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Effect of swirling inlet condition on the flow field in a stenosed arterial vessel model

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ABSTRACT

Blood flow in an artery is closely related to atherosclerosis progression. Hemodynamic environments influence platelet activation, aggregation, and rupture of atherosclerotic plaque. The existence of swirling flow components in an artery is frequently observed under *in vivo* conditions. However, the fluid-dynamic roles of spiral flow are not fully understood to date. In this study, the spiral blood flow effect in an axisymmetric stenosis model was experimentally investigated using particle image velocimetry velocity field measurement technique and streakline flow visualization. Spiral inserts with two different helical pitches (10D and 10/3D) were installed upstream of the stenosis to induce swirling flows. Results show that the spiral flow significantly reduces the length of recirculation flow and provokes early breakout of turbulent transition, but variation of swirling intensity does not induce significant changes of turbulence intensity. The present results about the spiral flow effects through the stenosis will contribute in achieving better understanding of the hemodynamic characteristics of atherosclerosis and in discovering better diagnosis procedures and clinical treatments.

1. Introduction

Circulatory vascular disease (CVD) is the leading cause of death in many developed countries. Atherosclerosis is one of the main causes of CVDs, which lead to high mortality and morbidity. Carotid atherosclerosis notably causes cerebral ischemia and infarction (stroke) [1]. The World Health Organization reported that 17.5 million people (30% of all global deaths) died from cardiovascular diseases in 2005 [2]. Heart attack and stroke caused 7.6 and 5.7 million of these deaths, respectively.

Blood flow characteristics in an artery is closely related to the progression of atherosclerosis. Once atherosclerotic plaques are developed in a blood vessel, the blood flow is significantly disturbed by the local contraction of the vessel diameter. This disturbance is characterized by high shear stress at the stenosis apex, flow separation, vortex shedding, and turbulent transition at the downstream region of the stenosis. These hemodynamic environments influence platelet activation, as well as aggregation and rupture of atherosclerotic plaque. Therefore, better understanding of the hemodynamic characteristics in a stenosis is important in identifying better diagnosis and clinical treatments of atherosclerosis.

In vitro experiments using artificial flow phantoms are effective for investigating the fluid mechanical aspects of the circulatory system without ethical and safety problems arising from animal and human experiments. Giddens [3,4] investigated mean flow field and frequency contents as functions of Reynolds number (*Re*) in both steady and pulsatile flows using *in vitro* stenosis models. The sinusoidal shape of stenosis with 50% reduction in diameter (75% reduction in area) used by Giddens became a canonical stenosis model. The mean and fluctuating flow fields in this model have been widely investigated as baseline flow [5–8]. Vétel et al. [9] carried out in-depth study of asymmetry and transition to turbulence in a smooth axisymmetric stenosis using both stereoscopic particle image velocimetry (PIV) and time-resolved PIV technique. Jet deflection toward the wall due to Coanda effect, flow asymmetry downstream of the stenosis, and unsteady turbulent flow above a critical Reynolds number (*Re*) have been well characterized through *in vitro* experiments.

Although most previous studies have employed fully developed Poiseuille flow with a parabolic velocity profile as an inlet condition, evidence shows that blood flow in an artery has a spiral corkscrew pattern. For example, Stonebridge and Brophy [10]

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reported the existence of a spiral blood flow in human femoral artery. Numerous reports also support that the spiral flow is a normal physiological flow phenomenon in circulatory system [11–14]. The three-dimensional (3D) complex squeezing motion of the heart and the tapered and curved geometrical configurations with rifled endoluminal surfaces of arteries are suspected to induce spiral flows. Despite abundant evidence of spiral flows, their exact roles in arteries are not fully understood to date.

Understanding the detailed mechanism of spiral flow in the circulatory system requires an in-depth study of their fluid dynamic characteristics. Computational fluid dynamics (CFD) techniques have been used to investigate the spiral flow effects on the stenosis and sudden expansion channel [15–17]. According to previous numerical simulations, the spiral flow provides relatively uniform distribution of wall shear stress, as well as reduces flow stagnation, separation, and instability. Experimental studies are relatively lacking compared with numerical simulations.

The main objective of the present study is to investigate the effect of spiral blood flow in a stenosis model with 50% reduction in diameter by means of experimental flow measurement techniques, especially PIV technique. This research specifically focuses on the changes of the recirculation length, location of turbulent transition, and turbulence intensity with regard to the swirling intensity of the inlet flow.

2. Materials and methods

2.1. Fabrication of stenosis model and spiral inserts

The stenosis model depicted in Fig. 1A was created using the following cosine-form formula [18]:

$$\frac{r(z)}{R} = 1 - \delta_c \left[1 + \cos\left(\frac{z\pi}{D}\right)\right], \quad -D \le z \le D \tag{1}$$

where R and D are the radius and diameter of the normal vessel, respectively; r and z represent the radial and axial coordinates; parameter δ_c denotes the percentage of the vessel constriction. The δ_c of this model is 0.25, indicating 50% and 75% reductions in the diameter and cross-sectional area, respectively. The total length of the stenosis model is 220 mm (22D), where diameter D = 10 mm. The length of the upstream and downstream of the stenosis is 10D, and the length of the stenosis region is 2D.

Fig. 1B shows a schematic diagram of the spiral insert. The spiral insert has the same diameter as the stenosis channel (D = 10 mm). The length of the insert is 5D. The spiral insert has a four-arm type of spiral structure having 0.3D of inner core and four fan-shaped arms with 45° angle. Two spiral inserts (spiral types I and II) with different helical pitches were designed to change the swirling intensity of the flow. Helical pitches of spiral types I and II are 10D and 10/3D, respectively. The spiral insert was placed at the inlet of the channel, after which it produced spiral flow 5D upstream of the stenosis region.

The stenosis model and spiral inserts were fabricated using acrylonitrile butadiene styrene (ABS) thermoplastic using a 3D printer (Fortus 400mc, Stratasys). Then, the silicone mold of the stenosis model was made using a silicone rubber compound (Silastic 3481, Dow Corning). The ABS stenosis model was removed from the silicone mold by slicing the mold open. The mold was then refilled with a low melting-point alloy metal (melting point 70 °C) to produce a metal replica of the original stenosis model. The fabricated metal replica was lightly sanded and casted using a polydimethylsiloxane (PDMS) silicone compound. The metal replica was then melted out from the PDMS stenosis channel. The final model was washed with 7% nitric acid solution for 5 h and with DI water for 30 min to remove any debris that adhered on the channel surface.

2.2. Flow circuit system

Fig. 2A shows a schematic diagram of the experimental setup used in this study. A total of 5 L working fluid was prepared in an acryl reservoir; a 15 W centrifugal pump circulated the fluid at a constant flow rate through the circuit. As a fluidic low-pass filter, a 0.5 L air container was installed using a three-way connector to stabilize possible fluctuations of a flow rate [19,20]. A variable-area type flow meter (Visi-Float®, Dwyer instruments, Inc.) was installed to measure the flow rate after properly re-calibrating with the working fluid. The flow rate can be controlled by the internal fluid valve of the flow meter.

Silicone tubes 11 mm in internal diameter were connected to the inlet and outlet of the PDMS stenosis channel model. The inlet part was connected to a 1 m-long straight tube coaxially aligned with the stenosis channel to establish a fully developed laminar flow before entering the stenosis channel. The outlet tube was connected vertically because camera B (Fig. 2B) was positioned to take images of the cross-sectional flow through the outlet window. The flow phantom was placed in the acryl container filled with index-matched working fluid to remove optical distortion at the surface of the model due to the difference of the refractive indexes between the stenosis model and the surrounding fluid.

2.3. Working fluid

The blood mimicking working fluid in this study was prepared according to Yousif et al. [21]. First, a mixture of glycerol and water mixture (44:56 by weight) was prepared, and then 15% (by weight) of sodium iodide was dissolved. Based on the Abbe refractometer (ATAGO, Japan), the refractive index of the resultant working fluid is 1.4130 ± 0.0005. This refractive index matches that of the present PDMS stenosis phantom well. In addition, the dynamic working viscosity of the fluid is 4.30 ± 0.05 cP, which lies within the range of human blood viscosity (4.4 ± 0.6 cP) [21]. The working fluid was seeded with silver-coated hollow glass spheres (Conduct-O-Fill SH400S20 silver hollow, Potters Industries, Inc.) with mean diameters of $d_p \sim 13\ \mu$m and then circulated through the flow loop. The index-matching enclosure was filled with unseeded working fluid.

2.4. PIV measurement

As shown in Fig. 2B, Q-switched double-pulse Nd:Yag laser (Gemini, New wave) generated a thin laser sheet to illuminate the measurement plane. The 4.2-megapixel high-resolution 14-bit charge-coupled device camera (PCO2000, PCO) was synchronized with the laser pulses to run at frame rates of up to 10 Hz. Up to 75 instantaneous velocity fields were obtained from a set of image acquisitions. Measurements were repeated six times to obtain a total of 450 instantaneous velocity fields. The obtained velocity fields were statistically analyzed to obtain their mean and fluctuation components. Each instantaneous velocity vector $U(t)$ [Eq. (2)] in steady flow can be decomposed into a time-averaged mean velocity component $\bar{U}$ [Eq. (3)] and fluctuating velocity component $u'(t)$.

$$U(t) = \bar{U} + u'(t) \tag{2}$$

$$\bar{U} = \frac{1}{N}\sum_{0}^{N} U(t) \tag{3}$$