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Microstructural modelling of cerebral aneurysm evolution through effective stress mediated destructive remodelling

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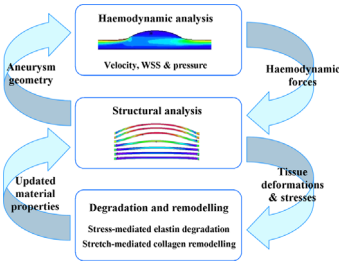
HIGHLIGHTS

- We presented a fluid-solid-growth model for the formation of a cerebral aneurysm.
- The arterial wall was defined with the ability of degradation, growth and remodelling.
- The degradation of elastin was defined as a function of vascular effective stress.
- The model was consistent with other computational or clinical studies.
- The evolving microstructural properties of the wall were predicted.

GRAPHICAL ABSTRACT

A fluid-solid-growth model for the formation of a cerebral aneurysm is presented. The degradation of elastin was defined as a function of vascular effective stress.



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ABSTRACT

Recently, researchers have shown an increased interest in the biomechanical modelling of cerebral aneurysm development. In the present study a fluid-solid-growth model for the formation of a fusiform aneurysm has been presented in an axi-symmetric geometry of the internal carotid artery. This model is the result of two parallel mechanisms: first, defining arterial wall as a living tissue with the ability of degradation, growth and remodelling and second, full coupling of the wall and the blood flow. Here for the first time the degradation of elastin has been defined as a function of vascular wall effective stress to take into account the shear dependent nature of degradation and the mural-cell-mediated destructive activities. The model has been stabilized in size and mechanical properties and is consistent with other computational or clinical studies. Furthermore, the evolving microstructural properties of the wall during the evolution process have been predicted.

1. Introduction

Cerebral aneurysms are pathological dilations of cerebral arteries which generally occur near or inside the circle of Willis. Rupture of the cerebral aneurysm may cause sudden subarachnoid haemorrhage (SAH) which is associated with a high mortality rate (Humphery and Canham, 2000; Krex et al., 2001; Valencia and Solis, 2006).

The natural history of a brain aneurysm can be divided into three phases: pathogenesis, progression and rupture that are complicated physiological procedures which are still quite unknown, but scientists believe that haemodynamic parameters play an important role in their formation (Krex et al., 2001; Shojima et al., 2004; Feng et al., 2005).

Most of the previous biomechanical studies on cerebral aneurysms were computational simulations of already existing aneurysms that analysed the flow characteristics or estimated the risk of

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rupture (Valencia and Solis, 2006; Shojima et al., 2004; Isaksen et al., 2008; Baharoglu et al., 2010; Torii et al., 2006). This category has not been discussed here because it is not directly related to the subject of this research.

Some researchers have focused on structural modelling of the developmental stages of disease. Baek et al. (2006) studied stress-mediated matrix turnover and examined different theories for the arrangement of new collagen fibres. Kroon and Holzapfel (2008) proposed a 3D model for saccular cerebral aneurysm growth by collagen fibre remodelling. Wulandana and Robertson (2005a, 2005b) presented a new multi-mechanism constitutive equation for the isotropic cerebral arteries and made use of it to predict the formation of an aneurysm in their later works. Li and Robertson (2009) modified the previous model of Wulandana and Robertson (2005a) and added the anisotropic effect and the damage mechanism. Watton et al. (2009a) modelled membranes subjected to 1D enlargement with continuous mechanical responses related to the elastin and collagen fibres. Kindo (Kindo, 2006) extracted the material constants of a cerebral artery by fitting different constitutive models to the experimental data of Scott et al. (1972) and presented a 1D growth model following Gasser et al. (2006) and Watton and Hill (2004). Vena et al. (2008) presented an anisotropic model of the early progression of brain aneurysms also following Gasser et al. (2006) and Watton and Hill (2004). Machyshyn and Van de Voss (2010a, 2010b) analysed the stability of the arterial wall against dynamic remodelling and modelled the formation of an aneurysm following Gasser et al. (2006) and Watton and Hill (2004).

Few numbers of studies have considered the effects of haemodynamic parameters on the aneurismal development process. Chatziprodromou et al. (2007) developed a model of aneurismal growth in a linear elastic straight artery with fluid structure interaction (FSI). Feng et al. (2005, 2008) constructed FSI aneurismal growth models with the hypothesis that high local wall shear stress decreases the module of elasticity. Figueroa et al. (2009) proposed a new FSI computational framework and introduced the remodelling rate as a function of stress or shear stress of the wall. Watton et al. (2009b, 2011) directly entered the haemodynamic parameters into the aneurysm formation mechanism for the first time. They defined elastin degradation as a function of wall shear stress (WSS) and wall shear stress gradient (WSSG).

In this paper a fluid-solid-growth (FSG) model of the formation of a fusiform brain aneurysm is presented in an axi-symmetric geometry of the internal carotid artery (ICA) as one of the most probable positions of cerebral aneurysms. This model is the result of two parallel mechanisms: First, writing a code to define the arterial wall as a living tissue with the ability of growth and remodelling (G&R). Second, performing full coupling between the arterial wall and the blood flow as the source of the forces and stresses exerted on the wall. This way, a complete interaction between the continuous changes in mechanical properties and contents of the arterial wall components and haemodynamic forces will be possible during the genesis of brain aneurysms.

2. Material and method

2.1. Model geometry

A straight cylinder with the dimensions of ICA was considered axi-symmetrically as the model geometry for simulating the formation of a fusiform aneurysm. Dimensions were obtained from references (Valencia and Solis, 2006; Watton et al., 2009b, 2011). The inner radius of the artery was 1.2 mm and its thickness 0.3 mm. The Disease-prone area was 10 mm long and the healthy areas of the same dimensions were connected to both sides.

2.2. Governing equations

The flow was assumed to be steady, laminar, Newtonian and incompressible. The incompressible Navier–Stokes equations with arbitrary Lagrangian–Eulerian (ALE) formulation were used as the governing equations for the fluid model.

Continuity equation:

$$\nabla.\boldsymbol{V}=0 \tag{1}$$

Navier–Stokes equation:

$$\rho_f\left(\frac{\partial \boldsymbol{V}}{\partial t}+((\boldsymbol{V}-\boldsymbol{V_g}).\nabla)\boldsymbol{V}\right)=-\nabla p+\mu\nabla^2\boldsymbol{V} \tag{2}$$

where p is the pressure, $\boldsymbol{V}$ is the fluid velocity vector, and $\boldsymbol{V_g}$ is the moving coordinate velocity. In the ALE formulation, $(\boldsymbol{V}-\boldsymbol{V_g})$ is the relative velocity of the fluid with respect to the moving coordinate velocity. Blood is assumed to be homogeneous with a constant density ρ_f=1050 kgm^{-3} and viscosity μ=3.5 mPa s (Valencia and Solis, 2006).

The governing equation for the solid domain is based on the momentum conservation equation. In contrast to the ALE formulation of the fluid field, here a Lagrangian coordinate system is adopted:

$$\nabla.\boldsymbol{\sigma_s}=\rho_s\boldsymbol{V_g^{\bullet}} \tag{3}$$

where $\boldsymbol{\sigma_s}$ is the solid stress tensor, and $\boldsymbol{V_g^{\bullet}}$ is the local acceleration.

2.3. Material properties of the arterial wall tissue

The hyperelastic model of Gasser et al. (2006) in its isotropic form has been adopted for the arterial tissue. According to this model, the energy function of the arterial wall consists of two main parts: The hyperelastic Mooney–Rivlin (or Neo-Hookean) term for the ground matrix including elastin fibres and the exponential term for collagen fibres:

$$W=n_e\frac{c_1}{2}(I_{1e}-3)+n_c\frac{k_1}{2k_2}(e^{k_2(I_{1c}-3)^2}) \tag{4}$$

where c_1 is the material constant of elastin and k_1 and k_2 are material constants related to the collagen fibres. n_e and n_c, respectively, represent the mass fraction of elastin and collagen. The magnitudes of the above constants were obtained from Watton et al. (2009b, 2011) as follows: $C_e=n_e\ c_1/2=66.9$ kPa, $C_c=n_c\ k_1/2k_2=1964$ Pa and $k_2=10$. I_{1e} and I_{1c} are the first invariants of the Cauchy–Green deformation tensor related to elastin and collagen respectively. The cause of this difference is the initial waviness of collagen fibres that lead to their delayed involvement.

In a morphological study, Finlay et al. (1998) showed that the collagen fibres at the centre of bifurcations, the most probable location of cerebral aneurysms, are arranged in a complex network that does not specify any particular fibre direction. On the other hand, Kindo (2006) observed the best agreement with experimental data in the model with both isotropic collagen and elastin. According to the above, both collagen and elastin have been considered isotropic.

In the isotropic modelling it is necessary to define a scalar criterion for the entire tissue deformation instead of fibre stretch to determine the status of collagen in terms of recruitment or remodelling. Following Wulandana and Robertson (2005a) we considered this scalar as a function of the first invariant of Cauchy–Green deformation tensor related to the ground matrix including elastin:

$$s_e=I_{1e}-3.0 \tag{5}$$

Therefore collagen should have one of the following conditions:

- $(s_e<s_{e,rec0})$ wavy collagen without load bearing
- $(s_{e,rec0}<s_e<s_{e,rem})$ load bearing collagen without remodelling
- $(s_{e,rem}<s_e)$ load bearing collagen with remodelling