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Influence of isotropic and anisotropic material models on the mechanical response in arterial walls as a result of supra-physiological loadings

Thomas Schmidt [a], Devdatt Pandya [c], Daniel Balzani [b]

[a] Institut für Mechanik, Fakultät für Ingenieurwissenschaften, Abteilung Bauwissenschaften, Universität Duisburg-Essen, Universitätsstr. 15, 45141 Essen, Germany

[b] Institut für Mechanik und Flächentragwerke, Fakultät Bauingenieurwesen, Technische Universität Dresden, 01062 Dresden, Germany

[c] University of Duisburg-Essen, Computational Mechanics, Essen, Germany

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ABSTRACT

As accepted in the literature, arterial tissues have in principle anisotropic material properties. Although some very special situations in arteries exist where isotropic constitutive models may approximate the real material behavior with sufficient accuracy, the larger part of analyses requires an anisotropic model. In particular for overstretched arteries, as e.g. a result of a balloon angioplasty, an accurate representation of the complex softening phenomena is important and then the consideration of anisotropy may be necessary. However, a variety of publications found in the literature, where such supra-physiological loading situations are analyzed to optimize e.g. stent designs, consider isotropic models. Therefore, in this contribution, the response of an isotropic and an anisotropic material model is compared in numerical calculations where arteries are subjected to supra-physiological loading. The constitutive formulations include the typical nonlinear stiffening of the fiber response as well as softening due to microscopic damage. In detail, the isotropic and the anisotropic model are adjusted to the same experimental stress–stretch curves of different arterial layers and then both models are applied to finite element simulations of overstretched arterial walls. As it turns out a significant difference is obtained for both calculations showing the importance of anisotropic models for these loading situations.

1. Introduction

Arterial tissue can in general be interpreted as a composite material, wherein families of collagen fibers and smooth muscle cells are embedded in an isotropic ground matrix. The fibers are dispersed around preferred directions and of wavy structure. In the high loading range arterial tissues exhibit a characteristic stiffening (Roach and Burton [34], Samila and Carter [38]), which arises due to the successive recruitment of collagen fibers. The anisotropic properties of arterial tissues are mainly traced back to the anisotropic distribution of collagen. Patel and Fry [30] identified cylindrical orthotropic behavior of the arterial wall by inflating dog aortas. Weizsäcker and Pinto [45] conducted a similar study for cylindrical segments of carotid arteries of Wistar rats for a broader range of pressures and axial prestretches. The authors revealed, that the three incremental elastic moduli with respect to cylindrical coordinates are complex functions of axial strain and internal pressure. However, for a narrow range about an axial extension of 1.74 (recovering the measured *in situ* length at excision) and under physiological pressures, the incremental elastic moduli in radial, circumferential and axial direction took comparable values and the Poisson's ratios were close to 0.5. Therefrom, it was concluded, that for the range of deformations that occur *in vivo*, the behavior of arterial tissue may be approximated as incrementally isotropic. Similar results were obtained by Dobrin [14] for carotid arteries of dogs, whereby only biaxial anisotropy was investigated by identifying axial and circumferential incremental moduli, which were about equal at an axial stretch of 1.7. However, in the latter work it is noted, that the carotid artery is subject to distinct variations of its *in situ* length, and therefore also much shorter lengths occur, for which anisotropic behavior was identified (see [14] and references therein). Moreover, as argued in Zhou and Fung [48], biaxial isotropy cannot be deduced from identifying the same Young's modulus (E) in two perpendicular directions, since a further requirement is, that the shear modulus (G) is connected to the Young's modulus (E) and to the Poisson's ratio (ν) via the relation $G = E/(2 + 2\nu)$. In the latter work [48] a nonlinear strain-energy

E-mail address: t.schmidt@uni-due.de (T. Schmidt).

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function was adjusted to biaxial tension tests of radially cut thoracic aortas of mongrel dogs. Thereby, different material parameter values were identified in circumferential and axial direction under *in vivo* stress and strain values, and thus anisotropic material behavior was deduced for that loading range. Lally et al. [26] observed anisotropy in biaxial tests of non-*in situ* porcine coronary arteries, but suggested the existence of a unique prestretch value for each artery *in vivo* necessary to produce an isotropic response, which may explain the contradictory findings in the literature. Note that the above mentioned publications which encourage the usage of isotropic models are restricted only to the regime of physiological loadings.

Motivated by [14,45], isotropic strain-energies are used for the mechanical description of arterial walls, cf. Raghavan et al. [32], Baldewsing et al. [4], Lally et al. [25], Early et al. [15], Pericevic et al. [31], Auricchio et al. [2] or Conti et al. [11]. One of the first structural models is the one by Lanir [27] considering statistically distributed fibers, which are embedded in a continuous ground matrix. Structural models, which include anisotropy by considering aligned fibers in preferred directions were proposed by Wuyts et al. [47], Holzapfel et al. [22], Holzapfel and Gasser [21], Zullinger et al. [49], Balzani et al. [7,50], and many others. A variety of these models are formulated in the framework of structural tensors, cf. Spencer [42]. Gasser et al. [18] introduced an improved model, wherein the deformation measure is linearly interpolated between the states of isotropy and perfect fiber alignment by means of a dispersion parameter. Beside this analytical approach to fiber dispersion in soft tissues also numerical approaches (see e.g. Alastrué et al. [1]) have been proposed, which take into account a variety of fiber directions and integrate their response over a unit sphere. This method is however computationally more expensive.

In this work, supra-physiological loadings and their impact on the overall behavior are going to be analyzed by means of arterial wall simulations. Here, the notion *supra-physiological* is used to characterize situations where the arterial wall is subjected to internal pressures higher than the normal blood pressure. Thereby, a softening of the tissue is induced, which is believed to arise from (i) microscopic damage mainly occurring in the collagen fibers and (ii) loss of smooth muscle cells which become necrotic by a large part, cf. e.g. Dirsch et al. [13]. These supra-physiological loading states occur for example during clinical interventions such as balloon angioplasty. There, also the situation after the overinflation is of particular interest for the analysis of stents inserted in the artery. In order to optimize the above treatment methods, it is crucial to predict the behavior of the arterial tissue during these interventions. Therefore, a lot of research is performed towards predictive arterial wall simulations. In order to account for the softening, various models are proposed within a continuum damage mechanics framework (see e.g. Simo [41], Miehe [28] for first isotropic approaches in the large strain setting). Hokanson and Yazdani [20] modeled anisotropic damage in arteries by applying a fourth order damage tensor to an isotropic Ogden-type material. Balzani et al. [9] proposed an approach of modeling anisotropic damage in soft biological tissues by considering a decoupled representation of the strain-energy [22] and associating a one-dimensional damage variable to the transversely isotropic fiber energy. Rodríguez et al. [35] considered additionally damage in the matrix and proposed a statistical formulation for the damage evolution. Another approach for anisotropic damage is given by Volokh [43,44], who included energy limiters into anisotropic strain-energy functions. Dargazany and Itskov [12] described anisotropic damage in rubbers with help of a network evolution model. In Ehret and Itskov [16] an evolution of structural tensors was taken into account to model anisotropic damage in soft biological tissues. Alternative approaches for anisotropic damage in soft biological tissues, where the response of a one-dimensional formulation for the individual fiber response is integrated over the unit sphere, are for example given in Gasser [17], Sáez et al. [37] or Rebouah and Chagnon [33]. These approaches are however accompanied by increased numerical costs.

Based on different isotropic and anisotropic model formulations comparative studies regarding their influence on arterial wall mechanics are published in the literature, in particular concentrating on the physiological loading regime. A comparison between nonlinear isotropic constitutive models (with and without residual strains, respectively) and a linear anisotropic model was conducted by Williamson et al. [46] for a two-dimensional diseased arterial wall simulation up to pressures of 14.6 kPa (110 mmHg). Therein, it was concluded that all investigated models provide useful estimates, however deviations of approximately 30% between isotropic and anisotropic models were identified in high stress regions. The residual strains turned out to have only negligible effect on the nonlinear isotropic models. Residual strains, which are associated to residual stresses, are expected to reduce stress gradients and the magnitude of maximum stress in arterial walls, see e.g. Chuong and Fung [10]. Holzapfel et al. [24] investigated 3D stress states resulting from an anisotropic model which incorporates plastic strains and a Neo-Hookean model during balloon angioplasty and stent deployment. Thereby, stress deviations up to 600% were identified between both model responses. In Haughton and Merodio [19] and Rodríguez and Merodio [36] bifurcation of inflated thin-walled tubes under axial loading was examined with reference to the mechanical response of arteries for an isotropic and an anisotropic strain-energy function, respectively. Thereby, it was found, that the anisotropic mechanical response associated to the collagen fibers stabilizes the artery avoiding the bulging mode of bifurcation. Auricchio et al. [3] compared an isotropic model against and anisotropic hyperelastic model in a carotid artery stenting analysis and found similar stress patterns for both models, however, the high vessel straightening and discontinuity of the lumen area near the stent ends, as known from clinical practice, were better reflected by the anisotropic model.

None of the comparative studies given in the literature (to the best of the authors' knowledge) considers the softening observed during supra-physiological loadings. This however is essential if situations after a balloon angioplasty, including the analysis of stent performance, are to be investigated. Therefore, in this work, a nonlinear isotropic model is compared with a nonlinear anisotropic model after having applied supra-physiological loadings in three-dimensional arterial wall simulations. For this purpose the constitutive framework proposed in (Balzani et al. [6]) is considered, which incorporates remanent strains in the fibers as a result of microscopic damage due to supra-physiological loading. Both constitutive models are adjusted to the same set of experimental stress–stretch curves in circumferential and axial direction of several layers of human arteries. Here, the finite element solutions yield non-uniform distributions of mechanical fields throughout the arterial wall and thereby provide additional insight compared with various experimental studies, e.g. [14,45,48,26], where uniform stress and stretch values are assumed.

2. Continuum mechanical modeling

A continuum mechanical framework is used considering the deformation gradient $\boldsymbol{F}$ with $J = \det \boldsymbol{F} > 0$ and the right Cauchy–Green tensor $\boldsymbol{C} = \boldsymbol{F}^T \boldsymbol{F}$ as deformation measures, respectively. In order to account for deformation measures in specific preferred directions of collagen fibers, structural tensors of the type $\boldsymbol{M}_{(a)} = \boldsymbol{A}_{(a)} \otimes \boldsymbol{A}_{(a)}$ are considered. Hereby, $\boldsymbol{A}_{(a)}$ denotes a preferred direction describing fiber family a in the undeformed configuration. A detailed description of the concept of structural tensors is for