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One dimensional blood flow in a planetocentric orbit

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Abstract

All life on earth is accustomed to the presence of gravity. When gravity is altered, biological processes can go awry. It is of great importance to ensure safety during a spaceflight. Long term exposure to microgravity can trigger detrimental physiological responses in the human body. Fluid redistribution coupled with fluid loss is one of the effects. In particular, in microgravity blood volume is shifted towards the thorax and head. Sympathetic nervous system-induced vasoconstriction is needed to maintain arterial pressure, while venoconstriction limits venous pooling of blood prevents further reductions in venous return of blood to the heart. In this paper, we modify an existing one dimensional blood flow model with the inclusion of the hydrostatic pressure gradient that further depends on the gravitational field modified by the oblateness and rotation of the Earth. We find that the velocity of the blood flow V_B is inversely proportional to the blood specific volume d, also proportional to the oblateness harmonic coefficient J_2, the angular velocity of the Earth ω_E, and finally proportional to an arbitrary constant c. For $c = -0.39073$ and $\xi_H = -0.5$ mmHg, all orbits result to less blood flow velocities than that calculated on the surface of the Earth. From all considered orbits, elliptical polar orbit of eccentricity $e = 0.2$ exhibit the largest flow velocity $V_B = 1.031$ m/s, followed by the orbits of inclination $i = 45°$and 0°. The Earth's oblateness and its rotation contribute a 0.7% difference to the blood flow velocity.

Keywords: Blood flow velocity; Zonal harmonics; Gravitational acceleration; Hydrostatic pressure

1. Introduction

One dimensional blood flow models are used to model blood flow in an extended or large section of the arterial system, with the goal of studying the wave propagation nature of the arterial pulse. Arteries are blood vessels that carry oxygenated blood away from the heart and distribute it to body tissue. Blood flow is driven by the pressure gradient along the arterial tree, which accelerates or decelerates the flow in these low resistance vessels. It is of great importance to be able to evaluate the velocity profile at any given location across an artery at any given time during the cardiac cycle. Having access to information related to the physiological parameters of a patient, is a way of non-invasively identifying the mechanics of the arterial system under different environmental variables. The regulation of an organism's internal environment is expressed in the change of physiological activities, so that they meet the conditions within the range of survival. This might be achieved by numerically solving the 3-D fluid structure interaction problem on real geometries coming from real patients; however this is a complex kind of modeling that requires a variety of resolutions. Moreover, modeling of the blood flow parameters can be used when modeling the transport of solute throughout artery tree, the transport of lipids to and from the arterial wall. In the case of a spacecraft orbit, it might also constitute a primary factor when the mobility permitted by future planetary spacesuits is considered. Prolonged exposure to changes in microgravity associated with a spaceflight in orbit might be associated with an increased risk of poor blood circulation flow

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malfunction. Knowing the effects of microgravity on the circulation of blood of deep-space astronauts, aims at meeting the unique challenges of treatment and prevention posed by long duration missions. Our study focuses at the mechanisms of interaction between the fluid bio-distribution and the gravitational zonal harmonic coefficient at various orbital inclinations and eccentricities that correspond to different orbits, and the calculation of different blood flow velocities for these orbits for a spacecraft orbiting the Earth, which can be also applied for any planet in the solar system.

2. The model

In a previously published work by Karim (2006), a model that combines one dimensional conservation laws with the theory developed by Womersley was used to study the flow in arteries. In this model, a one dimensional theory generates the pressure gradient needed by the theory that drives the equations derived by Womersley. Woemrsley's theory provides the velocity profile, which is used in the computation of the friction and non-linear terms of the one-dimensional theory. The corresponding velocity filed comes from axisymmetric flow with a pressure gradient that is sinusoidal in time. A one-dimensional system of hyperbolic partial differential equations, derived from the Navier–Stokes equations is used and Womersley's theory provides the velocity profile needed for the system, which also generates the friction and the nonlinear model term. The assumption included in the model takes into account an infinite rigid cylinder vessel described in the $r-z$ plane by $-\infty < z < \infty$ and $0 < r < R$. Also the axial velocity of the moving fluid is given in cylindrical coordinates $V(r, t, R)$, and the radial velocity is zero so that the flow is only in the z-direction. The pressure gradient driving the flow is given by $\frac{\partial P}{\partial z} = (A+B)\sin(\omega t) + C$ and preserves the pulsating character of the cardiac cycle. The solution of the axisymmetric flow along the pipe is given by:

$$\frac{\partial V}{\partial t} + \frac{1}{\rho}\frac{\partial P}{\partial z} = \frac{\mu}{\rho}\left(\frac{\partial^2 V}{\partial r^2} + \frac{1}{r}\frac{\partial V}{\partial r}\right), \tag{1}$$

and satisfies the no slip condition at the boundary $V|_R = 0$, along with the condition that the flow is finite at $r = 0$. When we talk about the pressure within blood vessels, venous or arterial, we are dealing with three different concepts (and the interaction of these concepts must be taken into account). The mean systemic filling pressure is related to the volume in the vessel, as well as to the properties of the wall, the dynamic pressure that is further related to the blood flow velocity and the resistance, and finally on the hydrostatic pressure, which is related to gravity. On the Earth the heart has to pump the blood against the gravitational field and therefore it will have to work harder. In the microgravity environment of space, the pumping of the heart will reduce. In the case of a spaceflight mission or a satellite laboratory setup, we want to estimate the gravitational effect on the velocity profile, and therefore we replace the equation of the dynamic pressure gradient by $\frac{\partial P}{\partial z} = \left(\xi_H + \frac{g(\vec{r}')}{d}\right)\sin(\omega_H t) + c\rho g(\vec{r}')$, where the $\sin(\omega_H t)$ will act as a pulsating factor of the heart's pumping effect, c is the slope of the vessel with the acceleration of gravity vector with the horizontal, and d is a constant that has the units m^3/kg or inverse density also called specific density, ξ_H the cardiac pressure gradient amplitude in the absence of gravity, $\vec{r}'$ is the radial distance from the center of the planet to the spacecraft, and $g(\vec{r}')$ is the gravitational acceleration at the spacecraft orbital altitude. The value of the slope c varies between -1 and 1 for vessels inclined vertically upwards and downwards, respectively, in the direction of the flow, and it is equal to the sine of the angle of the vessel axis with the horizontal. Gravity interaction with the blood flow may be transformed and amplified by modulation of the velocity profile. Turbulence and eddies are not taken into account in this one-dimensional flow model considered in this contribution. In particular, following (Stergiopoulos et al., 1992) we assume that the axisymmetric flow is parallel to the gravitational acceleration. We slightly modify the solution in Karim (2006), derived by the method of Fourier transform in the following way, by letting $v = \frac{\mu}{\rho}$ and introducing the dimensionless variables:

$$\zeta = \frac{r}{R}, \quad \tau = \frac{\mu t}{\rho R^2} = \frac{vt}{\rho}, \quad \text{and} \quad V_B = \frac{1}{4\mu}\left(\frac{dP}{dz}\right)R^2\theta(\zeta), \tag{2}$$

and therefore we find that

$$\frac{\partial \theta}{\partial \tau} = 4 + \frac{1}{\zeta}\frac{\partial}{\partial \zeta}\left(\zeta\frac{\partial \theta}{\partial \zeta}\right). \tag{3}$$

The boundary condition is $\theta = 0$ at $\zeta = 1$. The initial condition is taken $\theta = 0$ at $\tau = 0$. The method of separation of variables leads to the following solution:

$$\theta = 1 - \zeta^2 - 8\sum\nolimits_n \frac{J_0(a_n\zeta)}{a_n^3 J_1(a_n)}e^{-a_n\tau}, \tag{4}$$

and therefore the blood flow velocity becomes:

$$V_B = \frac{1}{4\mu}\frac{dP}{dz}R^2\left(1 - \frac{r^2}{R^2} - 8\sum\nolimits_n \frac{J_0\left(\frac{a_n r}{R}\right)}{a_n^3 J_1(a_n)}e^{-a_n\tau}\right), \tag{5}$$

where $a_n$are the roots of the 0th order Bessel function J_0, and finally Eq. (5) takes the final form:

$$V_B = \frac{1}{4\mu}R^2\left(\left(\xi_H + \frac{g(\vec{r}')}{d}\right)\sin(\omega t) + c\rho g(\vec{r}')\right) \times \left(1 - \frac{r^2}{R^2} - 8\sum_n \frac{J_0\left(\frac{a_n r}{R}\right)}{a_n^3 J_1(a_n)}e^{-a_n\tau}\right), \tag{6}$$

where J_0 is the 0th order Bessel function, ρ is the density of blood, $\omega_0 = 2\pi/T$ is the frequency of cardiac pulse and μ is the viscosity of blood. At this point we write the total gravitational acceleration $g_{tot}(r') = g_{cen}(r') + g_{J_2}(r') + g_{rot}(r')$ including thus the effect of the oblateness and rotation of