

5-2012

Simulation of Lower Limb Axial Arterial Length Change During Locomotion

Melissa D. Young

Cleveland Clinic, youngm7@ccf.org

Matthew C. Streicher

Cleveland Clinic

Richard J. Beck

Cleveland Clinic

Antonie J. van den Bogert

Cleveland State University, a.vandenbogert@csuohio.edu

Follow this and additional works at: https://engagedscholarship.csuohio.edu/enme_facpub

 [Digital Commons](#) Biomechanics and Biotransport Commons

Open Access to this work benefit you? Let us know!

Publisher's Statement
See next page for additional authors

NOTICE: this is the author's version of a work that was accepted for publication in Journal of Biomechanics. Changes resulting from the publishing process, such as peer review, editing, corrections, structural formatting, and other quality control mechanisms may not be reflected in this document. Changes may have been made to this work since it was submitted for publication. A definitive version was subsequently published in Journal of Biomechanics, 45, 8, (05-01-2012); 10.1016/j.jbiomech.2012.02.011

Original Citation

Young, M. D., Streicher, M. C., Beck, R. J., van den Bogert, A.J., Tajaddini, A., & Davis, B. L. (2012). Simulation of lower limb axial arterial length change during locomotion. Journal of Biomechanics, 45(8), 1485-1490. doi: 10.1016/j.jbiomech.2012.02.011

This Article is brought to you for free and open access by the Mechanical Engineering Department at EngagedScholarship@CSU. It has been accepted for inclusion in Mechanical Engineering Faculty Publications by an authorized administrator of EngagedScholarship@CSU. For more information, please contact library.es@csuohio.edu.

Authors

Melissa D. Young, Matthew C. Streicher, Richard J. Beck, Antonie J. van den Bogert, Azita Tajaddini, and Brian L. Davis

Simulation of lower limb axial arterial length change during locomotion

Melissa D. Young^{a,b,*}, Matthew C. Streicher^b, Richard J. Beck^b, Antonie J. van den Bogert^c, Azita Tajaddini^c, Brian L. Davis^d

^a Department of Biomedical Engineering, Lerner Research Institute, Cleveland Clinic, 9500 Euclid Avenue, Cleveland, OH 44195, USA

^b Medical Device Solutions (MDS) Core, Lerner Research Institute, Cleveland Clinic, Cleveland, OH 44195, USA

^c Orchard Kinetics, Cleveland, OH 44195, USA

^d Medical Device Development Center, Austen BioInnovation Institute in Akron, Akron, OH 44308, USA

1. Introduction

The peripheral arterial disease (PAD) is common in patients with diabetes and is a major risk factor leading to lower extremity amputations. Beyond morbidities, PAD significantly increases the risk of mortality from coronary and cerebrovascular events (Criqui, 2001). Investigation of arterial physiology is crucial in reducing the morbidity and mortality caused by PAD. Correction of arterial stenosis using stents can maintain patency in affected vessels; however, these devices are susceptible to mechanical problems. For example, the structural frame of the stent can fracture, broken struts can lead to thrombus formation, or the entire stent can migrate in the vasculature. The incidence of fractures corresponds to the length of the stented segment and the number of implants (Scheinert et al., 2005). This may be influenced by stress concentrations at the device–artery interface imposed by axial shortening and elongation during activities of daily living. Nitinol stent fractures have been observed more frequently in patients with PAD who walked more than 5000 steps a day compared to those patients who did not exercise ($p=0.0027$; Iida et al., 2006). This higher rate of stent fracture

could be related to mechanical fatigue as a result of repetitive arterial deformations.

Movement of the skeleton is not only an issue with stented arteries, but also joint excursions can have a direct impact on vascular grafts. Considering the arterial anatomy and joint mechanics, vascular tissue must withstand axial demands placed on blood vessels. This arterial shortening and elongation can directly influence the design of vascular grafts, both at the macro-level (as with stents) and at the micro-level where cells respond to oscillatory stresses. Studies imposing cyclic strain on cell-seeded constructs have shown that mechanical stimulation is an essential regulator of remodeling in tissue-engineered vessels (Seliktar et al., 2003). Controlled axial strain is considered vital in the conditioning of tissue engineered vascular grafts (Mironov et al., 2003; Gleason et al., 2004). Lack of compliance in synthetic vascular grafts is associated with intimal hyperplasia (thickening of the tunica intima) around the suture line (Ballyk et al., 1998). Therefore, implant axial compliance is crucial for accurate reproduction of physiological conditions.

Modern vascular implants do not replicate the mechanical properties of the native tissue, and provide undesirable stress concentrations at their interface. Specifically, physiological compliance of arteries in the longitudinal direction is not equivalent to those of synthetic devices. Metal alloy stents are stiffer than the vascular tissue and do not match the properties of a diseased vessel. For vascular grafts, the graft–vessel junction may be a potential site where compliance of the

* Corresponding author at: Department of Biomedical Engineering, Cleveland Clinic, Cleveland, OH 44195, USA. Tel.: +1 216 444 1793; fax: +1 216 444 9198.
E-mail address: youngm7@ccf.org (M.D. Young).

graft does not match the arterial compliance. Axial shortening and elongation may therefore play an important role in implant failure. Following stent placement in a given artery such as the femoropopliteal segment, the stented portion of the artery experiences less shortening than an unstented segment during hip and knee joint flexion. An unstented artery will shorten and/or bend more to accommodate a more rigid and stented portion of the femoropopliteal artery. With overlapping stents, the stiffness of the arterial segment is further increased. The bare portion of the artery may shorten more during flexion (Smouse et al., 2004), causing an uneven loading distribution that may lead to earlier implant failure.

Several imaging approaches have been used to describe arterial mechanical behavior. For example the superficial femoral artery (SFA) was examined using gadolinium-enhanced magnetic resonance angiography (MRA) during musculoskeletal manipulation involving knee and hip flexion. SFA shortening measured with this technique reached $13 \pm 11\%$ ($p < 0.001$) on average with a maximum of 25% under extreme knee and hip flexion (Cheng et al., 2006). Under knee and hip flexion manipulation, producing similar angles to those seen during normal gait, the SFA experienced $6.9 \pm 1.9\%$ shortening with a greater percentage coming from the distal third of the SFA ($8.1 \pm 2.0\%$), closer to where knee flexion was occurring (Cheng et al., 2010). The iliac arterial length was found to be the most affected by flexion in the hip joint. The common iliac artery was evaluated similarly, and shortening of the iliac artery (2.2–5.2%) was observed when contorting the human body from a supine to the fetal position, thus exaggerating hip flexion (Choi et al., 2009a, 2009b). While arterial properties determined using imaging are valuable, the standard deviation for such approaches is large and the results are dependent on the imaging equipment quality and resolution.

Other techniques used to determine arterial properties include (i) performing in situ stretch measurements and (ii) comparing the artery length before it is excised from an animal or cadaver with the length of the *ex situ* artery. The main limitation of these methods is that the arterial data acquired is specifically for non-load bearing weight activities, and does not accurately depict the artery during every day activities.

Since vascular implants are intended to be subjected to a range of loading conditions associated with routine activities of daily living, this study focused on (i) using a non-invasive approach to quantifying arterial stretch and (ii) developing a method that was allowed for dynamic activities to be assessed without the need for additional imaging. The goal was to quantify biomechanical parameters that could improve implant performance and durability. This study provided a means for comparing kinematic quantification of the femoropopliteal and iliac shortening/elongation for the following configurations: 1) gait, 2) ascending a step, and 3) sit-to-stand.

2. Methods

Motion data were obtained at 60 Hz, using an eight-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA), for upright standing, normal gait, stair climbing, and progression from sitting to standing positions in accordance with protocols approved by the Institutional Review Board. Lower extremity joint angles were calculated from the post processed data for comparison purposes using the Motion Analysis® Orthotrak software. The analysis involved the following two steps: 1) inverse kinematic analysis to solve joint angles based on marker coordinates and 2) forward kinematic simulation, using these joint angles, to compute artery length change during movement. Eleven subjects, seven male and four female, performed the aforementioned activities. The tracked data for the first four patients were loaded into SIMM (Software for Interactive Musculoskeletal Modeling, V.4.1, MusculoGraphics, Inc., Motion Analysis Corp., Santa Rosa, CA), and the remaining data was loaded into OpenSim (V.2.2.1) (Delp et al., 2007) to scale a musculoskeletal model. Realistic geometrical constraints were set on “muscle objects” by conforming their length and

orientation to the motion of the joints (Delp and Loan, 1995). Objects created as muscles and placed in the location of an artery followed similar geometric rules.

The model was modified to include representatives of the iliac, femoral and popliteal arteries (Fig. 1), following their general anatomic route. The main bifurcations were modeled as origin and insertion. Resting lengths of the model arteries were defined with the body in the upright standing position. The iliac artery was fixed at the fifth lumbar vertebrae, the location of aortic abdominal bifurcation. It then curved along the pelvic brim and passed under the inguinal ligament to become the second model artery: the femoropopliteal arterial segment. The main branching of the femoral artery near the lesser trochanter of the femur was set as a fixed point. In the popliteal fossa, the artery was deep near the femur and more superficial along the tibia. The femoropopliteal arterial segment was fixed at the bifurcation of the tibial arteries.

The described anatomy defined the static origin, whereas joint motion governed the change in length. The simulation was geometrical and was independent of regional properties within the arterial segments. The Motion Analysis® data defined the joint kinematics data and the internal body structures conformed to the orientation of the bones. Hip and knee flexion angles were more specifically looked at Fig. 2, and for this continuous arterial model, the arterial length was predicted in all subjects as a function of upright standing, walking, stair climbing, and sitting-to-standing using SIMM and OpenSim.

The static length was considered to be the length of the artery when the body is in the upright standing position. Axial shortening and elongation were calculated as the percent length change less than or higher than the static length, respectively. In some activities, elongation did not occur. All measurements are provided as means \pm standard deviation. Regression and ANOVA analysis using Minitab Statistical Software (Minitab Inc., State College, PA, USA) was used in Figs. 3 and 4 to identify statistical significance ($p < 0.05$).

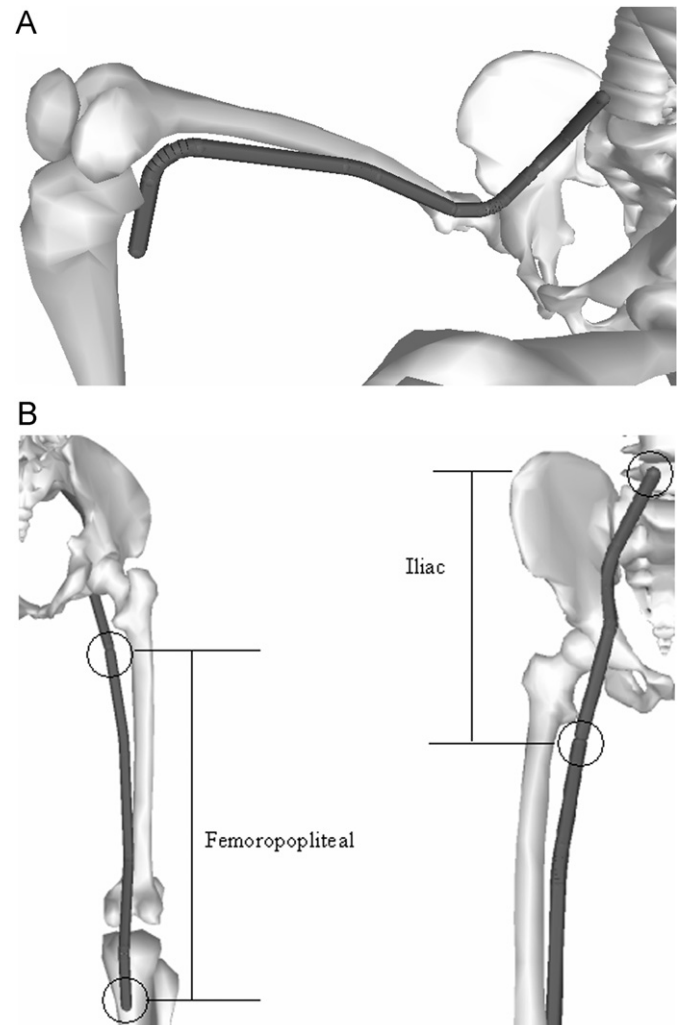


Fig. 1. Views of the adapted SIMM graphics-based model: sitting (A), posterior and anterior (B).

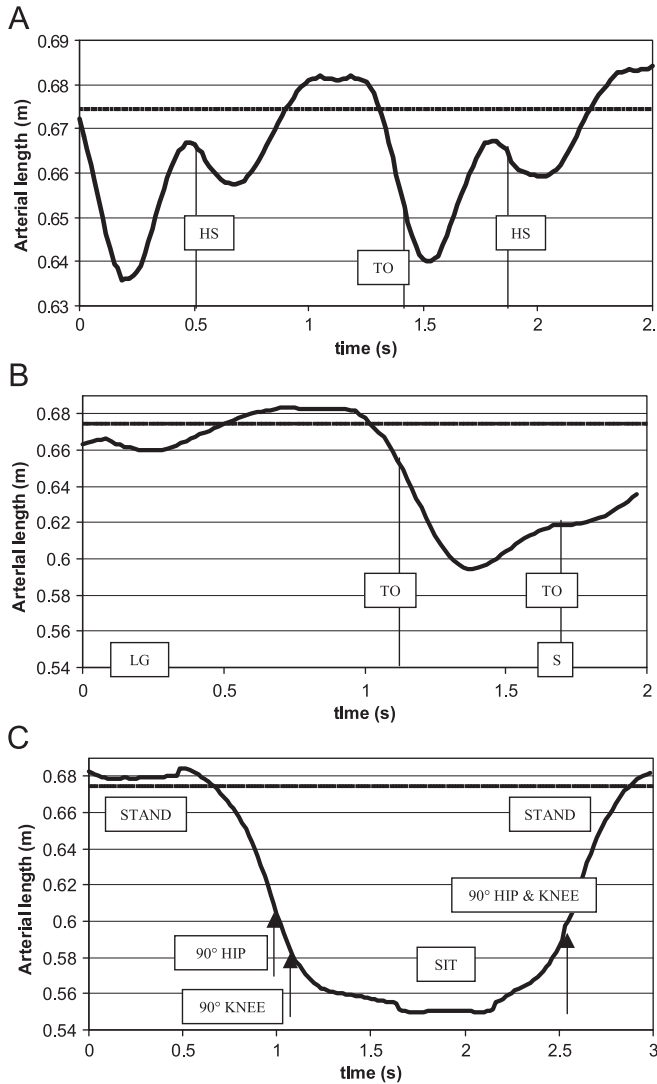


Fig. 2. Plots from the SIMM model of arterial length as a function of time for subject 2. The length shown is the total artery length, which was found by summing arterial length from each arterial segment for each point in time. The extremes of the iliac and femoropopliteal arteries did not usually occur at the same instant in time; therefore this total was not a simple addition of extremes. The resting arterial length of the combined iliac artery and femoropopliteal arterial segment is shown as the bold dotted horizontal line in each graph. Arterial length as a function of time is displayed for (A) walking, (B) stair climbing, and (C) progressing from a standing to a sitting position. References describing movement and points of 90° flexion of the hip and/or knee are provided. The markers for motion are as follows: HS indicates heel strike, TO indicates toe-off, and TC indicates toe contact. Stepping from level ground (LG) to one step (S) is specified on the time axis.

3. Results

The focus of the current study was on (i) modeling key physical (i.e., anatomical and kinematic) characteristics of lower extremities during various activities of daily living and then (ii) using the model to simulate global changes in arterial length for each activity. Descriptive statistics for height, mass, and age are summarized in Table 1. Height ranged from 156 to 190 cm, mass ranged from 54.2 to 99.5 kg, and age ranged from 23 to 69 years old. When comparing the arterial shortening percentages observed between the male and female subjects, there was no significant difference in iliac or femoropopliteal arterial shortening. Additionally, according to the models, there was no variation in arterial shortening of either vessel relative to the subject's age.

Gait ($8.3 \pm 0.6\%$), stair climbing ($11.9 \pm 2.2\%$), and the sitting and standing motion ($14.7 \pm 3.8\%$) all exhibited significant shortening of the femoropopliteal arterial segment among the 11 subjects relative to the static length obtained from upright standing. Some elongation was seen in each activity, but not with any consistency among the subjects (Table 2). Elongation values ranged from 0 to 0.9% for gait, 0 to 0.6% for stair climbing, and 0 to 1.0% for sitting/standing. In comparison to the average static length of the femoropopliteal artery (43.1 cm), the elongation percentages of less than 1.0% (approximately 0.4 mm) showed that elongation was not a common characteristic of the femoropopliteal artery during these activities of daily living; there were multiple subjects in each activity that did not experience any calculated arterial elongation at all.

The iliac arterial segment experienced a similar trend of shortening during gait ($5.6 \pm 1.0\%$), stair climbing ($12.8 \pm 2.0\%$), and sitting and standing ($16.0 \pm 1.9\%$), but also had measurable elongation (gait: $4.6 \pm 1.5\%$; stair climbing: $3.6 \pm 1.7\%$; sitting/standing: $1.4 \pm 1.6\%$) compared to the static arterial length (Table 3). The elongation observed for the iliac artery ranged from approximately 0.4 mm of stretch during sitting/standing, to 1.2 mm elongation during normal gait for the 11 subjects tested.

The femoropopliteal arterial region shortened as much as 23% (9.8 cm) and the iliac arterial segment had a maximum shortening of 21% (5.3 cm) during the prescribed activities of daily living. As expected, both overall maximum arterial shortenings occurred during the sitting and standing activity when hip and knee joint flexion angles were at their highest values. The percent shortening of the iliac arterial region compared to the maximum hip flexion angle during the various activities showed a strong correlation ($r^2=0.9$), and can be described by Eq. (1), where $I_{shortening}$ is the iliac shortening and $\theta_{HF_{max}}$ is the maximum hip flexion angle. The femoropopliteal arterial region shortening and maximum knee flexion angle also exhibited a strong correlation ($r^2=0.8$) shown by Eq. (2), where $F_{shortening}$ is the femoropopliteal shortening, and $\theta_{KF_{max}}$ is the maximum knee flexion angle.

$$\%F_{shortening} = (0.0016)\theta_{HF_{max}} - 0.0289 \quad (1) \leftarrow$$

$$\%I_{shortening} = (0.0016)\theta_{KF_{max}} + 0.0065 \quad (2) \leftarrow$$

4. Discussion

Previous research has only looked at passive joint motion. The ability to relate external kinematics to internal tissue mechanics during activities of daily living provides a new perspective on the axial deformation properties of these arteries. The geometrical demands associated with daily activities illustrate the necessity for arterial compliance in the longitudinal direction. The flexions of the hip and knee joints have been found to have a significant impact on the shortening of the two arteries modeled relative to the static arterial length calculated while standing upright. Although the simulation does not reflect arterial properties, it estimates the mechanical burden on the arteries while the body is weight bearing and in motion.

The extent of shortening calculated from the SIMM and OpenSim model is informative, demonstrating femoropopliteal shortening as much as 23%. This is comparable to the 20% mean shortening of the popliteal artery observed by radiograph in vivo during knee flexion (Browne et al., 1979) and approximately 25% shortening of the popliteal arterial axis based upon radiologic findings and theoretical calculations during 90° knee flexion (Vernon et al., 1987). Compared to the joint motions looked at previously (Smouse et al., 2005), the computer software models showed similar arterial shortening in the femoropopliteal segment. For gait-like angles approximately 9.3%

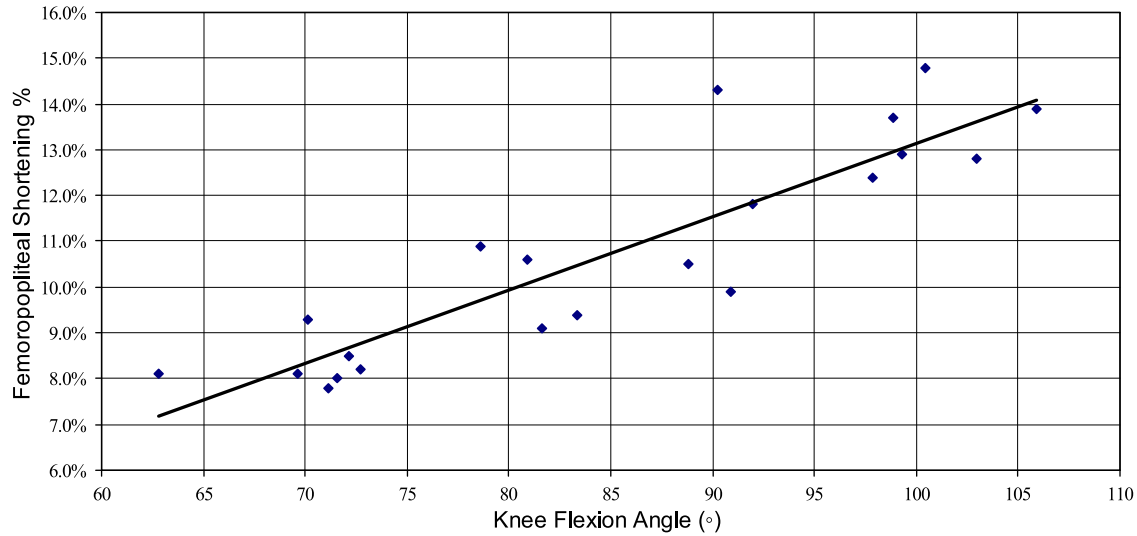


Fig. 3. Regression plot comparing the femoropopliteal arterial shortening derived from SIMM and OpenSim to maximum hip flexion angle observed during activities of daily living.

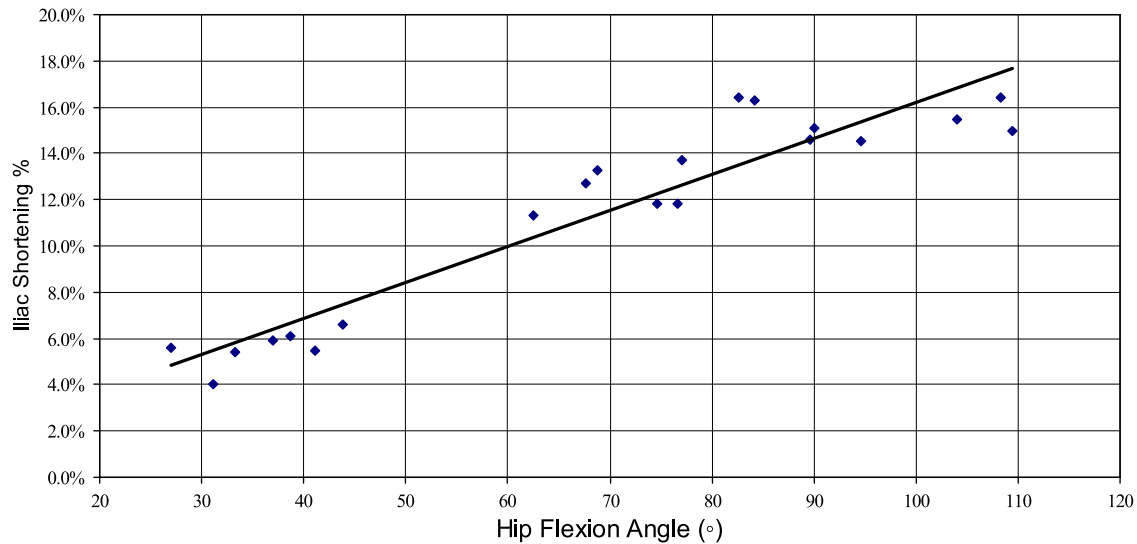


Fig. 4. Regression plot comparing the iliac arterial shortening derived from SIMM and OpenSim to maximum knee flexion angle observed during activities of daily living.

Table 1
Descriptive statistics of the test demographics.

Characteristics for participants (N=11)	Mean (SD)
Height (cm)	173.9 (10.9)
Males (N=7)	179.1 (8.9)
Females (N=4)	164.8 (7.7)
Mass (kg)	71.9 (13.0)
Males (N=7)	78.4 (11.3)
Females (N=4)	60.4 (5.8)
Age	43.5 (16.3)
Males (N=7)	39.6 (17.3)
Females (N=4)	50.5 (13.5)

femoropopliteal artery shortening and for sitting and standing or stair climbing roughly 15.7% shortening was observed. Compared to the 8.3% shortening observed from the gait activity and 13.3% shortening from the combined sitting/standing and stair climbing in this study, the computational model appears to be a valid resource for estimating femoropopliteal artery axial deformation.

However, it is difficult to do a true comparison due to the fact that the data presented by Smouse et al. (2005) is much more localized by breaking down the femoropopliteal artery into three segments: mid-SFA, distal SFA, and popliteal. The SIMM and OpenSim models estimate a regional rather than local shortening and elongation.

The model provided a means for quantifying the significant factors for vessel mechanics during day to day activities, but the model had some limitations as well. The model is intended to represent the femoropopliteal and iliac vessels; however because of the complexity of the vasculature the model assumed the vessel curvature as a straight line following the path of the muscle. The simulation does not account for bending, axial twisting motion, or branching that is present in these vessels during gait, ascending stair, and sit-to-stand activities. Arterial bifurcations in vivo would have external forces acting on each branch, resulting in complex stresses, while iliac artery axial deformation appears to be less conclusive based on the limited data available. For the most part, the former research looked at only the common iliac artery, while the SIMM and OpenSim model looked at the continuous segment of the common iliac artery and the external iliac artery.

Table 2

Axial geometric properties of the femoropopliteal arteries derived from SIMM and OpenSim length change data for each subject. Shortening and elongation are expressed as percentages of the corresponding resting (static) length.

Subject	Static (cm)	Gait			Ascending a step			Stand-to-sit		
		Shortening (%)	Elongation (%)	Range	Shortening (%)	Elongation (%)	Range (%)	Shortening (%)	Elongation (%)	Range (%)
S1	46.1	7.5	0.9	8.3	10.5	0.0	8.9	14.4	0.0	14.3
S2	42.6	9.4	0.2	9.6	14.1	0.0	13.6	16.9	0.0	16.9
S3	40.8	8.3	0.2	8.5	11.6	0.1	11.8	19.0	0.0	18.9
S4	42.4	8.5	0.0	8.5	16.2	0.4	16.6	23.1	0.0	23.2
S5	48.8	9.3	0.6	9.9	14.3	0.0	13.5	14.8	0.0	12.8
S6	41.2	8.1	0.0	7.9	10.9	0.0	10.7	10.6	0.2	10.8
S7	43.5	8.2	0.9	9.1	10.5	0.0	10.2	12.9	0.1	13.0
S8	40.3	7.8	0.3	8.2	9.1	0.0	8.8	9.9	1.0	10.9
S9	41.0	8.0	0.7	8.7	9.4	0.1	9.5	13.7	0.0	13.2
S10	46.9	8.5	0.5	9.1	11.8	0.6	12.4	12.8	0.0	12.9
S11	40.2	8.1	0.6	8.7	12.4	0.1	12.5	13.9	0.0	13.3
Average	43.1	8.3	0.4	8.8	11.9	0.1	11.7	14.7	0.1	14.6
St. dev.	2.9	0.6	0.3	0.6	2.2	0.2	2.4	3.8	0.3	3.7

Table 3

Axial geometric properties of the iliac arteries derived from SIMM and OpenSim length change data for each subject. Shortening and elongation are expressed as percentages of the corresponding resting (static) length.

Subject	Static (cm)	Gait			Ascending a step			Stand-to-Sit		
		Shortening (%)	Elongation (%)	Range (%)	Shortening (%)	Elongation(%)	Range	Shortening	Elongation	Range
S1	26.1	7.2	2.7	9.9	14.1	0.9	15.0	16.0	0.0	15.2
S2	24.8	3.8	5.7	9.5	9.6	5.5	15.1	21.2	4.7	25.9
S3	29.2	5.4	2.5	7.9	10.8	1.1	11.9	14.4	1.4	15.8
S4	23.8	5.8	3.3	9.1	15.4	4.1	19.5	17.1	2.3	19.4
S5	27.3	5.4	5.4	10.8	13.7	5.5	19.2	14.6	3.2	17.7
S6	21.7	4.0	7.3	11.3	12.7	5.0	17.7	15.0	2.8	17.8
S7	23.3	6.6	4.5	11.1	11.8	4.1	15.9	14.5	0.7	15.3
S8	21.8	5.6	5.0	10.5	13.3	1.6	14.8	15.1	0.0	15.2
S9	23.6	5.5	4.8	10.2	11.8	2.8	14.6	15.5	0.0	15.1
S10	27.7	5.9	3.7	9.6	11.3	4.7	16.0	16.4	0.4	16.8
S11	23.8	6.1	6.2	12.3	16.4	3.8	20.2	16.3	0.4	16.6
Average	24.8	5.6	4.6	10.2	12.8	3.6	16.4	16.0	1.4	17.3
Std. dev.	2.5	1.0	1.5	1.2	2.0	1.7	2.5%	1.9%	1.6%	3.2%

The model in its current form does not account for pathological states. In diseased vessels, the atherosclerotic lesion is not as distensible as normal vascular tissue (Giannattasio et al., 2001). The calcified plaque requires compensatory axial compliance from the adjacent healthy vascular segments to submit to the geometric demands imposed by locomotion. It was shown in cadaveric studies that examined lower limb extremity movements and their effects on stenting that longer and more rigid stents severely compromised the ability of the SFA or popliteal artery to axially compress (Smouse et al., 2005). In addition, arteries associated with PAD experience other deformation changes that may have a potential impact on stent failure. The diseased artery has stress concentrations around plaques and implants due to the abrupt change in properties. The model used in this study does not take into account these potential changes in mechanical properties of the arteries throughout its length; rather it assumes uniformity spanning the entire length of the vessel.

Both vessel curvature and axial twist are not calculated from the SIMM or OpenSim software. The amount of vessel curvature is affected by the surrounding tissue support and vessel compliance. Changes may be observed in this curvature as a result of pulsatile or nonpulsatile forces that produce a bending moment in the artery. Axial twisting of the vessel is a result of specific musculoskeletal movements or cardiac pulsatility which can produce a torsional moment (Choi et al., 2009a). Curving on the inner surface of the bending popliteal artery [15], morphological changes occurring in the femoral artery during knee flexion

(Cheng et al., 2006, 2010; Choi et al., 2009b; Smouse et al., 2005), and the tortuosity of the vessel being dependent upon age (Wensing et al., 1995) have all been noted.

The results provided in this study demonstrate that the kinematic model approach has potential for further exploration of the arterial mechanics under load bearing conditions. Future work involves acquiring a larger population and sample size, specifically because the motion analysis markers tend to move during data capture due to 1) soft tissue artifact, 2) difficulty of finding the bony landmarks, and 3) specific challenging positions such as near the hip (Andriacchi and Alexander, 2000; Stagni et al., 2005). The calculations in this study showed that age was not a factor for influencing shortening or elongation; however due to the limited subject data, a larger sample size would be needed to determine the true impact of age.

Other future work includes a validation of this technique by comparing it to imaging (ultrasound/x-ray, CT) on a flexed and straight leg. The imaging can also be used to provide more detailed anatomical reference markers such as exact centerline and curvatures of the vessel that may then be implemented into the model. Exploration of how to incorporate buckling, torsion, and twisting motion is still under consideration and future work will investigate these issues. Nevertheless, predictions from the model suggest that lower extremity arteries undergo substantial axial shortening, and that stent and vascular graft designers should be cognizant of these biomechanical demands. This simulation provides a reference point for investigating the effects

of axial shortening due to joint mechanics as an addition to distension due to hemodynamics.

5. Conclusion

This study provided data for vessel shortening while the patient performs activities of daily living. These activities could vary for different populations (athletes, astronauts, elderly, unhealthy patients), and therefore could directly impact how the disease progresses. Through innovative software modeling and analysis, the shortening and elongation could be captured for the iliac and femoropopliteal vessels, compared to previous studies where a collection of images at one angle with one perspective was used to obtain measurements. The results were comparable for the femoropopliteal position and can assist device design developers to accurately accommodate arterial mechanical behavior on vascular implants.

Conflict of interest

All authors listed on the title page were working in the Biomedical Engineering Department of the Cleveland Clinic when they participated in the research work described in this manuscript. There is no conflict of interest from any of the authors.

Acknowledgments

The National Science Foundation funded this work through the REU mechanism: Undergraduate Engineering in Medical Research Grant Award EEC 0244048. The authors are grateful to Peter Loan (MusculoGraphics, Inc.) for technical assistance in the adaptation of SIMM for this model. Supported in part by the National Institutes of Health, National Center for Research Resources, CTSA KL2RR024990, Cleveland, Ohio.

References

Andriacchi, T.P., Alexander, E.J., 2000. Studies of human locomotion: past, present and future. *Journal of Biomechanics* 33, 1217–1224.

Ballyk, P.D., Walsh, C., Butany, J., Ojha, M., 1998. Compliance mismatch may promote graft–artery intimal hyperplasia by altering suture-line stresses. *Journal of Biomechanics* 31, 229–237.

Browse, N.L., Young, A.E., Thomas, M.L., 1979. The effect of bending on canine and human arterial walls and on blood flow. *Circulation Research* 45, 41–47.

Criqui, M.H., 2001. Peripheral arterial disease: epidemiological aspects. *Vascular Medicine* 6, 3–7.

Cheng, C.P., Choi, G., Herfkens, R.J., Taylor, C.A., 2010. The effect of aging on deformations of the superficial femoral artery due to hip and knee flexion: potential clinical implications. *Journal of Vascular and Interventional Radiology* 21 (2), 195–202.

Cheng, C.P., Wilson, N.W., Hallet, R.L., Herfkens, R.J., Taylor, C.A., 2006. In vivo MR angiographic quantification of axial and twisting deformations of the superficial femoral artery resulting from maximum hip and knee flexion. *Journal of Vascular and Interventional Radiology* 17, 979–987.

Choi, G., Cheng, C.P., Wilson, N.M., Taylor, C.A., 2009a. Methods for quantifying three-dimensional deformation of arteries due to pulsatile and nonpulsatile forces: implications for the design of stents and stent grafts. *Annals of Biomedical Engineering* 37 (1), 14–33.

Choi, G., Shin, L.K., Taylor, C.A., Cheng, C.P., 2009b. In vivo deformation of the human abdominal aorta and common iliac arteries with hip and knee flexion: implications for the design of stent-grafts. *Journal of Endovascular Therapy* 16, 531–538.

Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E., Thelen, D.G., 2007. OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE Transactions on Biomedical Engineering* 54 (11), 1940–1950.

Delp, S.L., Loan, J.P., 1995. A graphics-based software system to develop and analyze models of musculoskeletal structures. *Computers in Biology and Medicine* 25, 21–34.

Giannattasio, C., Failla, M., Emanuelli, G., Grappiolo, A., Boffi, L., Corsi, D., Mancina, G., 2001. Local effects of atherosclerotic plaque on arterial distensibility. *Hypertension* 38, 1177–1180.

Gleason, R.L., Gray, S.P., Wilson, E., Humphrey, J.D., 2004. A multi-axial computer-controlled organ culture and biomechanical device for mouse carotid arteries. *Journal of Biomechanical Engineering* 126, 787–795.

Iida, O., Nanto, S., Uematsu, M., Morozumi, T., Kotani, J., Awata, M., Onishi, T., Ito, N., Sera, F., Minamiguchi, H., Akahori, H., Nagata, S., 2006. Effect of exercise on frequency of stent fracture in the superficial femoral artery. *The American Journal of Cardiology* 98, 272–274.

Mironov, V., Kasyanov, V., McAllister, K., Oliver, S., Sestino, J., Markwald, R., 2003. Perfusion bioreactor for vascular tissue engineering with capacities for longitudinal stretch. *Journal of Craniofacial Surgery* 14, 340–347.

Scheinert, D., Scheinert, S., Sax, J., Piorkowski, C., Bränlich, S., Ulrich, M., Biamino, G., Schmidt, A., 2005. Prevalence and clinical impact of stent fractures after femoropopliteal stenting. *Journal of the American College of Cardiology* 45, 312–315.

Seliktar, D., Nerem, R.M., Galis, Z.S., 2003. Mechanical strain-stimulated remodeling of tissue-engineered blood vessel constructs. *Tissue Engineering* 9, 657–666.

Smouse H.B., Nikanorov A., LaFlash D., 2004. Changes in major peripheral arteries during joint movement before and after stent placement in the cadaver model. In: Program and Abstracts, Transcatheter Cardiovascular Therapeutics, September 27–October 4, 2004, Washington, DC.

Smouse, H.B., Nikanorov, A., LaFlash, D., 2005. Biomechanical forces in the femoropopliteal arterial segment. *Endovascular Today* June, 60–66.

Stagni, R., Fantozzi, S., Cappello, A., Leardini, A., 2005. Quantification of soft tissue artifact in motion analysis by combining 3D fluoroscopy and stereophotogrammetry: a study on two subjects. *Clinical Biomechanics* 20, 320–329.

Vernon, P., Delattre, J.F., Johnson, E.J., Palot, J.P., Clément, C., 1987. Dynamic modifications of the popliteal arterial axis in the sagittal plane during flexion of the knee. *Surgical Radiologic Anatomy* 9, 37–41.

Wensing, P.J., Scholten, F.G., Buijs, P.C., Hartkamp, M.J., Mali, W.P.T.M., Hillen, B., 1995. Arterial tortuosity in the femoropopliteal region during knee flexion: a magnetic resonance angiographic study. *Journal of Anatomy* 186, 133–139.