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Twist Buckling Behavior of Arteries

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Abstract

Arteries are often subjected to torsion due to body movement and surgical procedures. While it is essential that arteries remain stable and patent under twisting loads, the stability of arteries under torsion is poorly understood. The goal of this work was to experimentally investigate the buckling behavior of arteries under torsion and to determine the critical buckling torque, the critical buckling twist angle, and the buckling shape. Porcine common carotid arteries were slowly twisted *in vitro* until buckling occurred while subjected to a constant axial stretch ratio (1.1, 1.3, 1.5 (in vivo level), and 1.7) and lumen pressure (20, 40, 70, and 100 mmHg). Upon buckling, the arteries snapped to form a kink. For a group of six arteries, the axial stretch ratio significantly affected the critical buckling torque ($p < 0.002$) and the critical buckling twist angle ($p < 0.001$). Lumen pressure also significantly affected the critical buckling torque ($p < 0.001$) but had no significant effect on the critical twist angle ($p = 0.067$). Convex material constants for a Fung strain energy function were determined and fit well with the axial force, lumen pressure, and torque data measured pre-buckling. The material constants are valid for axial stretch ratios, lumen pressures, and rotation angles of 1.3 – 1.5, 20 – 100 mmHg, and 0 – 270 degrees, respectively. The current study elucidates the buckling behavior of arteries under torsion and provides new insight into mechanical instability of blood vessels.

Keywords

stability; instability; kink; twist; torsional buckling; blood vessel; porcine; carotid artery

1. INTRODUCTION

Arteries and veins *in vivo* are often subjected to large degrees of torsion, i.e. twist along its longitudinal axis (Cheng et al. 2006; Klein et al. 2009; Salgarello et al. 2001; Wong et al. 2007). The twisting of blood vessels and vascular grafts may increase flow resistance and disrupt blood flow, causing ischemia to distal organs (Barton and Margolis 1975; Endean et al. 1989; Macchiarelli et al. 1991). It is essential that arteries remain mechanically stable and patent under torsion in order to maintain their normal function (Fung 1997; Han 2012). However, the mechanical stability of arteries under torsion is poorly understood.

Arteries, especially those located within mobile anatomical regions, are often subjected to torsion with body movement. For example, the vertebral artery is vulnerable to twist due to a sharp angulation as it travels through the upper cervical vertebrae and into the skull (Norris et al. 2000). In vivo arteriographs have shown that rotation of the head can initiate obstruction or kinking of the vertebral artery at the atlantoaxial joint and may lead to

vertebrobasilar symptoms (Barton and Margolis 1975). Rotation of the head also places considerable torsion on the common carotid and internal carotid arteries, compounded by their tendency to twist around each other (Vos et al. 2003). Mobile arteries within the torso and lower extremities such as the abdominal aorta, iliacs, superficial femorals, and femoropopliteals are also subjected to torsion with hip and knee flexion, which is thought to contribute to stent failure (Cheng et al. 2006; Choi et al. 2009b; Klein et al. 2009; Laird 2006; Scheinert et al. 2005). Twisting of the right coronary arteries has also been observed with the cardiac cycle (Choi et al. 2009a; Ding et al. 2002).

Arteries can also experience torsion during vascular surgeries. Surgical implantation of vascular grafts and microanastomosis procedures may result in inadvertent twisting of the arteries or grafts causing the vessel to mechanically buckle into a kinked or waved configuration (Dobrin et al. 2001; Selvaggi et al. 2004). Twisted vessels can create alterations in blood flow and impaired endothelial regeneration on the lumen surface at the anastomosis site leading to vascular degeneration or thrombosis (Endean et al. 1989; Macchiarelli et al. 1991). Artery twisting also occurs in perforator-based propeller flap procedures for skin grafting in which a skin island, still connected to its' perforating artery and vein, is elevated and rotated like a helicopter propeller up to 180° using the perforating vessels as a pivot point (Hyakusoku et al. 1991; Jakubietz et al. 2007). Although positive results for this procedure have been reported, patency of the perforating vessels remains a concern and long term studies are limited (Wong et al. 2007). Therefore, better understanding the mechanical stability of arteries under torsion is needed for improving vascular surgery techniques.

The goal of this work was to experimentally investigate the mechanical stability of arteries when subject to torsion and determine the critical buckling torque, critical buckling twist angle, and the buckling shape.

2. MATERIALS AND METHODS

We tested artery stability under torsion while subjecting them to a constant lumen pressure and axial stretch using a torsion machine designed in our laboratory. Through these tests, we determined both the critical loads and the material properties of the arterial wall under torsion.

2.1 Specimen preparation

Porcine common carotid arteries were harvested from domestic farm pigs (100–150 kg) at a local slaughterhouse postmortem with approval from the Texas Department of State Health Service and the University of Texas at San Antonio (UTSA) Institutional Biosafety Committee. The common carotid artery was chosen for our model due to its cylindrical shape, long segment availability, and because of the large rotation angles encountered within the neck *in vivo*. Immediately following dissection, the arteries were rinsed with an isotonic saline solution (0.9% w/v, Thermo Scientific), placed in test tubes containing ice cold PBS (Dulbecco's phosphate buffered saline, Sigma), and transported to our laboratory in an iced cooler within 1 hour of harvesting. Once in the laboratory, the test tubes containing the arteries were stored at 4° C and removed one at a time for testing. First, the artery was placed in a petri dish containing PBS at room temperature and excess connective tissue was removed. Next, female mounting luers (Cole-Parmer) were attached at both ends of the artery by inserting the hose barb end of the luer into the artery lumen and suturing the artery around the barb in order to hold it in place. Then, a syringe was connected to a luer at one end of the segment and a leak test was performed by briefly inflating the artery with air while submerged in PBS. Only those artery segments free of leaks and branches were

chosen for experimentation and all tests were completed within 72 hours of tissue harvesting.

The arteries were preconditioned in two steps. First, one end of the artery was attached to a syringe filled with PBS while the other end was plugged but able to move freely. The lumen pressure was then slowly increased to 200mmHg and deflated to zero for four cycles (Lee et al. 2012). After the first preconditioning step, a digital image of the artery at zero pressure was taken and the *in vitro* free length and outer diameter were measured with Image-Pro Plus software (Media Cybernetics, Inc.). The second pre-conditioning step, for twist preconditioning, was carried out on the torsion machine (process described in detail in section 2.3 Mechanical testing). While vessel dimensions were measured after pre-conditioning, our preliminary test showed that preconditioning had little effect on vessel dimensions and the zero-stress configurations, as the opening angle from arterial segments before and after preconditioning showed no statistical difference.

2.2 Torsion machine

The torsion machine was designed to mount arteries in the vertical position between a hollow aluminum cylinder fixed to the machine base at the bottom and a hollow square shaft which is adjustable from the top (Fig. 1). A rotating rod fits inside the adjustable hollow square shaft and connects to a cannula at the bottom for mounting the top end of the artery. This rod is driven by a SI programmable stepper motor (type 4023–820, Applied Motions) through a gear belt mechanism that applies a prescribed rotation angle to the artery segment. The bottom cannula for artery mounting is attached to the top of the hollow aluminum cylinder fixed to the base. The organ bath contains PBS at room temperature and is designed to immerse the artery. The axial stretch ratio is applied to the artery by adjusting the position of the top shaft. A miniature reaction torque transducer (Sensotec model: Sensotec error at 0.1% at full scale) and precision miniature load cell (Sensotec model: Sensotec error at 0.15–.25% at full scale) connected between the rotating rod and the top cannula measure the moment and axial force, respectively. The artery is filled with PBS and the static lumen pressure is controlled by adjusting the height of a PBS reservoir. PBS is pumped from the bottom of the base cylinder, through the artery lumen, and out a relief valve until all air has been removed. A digital pressure gauge (Ashcroft model: error at 0.5% at full scale) measures the lumen pressure. The clear lexan bath tank can slide up and down as well as rotate along the base cylinder to facilitate artery mounting and photographing. Labview 2009 (National Instruments) was used to record data for axial load and torsional resistance as a function of rotation angle.

2.3 Mechanical testing

Torsional tests were performed for porcine common carotid arteries using the torsion machine described above. Before testing, the top mounting cannula was positioned directly over the bottom mounting cannula and alignment screws within the machine housing were adjusted as necessary until the cannulas were aligned vertically and no gyration was visible upon rotation. Arteries (mounting luers sutured in place at both ends) were mounted vertically onto the torsion machine so that the distal end was always positioned towards the top (by mounting the luers to the cannula with tight-fittings). The arteries were tested at various combinations of lumen pressures (20, 40, 70, 100 mmHg) and axial stretch ratios (1.1, 1.3, 1.5, 1.7). These axial stretch ratios cover the sub-physiological to hyper-physiological range since the physiological level has previously been determined at 1.5 (Lee et al. 2012; Lee et al. 2008). At lower axial stretch ratios, artery segments became curved with increased lumen pressure (bend buckling (Han 2007)) and these load combinations were not included in the analysis. At each tested axial stretch ratio and lumen pressure combination, the artery was twisted gradually by rotating the top cannula (driven by the

step-motor) slightly past the point at which buckling was visually evident. The torque, rotation angle, and axial force were recorded during the process. The critical buckling torque and twist angle were obtained from the torque vs. rotation plots. Here, the critical buckling twist angle is defined as degree of rotation divided by the stretched artery length. Twist buckling tests were repeated three times for each combination of axial stretch ratio and lumen pressure and the results were averaged.

In total, torsional buckling tests were performed for nine porcine arteries. However, the first three arteries were tested in ascending order starting with the combination of lowest axial stretch ratio and lumen pressure, 1.1 and 20 mmHg, respectively. At this loading condition, arteries were pre-conditioned to twist by applying rotation to their buckling point a total of three times. At a maximum stretch ratio of 1.7, the axial load at the onset of rotation dropped with each repeated torsion test indicating that arteries required further pre-conditioning for the higher axial stretch ratio. Therefore, the remaining six arteries were first stretched to the maximum axial stretch ratio of 1.7 and lumen pressure of 100 mmHg, and left to sit for approximately twenty minutes. The arteries were then rotated to buckling until reproducible results in the axial load were obtained. Accordingly, only the torsional buckling results from the last six arteries were included in the analysis.

Upon completing the torsional buckling tests, the artery was removed and short ring segments were cut from the proximal, middle, and distal portions of the artery and photographed to measure its inner diameter (Image-Pro Plus software, Media Cybernetics, Inc.). Next, a radial cut was made for each ring segment to allow it to pop open into a sector and set to relax in PBS for approximately twenty minutes to fully release its circumferential residual stress. The opening angle (θ_o) of the segment was measured as the angle subtended by two radii connecting the midpoint of the inner wall of the open sector (Fig. 2A). For each artery, the inner diameter and opening angle were averaged from the three ring segments.

2.4 Biomechanical Principles

The porcine common carotid arteries are assumed to be straight cylinders with a circular cross section. We consider an artery with an opening angle θ_o to account for circumferential residual stress. The dimensions of the sector at zero-stress configuration are defined by initial inner radius (R_i), outer radius (R_o), and the opening angle (θ_o) (Fig. 2A) while the initial segment length is defined by L . A cylindrical coordinate (r, θ, z) was used to describe the position and deformation in the artery. The dimensions of the artery at the loaded configuration are defined by the inner radius (r_i), the outer radius (r_o), and the stretched length (λ). The artery is acted upon by a longitudinal stretch (λ), axial force (N), lumen pressure (P_i), and torque (T), which twists the artery by an angle (γ) (Fig. 2B). Using the cylindrical coordinates, the deformation gradient matrix (F) for arteries under torsion is

$$F = \begin{bmatrix} \frac{\partial r}{\partial R} & 0 & 0 \\ 0 & \frac{\pi r}{\Theta_o R} & r\gamma \\ 0 & 0 & \lambda \end{bmatrix} \quad (1)$$

where γ is the twist per unit unloaded length and Θ_o is equal to $2\pi - \theta_o$ (Humphrey 2002). Enforcing incompressibility, that is $\det F = 1$, requires that

$$\frac{\partial r}{\partial R} = \frac{\Theta_o R}{\pi r \lambda} \quad (2)$$

The deformation at any given point in the artery can be characterized by Lagrangian Green's strain tensor (E), given by

$$E = \frac{1}{2}(F^T \cdot F - I) = \frac{1}{2} \begin{pmatrix} \left(\frac{\Theta_o R}{\pi r \lambda}\right)^2 - 1 & 0 & 0 \\ 0 & \left(\frac{\pi r}{\Theta_o R}\right)^2 - 1 & \frac{r^2 \pi \gamma}{\Theta_o R} \\ 0 & \frac{r^2 \pi \gamma}{\Theta_o R} & r^2 \gamma^2 + \lambda^2 - 1 \end{pmatrix} \quad (3)$$

where I is the identity matrix (Humphrey 2002). The diagonal elements in the matrix are the radial, circumferential, and axial components of Green's strain tensor, E_r , E_θ , and E_z , and the only non-zero non-diagonal component are $E_{rz} = E_{zr}$. Accordingly, the generalized form of the exponential Fung strain energy function, with the incompressibility assumption, becomes (Chuong and Fung 1983; Humphrey 2002)

$$w = \frac{1}{2} C (e^Q - 1) + \frac{H}{2} [(1 + 2E_r)(1 + 2E_\theta)(1 + 2E_z) - (1 + 2E_r)4E_{\theta z}^2 - 1] \quad (4)$$

Where

$$Q = b_1 E_\theta^2 + b_2 E_z^2 + b_3 E_r^2 + 2b_4 E_\theta E_z + 2b_5 E_z E_r + 2b_6 E_r E_\theta + 2b_7 E_{\theta z}^2 \quad (5)$$

and w represents the strain energy density, C is a material constant which has the same unit as the stress, H is a Lagrangian multiplier, b_1 to b_7 are dimensionless material constants. Therefore, the Cauchy stresses are given by (Humphrey 2002)

$$\sigma_r = (1 + 2E_r)(b_3 E_r + b_5 E_z + b_6 E_\theta) C e^Q + H$$

$$\sigma_\theta = [(1 + 2E_\theta)(b_1 E_\theta + b_4 E_z + b_6 E_r) + 8b_7 E_{\theta z}^2 + (r\gamma)^2(b_2 E_z + b_4 E_\theta + b_5 E_r)] C e^Q + H$$

$$\sigma_z = \lambda^2 (b_2 E_z + b_4 E_\theta + b_5 E_r) C e^Q + H$$

$$\sigma_{\theta z} = \left[\frac{\pi r \lambda}{\Theta_o R} (2b_7 E_{\theta z}) + r\gamma \lambda (b_2 E_z + b_4 E_\theta + b_5 E_r) \right] C e^Q$$

The equation of equilibrium in a straight axisymmetric artery is

$$\frac{\partial \sigma_r}{\partial r} + \frac{\sigma_r - \sigma_\theta}{r} = 0 \quad (10)$$

with boundary conditions

$$\sigma_r(r_i) = -Pi; \sigma_r(r_e) = 0 \quad (11)$$

Integration along with boundary conditions yields

$$Pi = \int_{r_i}^{r_e} ((1+2E_\theta)(b_1E_\theta + b_4E_z + b_6E_r) + 8b_7E_{\theta z}^2 + (r\gamma)^2(b_2E_z + b_4E_\theta + b_5E_r) - (1+2E_r)(b_3E_r + b_5E_z + b_6E_\theta)) ce^Q \frac{\partial r}{r}$$

The axial force (N) and twisting moment (i.e. torque T) can be determined by (Humphrey 2002)

$$N = \pi \int_{r_i}^{r_e} [2\lambda^2(b_zE_z + b_4E_\theta + b_5E_r) - (1+2E_r)(b_3E_r + b_5E_z + b_6E_\theta) - (1+2E_\theta)(b_1E_\theta + b_4E_z + b_6E_r) - 8b_7E_{\theta z}^2 - (r\gamma)^2(b_2E_z + b_4E_\theta + b_5E_r)] Ce^Q r dr$$

$$T = 2\pi \int_{r_i}^{r_e} [(\frac{\pi r \lambda}{\Theta_o R} 2b_7E_{\theta z} + r\gamma \lambda (b_2E_z + b_4E_\theta + b_5E_r))] Ce^Q r^2 dr$$

2.5 Determination of material constants

Material constants were determined by minimizing the error associated with the sum of the squares of the differences between theoretically calculated and experimentally measured quantities while normalizing for residual errors. The minimization equation is

$$error = \sum_{k=1}^m \left[\left(\frac{Pi_{th} - Pi_{exp}}{Pi_{exp}} \right)_k^2 + \left(\frac{N_{th} - N_{exp}}{N_{exp}} \right)_k^2 + \left(\frac{T_{th} - T_{exp}}{T_{exp}} \right)_k^2 \right] \quad (15)$$

where m is the number of quasi-static equilibrium configurations measured, the subscripts th and exp represent the model predicted and experimental values, respectively, for lumen pressure (Pi), axial force (N) and torque (T). The material constants which optimally minimized the error function were determined via a hybrid formulation consisting of a genetic algorithm and a constrained nonlinear optimization function (fmincon), implemented in the optimization toolbox available in MATLAB (Mathworks). The genetic algorithm minimization technique has previously been utilized by Pandit et al. and Wang et al. to determine material constants for porcine coronary arteries based on a Fung stain energy function with good fitting (Pandit et al. 2005; Wang et al. 2006). In order to ensure

convexity, a constraint was given so that the eigenvalues of the matrix $\begin{pmatrix} b_1 & b_4 & b_6 & 0 \\ b_4 & b_2 & b_5 & 0 \\ b_6 & b_5 & b_3 & 0 \\ 0 & 0 & 0 & b_7 \end{pmatrix}$ were always positive (Holzapfel et al. 2000).

Experimental data for lumen pressure, axial force, and moment were obtained from buckling tests between axial stretch ratios of 1.3 and 1.5 (*in vivo* level), lumen pressures between 20 and 100 mmHg, and moment values for the first 270 degrees of rotation. The first 270 degrees of rotation were chosen because it was the maximum range of rotation that all vessels had in common without buckling. Moment values were first smoothed by fitting a linear line to the torque vs. rotation angle plot starting at the onset of rotation to the point of

buckling (maximum point on the curve). Data points on the lines at 30 degree angle intervals were then used in fitting the moment. For simplification, the axial force (N) was assumed to remain constant during rotation. Material constants were determined for those arteries for which buckling tests were started at the maximum tested axial stretch ratio of 1.7 and at least 5 different lumen pressure conditions were obtained between axial stretch ratios of 1.3 and 1.5.

2.6 Statistical analysis

Experimental data values for the critical buckling torque and the critical buckling twist angle at all combinations of lumen pressure and axial stretch were analyzed. An unbalanced two-way ANOVA with type III sum of squares was used to test for significance between the independent variables (axial stretch ratio and lumen pressure) and the dependent variables (critical buckling torque and critical buckling twist angle). After determining significance, a Tukey's HSD post hoc analysis was used to determine whether an incremental increase in axial stretch ratio or lumen pressure were significantly different from one another. All statistical analyses were carried out using SPSS statistical software (IBM). A p value less than 0.05 was considered to be statistically significant.

For the material constants, the "goodness of fit" between the predicted and experimental values for lumen pressure, axial force, and moment were determined using the coefficient of determination (R^2) and the percentage of root square mean error with respect to the mean, also known as the coefficient of variation of the root mean square error (CV-RMSE).

3. RESULTS

Results obtained for a group of six porcine common carotid arteries are presented below. The average length of vessels tested was 47.9 ± 11.2 mm with an average opening angle of 92 ± 27 degrees. The mechanical properties, buckling shape, and critical loads were determined for these arteries.

3.1 Torsional behavior of arteries

All arteries buckled when twisted under torsion at given axial stretch ratios and lumen pressures, characterized by the onset of a kink. In most cases, the arteries buckled suddenly into a twisted kink configuration at approximately the middle portion of the artery segment (Fig. 3, a video clip of the twist behavior can be seen in the supplement). However, there were instances, especially at higher stretch ratios and lumen pressures, where the kink occurred more gradually and shifted closer to one end of the artery. In all cases, the torque increased linearly with rotation angle until a sudden drop when buckling or kinking of the artery occurred (Fig. 4). Careful examination of the video recordings of the experiments verified that the sharp drop in torque occurred simultaneously with the kinking of the artery. The critical torque and twist angle represents the maximum torque and twist angle that were reached before buckling occurred for a given combination of axial stretch ratio and lumen pressure.

3.2 Effect of axial stretch and lumen pressure

Our results showed that an increase in axial stretch ratio very significantly affected the critical buckling torque ($p < 0.002$) and the critical buckling twist angle ($p < 0.001$). Comparisons of critical buckling torque and critical buckling twist angle as functions of axial stretch ratios at different given lumen pressures are illustrated in Figure 5. The critical buckling torque at an axial stretch ratio of 1.7 was significantly higher than the values at all other tested axial stretch ratios (1.5, 1.3, and 1.1) ($p < 0.002$). In addition, the critical twist

angles at all tested axial stretch ratios were significantly different from each other ($p < 0.001$).

Changes in lumen pressure had a significant effect on the critical buckling torque ($p < 0.0001$) but not on the critical buckling twist angle ($p = 0.067$). The critical buckling torque increased with increasing lumen pressure, both for individual arteries and the mean of the group (Fig. 6 Top). The critical buckling torque values at lumen pressures of 100 mmHg and 70 mmHg were significantly higher than at all other tested pressure values and were also significantly different from each other ($p < 0.02$). There was no difference however, in critical buckling torque between the two lowest lumen pressures of 20 mmHg and 40 mmHg ($p = 0.222$). The critical buckling rotation angle and twist angle at given axial stretch ratios, however, demonstrated biphasic change with increasing lumen pressure (Fig. 6 bottom). The twist angle increased when pressure increased from 20 to 40 mmHg and then decreased when pressure increased to 70 and to 100 mmHg. However, the effect was statistically insignificant. A possible explanation is given in the Discussion.

3.3 Effect of twist on axial tension

While twisting arteries to a small rotation angle ($< \sim 120^\circ$) had nearly no effect on the axial force, larger rotation angles increased the axial force nonlinearly (Figure 7). The actual transition point depended on the axial stretch ratio and the lumen pressure. In general, for a given lumen pressure the axial force was higher at larger axial stretch ratios. For axial stretch ratios of 1.5 and lower, the axial force was lower at a higher lumen pressure. The axial force was higher with increased lumen pressure at an axial stretch ratio of 1.7. Furthermore, the changes of axial force due to twisting were more pronounced at a higher lumen pressure and higher axial stretch ratio (see Fig. 7). These results suggest that the axial force was coupled non-linearly with the twist at the large rotation angle range ($> \sim 120^\circ$).

3.4 Mechanical properties

By fitting the pressure-diameter, pressure-axial force, and torque-rotation angle data with the theoretical equations, we obtained the material constants of the generalized Fung strain energy function for five porcine common carotid arteries (Table 1). These constants are convex (Figure 8A) and include a constant (b_7) which describes the torsional shear deformation. These material constants were determined for axial stretch ratios in the range of 1.3 and 1.5, lumen pressures between 20 – 100 mmHg, and twist within 270 degrees of rotation angle. The coefficient of determination (R^2) and the percentage of root-mean-square error with respect to the mean between the predicted and experimental values for lumen pressure, axial force, and moment are also listed in Table 1. For one artery loaded to an axial stretch ratio of 1.5 and a lumen pressure of 100 mmHg (*in vivo* values), a good correlation between the experimental and predicted twist behavior is seen (Fig. 8B), indicating that the material constants well represent the behavior of the artery.

4. DISCUSSION

In summary, we investigated the torsional behavior of porcine common carotid arteries and demonstrated that the torque load increased linearly with twist angle until arteries suddenly lost stability and snapped to form a kink. The pre-buckling torsional deformation was analyzed using a Fung strain energy function and convex material constants, including a shear constant, were determined. The critical buckling torsion loads and twist angles at various combinations of axial stretch ratios and lumen pressures conditions were obtained. Our results demonstrated that the critical buckling torque increased with increasing axial stretch and lumen pressure. The critical twist angle at buckling decreased with axial stretch but lumen pressure had very little effect on the critical twist angle.

4.1 Twist and rotation angle

Twist buckling of blood vessels was previously described by Selvaggi et al. to explain the twisted geometry of microvessels subjected to excessive torsion during microanastomosis (Selvaggi et al. 2004). It was observed that vessels developed a single wave distributed along their longitudinal axis or a concentrated twist at a focal point (kink) upon buckling. Studies for rat femoral arteries have however, demonstrated some discrepancy in the critical rotation angle at which patency is significantly reduced. For example, Salgarello et al. reported that as little as 90 degrees of rotation significantly reduced patency while Izquierdo et al. found the critical limit to be 270 degrees (Izquierdo et al. 1998; Salgarello et al. 2001). Topalan et al. found no significant reduction in patency even at 360 degrees of rotation (Topalan et al. 2003).

The large variability in results may be attributed to the differences in the species of rats used (Salgarello et al. used Wistar rats while Izquierdo et al. and Topalan et al. used Sprague-Dawley rats) and the differences in the dimensions (i.e. length, diameter, thickness) of arteries tested. It was shown by Dobrin et al. that the length, diameter, thickness, and material of vein grafts have a significant effect on the amount of rotation required to generate a kink and concluded that grafts having shorter lengths, larger diameters, thinner walls, and stiffer materials were more susceptible to kinking at lower rotation angles (Dobrin et al. 2001). Therefore, we used twist angle (rotation angle per unit stretched length) in describing the twist deformation.

An interesting new observation of this study is the kinking of arteries under torsion. This is similar to the kinking of veins and vein grafts reported by Dobrin and colleagues (Dobrin et al. 2001; Endean et al. 1989). While the mechanism needs further investigation, we believe the kinking is a buckling pattern for tubular vessels and might be triggered by local wall imperfection. Similar kinking pattern has been demonstrated previously in very thin membrane tubes (Hunt and Ario 2005). However, since the arterial wall is not a “thin-membrane”, the kink location did not show the sharp folding as seen in the kink area of the thin membrane tubes.

4.2 Effect of axial stretch ratio and lumen pressure

Another new finding from the current study is that axial stretch also has a significant effect on the amount of critical torque and twist angle required to generate buckling (kinking). Increased axial stretch ratio increases critical buckling torque but reduces the critical twist angle (and the total rotation angle). A possible explanation is that increased axial stretch causes the vessel wall to stiffen as a result of material nonlinearity. However, the level of axial stretch was not specified in previous studies. It was noted in the current study that a significant reduction in critical twist angle was only found at the highest level of axial stretch ratio (1.7), indicating a stronger coupling between twist and axial tension at high axial stretch ratios. In fact, the axial force required to maintain the artery to constant length remained small and stable at low axial stretch ratios and at a rotation angle less than 120° and then increased nonlinearly when the rotation angle was larger (see Fig.7), showing a stronger coupling effect at increased lumen pressures and axial stretch ratios. This may be due to the helical alignment of the collagen fibers in the arterial wall and the observation that collagen engages in load bearing at higher stress levels (Holzapfel et al. 2000; Humphrey 2002).

On the other hand, increasing lumen pressure significantly increased the critical buckling torque. This increasing trend can also be due to material nonlinearity. Increasing lumen pressure increases wall stress and the wall stiffness, thus increasing the critical torque needed for buckling to occur. However, the lumen pressure had little effect on the critical

twist angle overall (see Fig 5B) and can be explained as following: Twist buckling is the result of local buckling of wall elements under shear stress/strain –which is linked to the maximum compressive stress/strain in an oblique direction. Therefore, the critical load is decided by this local shear stress/strain. According to the deformation analysis, the circumferential-axial shear strain in the wall depends on the twist angle (rotation angle over axial length) and vessel radius (Gere 2004).

$$\eta = R \frac{d\varphi}{dz} \quad (16)$$

It is seen that an increase in diameter under lumen pressure would reduce the twist angle needed to reach the shear strain for buckling to occur. In addition, the reduction of wall thickness associated with diameter increase further reduces the shear strain needed to generate buckling in the wall. Thus, at the normal pressure range (70 to 100 mmHg), the diameter increase reduced the critical twist angle. However, the diameter changes were relatively small in the pressure range and thus the overall effect was small. At higher axial stretch ratios, the changes in diameter were even smaller, further reducing the effect of lumen pressure. At the lowest pressure of 20 mmHg, arteries were barely fully inflated to circular shapes and thus the critical twist angle was lower than at 40 mmHg. These observations are consistent with a previous report by Dobrin and colleagues showing no significant difference in kinking behavior of vein grafts with the addition of flow or varied lumen pressure (in the range of 50 – 150 mmHg) (Dobrin et al. 2001).

A buckling equation was given by Flugge to estimate the critical torque for thin-walled, isotropic, linear elastic long cylinders under an end torque only (Flugge 1973).

$$T_{cr} = \frac{\pi \sqrt{2} \mu}{3(1-\nu^2)^{\frac{3}{4}}} \sqrt{Rt^5} \quad (17)$$

where μ is the modulus of elasticity, R is the mid-wall radius, t is the wall thickness, and ν is poisson's ratio. This equation does not apply to arteries since the arterial walls are generally considered thick-walled cylinders of orthotropic, nonlinear elastic materials, and are under both lumen pressure and axial stretch. For the arteries tested, we calculated the ratio of $T_{cr} / \sqrt{Rt^5}$ using the media radius and wall thickness under load as well as the critical torques from our experimental data. Our results demonstrated that the ratio correlated well with the shear modulus G determined from:

$$\frac{\varphi_{cr}}{L} = \frac{T_{cr}}{GI_p} \quad (18)$$

where I_p is the polar moment of inertia and φ_{cr} is the critical rotation angle. Accordingly, we obtained that

$$T_{cr} = \alpha G \sqrt{Rt^5} \quad (19)$$

with a coefficient $\alpha = 17.61$ (Fig. 9, $R^2 = 0.85$). This empirical equation estimated the critical torque at a reasonable accuracy for individual arteries (see Fig. 9 bottom). This relationship provides insight for future work to establish the theoretical torsional buckling equation.

4.3 Edge effect and vessel imperfections

While most arteries buckled into a kink at the middle of the segment, some arteries buckled less dramatically and appeared to roll over into the kinked configuration or buckled closer to the edge. The drop in torque was still noticeable in these cases. The more gradual kinking typically occurred at a combination of high axial stretch and low lumen pressure while edge effects were more noticeable at increased levels of both axial stretch and lumen pressure. Even though only artery segments free of branches were included in the experiments, variables such as edge effects, arterial wall imperfections, and tapering can influence failure location and the critical torsion limits (Dahir et al. 2011). Despite this, clear trends were seen when examining the effect of axial stretch ratio and lumen pressure on the critical torsion limits.

4.4 Repeatability of twist buckling

Since the arterial walls are viscoelastic, we tested the arteries after preconditioning and under very slow loading steps to approximate quasi static conditions under large rotation angles. We also tested the artery three times at each loading combination and the results showed good repeatability and were averaged. To further determine repeatability of the twisting behavior after different loading sequence, we compared the torsion curve obtained from two different sequences of twist loading for one artery. First, a sequence of buckling tests was performed starting from the combination of highest axial stretch ratio (1.7) and lumen pressure (100 mmHg) until the final combination of lowest axial stretch ratio (1.1) and lumen pressure (20 mmHg). Immediately following, the second sequence of buckling tests were carried out in the reverse order, ending with the combination of highest axial stretch ratio (1.7) and lumen pressure (100 mmHg). The buckling (kink) occurred at the same location in the artery for repeating tests. Comparing the results from the two sequences indicated that the twist buckling data was reversible and reproducible. The difference between the first and last testing results (after over 50 cycles of twisting) was less than 7%. Additionally, histology examination of cross sections from the top, middle, and bottom portion of the artery did not identify any noticeable damage to the arterial wall structure.

4.5 Meaning of the material constants

We determined convex material constants, including the axial-circumferential shear strain constant (b_7), of a generalized Fung strain energy function for porcine common carotid arteries. The convexity constraint was applied to ensure that the function exhibits mathematically stable behavior which is important for use in finite element analysis (Holzapfel et al. 2000). Previously, Van Epps and Vorp (Van Epps and Vorp 2008) determined the constant (b_7) for the porcine coronary artery based on the experimental results from Lu et al. and Wang et al. and found good correlation between experimental and predicted results (Lu et al. 2003; Wang et al. 2006). They used a linear assumption between shear modulus and hoop stress to determine the shear parameter and used numerical modeling to demonstrate the validity of the constant at axial stretch ratio, lumen pressure, and rotation angle values in the range of 1.2 – 1.4, 0 to 120 mmHg, and –25 to +25 degrees of rotation respectively. In the current study, a minimization algorithm between experimental and predicted values of lumen pressure, axial force, and moment were used to determine the material constants. We found this algorithm worked better for porcine common carotid arteries in the much larger range of rotation. The material constants presented for these arteries are valid for an axial stretch ratio of 1.3 to 1.5, lumen pressure of 20 to 100 mmHg, and rotation angle of 0 to 270 degrees. The large number of material parameters and model nonlinearity led to large variations in individual constants from artery to artery because they are sensitive to small changes in data and thus are not unique (Fung 1993). However, these material constants as a set do fit the experimental data well for each individual artery and they are valid for describing the anisotropic behavior of arteries under

axial stretch, inflation, and torsion. The significance of determining the material constants is two-fold. First, these material constants will be useful for general stress analysis involving twist deformation. Second, once a theoretical model for artery buckling under tension, inflation, and torsion is developed, we can then use these experimental results to validate the model.

4.6 Limitations and future work

In the current study, arteries were placed in a PBS bath and tested under static lumen pressure for its simplicity since our main goal was to gain a basic understanding of artery buckling under torsion. An advantage of this setup is that it provides a well controlled environment in which the mechanical properties of the artery tissue and buckling loads can be obtained simultaneously. The material properties are for passive arterial tissue. The buckling behavior of arteries under pulsatile pressure and dynamic twisting requires further investigation (Liu and Han 2012). The axial force was assumed to remain constant in the minimization algorithm used to determine the material constants. This simplification is reasonable since the average increase in axial force was small (0.08 N) within the tested range.

While the current study demonstrated the twist buckling behavior of arteries and the effect of lumen pressure and axial stretch, a theoretical model that incorporates material nonlinearity and the large deformation feature of the artery wall is needed to describe the buckling behavior and predict the critical loads under torsion. The data presented here will serve as guidance for developing such a theoretical model to accurately predict twist buckling of blood vessels.

Additionally, we have observed that blood vessels that were curved due to bent buckling (under hypertensive pressure or reduced axial tension) (Han 2009) twisted into a helical configuration prior to kinking. The twisting of curved vessels needs further study.

4.7 Clinical relevance

Arteries are subjected to lumen blood pressure and axial tension from surrounding tissue tethering as well as torsion loads due to body movement. It is essential that arteries remain mechanically stable and patent under these loads in order to maintain their normal function. However, arteries are often observed to be tortuous and kinked in association with clinical complications and vascular diseases such as diabetes, high blood pressure, transitory ischemic attacks, and atherosclerosis (Del Corso et al. 1998; Han 2012; Milic et al. 2007; Pancera et al. 2000; Smedby and Bergstrand 1996). Tortuosity, twisting and kinking has been reported in vertebral, carotid, iliac, and femoral arteries which experience torsion due to head rotation or hip and leg movements (Cheng et al. 2006; Choi et al. 2009b; Grego et al. 2003; Klein et al. 2009; Laird 2006; Norris et al. 2000). It has been previously demonstrated that blood vessels will lose stability and mechanically buckle into tortuous geometries when subject to hypertensive lumen pressure or a reduction in longitudinal tension (Han 2009; Jackson et al. 2005; Martinez et al. 2010). Current results of twist buckling compliment previous studies and enrich our understanding of artery instabilities.

The critical buckling twist angles obtained in the current study are beyond the degree of twist that is normally encountered *in vivo*, indicating that normally arteries are safe *in vivo* against twist buckling. However, pathological changes due to cardiovascular diseases may compromise the structural integrity of an artery segment and reduce the critical twist angle needed for buckling to occur. Our additional experiments demonstrated that hypertensive pressure and the inclusion of a surrounding tissue support reduced the critical buckling twist angle (while increasing the critical torque). Furthermore, elastin degradation by elastase

treatment weakens arterial wall and reduces the critical buckling torque and twist angle, similar to its effect on bent buckling (Lee et al. 2012). On the other hand, artery twist and kinking have also been reported in clinical studies in association with pathological conditions such as hypertension, atherosclerosis, aging, or degenerative diseases due to genetic defects related to elastin deficiencies (Barton and Margolis 1975; Del Corso et al. 1998; Franceschini et al. 2000; Han 2012). Obstruction of arterial blood flow resulting from twisting or kinking has also been observed in the carotid and vertebral arteries (Barton and Margolis 1975; Han 2012; Pancera et al. 2000; Vos et al. 2003). Our current results provide insight into the twisting behavior of arteries and set a basis for further studies regarding the twisting behavior of arteries under pathological conditions.

Additionally, it has been shown that stented arteries suffer a reduced capacity for rotation since the stented portion of the artery remains stiff and inflexible under torsion loading and the un-stented portion is left to accommodate the full twist load (Vos et al. 2003). The large rotation angles encountered in some surgery techniques also increase the risk of for artery twist kinking (Jakubietz et al. 2007; Wong et al. 2007). Arteries may also become twisted after cross-limb grafting (Ramaiah et al. 2002). Therefore, understanding of arterial twisting behavior is important for the design and improvement of these treatments.

4.8 Conclusions and significance

In conclusion, arteries buckle under torsion and form a kink. The critical torque at buckling increased with increasing axial stretch and lumen pressure. The critical buckling twist angle decreased with increasing axial stretch but only slightly decreased with increasing lumen pressure. These results shed light into the structural stability and anisotropic behavior of arteries under torsion. Study of twist buckling is complementary to previous studies of artery collapse under low lumen pressure and bending buckling under high lumen pressure. Understanding of twist buckling behavior enriches our knowledge of vascular biomechanics, provides insight into pathology of twisted blood vessels and vein grafts, and will also be useful in planning new surgical treatment techniques.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Fig. 1.
Schematic of the torsion machine.

A

B

Fig. 2.

(A) Schematic of an arterial ring in zero-stress configuration with opening angle Θ_0 or arc angle Φ_0 , and inner and outer radii R_i and R_e respectively. (B) Schematic of an arterial segment acted upon by axial tension N , lumen pressure P_i , and torque T , which produces a rotation angle ϕ .

Fig. 3.

A porcine common carotid artery kinked from torsional buckling. The artery was initially loaded to 1.5 axial stretch ratio and 70 mmHg lumen pressure. The artery buckled after 1.7 rotations.

Fig. 4. Relationship between torque and rotation angle of a porcine carotid artery. The torque increases linearly with rotation angle until a critical buckling point results in a sharp drop in torque. The maximum torque and rotation angle are recorded at this point of instability. (A) Torque plotted against rotation angle for a carotid artery under four lumen pressures (20, 40, 70, and 100mmHg) at a constant axial stretch ratio of 1.5. (B) Torque plotted against rotation angle for three axial stretch ratios (1.3, 1.5, and 1.7) at a constant lumen pressure of 70 mmHg.

Fig. 5.

(A) The critical buckling torque (mean $\pm$ SD; $n = 6$) and (B) critical buckling twist angle (normalized to stretched vessel length) (mean $\pm$ SD; $n = 6$) plotted against axial stretch ratio for lumen pressures of 20, 40, 70, and 100 mmHg. The axial stretch ratio significantly affected both the critical buckling torque ($p < 0.002$) and critical buckling twist angle ($p < 0.001$).

Fig. 6. (A) The critical buckling torque (mean $\pm$ SD; $n = 6$) and (B) critical buckling twist angle (normalized to stretched vessel length) (mean $\pm$ SD; $n = 6$) plotted against lumen pressure for axial stretch ratios of 1.1, 1.3, 1.5, and 1.7. Lumen pressure significantly affected the critical buckling torque ($p < 0.001$) but had no significant effect on the critical buckling twist angle ($p = 0.067$).

Fig. 7.

Variations of axial force with rotation angle in a porcine carotid artery. (A) Axial force plotted against rotation angle for four lumen pressures (20, 40, 70, and 100mmHg) at a constant stretch ratio of 1.5. (B) Axial force plotted against rotation angle for three stretch ratios (1.3, 1.5, and 1.7) at a constant lumen pressure of 70 mmHg. While the axial force remains constant at lower rotation angle, an increase is seen at higher rotation angle and the non-linearity is more pronounced at higher levels of lumen pressure and stretch ratio.

Fig. 8. (A) Contour plot of the strain energy function for one porcine common carotid artery (artery 2). (B) A comparison between experimental torque (diamonds) and predicted torque (line) during the first 270 degrees of rotation for one artery (artery 2). The values shown are based upon the torque obtained during rotation at a constant axial stretch ratio of 1.5 and a lumen pressure of 100 mmHg.

Fig. 9.

Top: $T_{cr} / \sqrt{Rt^5}$ plotted as function of shear modulus for all six arteries under all tested lumen pressure and axial stretch ratio combinations. A good linear relation is seen for all six arteries (a fitting line of a slope of 17.61 with $R^2 = 0.85$ and $p < 0.001$). Bottom: Comparison of critical torque estimated by empirical equation (19) and experimental data at various lumen pressure (20, 40, 70, 100mmHg) and axial stretch ratio (1.1, 1.3, 1.5, and 1.7) for one artery.

Convex material constants of a generalized Fung strain energy function obtained from fitting the experiment data of porcine common carotid arteries under extension, inflation, and torsion.

Table 1

Artery	Material Constants							$\underline{R^2}$			%CV-RMSE			
	b_1	b_2	b_3	b_4	b_5	b_6	b_7	C (kPa)	P_i	N_i	T_i	P_i	N_i	T_i
1	3.766	0.267	5.000	0.001	0.001	0.950	0.841	9.955	0.76	0.85	0.92	24.45	33.30	34.18
2	1.502	0.810	5.000	0.760	1.750	1.744	0.258	152.236	0.96	0.92	0.98	11.54	19.07	14.22
3	5.000	0.436	4.950	0.576	0.001	2.758	0.865	28.029	0.92	0.95	0.91	16.16	9.29	49.36
4	4.085	0.071	5.000	0.437	0.001	2.634	0.550	19.276	0.87	0.70	0.90	22.64	46.83	42.65
5	0.910	0.323	1.506	0.130	0.131	0.648	0.167	83.362	0.96	0.92	0.93	12.43	9.00	45.16

R^2 and % CV-RMSE were computed for lumen pressure, axial force, and moment.