with the pulsatile velocity waveform. As expected high WSS values correspond to the systolic phase. The difference between the rigid and the elastic model is that higher and earlier values are observed for the rigid model.

Figure 3 presents a comparison of the magnitude variability of functional $f_2(\boldsymbol{b})$ associated to fluid velocity distribution at the cross-section of the distal graft-artery junction between the elastic and the rigid artery model. Contrarily to the previous functional, the rigid model presents higher values at the peak systolic phase as compared to the elastic model. This result can be explained due to lack of compliance between artery and prosthetic graft. Mismatched biomechanical properties between the graft and native surrounding tissue are commonly cited as a cause of graft failure.

## 4 RESULTS AND CONCLUSIONS

Pareto optimality is a concept that formalizes the trade-off between a given set of possible contradicting objectives. By only one time global search procedure all the Pareto optimal solutions are found managing the drawing of the Pareto front and then extracting optimal solutions according to selected preferences. As a compromise between computer time and population diversity, parameters for the genetic algorithm were taken as 12 and 5 for the population and elite group size, respectively. The number of bits in binary codifying for each design variable is 5. The GA termination has been defined by fixing the total number of generations as 300. For each generation 6 new individuals (five from crossover plus one from mutation) are created. Simulation results for the optimized bypass graft are presented in Figures 4 and 5.

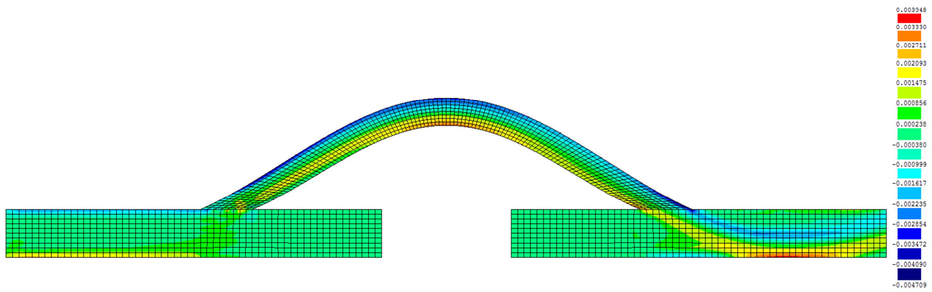

Figure 4: WSS distribution for the optimal elastic artery model at peak systolic phase

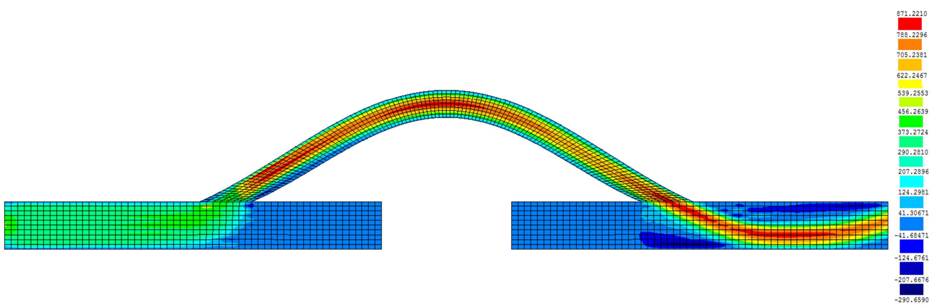

Figure 5: Velocity distribution for the optimal elastic artery model at peak systolic phase

Qualitatively, the distributions of wall shear stress as well as velocity in the elastic artery model do not change significantly compared to the rigid artery model, despite the quite large wall variations. However, less flow separation and reversed flow is observed in the elastic model. At peak systolic phase (t/tp=0.16), Figure 4 shows regions of high WSS appear around stenosis at the distal corner of the proximal bifurcation and at the toe of the distal anastomosis suggesting that both the proximal and distal regions are responsible for early bypass graft failure.

The velocity values are given in Figure 5 (t/tp=0.16) demonstrating a very good quality of the finite element simulation being capable of capturing the flow acceleration as it emerges from the graft to the artery and the flow recirculation at the floor of the host artery, consistently with the expectations. The abrupt connection between artery and graft induces large velocity variations. Long residence times usually observable immediately after the toe of the distal anastomosis are quite undetectable. The importance of designing optimized bypass graft cannot be neglected.

Transient hemodynamic flow induces minimum artery diameter during the low-pressure end-diastolic cardiac phase and the maximum carotid artery diameter during the peak systolic phase and the IMT has an opposite behavior [3]. On the other hand, the prosthetic graft is stiff and the hemodynamic flow changes due to compliance differences across an anastomosis cause increased shear stress to damage endothelial cells and also reduced shear stress leading to areas of relative stagnation and increasing interaction between platelets and vessel wall. The aim of geometry design is to minimize these disruptive flow characteristics. The optimization process manages to achieve geometries presenting wall shear stress values with the expected variability for the blood behavior in the systemic arterial tree.

The problem reported here is related with both shape design and flow control that are involved in the simulation of the bypass system. Improving blood flow dynamics in the graft system is an important element for the long-term success of bypass surgeries. Comparison between the rigid and compliant models illustrates a considerable reduction of reversed flow in the distensible model [15]. This is because when the graft flow rate decreases during the deceleration phase, in order to satisfy the fluid's mass conservation in the rigid wall model, this decrease can be compensated for, only by means of reducing the flow velocity. However, in the compliant model, the vessel's contraction partially compensates the flow rate drop by reducing the cross sectional area; consequently, less would be left to be compensated for, by reducing the velocity. As a result, less reversed flow is present in the distensible-wall models.

Further investigation on the hemodynamic benefits of the blood flow in graft artery compliance and configurations and in more realistic bio-mechanical conditions will be addressed by the authors.

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