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A numerical model of pulsatile blood flow in compliant arteries of a truncated vascular system☆

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ABSTRACT

Three dimensional blood flow in compliant, tapering vessels of a truncated vascular system model containing two levels of bifurcation was investigated numerically using a commercially available finite element analysis and simulation software. Although the branching pattern and the geometry of the human vascular system are complex, they can be specified for small arteries using Murray's hypothesis that the structure of the vascular system obeys the principle of minimum work. Accordingly, in the current vascular system model, the parent/daughter diameter ratios and the angles of bifurcation were specified according to Murray's law. Another geometrical parameter, the ratio of blood vessel length to diameter, was determined according to data found in the literature. The vascular system model also includes a 5 mm thick layer of tissue surrounding the vessels. This tissue layer helps to resist artery deformation during the cardiac cycle. Experimentally measured time dependent blood velocity data, available in the literature, were used as the inlet boundary condition to represent the cardiac cycle. An outflow boundary model, consisting of an elastic tube followed by a contraction tube, was used at the four outlets to represent both the compliance and the pressure drop of the small arteries, arterioles, and capillaries that would follow the truncated vascular system. The results show that, at each bifurcation, the blood flow velocity decreases significantly in the transition from the parent vessel to the daughter vessels due to the higher total cross-sectional area of the daughter vessels as compared to the parent vessel. This decrease in velocity is partially recovered along the arteries due to the tapering of the blood vessels. It can also be observed from the results that the pressure distributions and pressure drops along the vascular system are in good agreement with the physiological data found in the literature. The results also show that the velocity profiles immediately following a bifurcation are not initially symmetric, with their maxima shifted toward the inner part of the bifurcation in the daughter vessels. Finally, the results show that the maximum deformation is about 2% of the average vessel radius, which is relatively small and typical for small arteries.

1. Introduction

The cardiovascular system is one of the most vital systems in the human body. It transports nutrients, oxygen and vital hormones and neurotransmitters to all parts of the body. The cardiovascular system consists of two fluid pumps in series and a complicated, branched pipeline network made up of large arteries, small arteries, arterioles, capillaries and veins [1]. The diseases related to the cardiovascular system such as atherosclerosis, aneurysms, heart attack, heart failure, high blood pressure, and stroke are some of the leading causes of death in the world. The dynamics of blood flow are an underlying factor in many of these diseases and a better understanding of blood flow dynamics plays an especially important role in the diagnosis and treatment of these diseases. A better understanding of blood flow dynamics is also essential in the design and performance evaluation of cardiovascular devices such as left ventricular assist devices, artificial heart valves, stents, and grafts [2]. There are two methods that can be used to investigate blood flow in the cardiovascular system: in vivo studies and computational studies. In vivo studies are important because they yield actual measured data that ground theoretical solutions, but at the same time, in vivo studies are difficult, time consuming and expensive. Therefore, computational studies have been a necessity and have gained popularity in the investigation of blood flow in the cardiovascular system.

Although many attempts have been made to model blood flow in the cardiovascular system during the past decades, three dimensional simulation of the whole vascular system is still an unfeasible task because of the complexity of the cardiovascular system that consists of millions of vessels spanning several orders of magnitude in size. Therefore, previous studies reported in the literature include simplified lumped parameter and one dimensional flow models [3–6]. The lumped parameter models are based on an analogy between the current flow in an

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electrical circuit and the blood flow in the human circulatory system. The lumped parameter and one dimensional models are generally used to simulate the peripheral vascular network in order to obtain a relationship between the blood flow rate and pressure. On the other hand, three dimensional flow modeling is essential to capture all the details of the blood flow in the vascular system [7].

In three dimensional blood flow modeling, only a small truncated system of bifurcated vessels is considered and the remainder of the vascular system is usually represented by boundary conditions. There are a number of studies reported in the literature on three dimensional numerical simulations of blood flow [8–13]. However, nearly all of these simulations contain only a single level of bifurcation in the modeled vascular system. In addition, many of the studies focus only on blood flow in large diameter arterial level vessels and ignore the smaller vessels [12–14].

Blood flow in the cardiovascular system is pulsatile in nature. The pulsatility is generated by the intermittent ejection of blood from the heart during the systolic and diastolic phases. Therefore, the blood flow is both unsteady and quasi-periodic. However, the pulsatile flow from the heart is gradually damped as the blood flows from large to small vessels and eventually becomes steady flowing at the capillary level by a combination of the compliance of the large vessels and the frictional resistance in the small vessels. Therefore, the flow through the smaller arterioles and capillaries can be analyzed based on the assumption of steady state flow. However, there is still a significant pulsatile effect present in the blood flow through small arteries and larger arterioles. Even so, this pulsatile effect has been ignored in a number of studies found in the literature wherein the steady state assumption has been used for the sake of simplicity [15–17].

Most of the blood vessels in the vascular system are elastic and their diameter changes depending on the pressure during a heartbeat. However, the elasticity of blood vessels varies with the diameter of the vessels, that is, blood vessels become gradually more rigid as their diameter becomes smaller. The capillaries are essentially rigid and can be modeled with a rigid wall assumption. On the other hand, the rigid wall assumption fails to predict some essential characteristics of the blood flow through vessels that are larger than capillaries. Even so, the rigid wall assumption has been used in a number of studies [1,18,19] because modeling the fluid–solid interaction of an elastic wall dramatically increases the computational complexity and cost.

The vessels in the cardiovascular system are interconnected tubes of different diameters. Therefore, the diameter of a blood vessel tapers from beginning to end. In vivo studies show that the taper angle can be as large as 2° [20] and this conical nature of the blood vessels causes significant changes in blood flow [21–24]. The taper of the tube is also an important factor in the pressure development along the vessel [24]. Therefore, in order to obtain results consistent with physiological blood flow, the tapering effect of the blood vessels has to be taken into consideration. However, many of the studies found in the literature ignore the tapering effect and assume that the vascular system is composed of straight constant diameter vessels.

Therefore, the study presented in this paper is unique in that three dimensional pulsatile blood flow along small elastic arteries is simulated numerically in a truncated vascular system consisting of two levels of bifurcation. Furthermore, the tapering effect of the blood vessels as they approach the next bifurcation point is also taken into consideration in this study.

2. Description of the vascular system model

2.1. Geometry of the vascular system model

The vascular system model considered in this study is composed of small arteries and has two levels of symmetric bifurcation as shown in Fig. 1. Although the branching pattern and the geometry of the human vascular system are complex, they can be specified for small arteries

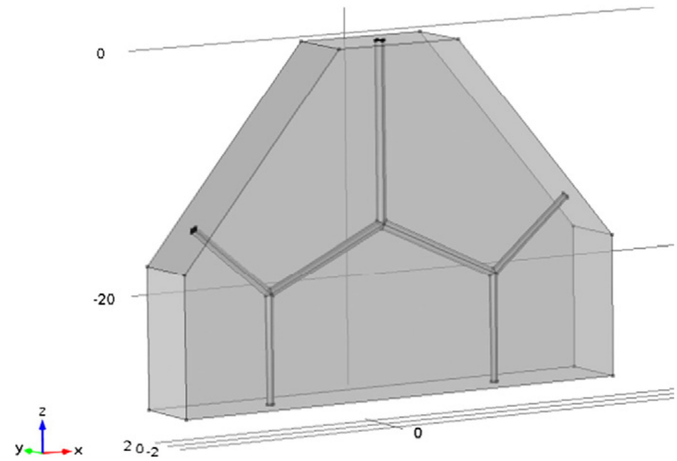

Fig. 1. The geometry and coordinate system for the vascular system model.

using Murray's hypothesis that the structure of the vascular system obeys the principle of minimum work. Accordingly, in the current vascular system model, the parent/daughter diameter ratios and the angles of bifurcation were specified according to Murray's law [25]. The tapering effect of the blood vessels as they approach the next bifurcation point is also accounted for in this study. This structured tree assumption is valid only for small arterial vessels and, accordingly, the diameter of the root vessel in the tree is taken as 0.6 mm.

The diameters of the daughter vessels after each bifurcation and the angles between the main vessel and the daughter vessels, as shown in Fig. 2, are specified according to a modified version of Murray's law [25]:

$$d_0^n = d_1^n + d_2^n\,;\; cos\theta_1 = \frac{d_0^4 + d_1^4 - d_2^4}{(2d_0d_1)^2}\,;\ \text{and},\ cos\theta_2 = \frac{d_0^4 - d_1^4 + d_2^4}{(2d_0d_1)^2}; \tag{1}$$

wherein subscript 0 refers to the parent vessel, and subscripts 1 and 2 refer to the two daughter vessels at the bifurcation, θ_1 and θ_2 are the angles between the parent and the daughter vessels, and n is the bifurcation exponent. The total branch angle $(\theta_1 + \theta_2)$ between the two daughter vessels is known as the bifurcation angle.

Murray proposed that physiological vascular systems, through evolution by natural selection, must have achieved an optimum arrangement such that in every segment of the vascular structure, flow is

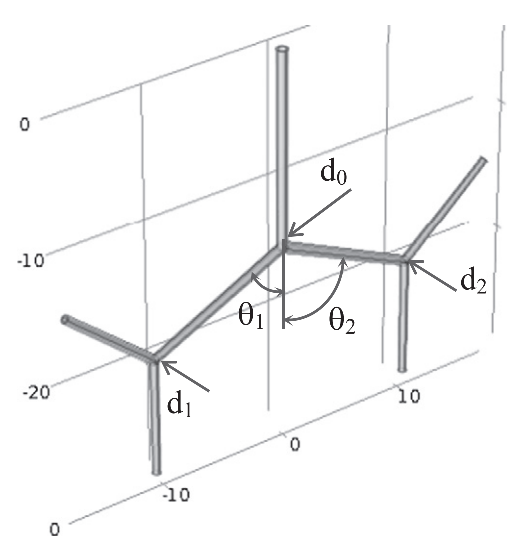

Fig. 2. The geometric parameters of the main and daughter vessels.