SIMULATION OF VASCULAR VOLUME PULSATION OF RADIAL INDEX ARTERY

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ABSTRACT

This paper presents an application of finite element simulation with the analysis of arterial stiffness. The influences of the intravascular pressures on arterial wall which behaves like hyperelastic material was investigated by using Mooney-Rivlin hyperelastic constitutive model with the finite element solver of LS-DYNA. The results were obtained in the forms of nonlinear pressure-diameter relationship. Moreover, the diameter variation of arterial model corresponds to the pulsatile blood pressure. But the distensibility of artery reduces when the level of pulsatile blood pressure increases. These numerical results are expected to clarify an assessment of the arterial stiffness using photoelectric plethysmograph.

Keywords: arterial stiffness, vascular volume change, finite element simulation

1. INTRODUCTION

Arterial stiffness is associated with the development of cardiovascular risk factors. It is one of the indices which are used to diagnose the pathophysiology of cardiovascular system in both of research and clinical applications.

Arterial stiffness is typically investigated by monitoring the arterial motion in the circumferential direction. The instantaneous change of vessel circumference corresponds to the arterial pressure pulse which can be seen in the form of pressure-diameter relationship or pressure-volume relationship from many in vitro tests (Cox 1978a-c; Carew, Vaishnav and Petal 1968). They investigated the volume change of blood vessel by increasing the intravascular pressure and the transmural pressure. The intravascular pressure was measured by using pressure transducer. Meanwhile, in vivo tests, the pressure-volume relationship can be obtained from photoelectric plethysmographic (PPG) technique (Kawarada et al. 1986). PPG system is compatible with the clinical application. It is a non-invasive measurement system and easy to use. To evaluate the arterial blood volume change, the other tissues are considered to be incompressible. The venous system is collapsed by the exertion of external pressure. At the same time, the changes of arterial blood volume can be controlled by the decrease in external pressure and the increase in transmural pressure. With these assumptions and the Lambert-Beer's Law, Kawarada et al. (1986) and Ando et al. (1991) could investigate arterial elasticity by detecting the arterial volume change at any changes in transmural pressure from DC signal of PPG.

In Biomechanics, arterial stiffness associates with the mechanical properties of arterial wall which consists of three main layers; tunica intima, tunica media and tunica adventitia. It is believed that elastin and smooth muscle cell in media layer assist artery to resist high loads in the circumferential direction. At the same time, the thick bundles of collagen fibres in adventitia also contributes significantly to protect artery from overextension and rupture when artery is exerted by force from blood pressure. With the structure of arterial wall, it makes artery exhibits hysteresis under cyclic loading, stress relaxation under constant extension and creep under constant loads. This behavior can identify artery to be viscoelastic. However, arteries are frequently considered simply as hyperelastic material and all inelastic phenomena are neglected. Therefore, the constitutive models of arterial wall have been developed regarding the hyperelasticity and the distribution of collagen fibres which reflects an orthotropic property of arterial wall in both of microstructure (Bischoff, Arruda and Grosh 2002, Zhang et al. 2005) and macrostructure (Gasser, Ogden and Holzapfel 2006).

With the high performance of current computers, the constitutive models for arterial wall and finite element method have been widely implemented to observe the responses of arterial wall to various types of load. These observations have been utilized in many clinical application (Xia, Takayanagi and Kemomochi 2001; Zhang et al. 2005, 2007; Zhao et al. 2008).

In this study, the responses of arterial wall to the loads have been observed by coupling anisotropic hyperelastic constitutive models with the finite element method. These numerical results are expected to clarify the assessment of the arterial stiffness using PPG.

2. MATERIAL AND METHODS

2.1 The finite element model of artery

A radial index artery was considered in order to apply this study with PPG application. A tube with three layers of arterial wall; tunica intima, tunica media and tunica adventitia, was combined into one-layer to simplify anatomical structure of artery. The outer diameter of the tube was 1.54 mm (Bilge et al. 2006). The ratio of total wall thickness to outer diameter was 0.189 mm (Holzapfel et al. 2005). The length of the tube was specified to be 10 mm. This model was discretised into 15,360 hexahedron solid elements by using LS-Prepost version 2.1. All elements were assumed to be the constant stress solid element in order to avoid volumetric locking effect as shown in Figure 1.

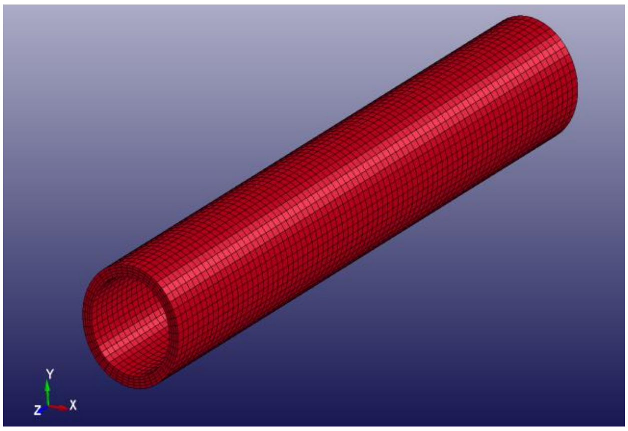


Figure 1: The geometrical model of radial index artery

2.2 The finite element method

The commercial explicit finite element solver of LS-DYNA 971 was employed to investigate the responses of arterial wall on various types of pressure which are given in the following form:

- Intravascular pressures which represented mean blood pressure were assigned to be 10, 20, 30, 40, 50, 70, 90, 100, 120, 140, 160 and 180 mmHg at the inner wall of the tube as the ramped loads. The pressure raised from 0 to the maximum pressure of each level within 100 ms. This was done to obtain the pressure-diameter relationship of the artery.
- Pulsatile loads which represented pulsatile blood pressures were given in the form of sinusoidal function to simplify the arterial pressure pulse.

$$P(t) = \overline{P} + P_{amp} \sin(2\pi f t) \tag{1}$$

where P(t) is the instantaneous pressure (mmHg), $\overline{P}$ is mean level of the pressure pulse (mmHg). The level of $\overline{P}$ was assigned to be 70, 100 and 120 mmHg which represented low blood pressure, normal blood pressure and high blood pressure conditions, respectively. P_{amp} is the amplitude of pulsatile pressure (mmHg) which relates to systolic pressure and diastolic blood pressure. In this study, the systolic/diastolic blood pressures were 90/50, 120/80 and 140/100 mmHg for low blood pressure, normal blood pressure and high blood pressure conditions, respectively. f is frequency (Hz) and t is time (s). Pulsatile pressures were also applied at the inner wall of the tube to investigate the responses of arterial wall on pulsatile pressure.

- External pressure was assume to be zero in this simulation.

These specifications were considered as traction boundary conditions of boundary value problem. Meanwhile, the displacement boundary condition was applied at the annulus of the tube which constrained movement in all direction.

The numerical models were based on solving the momentum balance equation and boundary conditions which are given as follows:

The momentum balance equation:

$$\nabla \cdot \boldsymbol{\sigma} + \rho_0 \mathbf{f} = \rho_0 \mathrm{a} \tag{2}$$

where $\boldsymbol{\sigma}$ is the Cauchy stress, $\mathbf{F}$ is the deformation gradient, ρ_0 is the mass density, $\mathbf{f}$ is the body force and $\mathbf{a}$ is the acceleration.

The traction boundary condition:

$$\mathrm{N} \cdot \boldsymbol{\sigma} = \overline{\mathbf{T}} \tag{3}$$

where $\overline{\mathbf{T}}$ is traction force and $\mathbf{N}$ is the unit normal vector.

The displacement boundary condition:

$$\mathbf{u} = \overline{\mathbf{u}} \tag{4}$$

where $\mathbf{u}$ is the displacement.

When $x^{+} = x^{-}$The contact discontinuity:

$$\mathbf{N} \cdot (\boldsymbol{\sigma}^{+} - \boldsymbol{\sigma}^{-}) = 0 \tag{5}$$

The Cauchy stress in the momentum balance equation related to the strain energy function through the constitutive equation.

Constitutive Equations for incompressible hyperelastic material can be expressed as follows:

$$\boldsymbol{\sigma} = \mathrm{F}\frac{\partial \mathbf{W}}{\partial \mathbf{F}} - p\mathbf{I} \tag{6}$$

where W is the strain energy function, F is the deformation gradient tensor, p is the Lagrange multiplier and $\mathbf{I}$ is the identity tensor.

The strain energy function for arterial wall was defined by a Mooney-Rivlin hyperelastic constitutive model as follow:

$$W = C_{10}(I_1 - 3) + C_{01}(I_2 - 3) + C_{20}(I_1 - 3)^2 \\ + C_{11}(I_1 - 3)(I_2 - 3) + C_{30}(I_1 - 3)^3 \tag{7}$$

where C_{10}, C_{01}, C_{20}, C_{11} and C_{30} are hyperelastic coefficients used for artery. I_1 and I_2 are invariants which can be expressed as

$$I_1 = tr\,\mathbf{B} \tag{8}$$

$$I_2 = \frac{1}{2}[I_1^2 - tr(\mathbf{B}^2)] \tag{9}$$

where $\mathbf{B}$ is the left Cauchy-Green deformation tensor.

The hyperelastic coefficients of arterial wall which were reported by Loree et al. (1994) are shown in Table 1

Table 1: hyperelastic coefficients of arterial wall

	C_{10} [KPa]	C_{01} [KPa]	C_{20} [KPa]	C_{11} [KPa]	C_{30} [KPa]
Artery	708.416	-620.042	0	2827.33	0

2.3 Analysis of results

In this simulation, the change in diameter of artery was obtained from the average movement of radial position of the inner wall at every 1 mm length in axial direction using post-processing of LS-DYNA software.

3. RESULTS AND DISCUSSIONS

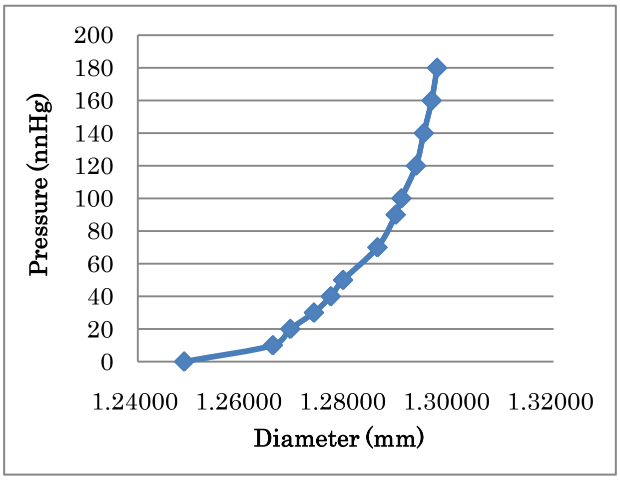
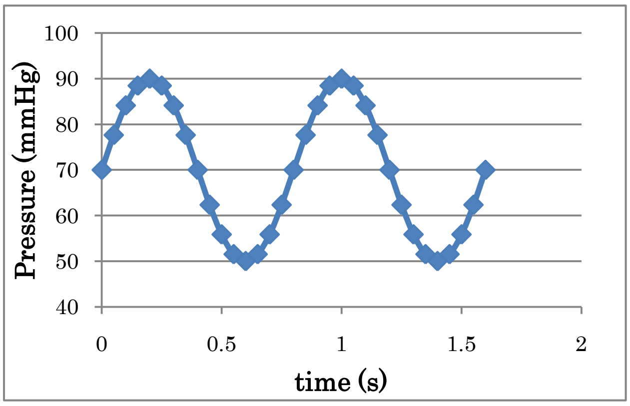


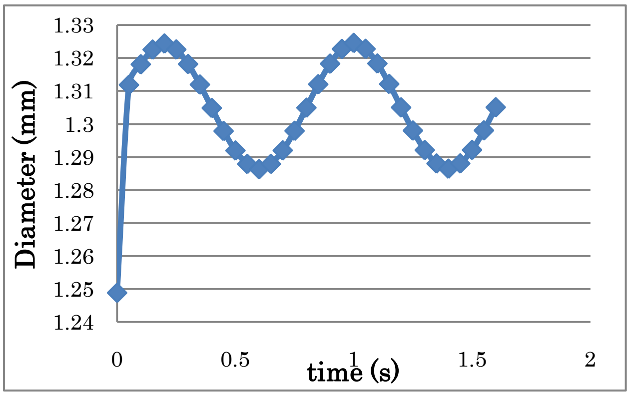
Figure 2: Pressure-diameter relationship of artery

Figure 2 shows the relationship between intravascular pressure and diameter of arterial tube which was obtained by varying the level of intravascular pressure as in section 2.1. This curve shows that using Mooney-Rivlin hyperelastic model generates non-linear pressure-diameter relationship. At the lower level of intravascular pressure, 10-70 mmHg, the distensibility of arterial model is higher than the upper level pressure, 90-180 mmHg. This shows that arterial model is stiffer at higher pressure level. This result corresponds to the relationship between the transmural pressure and volume elastic modulus of rabbit artery from the experiment of Kawarada et al. (1986) that the elastic modulus of artery increases with transmural pressure nonlinearly.

This result confirms that the simulation model of artery behaves similarly to the real artery, it is also suitable to apply this model with the investigation of the influence of pulsatile pressure on arterial wall.

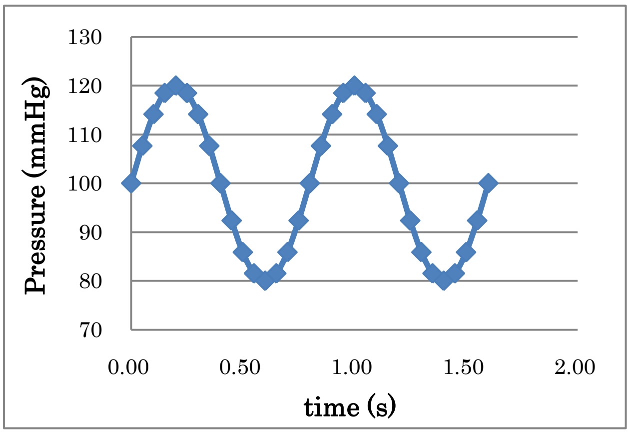


(a)

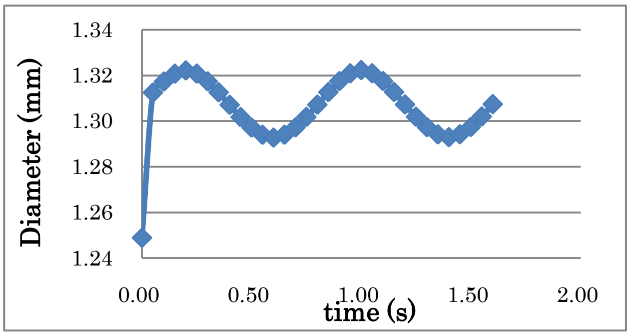


(b)

Figure 3: (a) Pressure-time relationship and (b) Diameter-time relationship of low blood pressure condition.

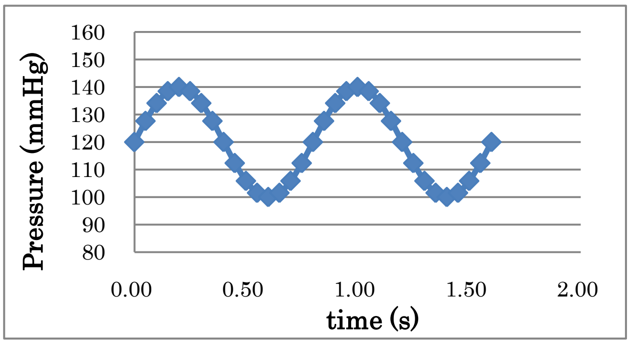


(a)

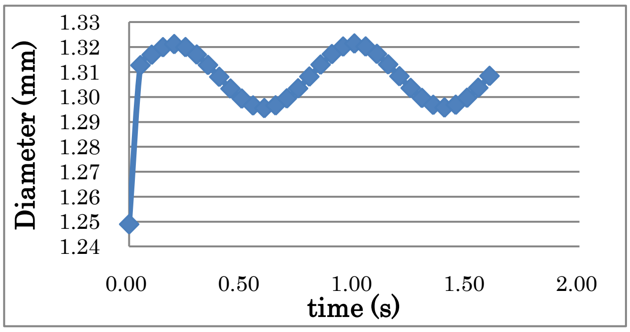


(b)

Figure 4: (a) Pressure-time relationship and (b) Diameter-time relationship of normal blood pressure condition.



(a)



(b)

Figure 5: (a) Pressure-time relationship and (b) Diameter-time relationship of high blood pressure condition.

Figure 3, 4 and 5 show the responses of arterial diameter on pulsatile pressure in low blood pressure, normal blood pressure and high blood pressure conditions, respectively. The results show that arterial diameters vary with pulsatile pressure in all conditions. The difference of arterial response in each blood pressure condition is the amplitude of diameter variation. Within the same level of the amplitude of pulsatile pressure, the amplitude of diameter variation in high blood pressure condition is the lowest one (1.14% of mean diameter). Meanwhile, the amplitude of diameter variation in normal blood pressure condition (1.28% of mean diameter) is also lower than the amplitude in low blood pressure condition (1.62% of mean diameter). These are because the arterial distensibility decreases when the level of intravascular pressure increases which correspond to the pressure-diameter relationship.

The investigation of the responses of arterial diameter on pulsatile pressure is just only the first step to apply finite element method with the analysis of arterial elasticity. To apply the simulation results with the analysis of arterial elasticity using PPG. The further simulations need to be carried on for the future work. For example, the pulsatile pressure should be modified to be similar to the real pulsatile blood pressure. Moreover, the influence of external pressure (or cuff pressure for PPG) on diameter of arterial model also need to be studied.

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