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A Pragmatic Approach to Understand Peripheral Artery Lumen Surface Stiffness Due to Plaque Heterogeneity

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Abstract

The goal of this study was to develop a pragmatic approach to build patient-specific models of the peripheral artery that are aware of plaque inhomogeneity. Patient-specific models using element-specific material definition (to understand the role of plaque composition) and homogeneous material definition (to understand the role of artery diameter and thickness) were automatically built from intravascular ultrasound images of three artery segments classified with low, average, and high calcification.

The element-specific material models had average surface stiffness values of 0.0735, 0.0826, and 0.0973 MPa/mm, whereas the homogeneous material models had average surface stiffness values of 0.1392, 0.1276, and 0.1922 MPa/mm for low, average, and high calcification, respectively.

Localization of peak lumen stiffness and differences in patient-specific average surface stiffness for homogeneous and element-specific models suggest the role of plaque composition on surface stiffness in addition to local arterial diameter and thickness.

Keywords

Peripheral Artery Disease; Finite Element; IVUS; Patient-Specific; Surface Stiffness; Plaque

Introduction

Peripheral artery disease (PAD) affects about 200 million individuals worldwide, almost 14 million in the Americas alone (Fowkes et al. 2013). Atherosclerosis, artery narrowing due to plaque buildup, is the leading cause of PAD in patients over the age of 40 (Creager & Loscalzo 2012). If untreated, PAD patients are at risk for developing critical limb ischemia,

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Conflict of Interest

AE owns innodof, LLC, a consulting company for modeling and simulation. Other co-authors do not have conflicts of interest to declare.

due to obstructed blood flow to the legs and arms, which may lead to amputation (25% of cases) (Creager, M 2003). Treatment of PAD includes mechanical interventions, such as balloon angioplasty and stent placement, to improve the degree of stenosis. Performance of these interventions rely on the mechanics of the arterial wall they interact with and therefore likely depend on the local surface stiffness at the lumen. Patient-specific properties of the artery (gross geometry, e.g. lumen diameter and thickness, and plaque heterogeneity) will likely dictate the apparent stiffness on the lumen surface and may indicate patient-specific variations in response to angioplasty and stent placement. Computational modeling, e.g., finite element analysis, may be used as a tool to estimate the surface mechanical properties of arteries in a patient-specific manner. With individualized models, design and selection of devices and solutions for treatment can be tailored to the specifics of the patient. More specifically, a surface stiffness metric will potentially assist in stent selection by quantifying the necessary radial stiffness of the stent, which can be calculated from the patient-specific finite element model, without having to simulate expansion of each stent within the artery.

Intravascular ultrasound (IVUS) imaging is used to render patient-specific arterial geometry for scientific and clinical analysis (Vukicevic et al. 2014; Majdoulina et al. 2014; Timmins et al. 2016). Plaque classification can also be extracted from IVUS images using advanced radio-frequency image processing techniques, known as IVUS virtual histology (VH) (Nair et al. 2002). These virtual histology (IVUS-VH) images provide color-coded maps of varying plaque types, including fibrous, fibro-fatty, adipose, and calcified plaques (of different stiffness respectively (Hiro et al. 1997)). Such information can be used in computational modeling of arterial plaque and has been ubiquitous, commonly relying on explicit description of plaque geometry (Guo et al. 2017). Explicit geometric description of plaque may not necessarily be feasible for highly inhomogeneous cases. Even when capturing such geometric detail is possible, modeling the intricate internal detail of the inhomogeneous plaque will require dense finite element meshes even when it is possible with tetrahedral meshing. This strategy was used by Brands et al. (Brands et al. 2008), but was only computationally feasible through implementation of parallel processing. Studies investigating plaque vulnerability have also used IVUS-VH to capture atherosclerotic plaque geometry in computational models of the coronary artery (Yang et al. 2009; Teng et al. 2014). While these investigations modeled plaque geometry explicitly, as we noted earlier, such an approach may be infeasible when plaque distribution is highly heterogeneous and representation of plaque boundary may be problematic. In addition, the model generation process was semi-automated, requiring manual user adjustments and the computational cost of running these models was not provided and may not be clinically acceptable. Development of patient-specific models may largely benefit from alternative strategies where segmentation and explicit geometric representation of plaque can be avoided and lumen surface stiffness can be represented with a decreased computational cost. Such an approach will be instrumental when simulation-based recommendations are needed within clinical time frames.

A limitation of IVUS-VH is that it cannot capture artery curvature or alignment as a stand-alone imaging dataset. Angiographic imaging has been used to capture the three-dimensional path of the IVUS probe so that image registration can be performed and the three-dimensional arterial geometry can be accurately reconstructed (Evans, JL et al. 1996;

Wahle et al. 1999; Slager et al. 2000; Schuurbiers et al. 2009; Vukicevic et al. 2014). However, association of angiography and IVUS-VH is not trivial. Temporal synchronization to ensure the starting position of the IVUS probe in the angiography image and three-dimensional spatial registration for reconstruction of artery geometry require additional setup, data collection, and/or post processing, which increases the time and effort required to build a computational model and launch a simulation. While several research groups have worked toward development of software tools to implement the post-processing process (Bourantas et al. 2008; Doulaverakis et al. 2013; Carlier et al. 2014), the tools are not fully automated, require manual intervention, and lack detailed registration of plaque placement relative to artery curvature. The focus of this study is to build peripheral artery models to describe the arterial mechanical environment in a clinically applicable time frame. Therefore, the goals of this study include automatic development of computationally efficient artery models from limited clinical data (i.e. only IVUS-VH images) and demonstration of surface stiffness as a potential metric of mechanical performance for patients with varying plaque compositions.

Material and Methods

Overview

IVUS-VH images were used to develop the gross geometry (local diameter and thickness) for patient-specific finite element models of the superficial femoral artery (SFA) (Figure 1). Element-specific material properties were defined using a weighted average of material property coefficients associated with the pixel composition of the IVUS-VH images. This strategy implicitly represents the influence of plaque inhomogeneity (and geometry). Models of arteries with varying calcified plaque composition (low, average, high) were developed and evaluated using luminal surface stiffness. All procedures followed were in accordance with the ethical standards of the Cleveland Clinic institutional review board and with the Helsinki Declaration of 2013. Informed consent was obtained from all patients for being included in the study.

IVUS-VH Imaging

Superficial femoral arteries (SFA) from 3 subjects, diagnosed with PAD, were imaged using an IVUS-VH system (Volcano Corp, Rancho Cordova, CA) (Table 1). Image segments, approximately 25 mm in length, of the SFA were acquired at the same blood pressure (between diastole and systole), gated by electrocardiogram. The pullback rate was set to 0.5 mm/s and the spacing between images was recorded (spacing ranged from 0.3–0.9 mm). IVUS-VH images overlay IVUS images with label maps of different material components within the SFA [arterial wall: gray, plaque: white (calcified), dark green (fibrous), light green (fatty), and red (adipose)]. The image field-of-view was 10 mm, with an image size of either 328×328 (low) or 400×400 (average, high) pixels, giving resolutions of 0.0305 and 0.025 mm/pixel, respectively. Example subsets of IVUS-VH images corresponding to each plaque composition group are shown in Figure 2. For convenience, the three arteries were ordered according to the averaged percentage (by cross sectional area) of calcified plaque within the imaged arterial segment.

Automated Model Generation

A series of Python scripts were used to automate the model generation process. To define the mesh, the inner and outer boundaries of the artery were extracted from the IVUS-VH images, using a raster scanning method (Figure 1). Boundary smoothing was initiated by the user, by specifying the number of boundary pixels to skip (e.g. 6, in this study). Next, a structured cylindrical hexahedral mesh was generated using the boundaries at a user-specified mesh density (.inp: input file for Abaqus, Simulia Corp., Providence, RI, USA). The mesh density was defined by the number of in-plane (xy) elements (radial x angular) and the number of elements between out-of-plane (z) slices (Figure 3). For re-sampling, the coordinate system of the inner and outer boundary points were translated to the average lumen centroid axis. The points were then re-sampled, based on the number of angular elements, about the average lumen centroid axis. Interior nodes (between inner and outer nodes) were determined by uniformly interpolating for the specified number of radial elements. This raw mesh was used to define element-wise material properties. Alignment is not inherently captured by IVUS imaging, and therefore, it was assumed to fall into one of three categories: no alignment (probe moves along a straight line), lumen centered (probe moves along the axis of the lumen), or wall centered (probe moves along the axis of the outer arterial wall). Since three-dimensional reconstruction may not be possible with IVUS imaging alone, these three alignment options were used to examine the model sensitivity to variations in alignment. A filtered mesh was also created by averaging the inner and outer nodes, respectively, for a user-specified number of neighboring slices (e.g. 5, in this study) to smooth arterial geometry generated between slices.

Material properties for each component of the model (artery and plaque types) were defined as hyperelastic and incompressible (Table 2). Determination of plaque properties, even in highly controlled in vitro testing, can be difficult due to separating of plaque types within each artery for sample preparation. Therefore, material properties of different plaque compositions were taken from literature. Arterial wall properties were described with the reduced form Mooney-Rivlin (special case 2nd order polynomial) coefficients of the human femoral artery presented by Prendergast et al (Prendergast et al. 2003). Material properties for each plaque type were determined by finding 3rd order reduced polynomial coefficients to fit experimental compression data presented by Hiro et al. (Hiro et al. 1997). Element-wise material properties were determined by the weighted average of the coefficients associated with the IVUS-VH color codes of the pixels lying within (or on the boundary of) each element. Finally, each end of the artery was constrained to contact a rigid plane restricting normal motion of the nodes relative to the plane but allowing frictionless in-plane motion. One rigid plane was fixed in place in all directions and the other was set free to move in the z-direction (Figure 4).

Simulations

Simulations using the automatically generated patient-specific arteries were performed in Abaqus (6.7–1, Standard, 4 CPUs) (Simulia Corp., Providence, RI, USA). A pressure of 120 mmHg (0.016 MPa) was applied to the luminal surface, with output variables requested every 10 mmHg. This pressure level allows us to expand past the physiological range of blood pressure in order to capture the mechanical loading levels that may be seen during

balloon expansion or stent deployment at the initial contact with the lumen surface. Different pressure levels can be used if desired. Yet, additional properties of the specific balloon or stent are required to more accurately investigate the stiffness curve that represents the arterial mechanical environment during complete deployment, however once known, the representative pressure may be implemented into the simulation as an updated lumen surface load. Surface stiffness was defined at each node as the change in pressure (MPa) divided by the magnitude of change of local radius (mm) between the two final increments of the simulation. The local radius at each node was calculated by first fitting a circle, using least squares, to the location of nodes in each z-slice to find the centroid, which was then subtracted from the in-plane coordinates node of interest. Mesh densities of 2×32 , 4×64 , and 8×128 (# radial $\times$ # angular) were used to determine the appropriate mesh to balance surface stiffness accuracy and computational time (mesh convergence) (Figure 3). Relative percent error in surface stiffness was used as a metric to determine accuracy between mesh densities. Mesh convergence was performed using each of the three models of various plaque compositions (“no alignment” models). This mesh convergence analysis not only indicated adequate representation of gross artery anatomy but also minimum required mesh size to determine the emergent mechanical response due to plaque inhomogeneity.

The SFA segments were classified by the amount of densely calcified plaque observed in the respective IVUS-VH images as: low, average, and high. One model representation (4×64 mesh) in each category was modeled with both heterogeneous and homogeneous material definitions (“no alignment” case). Heterogeneous models were used to demonstrate the combined effect of plaque composition and arterial geometry (diameter and thickness) on surface stiffness, while homogeneous models were used to examine the effects of local arterial diameter and thickness on surface stiffness. Homogeneous models utilized the arterial wall properties assigned to the whole model geometry. Simulations with heterogeneous models with 4×64 mesh density were conducted for all longitudinal alignment cases (“no alignment”, “lumen centered”, “wall centered”) in order to evaluate the role of alignment assumption on predictive capacity of the models.

Results

Mesh Density Convergence of Surface Stiffness

The luminal surface stiffness, from 110 mmHg to 120 mmHg, was used as a convergence metric for the three selected mesh densities. As the mesh density increases, the mesh resolution gets closer to VH resolution therefore becoming equivalent to explicit modeling of each pixel of plaque. As the mesh size is increased the contributions of each plaque pixel to apparent properties should stabilize at this limit case, if not earlier. The statistics relating successive mesh densities, including average error, standard deviation of error, and maximum error are included in Table 3. In addition, computational cost associated with each mesh density (2×32 , 4×64 , and 8×128) is given in Table 4.

Plaque Composition Effect on Surface Stiffness

Von Mises stresses at a corresponding blood pressure (120 mmHg) for the three different models (“no alignment”) exhibiting low, average, and high compositions of dense calcium

are shown in Figure 5. Material composition break-down, including arterial wall and all plaque types, varied across the three arteries (Figure 6). The average lumen diameter was 4.43 ± 0.15 mm, 4.55 ± 0.16 mm and 3.45 ± 0.21 mm for the low, average, and high model, respectively. The average thickness of combined plaque and arterial wall was 0.75 ± 0.15 mm, 0.57 ± 0.17 mm, and 0.75 ± 0.21 mm for the low, average, and high model, respectively.

Arterial surface stiffness maps, calculated from the final step of the simulation (110 mmHg to 120 mmHg), for each artery model are shown in Figure 7 & Figure 8. The local stiffness metric was not sensitive to the alignment option of the IVUS images (Figure 7). Using the models defined by element-specific averaged material properties, arterial average surface stiffness was 0.0735, 0.0826, and 0.0973 MPa /mm for low, average, and high calcification, respectively (“no alignment” models). Using the model geometries defined by homogeneous material properties, arterial average surface stiffness was 0.1392, 0.1276, and 0.1922 MPa/mm for low, average, and high calcification, respectively. Location of peak surface stiffness on the lumen were different when the same artery was modeled as homogeneous or as heterogeneous (incorporating plaque composition) (Figure 8). Variations in surface stiffness (average across lumen circumference at a given slice), plaque composition, initial thickness and initial diameter are observed along the length of the artery (Figure 9).

Discussion

The mechanical response of a stent is dependent on its mechanical environment, which is likely to be dictated by the arterial surface stiffness. Specimen-specific models were generated in this study to investigate how differences in arterial morphology and plaque composition affect its surface stiffness. We improved on the original surface stiffness metric that was defined in our previous work (Bennetts et al. 2013; Bennetts, CJ et al. 2013), which did not separate tissue deformation from rigid body motion and therefore the metric was highly sensitive to IVUS alignment (results not shown). By eliminating the contribution of rigid body motion from the surface stiffness metric defined in this study, we were able to minimize the variation in local stiffness across different alignment options (Figure 7). While this study introduces a new metric to examine arterial mechanics, previous studies have shown that plaque composition, distribution, and morphology effect stent performance (Timmins et al. 2008; Pericevic et al. 2009; Zhao et al. 2012; Gökgöl et al. 2017; Wei et al. 2018). If the surface stiffness metric is shown to correlate to stent performance, the need to accurately model the stent and stent-artery interactions, which increase computational cost, may not be necessary. Current regulatory standards indicate that radial stiffness and strength be reported for balloon expandable stents, while radial outward force should be reported for self-expanding stents (FDA 2010). The artery surface stiffness metric may be used in conjunction with these stent performance metrics, which are already reported, to assist in the stent selection and placement process for each clinical scenario. The relative surface stiffness changes, observed in this study, provide an understanding of the artery properties that may have a significant impact on stent performance, giving insight and motivation for future stent deployment and fatigue simulations.

Model convergence was determined using the surface stiffness as a comparative metric between mesh densities. The mesh convergence not only addressed the traditional discretization problem in finite element analysis but in this case, it also provided the required resolution to capture inhomogeneity of artery as it influences the mechanical response at the lumen surface. The average percent error in luminal surface stiffness between the 2×32 and 4×64 meshes was 3.4% (Table 3). This error in mechanical response suggests that the 2×32 mesh density may not be capturing the influence of inhomogeneity in full, so the relative response of the 4×64 and 8×128 meshes were determined. The average error between the denser meshes dropped to 1.0% (Table 3), suggesting that the 4×64 element mesh was sufficient to represent the luminal surface stiffness of these artery segments. In addition, all measures of computational cost were reduced by at least 73% when comparing the 4×64 element mesh to the 8×128 element mesh (Table 4). The average time for computation under 2.5 minutes for the 4×64 element mesh density (Table 4) is an indicator of the approach's computational feasibility to convert this model into a tool for potential clinical use. Being mindful to keep computational cost at clinically relevant levels and also acknowledging the low level of discrepancies between the results of the 4×64 and 8×128 mesh sizes, we utilized 4×64 mesh size as a practical and reliable compromise. In the literature, patient-specific diseased arteries are often modeled without the conscious effort to minimize computational cost, therefore wall clock time is not reported (Brands et al. 2008; Teng et al. 2014). The mesh size in this study is smaller than previous studies (measured by number of equivalent nodes for a 25 mm segment) (Brands et al. 2008; Teng et al. 2014), however mesh convergence was still observed with the implemented material averaging technique. If desired, the automated model generation process facilitates implementation of increased mesh densities, however increases should be aware of image resolution, i.e. further increasing mesh density therefore decreasing element edge length beyond voxel resolution may not provide additional gains.

Surface stiffness quantification between arterial segments exhibiting various plaque compositions can provide insight into the relative stent response for different patients. Intuitively, it seems plausible that plaque composition classified as high results in a stiffer arterial surface response. The average surface stiffness trends of the models defined with element-specific averaged material properties followed this intuition, showing a qualitatively positive correlation between average surface stiffness and dense calcium plaque level. However, this was not the case when looking at the three different segments of the homogeneous material models in this study. The model with low calcified plaque levels exhibited a higher average surface stiffness than the segment rated with average calcified plaque levels. While the lumen diameters of the low and average calcified plaque models were similar, the average thickness was larger in the low calcified model, suggesting the role of arterial thickness on arterial surface stiffness. The average surface stiffness for the homogeneous models showed a different trend than the element-specific models, suggesting that surface stiffness is dependent on relative contributions from arterial plaque composition. This confirms that metrics of material composition and distribution, as well as arterial geometry are all important factors to consider when defining arterial surface mechanics. Akyildiz et al. found similar results when examining homogeneous and heterogeneous two-dimensional models of arterial intima for insight into peak cap stress, which provides an

indication of plaque vulnerability (Akyildiz et al. 2018). Further, this conclusion indicates the need for patient-specific arterial representations to aid in selection of intervention devices, whose performance are likely to be determined by lumen surface stiffness.

Localized surface stiffness may be more indicative of the mechanical environment within the artery, providing more information on stresses placed on an implanted stent. The location of peak surface stiffness changes between the heterogeneous and homogeneous models, potentially causing different interpretation of high stress regions if one does not account for the contribution of plaque composition on arterial mechanics. In addition, the cross-section distribution of surface stiffness at the location of peak surface stiffness varies across arteries. The low and average arteries show a more homogeneous distribution along the circumferential direction (indicated by vertical bands), whereas the highly calcified artery demonstrates a more heterogeneous distribution (Figure 7). In the heterogeneous situations, the stent would likely be anchored at the stiff location(s) or edge(s) and tissue bulging during stenting may occur on the less stiff regions of the artery. Using a simplified model of arterial plaque, Morlacchi et al. found that localized calcifications impede the stent from expanding fully and therefore shift the location of the peak stress within the stent (Morlacchi et al. 2014). In addition, Wei et al. reported negative effects on stent performance for arteries exhibiting large plaque eccentricity or complex plaque composition, however there was not a focus placed on local stress or stiffness to examine stent anchoring (Wei et al. 2018). Other potential local stiffness scenarios include peak stiffness offsets and variations along the length (z) of the artery, which may introduce torsional or bending loads on the implanted stent, potentially increasing the risk of stent fracture or artery rupture. Surface stiffness is expected to be related to artery geometry (initial diameter and thickness) and plaque composition. Finite element representation of the artery essentially provides this relationship based on mechanical principles. There is also the possibility to establish statistical correlations between surface stiffness and artery anatomy and plaque composition. For all three arteries in this study, the initial diameter and fibrous plaque composition were strongly correlated with the surface stiffness (results not shown) and similar correlations may also be inferred from Figure 9. Yet, analysis of additional arteries will be necessary to provide a better understanding of which arterial characteristics dominantly contribute to surface stiffness and potentially stent performance. With this added information, the physician may be able to anticipate such issues and modify the treatment strategy or the implanted devices.

Several features should be implemented in the future to account for some limitations of this first generation patient-specific models. In this study, we assumed that all image slices were parallel and therefore no rotational realignment was performed within the planes. While this assumption may be accurate for low curvature regions of the peripheral arteries, it may not provide the appropriate three-dimensional structure for modeling of areas of high curvature (i.e. flexed leg position, distal portion of femoropopliteal artery) (Wood et al. 2006; Ghriallais et al. 2016). Although curvature was not captured for the arteries in this study, future work may incorporate simultaneous biplane fluoroscopy or angiography scans (Evans, JL et al. 1996; Schuurbiers et al. 2009; Doulaverakis et al. 2013; Vukicevic et al. 2014) to capture three-dimensional positioning of the artery, creating a more accurate spatial representation in the models. The multi-step registration process requires accurate transformations of the biplane angiography 2D positions to 3D space, completed by imaging

a calibration cube (with steel markers of known spatial location) from both perspectives (MacKay et al. 1982; Evans, JL et al. 1996). After calibration, the IVUS catheter path is selected in both views of the angiography (manual or automatic depending on image quality). Next, IVUS-VH images are rotated to be normal to the catheter path. In-plane rotation may also need to be determined to account for rotation of the IVUS probe during image acquisition (Wahle et al. 1999). Mathematical methods, using maximum likelihood and neighborhood similarity estimator optimizations, may also be used to register the vessel data spatially to account for motion between the vessel and catheter (Talou et al. 2017). Carlier et al. presents a real-time registration approach aimed for clinical use, however the rotational relationship between the plaque and curvature was not included due to the demanding nature of the post-processing and contour analysis workflow (Carlier et al. 2014). Due to the pragmatic focus of this study and generally low curvature observed in the peripheral arteries, we chose to neglect representation of artery curvature. In future studies, curvature variations may be explored further, using a three-dimensional registration procedure, to examine its effect on surface stiffness.

In addition, when IVUS images were captured, the artery was experiencing a degree of internal pressure somewhere between diastole and systole, creating initial mechanical strain that was ignored in the initial geometric state of the model (Wang et al. 2017). While the stiffness values reported in this study can be compared relative to each other (and can be considered as a biomechanical marker), *in situ* strains need to be determined experimentally and implemented to provide a more accurate stiffness of the luminal surface during simulation. Similarly, further experimentation is needed to determine the validity of the surface stiffness metric, which can be calculated using diameter measurements along the length of in vivo or in vitro arteries, e.g. pressurized at different levels (Vonesh et al. 1997). Finally, the material properties were defined from literature and therefore did not account for potential patient-specific outliers. In future studies, this pragmatic modeling framework may be used to perform inverse finite element analysis to investigate the variability of arterial wall and plaque material properties, further enhancing the patient-specificity of the model. Such investigations have been performed to obtain local artery properties using in vitro pressurization experiments (Vonesh et al. 1997) and can be expanded to in vivo studies when IVUS images are acquired for different pressure levels.

A fully automated patient-specific artery modeling strategy was developed in this study, utilizing IVUS-VH imaging for geometry and plaque composition definitions. In addition, a lumen surface stiffness metric was introduced to provide insight into the geometric or material properties that contribute to variations in the artery surface mechanics, which may be correlated to stent performance. Our pragmatic approach for quantification of *in silico* artery mechanics has the potential to be developed into a standardized translational tool for PAD, designed to assist in patient care and device selection. Since each model is built from patient-specific IVUS data, development of a tool to provide personalized selection of the appropriate intervention could be useful and may decrease the number of stent failures in PAD, especially for patients with unique disease patterns (i.e., extreme areas of calcification, large/small vessel diameters, etc.). Simulations of stent deployment and fatigue can be performed to evaluate potential failures based on local stress concentrations. Ultimately, the

goal is to transform these models from their current research role into clinical tools that can be used to assist physicians in patient assessment and personalized care.

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Figure 1. Flowchart of the model development process: A) IVUS imaging, B) boundary extraction, C) material mesh definition, and D) assigning material properties.

Figure 2.
Sample set of IVUS-VH images in arteries for low, average, and high calcified plaque compositions.

Figure 3.
Artery meshes of varying in-plane density.

Figure 4.
Finite element model with constraints and boundary conditions.

Figure 5. Von Mises stress distributions in FE models for low, average, and high calcified plaque compositions for the material-specific FE models (top row) and homogeneous FE models (bottom row). Results are for “no alignment” case.

Figure 6.
Composition breakdown of artery materials (wall and plaque).

Figure 7. Surface stiffness (MPa/mm) from element-specific FE models for arterial segments with low, average, and high density calcification (* = location of peak surface stiffness). The lumen surface is unwrapped for visualization purposes. Columns show the three alignment options of the IVUS images.

Figure 8.

Normalized surface stiffness (by average stiffness) from heterogeneous and homogeneous FE models for arterial segments with low, average, and high density calcification (* = location of peak normalized surface stiffness). Results are for “no alignment” case. The lumen surface is unwrapped for visualization purposes. Heterogeneous models had average surface stiffness values of 0.0735, 0.0826, and 0.0973 MPa /mm, whereas the homogeneous material models had average surface stiffness values of 0.1392, 0.1276, and 0.1922 MPa/mm for low, average, and high calcification, respectively.

Figure 9.

Measurements of average surface stiffness, material composition, initial diameter, and initial thickness (top to bottom) along the length of the arteries (right to left: low, average and high calcification).

Table 1.

Subject demographics for low, average and high peripheral arteries.

Subject Calcification	Age (years)	Gender	Race
Low	58	Female	Caucasian
Average	73	Female	Caucasian
High	56	Female	African American

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Table 2.

Arterial wall and plaque material coefficients.

	C₁₀ (MPa)	C₀₁ (MPa)	C₂₀ (MPa)	C₁₁ (MPa)	C₃₀ (MPa)
Arterial Wall	18.9 E-03	2.75 E-03	590 E-03	857 E-03	0
Calcified	2.49 E-03	0	16.4 E-03	0	6.98 E-03
Fibrous	1.41 E-03	0	7.34 E-03	0	5.83 E-03
Fatty	1.55 E-03	0	9.29 E-03	0	1.1 E-03
Adipose	0.70 E-03	0	2.16 E-03	0	−0.45 E-03

**

Note: The remaining coefficients were set to a value of zero.

Table 3.

Percent error in luminal surface stiffness among simulated arteries.

	Average error (mean $\pm$ SD)	Maximum error (mean $\pm$ SD)
2 $\times$ 32 vs. 4 $\times$ 64	3.4 $\pm$ 1.5 %	11.4 $\pm$ 2.3 %
4 $\times$ 64 vs. 8 $\times$ 128	1.0 $\pm$ 0.3 %	5.8 $\pm$ 0.3 %

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Table 4.

Computational cost (Intel Core 7, 3.4 GHz, 4 CPUs) among simulated arteries.

Mesh size	Minimum Memory Required (MB)	Memory to Minimize I/O (MB)	Wall Clock (seconds)	Number of Nodes
2×32	25 ± 4.4	78 ± 27	22 ± 13	9631 ± 3818
4×64	65 ± 28	297 ± 151	144 ± 92	32756 ± 13743
8×128	243 ± 140	1491 ± 874	1877 ± 1842	122143 ± 51920