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Comparison of a histology based multi layer artery model to its simplified axisymmetric model.

Abstract: Arteries are vessel structures that serve vital function of transportation of blood to different parts of the body. Researchers have experimented with some approaches to model the arterial behaviour and to analyse its biomechanical properties. To analyse the in-vivo arterial properties, at Furtwangen University an inflatable sensor-actuator system is being developed, which provides the basis for a decision support system for vascular surgeons. The capabilities of this sensor shall be evaluated in simulations which requires appropriate modelling of the arteries. The inverse problem, i.e. how to efficiently identify arterial wall properties from sensor readings is targeted. A histology motivated 3D artery model was implemented in FEM using COMSOL (v5.5). The geometry of one model was based on a cross section of a real artery. The second model was axisymmetric and of equal dimensions with respect to volume, layer thickness etc. A biomechanical pressure-stretch analysis was performed applying an inflating pressure inside the walls of the vessels. Stretch in different areas of the first model was evaluated and the circumferential strain was compared to the axisymmetric model. The results show variation of strains within the segments of the first model of upto 10 percent. In addition, its outer wall circumferential stretch was found to be 10 percent lower compared to the axisymmetric setup. This comparison sheds light upon whether a simplification of arterial models is possible, without loss of accuracy in the context of the novel sensor evaluation. It provides useful information whether e.g. standardizing vessel structures to axisymmetric models will still provide results within allowable tolerance limits. Simulations proved useful to evaluate different vessel model formulations in the context of arterial diagnostics.

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Keywords: modelling, artery, identification, simulation.

https://doi.org/10.1515/cdbme-2021-2150

1 Introduction

Vascular surgeons find it challenging to determine the appropriate pressure/force to open a constricted arterial vessel. Force should be low enough to prevent the vessel from plastic deformation or rupture. Similarly, research pertaining to in-vivo biomechanics of arteries is lacking due to missing of an appropriate minimal invasive sensor technology. At two research institutes of Furtwangen University - 'Institute for Microsystems Technology' and 'Institute of Technical Medicine'- an intra-luminal sensor-actuator is being developed. It would provide pressure-extension dynamics of the peripheral arterial vessel in-vivo during a clinical procedure. By processing data from this intraluminal sensor, the biomechanical behaviour of the artery could be predicted (identified). This can ultimately provide additional insights that may simplify decision making for the surgeon [1].

Forward modelling of the artery is the 'forward identification' of the arterial biomechanical response. Prediction/ identification of the actual arterial condition from the sensor stretch-strain data, can be understood as reverse or inverse modelling.

Studies regarding modelling of the artery have gathered pace since 2 decades. Histology motivated multilayer-fibre based descriptions have been used to model the anisotropic nature of these vessels. Histological studies show that the arterial vessels may not only have some thickness variation within its biomechanically relevant layers (Media, Adventitia) but also have a multifold structure inside its lumen [2,3].

A simplified artery (healthy) geometry structure is an axisymmetric one. Each of the layers present in such an artery does not have any thickness variation. An ideal geometric prior like this would likely simplify the reverse identification procedure. Therefore, it is of interest to compare the biomechanics of such an ideal vessel geometry to its non-

axisymmetric counterpart. Using Finite Element Method(FEM) techniques and mathematical modelling, this paper aims at comparing the biomechanical response of both the cases.

2 Methods

COMSOL's (v5.5) software with structural and non-linear modules were used for the simulations. The subsequent sections explain how the 2 arterial models were generated and later simulated to obtain their pressure-stretch responses.
Gasser et al., described a constitutive modelling for the arteries considering distribution of fibres. Their equation is an extension from the Holzapfel Gasser Ogden (HGO) model for arteries [2,3]. The strain energy density equation defined by Gasser et al., was chosen as the mathematical equation for the modelling. It is defined as follows:

$$W = \mu(I_1 - 3) + \frac{k_1}{2k_2}\sum_{\alpha}\left(e^{k_2\left[\kappa(I_1)+(1-3*\kappa)\left(I_{fib}^{\alpha}\right)-1\right]^2} - 1\right) \qquad (1)$$

As described by Gasser et al., W is the strain energy density (see eq 1). μ is the isotropic part constant of the neo-hookean material. I_1 is the first invariant of the tensor. It is a trace of the Cauchy normal stresses on the tensor. κ is a structural constant. The fibre network consists of two families of fibres with material properties $k_1$and k_2. $k_1$is the stress parameter expressed in kPa while k_2 is a dimensionless parameter that influences primarily the exponential function. $(I_{\text{fib}}^{\alpha})$ is the invariant which described the stretch of the fibre family in consideration. The fibres contribute only when there is no buckling and only a positive stretch in the fibres. i.e. I_{fib}^{α} and are greater than 1.

Table 1: Gasser equation parameters for Adventitia and Media regions. The parameter values chosen for each layer are taken from literature and within range of values appropriate for the pressure range [4].

	μ (kPa)	k_1 (kPa)	k_2	κ	Fibre family 1(°)	Fibre family 2(°)
Media	30	8.5	1	0.2	40°	140°
Adventitia	15	5	50	0.2	40°	140°

A tube like model representing a section of the urethra is modelled and described in subsequent parts. The tube consists of 2 fibre families. 'α'=1,2 depict the two fibre families. Incompressibility condition was applied. The model described by Gasser et al., is shown to adhere to the polyconvexity and stability conditions. The parameters for the Gasser equation for the 'adventitia' and 'media' parts are given (see Table 1).

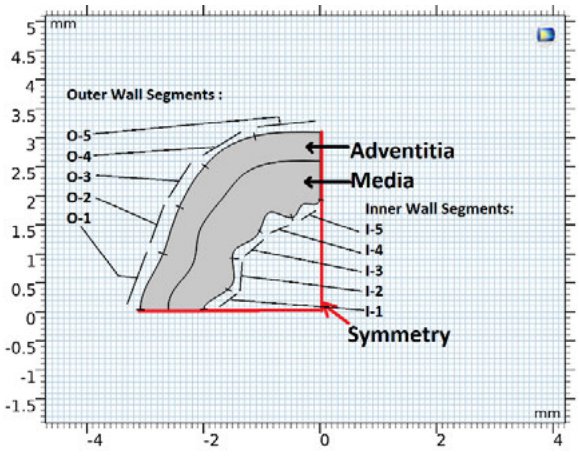

Figure 1: non-axisymmetric 'quarter' model indicating adventitia and media. I-1 to I-5 are inner wall segments while O-1 to O-5 are the outer wall segments. Symmetry planes indicated. Pressure is applied on the inner wall.

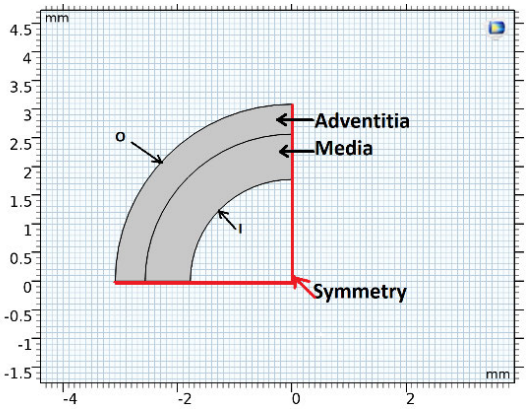

Figure 2: axisymmetric 'quarter' model indicating 'adventitia' and 'media'. 'I' is the inner wall segments while 'O' is the outer wall segment. Symmetry planes indicated. Pressure is applied on the inner wall.

To generate better convergence and less meshing issues, the models were created in 2D and extruded to 3D. Only a quarter of the model was simulated with necessary symmetry conditions (see Figure 1,2). The non-axisymmetric model was generated first with the biomechanically relevant media and adventitia (see Figure 1). It was based on histological descriptions and images of peripheral vessels. When computed, within this quarter, the hollow lumen occupied 2.48 mm^2, the media 2.68mm^2 while the adventitia measured 2.08 mm^2. The axisymmetric model was created with areas of layers that were equal as the non-axisymmetric model for each of the corresponding layer (see Figure 2). The radii of the innermost to the outermost boundary was calculated from the

areas and set to 1.77mm, 2.56mm and 3.09mm respectively. The length of the artery models for computation and relevance of fibre orientations was set to 0.5 cm. The fibres were oriented in the direction of the path of the lumen using 'diffusion method' and curvilinear description using the software.

An inflating ideal actuator (balloon) consideration was hypothesized. A pressure was thus applied on the inner boundary of the arterial model's lumen in steps of 0.1kPa for a range of 0kPa:21kPa. The stretch (current length divided by its original length) of the axisymmetric models inner and outer walls was evaluated for this pressure range. The non-axisymmetric model had its inner wall and outer wall divided into 5 equal-length segments (see Figure 1). The average stretch response for both the walls as well as for each individual segment was evaluated for the pressure range. These plots were compared, and an analysis was performed.

3 Results

The two models that were simulated based on the selected mathematical model and its parameterization according to Table 1 generated an hyperelastic response as expected for biological tissues like the arteries.

Both the inner and outer walls of both the models are compliant till 12kPa but later their response becomes relatively stiff (see Figure 3). The stretch for the inner wall is higher throughout the pressure range when compared to the outer wall. At applied pressure of 21kPa, a stretch of 1.54 was found for the 'N-AS' model, 1.72 for the 'AS' model while 1.28 for inner wall of both the 'AS' and 'N-AS' models. The total inner wall stretch response for both the models was quite similar. The outer wall stretch for the 'N-AS' model is stiffer around 10% when compared with the 'AS' model.

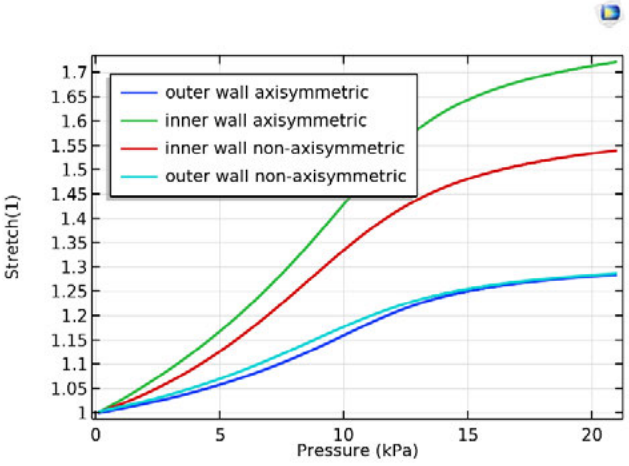

Figure 3: pressure-stretch graph for axisymmetric (AS) and non-axisymmetric (N-AS) models for their inner and outer walls. The inner wall is more compliant throughout the pressure range in comparison to the outer wall for both cases. The inner wall of AS model is more compliant compared to the N-AS model.

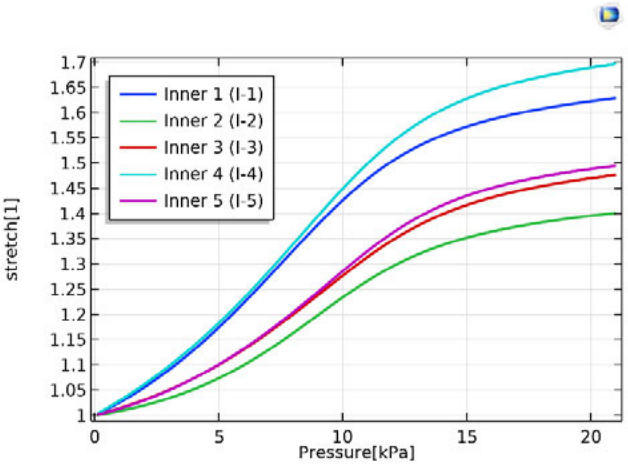

Figure 4: pressure-stretch response for segments of the inner wall in the non-axisymmetric (N-AS) model. The inner wall segments show a variation of stretch that ranges from 1.4 to 1.7 at pressure of 21 kPa.

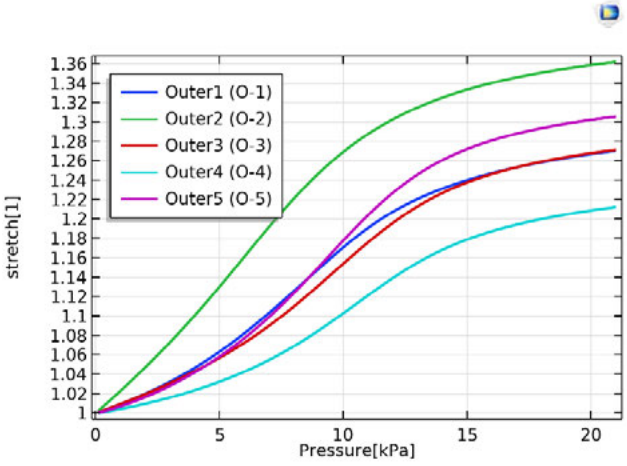

Figure 5: pressure-stretch response for segments of the outer wall in the non-axisymmetric(N-AS) model. The outer wall segments show a variation of stretch that ranges from 1.21 to 1.36 at pressure of 21 kPa.

While evaluating the pressure-stretch response for 'N-AS' model related to the segments within the inner (I-1 to I-5) and outer walls (O-1 to O-5), a variation is clearly visible from the outputs (see Figure 4-5).

At maximum applied pressure (see Figure 4), a maximum stretch of 1.7 was found (at segment 4), while the least stretch of value 1.4 was observed (at segment 2). A maximum variation of +/-10% was therefore observed when individual segment response was compared to the average stretch response of the inner wall at that pressure. Within the outer

wall segments (see Figure 5), at applied pressure of 21kPa, a maximum stretch of 1.36 was found (at segment 4), while the least stretch of value 1.21 was observed (at segment 2). A deviation of 8% was observed when the outer wall segments stretch response were compared to the that of the outer wall.

4 Discussion

All the pressure-stretch curves show a general response of a hyperelastic material. Beyond certain pressures (for each curve different), the slope of circumferential extension tapers (still being positive). The mathematical model has already shown a capability to tweak the response by changing the values of the parameters of the chosen mathematical model [1,3]. The pressure range was chosen based on general arterial pressures.

As shown in the results, when the pressure-stretch response of the inner wall of both models is compared to each other, the non-axisymmetric model is stiffer by about 10%. This means, if stretch at inner wall is important for measurements, a symmetric generalization might be an issue. There is a variation of +/-10% within the 5 segments of the inner wall, while it was +/-8% for the outer wall. Although the pressure-stretch response within outer wall segments of non-axisymmetric model is different, its averaged response in comparison to the axisymmetric model was almost identical. If the objective was to anaylze the average stretch at different portions of inner and outer segments, the non-axisymmetric model was suited. If evaluating the overall outer wall response was the primary objective, the symmetric model could be well suited. The simulations are indicative that expansion of the artery outer wall relies on its geometry and other constraints and generalized use of symmetric prior could be undertaken based on what outputs are of importance.

If the geometry of the tube were tweaked, the response could have been different. The presence of thickness redistribution may have an impact on how the pressure redistributes itself within the vessel. This might be the reason for obtaining a non-uniform response although both models have similar tissue properties.

The material description chosen used a mathematical model based on the fibre stretch. In simulation, while the tube expands, there may be sections where the pressure leads to buckling of the material at certain non-pressure portions. The fibres orientation was based on the geometry, path of the folds and histological descriptions. Tweaking some parameters and inducing conditions of pre-stress and pre-stretches within the artery model could give results of interests and remains a part of further investigation.

5 Conclusion

Approaches to model the behaviour of the tissue and the process of forward and reverse modelling is being undertaken. The use of axisymmetric priors for inverse modelling is very specific to the type of output that needs evaluation. Multiple test scenarios could be conceptualized to mimic in-vivo situations that are not a part of this study.

Verification and validation of these outputs with in-vivo experiments would be the next step towards refining the models.

Author Statement

This research was partly supported by the German Federal Ministry of Research and Education (BMBF) under grant no. 13FH5I05IA (CoHMed/Digitalization in the OR).

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