Published in final edited form as:

Ann Biomed Eng. 2008 December ; 36(12): 2042–2050. doi:10.1007/s10439-008-9582-0.

Mechanical Modeling of Stents Deployed in Tapered Arteries

Lucas H. Timmins, Clark A. Meyer, Michael R. Moreno, and James E. Moore Jr.

Department of Biomedical Engineering Texas A&M University 337 Zachry Engineering Center
College Station, TX 77843-3120

Abstract

The biomechanical interaction of stents and the arteries into which they are deployed is a potentially important consideration for long-term success. Adverse arterial reactions to excessive stress and the resulting damage have been linked to the development of restenosis. Complex geometric features often encountered in these procedures can confound treatment. In some cases, it is desirable to deploy a stent across a region in which the diameter decreases significantly over the length of the stent. This study aimed to assess the final arterial diameter and circumferential stress in tapered arteries into which two different stents were deployed (one stiff and one less stiff). The artery wall was assumed to be made of a strain stiffening material subjected to large deformations, with a 10% decrease in diameter over the length of the stent. A commercially available finite element code was employed to solve the contact problem between the two elastic bodies. The stiffer stent dominated over arterial taper, resulting in a nearly constant final diameter along the length of the stent, and very high stresses, particularly at the distal end. The less stiff stent followed more closely the tapered contour of the artery, resulting in lower artery wall stresses. More compliant stents should be considered for tapered artery applications, perhaps even to the exclusion of tapered stents.

Keywords

Stress; Restenosis; Vascular Interventions; Finite Element Analysis

Introduction

Cardiovascular disease is the leading cause of death in developed countries. In the U.S. it has been estimated that costs associated with this disease will be in excess of \$430 billion in 2007.¹ Atherosclerosis, one of the more common and serious forms of this disease, is characterized by focal accumulation of material within an artery that restricts blood flow. Spatial variations in mechanical factors such as blood flow patterns and artery wall stresses have been implicated in atherogenesis. Treatment options (e.g. stenting, balloon angioplasty, etc.) for arteries occluded by atheromatous lesions typically alter the local mechanical environment. Given the influence of mechanical factors in disease development, this is a consequence that must be considered when attempting to determine the efficacy of a given therapeutic option.

Vascular stenting has proven to be an effective treatment option for most patients with atherosclerosis; though complications such as restenosis, or the development of neointimal hyperplasia within the stented region following implantation, can result in the ultimate failure of the procedure. Stents are essentially tubular scaffolds comprised of a metallic mesh deployed inside the artery in order to prop the diseased region open and restore blood flow. To function

properly and remain in place, stents are designed to be larger in diameter than the healthy artery — this characteristic of the therapy is referred to as “stent over-sizing”. As the artery is distended beyond normal physiologic limits, pathologic stresses are induced in the artery wall and the artery may be irreversibly damaged. Even in cases wherein acute tissue damage is minimal, cellular response to the new stress environment created by the stent can be problematic.

The mechanical mismatch is a particular concern for situations in which arterial diameter changes to a significant degree over the length of a stent. Such diameter changes can be encountered due to tapering or the need to deploy a stent from a parent vessel into a branch. This significant diameter change is sometimes the case in carotid artery stenting, but can also occur in femoral arteries. The clinician must then decide which diameter on which to base the stent sizing decision. A large diameter mismatch at the distal end of the stent could lead to excessive damage in the artery wall and increased risk for clinical failure. While there are tapered stents available in some markets, the geometric variations require more careful consideration of stent choice.²

Modeling stent/artery interactions has mainly focused on applications in which arterial tapering is not a factor, either through simplifying assumptions or by modeling specific, non-tapered arteries. Rogers et al.³ used two-dimensional linear elastic models to investigate balloon expansion with stent and artery contact. It was determined that high inflation pressures, wide stent-strut spacings, and more compliant balloon materials cause markedly larger surface-contact areas and contact stresses between stent struts. Consequently, it was concluded that stent design and deployment protocols play an important role in stenting outcomes. The stress fields induced by stents are as unique as the stents themselves. Consistent with these concepts, evidence suggests restenosis rates vary by stent design.⁴ It is hypothesized that stent-induced stresses mediate processes conducive to the development of restenosis.⁵ Therefore characterization of the stress field induced by a given stent design or “design parameter” is of particular interest to efforts to reduce restenosis rates in stent therapy.

Finite element modeling is commonly employed to investigate stresses and strains in stents and stented arteries. The quality of information gained is typically limited by challenges associated with establishing an appropriate constitutive models and the need to impose simplifications. Holzapfel et al.⁶ modeled balloon expansion of a full 3-dimensional anisotropic diseased artery. It was proposed that the work provided tools with the potential to improve procedural protocols and the design of interventional instruments. Moreover, the techniques employed could be used as an aid in predicting post-angioplasty mechanical environments, which could subsequently be correlated with restenosis responses. Later, in works that presently constitute the most ambitious efforts to model diseased arteries, anisotropic plaque properties were characterized and a model of 3-dimensional stent artery interaction (incorporating parameterized commercially available stents) in a severely diseased iliac artery with 8 different components was developed.⁶ Given evidence that regions near the ends of a stent are particularly susceptible to restenosis, Berry et al.⁷ examined stresses in the artery wall in these locations. It was determined that high stress concentrations are imposed in these regions of compliance mismatch between the artery and the stent. Prendergast and colleagues⁸ modeled the stent-artery interaction of commercially available stents (NIR — Boston Scientific; S7 — Medtronic AVE) on an idealized stenosed artery. They proposed the testing methodology as a pre-clinical testing tool, which could be used to compare and contrast existing stent designs as well as aid in developing novel stent designs. The mechanical properties of stents have been investigated by Migliavacca and colleagues. In a study of the influence of geometry on the stent behavior, they determined that a stent with a low metal-to-artery surface ratio has higher radial and longitudinal recoil, but lower “dogboning”.⁹ It was

concluded that the strut thickness influences the stent performance in terms of foreshortening, longitudinal recoil and dogboning.

Bedoya et al.¹⁰ investigated the effects of varying specific design parameters. Using a cylindrical scaffold constructed with sinusoidal rings joined by axially oriented connector bars, finite element models of generic stents were constructed by varying three design parameters - the strut spacing (h), the radius of curvature (ρ), and the axial amplitude (f) (Figure 1).

The stent models were deployed in models of healthy circular cylindrical arteries. The effects of stent design on the stress field in the artery wall, radial displacement of the artery, and cyclic deflection of the artery were assessed. Designs that induced the greatest stresses in the artery wall, a characteristic considered unfavorable, also achieved the most favorable radial displacements. Conversely, designs that induced the lowest stresses in the artery wall produced the smallest radial displacements. Timmins et al.¹¹ provided an optimization scheme to reconcile the mutually exclusive nature of the low stress vs high radial displacement relationship. In the interest of advancing modeling techniques, it is important to address the use of cylindrical, healthy artery models.

While stents are not typically deployed in healthy arteries, the use of healthy artery models is purposeful in comparative studies aimed at elucidating the effects of varying specific design parameters, without additional complicating factors. Incorporation of disease elements, i.e. stenoses, is potentially convoluting. Like vessel geometry, material properties of atherosclerotic lesions vary greatly, from soft lipid pools to stiff calcified plaques. In a comparative study of the effects of stent design, the use of a specific geometry and combination of mechanical properties from one patient may not provide generally applicable results. In addition, animal studies can be carried out to verify results. A healthy artery model would be most similar to the animal model used to validate computational results. Further justification arises from the contact between the stent and healthy arterial tissue at the edges of the stented region, as stents are generally longer than the diseased region to ensure proper radial support. As restenosis commonly develops at this site, examination of the biomechanical environment in this region has a major clinical significance.

However, it has been acknowledged that to improve clinical relevance the effects of the stenosis must ultimately be addressed. Moreover, arteries in regions of the vasculature that are particularly vulnerable to disease typically vary in diameter along the axial direction, i.e. taper. Tapered vessels present unique challenges to stent therapy. Though not yet commonly utilized, stents incorporating a tapered design are available, e.g. Protégé RX (EV3, Plymouth, MN), designed for deployment in carotid arteries. In this case, the tapered form is intended to provide a more appropriate fit in the transition at the carotid bifurcation between the common carotid and smaller internal carotid artery. While rarely examined, coronary arteries also exhibit a degree of taper. Angiographic data suggested that left anterior descending (LAD) and right coronary arteries (RCA) taper approximately 14% and 9%, respectively, along their lengths in both men and women.¹² Even with the emergence of some tapered designs, the use of non-tapered stent designs in tapered vessels remains standard. The implications of this practice are the focus of the present investigation. Herein, we aim to employ the modeling techniques previously outlined¹⁰ to investigate the effects of specific stent design parameters on wall stresses and radial displacements in healthy tapered vessels.

Methods

In order to examine the effects of stent design on tapered vessels, the finite element method was employed to determine final lumen diameter, as well as estimate stresses in the artery wall of the elastic stent/artery system in mechanical contact. The two stent models investigated,

selected from the two extreme designs identified in Bedoya et al.¹⁰, are intended to represent designs that have high (1Z1) and low (2B3) radial rigidity (Table 1). The artery wall model features non-linear, strain-stiffening mechanical properties, along with geometrical dimensions which agree with general observations of large tapered arteries.

Model Geometry

The 3D stent geometries investigated in this study have been previously identified.¹⁰ Briefly, generic stent models were developed by varying three specific design parameters — strut spacing (h), radius of curvature at the crown junctions (ρ) and axial amplitude (f) (Figure 1). From the aforementioned investigation two stent designs were selected, representing the extreme cases (based on evaluation of the stresses induced in the artery wall) of the stent designs investigated. Stent 1Z1 is characterized by a tight strut spacing, zero radius of curvature, and low amplitude, while stent 2B3 incorporates a large strut spacing, large radius of curvature and large axial amplitude. Both designs have a constant strut thickness and width of 100 μm and an outer radius of 2.475 mm, which is approximately 10% larger than the systolic inner radius of the artery at the proximal end of the stented region. Since the biomechanical interest in stenting tapered arteries is at the regions where diameter differences are largest (i.e. edges of the stented region), stent models have been scaled accordingly to ensure proper investigation, but yet still be computationally feasible. Thus, only a portion (approximately 9 mm in length) of a typical full length stent was modeled. The stent material was modeled as 316L stainless steel ($E = 200 \text{ GPa}$, $\nu = 0.3$).

The tapered artery was modeled as an idealized, healthy, cylindrical vessel with proximal and distal radii values or 2.133 mm and 1.920 mm, respectively, at diastolic pressure. These radii values correspond to a 10% taper, by diameter, or a taper angle of 0.43° , both of which agree with values taken from angiographic data of large arteries.^{13, 14} The artery model was initially 18 mm in length under no axial load; with the stented region being the centermost 9 mm section. The artery material properties used in the computational models have been previously utilized,¹⁰ and were based on data obtained from pressure-diameter and force-elongation testing of porcine carotid arteries using a modified version of the Computer Aided Vascular Experimentation (CAVE) device described in Humphrey et al.¹⁵ Data from the mechanical testing was used to determine the constants for the constitutive relation, which took the form:

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

where, $C_{10} = 25,466 \text{ Pa}$, $C_{01} = -11,577 \text{ Pa}$, $C_{11} = -506 \text{ Pa}$, $C_{20} = 1703 \text{ Pa}$, and $C_{30} = 1650 \text{ Pa}$. Ultimately, the artery model employed herein was characterized as a straight homogeneous tapered cylinder with isotropic non-linear hyperelastic mechanical properties.

FEM Simulations

The stent and artery models used in this study were developed with MSC.Patran; MSC.Marc was employed as the non-linear solver (MSC Software, Santa Ana, CA). The element and nodal numbers for the simulations varied slightly due to differences in stent design. All tapered vessel models consisted of 15232 elements (74657 nodes). The more rigid 1Z1 stent design was comprised of 922 elements (8928 nodes), while the model incorporating the more compliant 2B3 design was represented by 938 elements (9036 nodes). The boundary conditions utilized included displacement boundary conditions, pressure and contact, and were applied in multiple stages. Initially an axial stretch ratio of 1.59 was applied to the vessel, simulating the axial tethering that was measured in vivo during carotid vessel harvesting. The vessel was then inflated by applying a pressure (35 kPa) sufficient to dilate the artery such that the oversized

stent could be “implanted”; i.e. the stent was originally positioned outside the artery and then translated in the axial direction such that the stent and artery mid-points along that direction coincided. Pressure was then reduced to systolic levels and subsequently, to diastolic levels. Displacement boundary conditions only allowing in-plane deformations for the stent struts and vessel symmetry edges were imposed on these quarter models. An analytical contact boundary condition restricting the motion of nodes in contact with the opposite body was enforced. This contact condition (“glue” option in MSC.Marc) enforces multipoint constraint equations that restrict the motion of the contacting body to be strictly in the normal direction. Displacements were interpolated using quadratic Lagrange functions, while the spherical stress was interpolated with a linear function. The contact bodies were defined by C^2 -continuous Non-Uniform Rational B-Splines surfaces (NURBS).

The computer cluster used to solve this boundary value problem consists of a head node with dual 2.8 Ghz 32-bit processors, 4GB of RAM. The slave nodes (15) consisted of single 2.8 GHz 32-bit processors, 2GB of RAM. The operating system of the computer cluster was RedHat 9. The version of Patran was 2005 release 2, and Marc 2005 release 1.

Evaluation Methods

The model results are all oriented such that the vessel tapers from left to right, i.e. the diameter at the right end of the stented portion is 10% smaller than the diameter at the left end of the stented portion. As a measure of the stent ability to prop the vessel open adequately, the final radial position achieved by each design was assessed. Further, it is hypothesized that the development of neointimal hyperplasia following stent implantation is mediated by the anomalous stresses induced in the artery wall by the stent. Herein the circumferential (hoop) stresses are presented since they are most likely to cause disruption of the internal elastic lamina; a phenomenon that has been directly linked with restenosis formation.^{16, 17} We have previously shown that circumferential stresses constitute the major contribution to the maximum principal stresses.¹⁰ For reference purposes, the average circumferential stresses at the proximal and distal ends of the stented region are estimated using the Law of Laplace as 34 kPa at diastolic pressure. While this formula is not strictly applicable to this situation, because it is not valid for thick walled tubes, the values serve as a general guideline to provide an indication of the degree to which stent implantation affects stress in the artery wall relative to the unstented case. Averaging nodal circumferential stress values through the vessel thickness at the proximal and distal ends of the stented region of the tapered artery models results in values of approximately 50 kPa at diastolic pressure.

Nodal values of radial displacement and circumferential stress values on the intimal surface of each tapered artery model were obtained at diastolic pressure. Radial displacement values were used to determine final radial positions after stent deployment. To assess the impact of the stent along the tapered vessel, these values were averaged and plotted against the undeformed axial position. Nodal values were averaged along the azimuth angle (circumference) of the quarter vessel at identical undeformed axial locations along the artery. This procedure was a convenient method to examine differences amongst stent designs, as well as compare against the unstented artery. Nodal circumferential stress values were also plotted as a colormap of the model for quantitative analysis. Both approaches are used herein to provide a more complete conception of the biomechanical impact of stent design on a tapered artery.

Convergence Criteria

Mesh convergence was evaluated as described in Bedoya et al.¹⁰ Briefly, the mesh convergence study consisted of a three-step process. The first step was to perform mesh refinements in the model of the artery alone — with no contact — observing the variation of maximum principal (i.e. circumferential) stress distributions. The criterion used was that the maximum principal

Cauchy stress field in the lumen and adventitia of the artery had to vary by less than 1%. The second step was to perform refinements in stents themselves by applying a pressure load on the outside of the stent and observing changes in displacements. The mesh was deemed converged when changes in displacement were less than 1% in radial displacement, which corresponded to stents with an element edge length of 0.10 mm. The third phase of the mesh convergence was to run stented artery models while increasing the mesh density of the artery to minimize variance in the circumferential stresses in the artery. Convergence was assumed when hoop stress and radial displacement values differed by <4.5% and <<1%, respectively, with increases in mesh density.

To optimize computational resources a non-uniform mesh of the vessel was constructed. The artery was divided into three regions in the axial direction (proximal and distal non-stented segments, central stented region). The mesh within the stented region was twice as dense as the non-stented region, thus allowing for a high mesh density only within the area of interest. The models were constructed incorporating 20-node hexahedral elements. Due to axisymmetry, only a quarter of the circumference of the artery and stent were modeled to save computational resources.

Results

The stiffer 1Z1 design produced radial displacements that increased in the direction of taper. The net effect is that the vessel assumes the shape of the stent, i.e. the taper is no longer evident in the stented segment. This is illustrated in Figure 2, which gives the final radial position of the intima averaged over the azimuth angle (θ -averaged) versus axial position at diastolic pressure. In comparing the radial position plots of the 1Z1 design with the unstented vessel, it can be seen that the increase in radius ranged from over 20% at the proximal (wide) end to near 25% at the distal (narrow) end of the stented region. With the more compliant 2B3 design, radial displacements were similar at both ends of the stent, with gains in radius near 15%; the greatest displacement occurred in the central portion of the stented region, where the increase in radius was greater than 18% (Figure 2). Thus, the gradual transition in compliance from stent to artery, a sign of “compliance matching”⁷ is evident near the ends of the stent. Differences between stent designs were small in the central stented region, where the stiffer 1Z1 design achieved displacements that were less than 3% greater than those achieved by the more compliant 2B3 design; and large at the distal end of the stented region, where the 1Z1 design achieved displacements that were more than 5% greater than those attained by the 2B3 design. Note that the stent over-sizing of 10% is determined at systolic pressure at the proximal end of the stented segment, while the aforementioned evaluation is conducted at diastolic pressure - as this would represent a “worst” case scenario, wherein the load bearing on the stent is at a maximum.

Examination of the tensile circumferential stresses imposed on the inner surface of the tapered artery model at diastolic pressure reveal that the highly rigid 1Z1 design induced diffuse stresses that covered a broad range. Peak stress values of approximately 700 kPa (>4X larger than the unstented vessel) were observed at the most distal strut (i.e. the narrowest section of the artery), with values >600 kPa covering more than 50% of the stented region (Figure 3). The average circumferential stress for all nodal values over the intima of the stented region was approximately 630 kPa for this design. Conversely, stent 2B3 induced stresses that were generally much lower throughout the stented region, with nodal hoop stress values averaging under 530 kPa. High stresses in the 2B3 design were more focal, localized at areas of mechanical contact between the stent struts and the artery (Figure 3). The peak stress for the 2B3 design was approximately 600 kPa (>3.5X larger than the unstented vessel) and was measured on the crowns of the most distal strut. Stress values >600 kPa, covered <<1% of the stented region. Following an approach similar to that described above, intimal circumferential

stress values were averaged about the circumference and plotted versus axial location at diastolic pressure (Figure 4). Here it is evident that the stresses induced by the stiffer 1Z1 stent increase in a quasi-linear fashion in the direction of taper. While the stresses are not as high with the 2B3 design, stress concentrations and high stress gradients near the stent struts are evident. Stresses associated with the 1Z1 design were 3-4 times higher than those associated with the unstented artery. Stresses associated with the 2B3 design were also high compared to the unstented artery but remained 100 – 200 kPa lower than those associated with the 1Z1 design.

For both designs there are high stress gradients present in the regions in between ring segments, as well as at the end of the stented regions. For the 1Z1 design, circumferential stress values vary abruptly by 50 – 75 kPa over distances of < 0.1 mm. The 2B3 design has larger differences, but over greater lengths, with stress variations of 80 – 90 kPa over distances of approximately 1 mm. The difference between average stress values at the proximal and distal regions (averaged over 1 mm in length) of the stents, were approximately 90 and 20 kPa for the 1Z1 and 2B3 design, respectively.

Discussion

Improving the biomechanical interaction between implanted devices and the surrounding tissues has the potential to reduce clinical failures. This reduction can be accomplished through implant design changes, as long as the desirable criteria are known. Given the extensive history of human implants of stents, this technology provides an opportunity to establish such criteria. The challenge in this case is to construct appropriate biomechanical models of a wide variety of in vivo conditions in a way that elucidates the important design criteria. The present study focused on the effects of vessel taper on stent/artery interaction.

Based on a previous study in non-tapered, healthy vessels,¹⁰ a relatively stiff stent design (1Z1) and a relatively flexible design (2B3) were selected for the present investigation. The stiffer design featured closely spaced rings with small crown amplitudes, while the more flexible design had a larger ring spacing and larger amplitude. In the non-tapered vessels, the stiffer design propped the artery open to a greater degree, but provoked much higher stresses throughout the artery wall. The more flexible stent also propped the artery open to a clinically successful (although smaller) final diameter, with much lower artery wall stress. The use of finite deformation contact mechanics was crucial, as assuming a perfectly rigid stent would not have given similar results. Furthermore, modeling of truly tapered arteries provides a more complete representation, relative to modeling several non-tapered stented arteries at different degrees of diameter mismatch. That is, while one could perhaps get some idea of the stress field in a tapered artery by constructing two straight artery models whose diameters correspond to the proximal and distal diameters of the tapered artery, the direct effects of tapering on the final stent/artery configuration (e.g. axial dependence of displacement) would remain unknown. Such a strategy would also require a greater number of computational cases to be run. In addition, the stented tapered arteries would have to be modeled to verify that the use of varying degrees of diameter mismatch is an appropriate representation.

Simulating the implantation of these stents in tapered vessels provides further indications that stiffer stents dominate over the artery wall in determining the final geometry and thus the artery wall stress field. The stiffer stent in this case resulted in a nearly constant final lumen diameter in the stented region, despite the natural taper of the artery by 10% along the length of the stent. Given the non-linear, strain stiffening behavior of the artery, the stent had to apply a much greater outward force on the distal end of the tapered vessel, resulting in particularly high artery wall stresses at this region. By contrast, the more flexible stent design followed the natural

contour of the artery, with the final deformed geometry still exhibiting some degree of taper. As a result, the stresses at the distal end were much lower.

While there is no known explicit stress level which can lead to an adverse biological response (inflammation, intimal hyperplasia, etc.), design characteristics which minimize stent induced stresses, possibly preventing post procedural complications, should be considered. Currently there are tapered stent designs on the market (e.g. Protégé RX; EV3, Plymouth, MN), but placement of these stents requires an additional degree of accuracy with regard to stent choice to ensure a correct geometric match with the artery wall. Based on our findings, a more flexible stent design can still achieve a clinically successfully final lumen diameter, while reducing the stress induced on the artery wall. This could possibly reduce the chance of post-procedural complications that can be associated with stent deployment. Furthermore, a unique solid mechanical environment occurs in tapered regions of the vasculature that are commonly stented (i.e. carotid, coronary, femoral). These regions see large deformations that result from pulse pressure, myocardial contraction, and limb movements such as hip and knee flexion.^{18, 19} Thus, a more flexible stent design would assist in preventing strut fracture, a common cause of clinical failure in stent therapy. It would also be plausible to develop an algorithm, similar to the optimization scheme developed by Timmins et al.¹¹, to allow for the prediction of the biomechanical environment after stent implantation based on stent design and patient specific vessel geometry. For example, if given adequate input data (i.e. sufficient computational models which encompass the range of parameters of interests), an algorithm that examined parameters such as stent length, stent diameter, and degree of stent or artery tapering could be developed. Such a scheme could provide clinicians with optimal stent designs when tapered geometries are encountered.

From a clinical standpoint, the results presented provide general guidelines for appropriate stent designs that can be used to treat tapered arteries. As most sites of stent deployment (e.g. coronary, carotid, femoral) experience a significant degree of taper, implantation of an unfavorable stent design could cause unnecessary trauma to the artery wall. Such injury could increase the risk of restenosis development or other major cardiac events; with a judicious choice of stent design such complications could be avoided. For example, our results indicate that if confronted with an artery with a high degree of taper, the clinician should choose a stent design that is more compliant and can conform to the natural taper of the artery. This will minimize the stress induced on the artery wall, with the possibility of increasing the success rate of the clinical procedure. The use of a more compliant stent could then serve the intended purpose of a tapered stent, without having to make the additional choice of the appropriate degree of taper for a particular artery.

Limitations associated with this study include the use of axisymmetric artery wall models without a simulated diseased segment. The use of a specific diseased geometry would have limited the applicability of the results. It would have also been more difficult to compare the results from the two different stent designs, as local differences in stress would have been created simply by some portion of the plaque being covered by struts from one design and not another. It was our desire to generate results that have the widest degree of applicability, along with the capability of directly comparing the two stent designs. This comparison is most effectively done with the non-diseased tapered model used here. While residual stresses were not incorporated into the models, the stent induced changes in stress are at least an order of magnitude larger than those associated with residual stress. Therefore, any influence that this would have on the values presented would likely not affect the relative comparison amongst the two stent designs. The use of relatively short stents is a minor limitation, in that it is not likely that the use of longer stents (with equivalent degrees of tapering in the arteries) would have resulted in any new information. It should also be noted that enforcement of the “glue” contact boundary prevented the stent from translating after “implantation”. Due to the high

radial displacements achieved by both stent designs, the stents were essentially embedded in the artery, as they would be during balloon inflation, and sliding, if any, would be minimal. Our previous investigations with non-tapered arteries showed that changing the contact condition to allow the stent to slide freely in the artery has a minimal effect on the stress environment.²⁰

In summary, we have modeled the mechanical interaction of two different stents (one less stiff than the other) with tapered arteries. The rigidity of the stent was found to be an important determinant of final artery wall diameter and stress. More compliant stents should be considered in cases where significant arterial tapering is found.

Acknowledgements

This research was supported by NIH Grant No. R01 EB000115.

References

1. Rosamond W, et al. Heart disease and stroke statistics--2007 update: a report from the American Heart Association Statistics Committee and Stroke Statistics Subcommittee. *Circulation* 2007;115(5):e69–171. [PubMed: 17194875]
2. Bosiers M, et al. Carotid artery stenting: which stent for which lesion? *Vascular* 2005;13(4):205–210. [PubMed: 16229793]
3. Rogers C, et al. Balloon-artery interactions during stent placement: a finite element analysis approach to pressure, compliance, and stent design as contributors to vascular injury. *Circulation Research* 1999;84(4):378–83. [PubMed: 10066671]
4. Kastrati A, et al. Restenosis after coronary placement of various stent types. *Am J Cardiol* 2001;87(1):34–9. [PubMed: 11137830]
5. Serruys PW, Kukreja N. Late stent thrombosis in drug-eluting stents: return of the 'VB syndrome'. *Nature Clinical Practice Cardiovascular Medicine* 2006;3(12):637.
6. Holzapfel GA, Stadler M, Schulze-Bauer CA. A layer-specific three-dimensional model for the simulation of balloon angioplasty using magnetic resonance imaging and mechanical testing. *Annals of Biomedical Engineering* 2002;30(6):753–67. [PubMed: 12220076]
7. Berry JL, et al. Hemodynamics and wall mechanics of a compliance matching stent: in vitro and in vivo analysis. *Journal of Vascular & Interventional Radiology* 2002;13(1):97–105. [PubMed: 11788701]
8. Lally C, Dolan F, Prendergast PJ. Cardiovascular stent design and vessel stresses: a finite element analysis. *Journal of Biomechanics* 2005;38(8):1574–81. [PubMed: 15958213][erratum appears in *J Biomech.* 2006;39(9):1760]
9. Migliavacca F, et al. Mechanical behavior of coronary stents investigated through the finite element method. *Journal of Biomechanics* 2002;35(6):803–11. [PubMed: 12021000]
10. Bedoya J, et al. Effects of stent design parameters on normal artery wall mechanics. *Journal of Biomechanical Engineering* 2006;128(5):757–65. [PubMed: 16995763]
11. Timmins LH, et al. Stented artery biomechanics and device design optimization. *Medical & Biological Engineering & Computing* 2007;45(5):505–13. [PubMed: 17375345]
12. Zubaid M, Buller C, Mancini GB. Normal angiographic tapering of the coronary arteries. *Can J Cardiol* 2002;18(9):973–80. [PubMed: 12368932]
13. Humphrey, JD. *Cardiovascular Solid Mechanics: Cells, Tissues, and Organs*. Springer; 2002.
14. Milnor, W. *Hemodynamics*. Vol. 2nd ed. Williams & Wilkins; Baltimore: 1989.
15. Humphrey JD, et al. Computer-aided vascular experimentation: a new electromechanical test system. *Ann Biomed Eng* 1993;21(1):33–43. [PubMed: 8434818]
16. Farb A, et al. Morphological predictors of restenosis after coronary stenting in humans. *Circulation* 2002;105(25):2974–80. [PubMed: 12081990]

17. Indolfi C, et al. Rat carotid artery dilation by PTCA balloon catheter induces neointima formation in presence of IEL rupture. *Am J Physiol Heart Circ Physiol* 2002;283(2):H760–7. [PubMed: 12124225]
18. Ding Z, Friedman MH. Quantification of 3-D coronary arterial motion using clinical biplane cineangiograms. *Int J Card Imaging* 2000;16(5):331–46. [PubMed: 11215918]
19. Cheng CP, et al. In vivo MR angiographic quantification of axial and twisting deformations of the superficial femoral artery resulting from maximum hip and knee flexion. *J Vasc Interv Radiol* 2006;17(6):979–87. [PubMed: 16778231]
20. Bedoya, J. Stent Design and Arterial Mechanics: Parameterization Tools Using the Finite Element Method, in Dept. of Biomedical Engineering. Texas A&M University; College Station, TX: 2005.

Figure 1. Stent Design Parameters

Illustrations of the stent designs employed, showing the three parameters used in their construction: h is connector bar length (or strut spacing), ρ is the radius of curvature at the crown junctions, and f is the axial amplitude.

Figure 2. Final Radial Position

Deformed radial position values were averaged around the circumference of the vessel at diastolic pressure. With the stiff 1Z1 stent design, the vessel essentially assumes the shape of the stent, i.e. the taper of the vessel is no longer evident through the stented region. With the more compliant 2B3 design a gradual transition in compliance from stent to artery is evident near the ends of the stent.

Figure 3. Circumferential Stress Distribution

Stent 1Z1 imposed high stresses (> 600 kPa) across larger portions of the artery, whereas stent 2B3 (less rigid) only imposed such high stresses in the most narrow part of the artery. Overall, the 2B3 design imposed stress values that were 100 – 200 kPa lower than the 1Z1 design.

Figure 4. Average Circumferential Stress

Intimal artery wall circumferential stress (averaged over the circumference) versus axial position plots reveal the dominance of the 1Z1 stent over the taper resulting in high stresses at the distal end. The less stiff 2B3 stent more naturally follows the arterial taper, resulting in lower stresses. Note circumferential stress values at the intimal are most likely to affect the internal elastic lamina, and restenosis rates have been linked to the disruption of this structure.
16, 17

Table 1

Stent Designs and Parameter Values.

Stent	Strut Spacing- h (mm)	Radius of Curvature- ρ (mm)	Axial Amplitude- f (mm)
1Z1	1.2	0	0.6
2B3	2.4	0.3	1.8