Journal of Biomechanics 47 (2014) 1082–1090

Contents lists available at ScienceDirect

Journal of Biomechanics

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

Numerical simulations of the pulsating flow of cerebrospinal fluid flow in the cervical spinal canal of a Chiari patient

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ARTICLE INFO

Article history:
Accepted 22 December 2013

Keywords:
CSF flow
Chiari I
Simulations
Turbulence
Flow instability
Local Reynolds number

ABSTRACT

The flow of cerebrospinal fluid (CSF) in a patient-specific model of the subarachnoid space in a Chiari I patient was investigated using numerical simulations. The pulsating CSF flow was modeled using a time-varying velocity pulse based on peak velocity measurements (diastole and systole) derived from a selection of patients with Chiari I malformation. The present study introduces the general definition of the Reynolds number to provide a measure of CSF flow instability to give an estimate of the possibility of turbulence occurring in CSF flow. This was motivated by the fact that the combination of pulsating flow and the geometric complexity of the spinal canal may result in *local* Reynolds numbers that are significantly higher than the commonly used global measure such that flow instabilities may develop into turbulent flow in these regions. The local Reynolds number was used in combination with derived statistics to characterize the flow. The results revealed the existence of both local unstable regions and local regions with velocity fluctuations similar in magnitude to what is observed in fully turbulent flows. The results also indicated that the fluctuations were not self-sustained turbulence, but rather flow instabilities that may develop into turbulence. The case considered was therefore believed to represent a CSF flow close to transition.

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1. Introduction

The Chiari I malformation is a condition characterized by tonsilar herniation through the foramen magnum. The herniation obstructs the pulsating flow of cerebrospinal fluid (CSF) through the foramen magnum and is believed to play a key role in the development of syringomyelia that commonly occurs in Chiari I patients. Computational fluid dynamics (CFD) has been used extensively for several decades and proven itself as a reliable approach to predict complex fluid flow phenomena. CFD has thus become a popular tool for investigating flows from a wide range of applications including complex biomedical flows.

The occurrence of complex flow phenomena, including bidirectional flow patterns, in the cervical canal has previously been demonstrated using PC-MR for Chiari I patients (Quigley et al., 2004) Despite the complexity of the flow, all CFD studies to date, to the authors' knowledge, rest on the assumption that the flow remains laminar (Loth et al., 2001; Linge et al., 2010, 2011; Roldan et al., 2009; Cheng et al., 2012; Clarke et al., 2013; Yiallourou et al., 2012; Rutkowska et al., 2012; Shaffer et al., 2011). This is based on a priori estimate of the Reynolds number, based on the peak systole velocity and the hydraulic diameter of the spinal canal, which usually lies in the range 150–570. Such a range is considered to be too low for turbulence to occur based on transition studies in pipe flow such as Stettler and Hussain (1986). However, the anatomy of the spinal subarachnoid space is far more complex than idealized pipes. Furthermore, recent 4D PC MR studies (Bunck et al., 2012) in which velocities as high as 20 cm/s were reported have demonstrated that even higher Reynolds numbers can occur. Also, the recent comparison of PC-MR reveals significant differences between the measured flow and the computed laminar flow (Yiallourou et al., 2012).

Previous estimates of the Reynolds-number, and thus also on the a priori assumption of laminar flow, are based on a global measure that does not provide information about local variations of the Reynolds-number within the spinal canal. The motivation for this study is that the combination of pulsating flow and the geometric complexity of the canal may result in *local* Reynolds-numbers that are significantly higher than the commonly used global measure such that flow instabilities may develop into

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http://dx.doi.org/10.1016/j.jbiomech.2013.12.023

chaotic turbulent flow in these regions. The objective of the present study is to use a state-of-the-art CFD methodology to investigate pulsating CSF flow in Chiari I patients in order to provide better understanding of the complexity of flow field, and to provide a measure of CSF flow stability that estimated the possibility of turbulence in CSF flow. The possibility of turbulent flow in the CSF has not to our knowledge been studied systematically. For this study, we created a 3D model of the spinal subarachnoid space in a Chiari I patient and we assumed CSF flow volumes in the upper range of normal.

2. Mathematical and computational modeling

The present study employs a method that is similar to what is known as direct numerical simulations (DNS) which is a branch of traditional computational fluid dynamics (CFD) devoted to high-fidelity solution of transitional and turbulent flows (Moin and Mahesh, 1998). Here, we take particular care in choosing a time stepping scheme (Simo and Armero, 1994) and to minimize any stabilization that may introduce numerical diffusion. Therefore, we are forced to resolve the finer details of the flow, which makes the simulation computationally demanding. Such high resolution simulations are often avoided in physiological flow under the assumption of laminar flow. However, to investigate whether flow may be transitional, such as our case, this approach is therefore ideal.

The explicit requirement of a DNS is that all temporal and spatial scales of the flow field are computationally resolved. Turbulence is inherently multi-scale fluid flow phenomena and it can be shown that the number of degrees-of-freedom of a turbulent flow scales as $N \sim Re^{9/4}$ (Sandham, 2002), where Re is the Reynolds-number which typically exceeds $O(10^3)$ in turbulent flows.[1] Due to the extreme computational costs, the applicability of DNS is limited to low Reynolds-number flows only which makes it a potentially viable approach in biomedical applications. A still unresolved issue, however, is that all inflow/outflow boundary conditions must be properly prescribed (in time and three-dimensional space). Lacking this information makes the application of DNS in biomedical flows challenging, and special attention has to be taken to ensure minimum influence of this error. Since the CSF flow considered in this study is subjected to an oscillatory motion, this issue was handled by defining the size of the computational domain such that the inflow/outflow boundaries had very small influence on the results in the area of interest.

2.1. Governing equations

The equations governing the flow of an incompressible Newtonian fluid, such as CSF fluid, are based on the fundamental principles of conservation of mass and momentum which can be written as

$$\frac{\partial \tilde{u}_i(\mathbf{x},t)}{\partial t} + \tilde{u}_k(\mathbf{x},t)\frac{\partial \tilde{u}_i(\mathbf{x},t)}{\partial x_k} = \nu \frac{\partial^2 \tilde{u}_i(\mathbf{x},t)}{\partial x_k \partial x_k} - \frac{1}{\rho}\frac{\tilde{p}(\mathbf{x},t)}{\partial x_i}, \quad (1)$$

$$\frac{\partial \tilde{u}_i(\mathbf{x},t)}{\partial x_i} = 0. \quad (2)$$

These are commonly referred to as the Navier–Stokes and continuity equations, respectively. Here $\tilde{u}_i(\mathbf{x},t)$ denotes the instantaneous velocity component in the x_i-direction, $\tilde{p}(\mathbf{x},t)$ is the instantaneous pressure, and ρ is the fluid density whereas $\nu = \mu/\rho$ denotes the kinematic viscosity and μ is the molecular viscosity. We apply the notation $[x_1, x_2, x_3] = [x, y, z]$, cf. Fig. 1, and $[\tilde{u}_x, \tilde{u}_y, \tilde{u}_z] = [\tilde{u}, \tilde{v}, \tilde{w}]$. Einstein's summation convention also applies, i.e. summation over repeated indices, e.g. $\partial \tilde{u}_i/\partial x_i = \partial \tilde{u}_1/\partial x_1 + \partial \tilde{u}_2/\partial x_2 + \partial \tilde{u}_3/\partial x_3$.

The CSF fluid is assumed to have the physical properties as water at 37 °C, i.e. $\rho = 1000\ (\mathrm{kg/m^3})$ and $\nu = 0.7 \times 10^{-6}\ (\mathrm{m^2/s})$. Gravity is neglected from Eq. (1) since its effect is implicitly contained in the inlet velocity boundary conditions which is obtained indirectly from measured data.

[1] In most applications $Re \sim O(10^3) - O(10^9)$.

2.2. Numerical simulations

From the registry of Chiari patients at the University of Wisconsin, with the approval of the local Institutional Review Board, a patient with Chiari I and with evidence of elevated CSF velocities was chosen for the patient-specific modeling.

A multi-echo 3D radial acquisition (VIPR – Vastly undersampled Isotropic Projection Reconstruction) was used for fully refocused, balanced SSFP imaging with isotropic spatial resolution and fat/water separation (Mistretta, 2009). The VIPR sequence samples data along radial lines evenly spaced through a spherical volume each intersecting the origin of k-space. Short acquisition times, large volume coverage, and small isotropic voxels are achieved with radial undersampling, which introduces negligible artifacts in high contrast imaging scenarios. Here, a $20 \times 20 \times 20\ \mathrm{cm^3}$ image volume of $256 \times 256 \times 256$ voxels (of the cervical spine and lower posterior fossa) was acquired in 5 min scan time, providing high SNR from the spinal canal due to its high T2. The high contrast between CSF and surrounding tissues and the high spatial isotropic resolution are beneficial for image segmentation. The surface model of the spinal canal was created using the Vascular Modeling Toolkit (Vascular Modeling Toolkit, 2013). Figs. 1 and 2 show the three-dimensional geometry used in the present study.

The three-dimensional and time dependent Navier–Stokes equations (1) and (2) were solved using the node-based finite volume method *cliff* from Cascade Technologies. This method is based on second-order accurate discretization in time and space, and with time integration based on the fractional-step method. The second-order operators used in the computations were also used when calculating derivatives for post-processing (such as the calculation of l^+ values, Eq. (3), and the local Reynolds number, Eq. (5)), except from the time derivative in Eq. (5) in which a first-order scheme was used.

2.2.1. Boundary conditions

The pulsating CSF flow in the subarachnoid space was simulated using a time varying velocity pulse defined at the lower boundary of the computational domain (see Fig. 1). The CSF pulse could equally well be specified on the upper boundary, but since the geometry close to the lower boundary varies considerably less than close to the upper boundary, the flow field close to the lower boundary exhibits less local variations. Unwanted effects associated with the imposed velocity boundary condition on the flow field are therefore minimized.

The form of the pulse varies between patients and with heart rate, but in general the systolic phase of the flow (downward fluid movement) has a greater amplitude and a shorter period compared to the diastolic phase (upward fluid movement). The CSF flow pulse used in this study was generated to agree with the peak velocity amplitudes derived from a selection of patients with Chiari I malformation measured by Shah et al. (2011), cf. Fig. 1. The pulse was generated to produce maximum systolic and diastolic velocities around 10 cm/s and 7.5 cm/s, respectively, with a cycle of 0.75 s. This corresponds to a heart rate of 80 beats per minute. Fig. 2 shows the modeled spinal channel. The outer (and inner) boundaries of the spinal canal were assumed to be rigid and

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