Effect of Axial Stretch on Lumen Collapse of Arteries

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The stability of the arteries under in vivo pressure and axial tension loads is essential to normal arterial function, and lumen collapse due to buckling can hinder the blood flow. The objective of this study was to develop the lumen buckling equation for nonlinear anisotropic thick-walled arteries to determine the effect of axial tension. The theoretical equation was developed using exponential Fung strain function, and the effects of axial tension and residual stress on the critical buckling pressure were illustrated for porcine coronary arteries. The buckling behavior was also simulated using finite-element analysis. Our results demonstrated that lumen collapse of arteries could occur when the transmural pressure is negative and exceeded a critical value. This value depends upon the axial stretch ratio and material properties of the arterial wall. Axial tensions show a biphasic effect on the critical buckling pressure. The lumen aspect ratio of arteries increases nonlinearly with increasing external pressure beyond the critical value as the lumen collapses. These results enhance our understanding of artery lumen collapse behavior.
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Keywords: mechanical instability, lumen collapse, critical buckling pressure, axial stretch, artery

1 Introduction

Blood vessels collapse when the external pressure is much higher than the lumen pressure, i.e., the transmural pressure, the difference between the lumen pressure and the external pressure, is highly negative [1,2]. While lumen collapse is commonly seen in vein due to their thinner wall and low-lumen pressure [1,2], arteries may collapse under certain conditions such as peripheral arteries compressed by pressure cuffs or intramyocardial coronary arteries during left ventricle contraction [2–4]. It has also been reported that negative transmural pressure is induced at the throat of the stenosis in arteries under cyclic pressure that may cause critical conditions related to heart attack and stroke [2,5–7]. Lumen collapse hinders or even blocks the flow in the artery that can lead to ischemia to distal organs and tissues. Therefore, a better understanding of mechanical stability of arteries under in vivo loads is of clinical importance.

Buckling is a mechanical instability that can lead to sudden occurrence of large deformation and structure failure. The minimum load that causes instability in the structure is the critical load [1]. Arteries can be considered cylindrical structures under complex mechanical loads due to blood pressure, tissue tethering, and

body movements [2]. Arteries may lose their mechanical stability under these loads, and different types of artery buckling have been reported including lumen buckling, bent buckling, twist buckling, and helical buckling [2,8,9]. This study examined the lumen buckling under the transmural pressure.

The buckling equation has been developed for thin-walled cylindrical vessels of linear elastic material subjected to a transmural pressure [1]. Arteries are thick-walled tubes under in vivo loads including lumen blood pressure, external surrounding tissue compression, and axial tension [2]. The thin-walled model described by Fung needs to be extended for thick, nonlinear arterial wall under axial stretch ratio. Using a linear elastic model, Ku and colleague demonstrated that buckling of thick-walled arteries with eccentric atherosclerotic plaque was significantly different from thin-walled veins [10]. Many studies have been done on flow-induced collapse of blood vessels that model blood vessels as elastic tube under a steady or pulsatile flow using fluid–structure interaction simulation [11,12]. Both computational simulations and experimental measurement have demonstrated that wall thickness influences the critical buckling pressure and flow pattern within the tube [11–13]. Therefore, it is important to investigate the buckling behavior of thick-walled arteries.

In addition, using a rubber tube model, Bertram showed experimentally that a small axial stretch could affect the critical buckling pressure and the pressure–lumen area curve of silicon rubber tubes [14]. With the significant axial stretch in arteries and veins in vivo that varies with aging, body movement, pregnancy, growth, and vascular surgery [15–18], it is important to understand its effects on the critical buckling pressure.

Therefore, the objectives of this study were to develop the lumen buckling equation for the nonlinear anisotropic thick-walled arteries and simulate the postbuckling behavior to determine the effect of axial tension on lumen buckling.

2 Methods

We determined the collapse of cylindrical vessel of nonlinear anisotropic material at various axial stretch ratios. Theoretical estimations then compared to computational results from a finite-element analysis for normal porcine coronary arteries.

2.1 Deformation Analysis of Cylindrical Arteries. The coronary artery is considered as a cylindrical tube with wall material assumed to be homogenous, orthotropic with Fung-type strain energy function [19]. Considering the residual strains in the arteries [19,20], we denoted the opening angle (central angle of the open sector) as $(2\Theta_0)$ for the artery [21]. The dimensions of the segment at zero stress configuration are designated by initial inner radius (R_i), outer radius (R_e), and length of the artery open sector (L). When the artery is under an internal pressure (p_i), external pressure (p_e), and axial tension (N_a), the inner radius, the outer radius, and the length of artery become (r_i), (r_e), and (l), respectively. The average axial stress is defined as (σ_z^0) , which is corresponding to an axial elongation of stretch ratio (λ_z^0). Using cylindrical coordinates, a material point (R, Θ, Z) in the stress-free state (open sector) deforms into the point (r, θ, z) in the loaded state

$$r = r(R, p_i, p_e); \quad \theta = \frac{\pi}{\Theta_0} \Theta; \quad z = \lambda_z^0 Z \quad (1)$$

Accordingly, in cylindrical coordinates, the Green's strains are

$$E_r = \frac{1}{2}(\lambda_r^2 - 1), \quad E_\theta = \frac{1}{2}(\lambda_\theta^2 - 1), \quad E_z = \frac{1}{2}(\lambda_z^2 - 1) \quad (2)$$

where

$$\lambda_r = \frac{\partial r}{\partial R}, \quad \lambda_\theta = \frac{\pi}{\Theta_0} \frac{r}{R}, \quad \lambda_z = \lambda_z^0 \quad (3)$$

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are the radial, circumferential, and axial stretch ratios, respectively.

The arterial wall is assumed to have the exponential Fung strain energy function [19,22]

$$w = \frac{1}{2} b_0 e^Q, \quad Q = b_1 E_r^2 + b_2 E_\theta^2 + b_3 E_z^2 + 2b_4 E_r E_\theta + 2b_5 E_z E_r + 2b_6 E_\theta E_z \quad (4)$$

where b_0 to b_6 are the material property constants. Accordingly, the Cauchy stress components are given by [19,22]

$$\sigma_i = \lambda_i^2 \frac{\partial w}{\partial E_i} + p, \quad i = r, \theta, z \quad (5)$$

where p is the static pressure for wall incompressibility. The equations of equilibrium for the straight cylindrical artery are [19,22]

$$\frac{d\sigma_r}{dr} + \frac{\sigma_r - \sigma_\theta}{r} = 0 \quad (6)$$

with boundary conditions

$$\sigma_r(r_i) = -p_i; \quad \sigma_r(r_e) = -p_e \quad (7)$$

Integrating Eq. (6) and using boundary conditions Eq. (7) and then combining with Eq. (5), the stress components can be expressed as [21,23]

$$\begin{aligned} \sigma_r &= -p_i + \int_{r_i}^r \left[\lambda_\theta^2 \frac{\partial w}{\partial E_\theta} - \lambda_r^2 \frac{\partial w}{\partial E_r} \right] \frac{dr}{r} \\ \sigma_\theta &= -p_i + \left[\lambda_\theta^2 \frac{\partial w}{\partial E_\theta} - \lambda_r^2 \frac{\partial w}{\partial E_r} \right] + \int_{r_i}^r \left[\lambda_\theta^2 \frac{\partial w}{\partial E_\theta} - \lambda_r^2 \frac{\partial w}{\partial E_r} \right] \frac{dr}{r} \quad (8) \\ \sigma_z &= -p_i + \left[\lambda_z^2 \frac{\partial w}{\partial E_z} - \lambda_r^2 \frac{\partial w}{\partial E_r} \right] + \int_{r_i}^r \left[\lambda_\theta^2 \frac{\partial w}{\partial E_\theta} - \lambda_r^2 \frac{\partial w}{\partial E_r} \right] \frac{dr}{r} \end{aligned}$$

Then, by applying the boundary conditions for the first equation in Eq. (8) and integrating the axial stress (σ_z) over the cross section area, the transmural pressure ($p_e - p_i$) and the axial force (N_0) in the artery are given by

$$p_e - p_i = \int_{r_i}^{r_e} \left(\lambda_r^2 \frac{\partial w}{\partial E_r} - \lambda_\theta^2 \frac{\partial w}{\partial E_\theta} \right) \frac{dr}{r} \quad (9)$$

and

$$N_0 = \pi r_i^2 p_i - \pi r_e^2 p_e - \pi \int_{r_i}^{r_e} \left(\lambda_r^2 \frac{\partial w}{\partial E_r} + \lambda_\theta^2 \frac{\partial w}{\partial E_\theta} - 2\lambda_z^2 \frac{\partial w}{\partial E_z} \right) r dr \quad (10)$$

2.2 Buckling Deformation Analysis. To investigate the buckling behavior of arteries, we considered the critical condition when the vessel starts to collapse. Based on the curved beam bending theory and the solution for thin-walled vessel collapse [1], the curvature change for the initially circular vessel was assumed to be a sinusoidal waveform at the neutral layer when the collapse occurs

$$\frac{1}{\rho_m} - \frac{1}{r_m} = \kappa = C \cdot \sin(n\theta) \quad (11)$$

where C is a small-value constant, n is a positive integer, and (r_m) and (ρ_m) are the radius of curvature of the neutral axis in the circular and buckled configurations with subscript m representing the neutral layer (Fig. 1). Using the plane assumption for curve beam bending, the circumferential stretch ratio ($\tilde{\lambda}_\theta$) at the layer with radius r due to the bending can be obtained as

$$\tilde{\lambda}_\theta = \lambda_\theta (1 + \kappa(r - r_m)) \quad (12)$$

where λ_θ is the circumferential stretch ratio before buckling given in Eq. (3). By neglecting the higher order terms of C , the incremental circumferential strain becomes

$$\Delta E_\theta = (\lambda_\theta^2) \kappa (r - r_m) \quad (13)$$

Accordingly, the incremental circumferential stress ($\Delta\sigma_\theta$) can be determined based on the incremental circumferential strain (ΔE_θ) using Eq. (8) [23]

$$\Delta\sigma_\theta = J_\theta \Delta E_\theta + \int_{r_e}^r J_\theta \Delta E_\theta \frac{dr}{r} \quad (14)$$

where

$$\begin{aligned} J_\theta &= [2g_\theta + (1 + 2E_\theta)b_2 - (1 + 2E_r)b_4] b_0 e^Q \\ &+ 2g_\theta [(1 + 2E_\theta)g_\theta - (1 + 2E_r)g_r] b_0 e^Q \quad (15) \end{aligned}$$

with

$$\begin{aligned} g_r &= (b_1 E_r + b_4 E_\theta + b_5 E_z) \\ g_\theta &= (b_4 E_r + b_2 E_\theta + b_6 E_z) \\ g_z &= (b_5 E_r + b_6 E_\theta + b_3 E_z) \quad (16) \end{aligned}$$

Note that though the stresses and strains are axisymmetric before buckling, the small bent at buckling leads to incremental stress and strain that are nonaxisymmetric. The bending moment (M_θ) per unit vessel length can be obtained by integrating the moment generated by the incremental circumferential stress ($\Delta\sigma_\theta$) across the arterial wall thickness

$$\begin{aligned} M_\theta &= \int (\Delta\sigma_\theta)(r - r_m) dr = \int_{r_i}^{r_e} J_\theta \Delta E_\theta (r - r_m) dr \\ &- \frac{1}{2} \int_{r_i}^{r_e} J_\theta \Delta E_\theta [r^2 - 2(r - r_i)r_m - r_i^2] \frac{dr}{r} \quad (17) \end{aligned}$$

Taking Eq. (13) into Eq. (17) yields

$$\begin{aligned} M_\theta &= \kappa \int_{r_i}^{r_e} J_\theta (1 + 2E_\theta) (r - r_m)^2 dr \\ &- \frac{\kappa}{2} \int_{r_i}^{r_e} J_\theta (1 + 2E_\theta) (r - r_m) (r - r_i) (r + r_i - 2r_m) \frac{dr}{r} \quad (18) \end{aligned}$$

Therefore, an “equivalent rigidity” becomes

$$\begin{aligned} EI &= \frac{M_\theta}{\kappa} = \int_{r_i}^{r_e} J_\theta (1 + 2E_\theta) (r - r_m)^2 dr \\ &- \frac{1}{2} \int_{r_i}^{r_e} J_\theta (1 + 2E_\theta) (r - r_m) (r - r_i) (r + r_i - 2r_m) \frac{dr}{r} \quad (19) \end{aligned}$$

Also, the incremental circumferential tension (per unit vessel length) can be determined by

$$\Delta N_\theta = \kappa r_i \int_{r_i}^{r_e} J_\theta (1 + 2E_\theta) (r - r_m) \frac{dr}{r} \quad (20)$$

Letting $\Delta N_\theta = 0$, we determined the neutral axis radius

$$r_m = \frac{\int_{r_i}^{r_e} J_\theta (1 + 2E_\theta) dr}{\int_{r_i}^{r_e} J_\theta (1 + 2E_\theta) \frac{dr}{r}} \quad (21)$$

with (EI) given in Eq. (19), using the small sinusoidal deflection given in Eq. (11) and the adjacent equilibrium approach [1], a critical buckling pressure was obtained in the form of

$$p_{cr} = (p_e - p_i)_{critical} = \frac{3EI}{r_m^3} \quad (22)$$

Note that although the expression is similar to the classic thin-walled tube buckling, both (EI) and (r_m) are complicated functions of material constants and strains, including the axial strain.

2.3 Numerical Simulations. For a set of five porcine coronary arteries with material constants reported in the literature (Table 1) [24], the critical buckling pressure was determined based on the model equations presented above. These material constants were obtained from pressurized inflation testing of five normal porcine left anterior descending arteries (LAD) freshly harvested postmortem. The arteries were tested in saline solution at room temperature (22°C), and more details can be found in Ref. [24]. Briefly, for each artery, the wall was divided into sub-layers, and the radii under pressure were determined using incompressibility equation as previously described

$$r = \sqrt{r_i^2 + \frac{\Theta_0}{\pi \lambda_z^0} (R^2 - R_i^2)} \quad (23)$$

Accordingly, the strains at each layer were calculated, and thus the stress and the pressure as well as modulus (EI) and neutral axis (r_m) were determined. The critical pressures (p_{cr}) at given axial stretch ratio were determined using try and error iteration approach which used Eqs. (9) and (22) to determine the pressures for each testing (r_i) until they converge to the same pressure [21,23].

2.4 Finite-Element Analysis of Buckling and Postbuckling Collapse. Buckling analysis was conducted using the commercial FEA package ABAQUS®. Cylindrical arterial models were created for the same five arteries analyzed above with the same Fung model and material constants (Table 1). Normal coronary artery was assumed to be cylindrical with opening angle of 2π (ignored residual stress) and length of 10 mm, inner diameter of 0.8 mm, and wall thickness of 0.5 mm in the unloaded condition. The models were meshed using quadratic hexagonal elements. Both ends

Fig. 1 Schematic of a sector of an artery cross section deforming from circular configuration into buckled configuration. When the artery is under an internal pressure (p_i) and an external pressure (p_e), the inner radius and the outer radius are (r_i) and (r_e), respectively. The deformation is based on the plane assumption of curved beam bending, with (r_m) and (ρ_m) denoting the radius of the neutral axis in the circular and buckled states.

Table 1 Material constants for the Fung's strain energy function of five porcine coronary arteries used in our model simulations (from Ref. [24])

Artery no.	b_0 (kPa)	b_1	b_2	b_3	b_4	b_5	b_6
1	11.39	1.39	2.62	0.51	0.32	0.06	0.06
2	13.86	1.26	1.89	0.52	0.27	0.05	0.14
3	4.16	1.58	2.66	0.73	0.3	0.03	0.08
4	5.94	1.27	2.78	0.62	0.45	0.13	0.13
5	11.64	1.03	2.09	0.38	0.37	0.02	0.06

of the cylindrical artery were allowed to expand axially without any lateral displacement or rotation. A static transmural pressure was applied to the outer surface of arterial models, while the internal pressure was set at zero [10]. Since arteries are under considerable longitudinal strain in vivo, axial displacements in longitudinal direction were applied at both ends to create various axial stretch ratios in the models. Then, an external pressure was applied to generate buckling and keep increasing to simulate the buckling behavior of all models at a series of designated stretch ratios [25].

An initial imperfection was used in our models to facilitate the buckling. There are various approaches to create imperfections including ovalization, variation in thickness, and material imperfections [26]. In this study, 1% variance in the thickness was created as an initial imperfection, while the lumen remained perfect circle. Different amounts of possible variations in thickness were examined, and the results showed that these variations did not have any significant effect on the critical buckling pressure results. Different mesh sizes were also compared to determine the mesh sensitivity and determine the final mesh size. Arteries were compressed (diameter decreased) initially by applying a gradual static pressure on their outer surface. When the external pressure exceeded a critical value, arteries buckled and started to become elliptic and aspect ratio increased nonlinearly (postbuckling). The compression was initially small and then suddenly increased to become visually large with further increase of the pressure. The lumen aspect ratio, the roundness, and the lumen area were calculated in each step to capture the cross-sectional shape change. The aspect ratio was defined as the major to minor axis ratio, and the roundness was defined as the percentage of the difference between the major and minor axes. The critical buckling pressures from the FEA simulations were defined as the pressure at which the aspect ratio increased by 1.7%. This value was selected based on the postbuckling analysis of all five arteries to capture the sudden change in shapes.

3 Results

We performed theoretical studies using the material parameters reported in Table 1 [24] with $R_i = 0.8$ mm, $R_e = 1.3$ mm, and an opening angle of 2π (ignored residual stress).

3.1 Location of Neutral Axis. The neutral axis radius was determined using Eq. (21) or by using a weighted average of (r_i)

Table 2 A comparison of the radius of the neutral axis of an artery (artery no. 2 in Table 1) obtained from Eq. (21) and weighted average of inner and outer radii

Stretch ratio	r_m (Eq. (21)) (mm)	$0.6r_i + 0.4r_e$ (mm)
1.1	0.984	0.962
1.2	0.945	0.925
1.3	0.906	0.887
1.4	0.866	0.845
1.5	0.824	0.798
Mean	0.905	0.8834

Fig. 2 Comparison of the critical buckling pressures obtained by theoretical model equations (theory) and by FEA simulations (FEA) for five porcine coronary arteries (artery 1–5 in Table 1) at various axial stretch ratios

and (r_e) based on histological observations. The neutral axis obtained from these two approaches for artery no. 2 is compared in Table 2. Our results indicated that the weighted average of (r_i) and (r_e) is a reasonable approximation for the neutral axis.

3.2 Critical Pressures and Effect of Axial Stretch. The critical buckling pressures were obtained for a group of five coronary arteries at stretch ratios of (1.1–1.5) using the theoretical model equations and FEA simulations (Fig. 2). The results showed that arteries buckled when the negative transmural pressure exceeded a

critical value. Also, with an increase in the axial stretch ratio, the simulations demonstrated that the value of critical pressure first dropped slightly and then rose (Fig. 2). The results from model equation and FEA simulations demonstrated good agreement.

3.3 Critical Pressures and Effect of Residual Stress. The effect of residual stress on the artery lumen buckling was investigated by varying the opening angle ($2\Theta_0$) in the model equations. The results showed that the opening angle (thus the residual stress) affects the critical buckling pressure (Fig. 3). It is seen that the impact can be dramatic at small opening angle (large residual stress) but is very little at large opening angle over 300 deg (small residual stress). It was reported that the opening angle for porcine LAD was very large ($2\Theta_0 = 340$ deg) [27]. So we ignored the residual stress in our model simulations and FE simulations.

3.4 Postbuckling Deflection of Arteries. Our simulations also captured the postbuckling behavior of the arteries. It was seen that when the pressure continued to increase beyond the critical buckling pressure, the lumen aspect ratio continued to increase nonlinearly, and the lumen area continued to decrease (Fig. 4). The axial stretch ratio does not have any noticeable effect on the postbuckling shape of the artery's cross section.

4 Discussion

In this study, we established the lumen buckling equation for arteries using a nonlinear anisotropic thick-walled cylindrical model and investigated the effect of axial tension on the critical buckling pressure. Postbuckling behavior of arteries was investigated using computational simulations. Our results demonstrated that arteries could buckle when the transmural pressure is negative and exceeded a critical value. This value depends upon the wall material properties, the axial stretch ratio, and the residual stress in the arteries. Our simulation results showed that the arterial lumen aspect ratio continued to increase nonlinearly with increasing the external pressure beyond the critical value. Furthermore, the critical buckling pressures of arteries predicted by FEA were in good agreement with the theoretical predictions.

To further validate the model results, we compared our model results and thin-walled elastic tube model equation given in Ref. [1]

Fig. 3 Change of the critical buckling pressure with opening angle ($2\Theta_0$): (a) artery no. 1 at an axial stretch ratio = 1.1 and (b) artery no. 2 at an axial stretch ratio = 1.5

$$p_{cr} = \frac{Eh^3}{4(1-\nu^2)r^3} \quad (24)$$

The Young's modulus E and the Poisson's ratio ν were determined from the material constants using the converting equations given in Ref. [28]. For the case of $\lambda_z = 1$, the results for the five arteries demonstrated a good agreement (results not shown).

Compared with previous studies [1,10,29,30], the advantage of the buckling equation obtained in this study lies in its capability to model nonlinear anisotropic material for thick-walled artery. This model overcomes the limit of thin-wall and linear elastic assumptions. A new finding from the current study is that the axial stretch ratio affects the critical buckling pressure. The biphasic change of critical pressure with the axial stretch ratio could be due to the competitive effects of the decreases in wall thickness and radius under axial stretch or could be related to the decoupled axial stretch-diameter change relations over the certain physiological

range [22,31]. Further studies are needed to elucidate the mechanism.

Both arteries and veins are exposed to axial stretch which varies with aging, body movement, pregnancy, growth, and vascular surgery [15–18,32,33]. For example, the axial stretch ratio in arteries decreases with aging but may increase due to vascular surgery [15,17,18,32]. The current model and results help us to better understand the buckling behavior of arteries under these pathological conditions. The model equation can also be used to analyze the buckling behavior of other tubular structures such as veins, lymphatic tubes, ureter, esophagus, or umbilical cord [2,34,35].

There were a few limitations in this study. One limitation is that our analysis was limited to the arterial wall as a single layer anisotropic material, although arterial walls are multilayered structures. Another limitation is that the model is limited to the critical buckling pressure, and the postbuckling analysis was only done with FEA simulations. While Fung strain energy function

Fig. 4 Deformation of an artery under increasing external pressure. (a) Cross-sectional views from left to right: under no load, axially stretched, under a small pressure without buckling, slightly buckled, and postbuckling. (b) Aspect ratio of the lumen (major to minor axis ratio) plotted with pressure for different axial stretch ratios. (c) Lumen area plotted with pressure for different axial stretch ratios.

was used in our current analysis, the model can be extended to other strain energy functions such as the four-fiber model [23,36]. While it is possible that the Fung material constants obtained under inflation tests may not well represent the behavior under compression, our simulations were to demonstrate the capability of the nonlinear elastic buckling model, not how accurate the model can predict the critical pressure, due to the lack of experimental results. Both model simulation and FE simulation used the same material constants, and the results match each other well. On the other hand, Yu and Fung's previous work demonstrated that the compressive behavior is similar to tensile behavior at small load range [37], and the critical pressure for artery lumen buckling happens in the small load range [10]. Interestingly, arteries are under radial compression in the inflation tests, and Fung model has been successfully used for these conditions. Furthermore, it has been shown that the Fung strain energy function can well describe the compressive behavior of brain tissues [38]. Further studies are needed to experimentally measure the critical buckling pressure at different axial stretch ratios and determine material constants for arterial wall under compression.

Collapse of arterial wall may cause compressive stresses in arteries which are normally constructed for tension. Due to the pulsatile changes of pressure load, the compressive stress could be cyclic and may lead to mechanical fatigue of arteries and cracks in the arterial wall [6,10,39]. Local collapse may occur in coronary arteries distal to stenosis [6] or due to myocardial bridge [3,4]. This causes vessel compression resulting in hemodynamic changes which probably lead to angina, myocardial ischemia, cardiac arrhythmias, syncope, or even sudden cardiac death [3,4]. Thus, understanding of artery cross section buckling (collapse) behavior is important in understanding these vascular disorders.

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