FINITE ELEMENT SIMULATION OF CAROTID ENDARTERECTOMY

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Abstract. *Carotid endarterectomy is the standard surgical procedure to remove the plaque from inside the artery. The current work is concerned with the mechanical behavior of human carotid arteries subjected to endarterectomy. A 3D tuning-fork-shaped model of the average carotid bifurcation geometry is analysed in the context of the finite element method. A patch is added in the model to cover the incised region of internal carotid artery. In this study, the non-linear and near incompressible mechanical behavior of the artery is described by an isotropic function of strain energy potential. The materials used for patches in arteries may be either biological (veins, bovine pericardium) or synthetic (PTFE, Dacron). A linear elastic material model is used to describe the behavior of the patch. The residual stresses, present in the unloaded state of the artery, are accounted for in the analysis. A crack is gradually opened in the region of the internal carotid artery in order to simulate the incision made at the beginning of the surgery. The patch is then added in the model and rigidly connected with the artery only at the boundary nodes where the sutures are placed during the surgery. Finally, the patched artery is loaded under physiological conditions. The stress and deformation fields that develop on the carotid artery wall during the endarterectomy process and subsequently during operation of the patched artery under physiological conditions are investigated.*

1 INTRODUCTION

Carotid artery has particular clinical and biomechanical concern because it is prone to atherosclerosis and often subjected to surgery for the prevention of stroke in humans. Atherosclerosis is caused by the deposition of atherosclerotic plaque on the walls of the artery, usually around carotid bifurcation, and is responsible for carotid stenosis. Carotid endarterectomy is the standard surgical procedure to remove the plaque from inside the artery. During endarterectomy, an incision is made across the affected region of the artery, and then the artery is opened and cleaned from the accumulated pathological material. A patch is finally sutured around the incised region in order to ensure blood-proofness during normal operation of the artery. The materials used for patches in arteries may be either biological (veins, bovine pericardium) or synthetic (PTFE, Dacron). Although a variety of patches have been made available for use during endarterectomy, there is minimal information on the biomechanical effects caused by the placement of specific patches on diseased carotids. Recently, Kamenskiy et al. [1] made the first attempt via Finite Element (FE) modelling towards producing evidence to aid the surgeon in choosing a patch while caring for a patient with carotid disease. However, the surgical process, as well as, the arterial wall residual stresses were not taken into account in their numerical calculations.

The current work is concerned with the mechanical behavior of human carotid arteries subjected to endarterectomy. The analysis involves the development of FE models in order to investigate stress and deformation fields that develop on the carotid artery wall during the whole endarterectomy process and subsequently during operation of the patched artery under physiological conditions.

2 MATERIALS AND METHODS

2.1 Patched carotid geometry and mesh

A 3D tuning-fork-shaped model of the average carotid bifurcation geometry is analysed using ABAQUS general purpose FE program (Figure 1). The total length of the arterial model was set at 48mm and a uniform wall thickness of 0.7mm was applied. Two model cases with Common Carotid Artery (CCA) internal diameter 4.6mm and 6mm were considered in the analysis. The rest dimensions of the model are dependent on CCA internal diameter [2] and presented in Table 1. A patch of 0.7mm thickness is added in the model to cover the incised region of Internal Carotid Artery (ICA) (Figure 2). The patched artery model was meshed with second-order ten-node tetrahedral hybrid elements. The model with CCA internal diameter 6mm consists of about 840.000 nodes, 330.000 elements and a total of 1.841.000 variables.

	Position in Figure 1						
	A	B	C	D	E	L	a
× A	1.00	1.04	1.11	0.72	0.69	2.40	50^o

Table 1 : Example of how to set a table

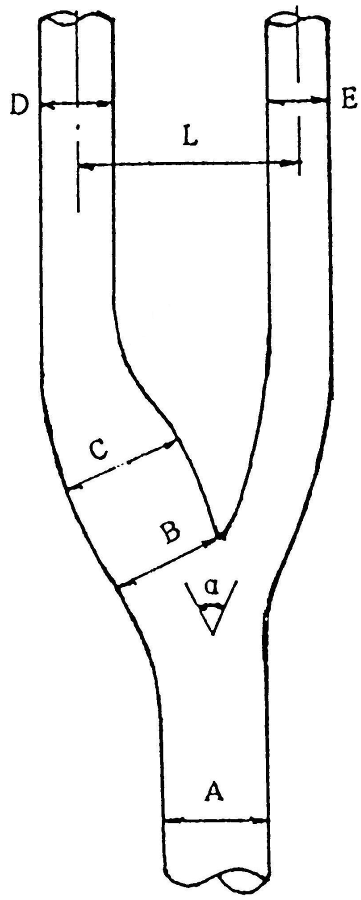

Figure 1: Schematic representation of a tuning-fork-shaped model of the average carotid bifurcation geometry [2].

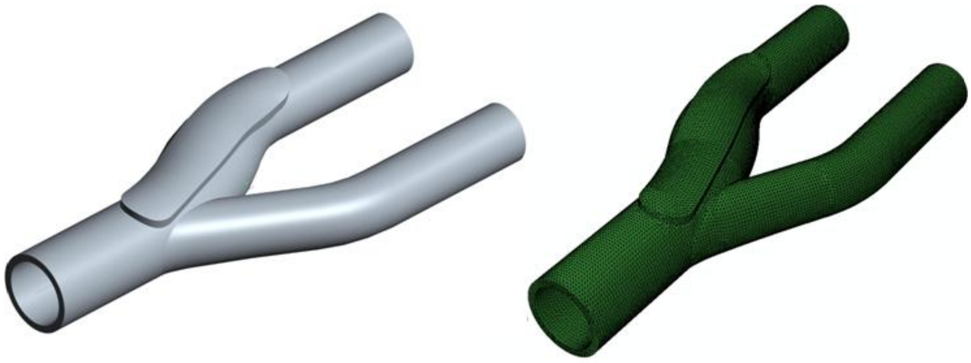

Figure 2: Patched carotid artery with 6mm ICA internal diameter: (a) Schematic representation of the 3D model, (b) FE mesh.

2.2 Arterial wall material

In this study, the non-linear and near incompressible mechanical behavior of artery is described by an isotropic function of strain energy potential [3]

$$U = \frac{c}{2}\left(\bar{I}_1 - 3\right) + \frac{a}{2b}\left(e^{b\left(\bar{I}_1 - 3\right)^2} - 1\right) + \frac{1}{D}\left(J - 1\right)^2 \tag{1}$$

where U is the strain energy per unit of reference volume, $c, a, b, D > 0$ are material parameters, $\bar{I}_1 = \bar{\lambda}_1^2 + \bar{\lambda}_2^2 + \bar{\lambda}_3^2$ is the first modified invariant of the modified left Cauchy-Green strain tensor, $\bar{\lambda}_i = J^{-1/3}\lambda_i$ are the deviatoric stretches, J is the volume ratio, and λ_i the principal stretches. The material parameters c, a, b were calculated using a least-square algorithm to fit the curve obtained via equation (1) within the range of variation of $\bar{I}_1$ with the experimental data of Delfino et al. [4]. The values of the calculated parameters are $c = 0.036312$ MPa, $a = 0.306546$ MPa, and $b = 11.6681$. An initial bulk modulus of $K_0 = \frac{2}{D} = 36.312$ MPa was set to consider for the near incompressibility of the arterial tissue. The strain energy potential for isotropic hyperelastic material behavior (1) was implemented in ABAQUS via a user subroutine UHYPER.

2.3 Patch material

The materials used for patches in arteries may be either biological (veins, bovine pericardium) or synthetic (PTFE, Dacron). In the current study a linear elastic material model is used to describe the mechanical behavior of the patch. Among the above mentioned materials used for patches, Dacron has the greatest stiffness with $E = 40$ MPa [5]. A lower elastic modulus $E = 10$ MPa, which approximates the behavior of biological patches, is also used in the current analysis. The material is also assumed near incompressible with a Poisson's ratio $\nu = 0.4999$.

2.4 FE analysis

The surgical procedure of carotid endarterectomy is simulated step-by-step via the finite element method. During the analysis the boundary nodes of CCA, ICA, and ECA remain fixed.

<u>STEP 1</u>: The residual strains, present in the unloaded state of the artery, are accounted for in the analysis. During the surgical procedure CCA, ICA, and ECA are clamped in order to prevent blood supply at carotid bifurcation; therefore the axial residual stretch of the arterial wall can be ignored in the analysis. The circumferential residual stretch is taken into account by applying internal pressure on the arterial wall. Based on experimental measurements [6] the circumferential residual stretch is about $\lambda_g = r/R = 1.04$ at an ICA segment located far from the bifurcation area.

STEP 2: A crack is gradually opened in the region of ICA in order to simulate the incision made at the beginning of the surgery (Figure 3).

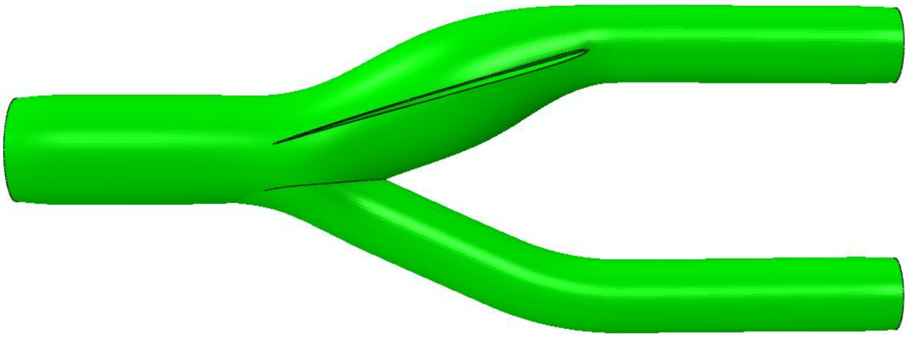

Figure 3: Incision in the region of ICA.

STEP 3: The patch is added in the model and rigidly connected with the artery only at nodes around the patch border, where the sutures are placed during the surgery, via a tie multi-point constraint. The rest part of the patch is allowed to interact with the arterial wall via finite sliding frictionless contact.

STEP 4: A systolic pressure of 120mmHg is applied in the patched artery under physiological conditions.

STEP 5: The patched artery is unloaded. The worst case scenario of a zero diastolic pressure is studied.

STEP 6: A systolic pressure of 240mmHg is applied in order to account for an over-pressurized artery.

3 RESULTS AND DISCUSSION

The arterial geometries with CCA internal diameter 4.6mm and 6mm as described in section 2.1 are analysed and the surgical procedure of carotid endarterectomy is simulated step-by-step as described in section 2.4. The same arterial geometries not subjected to a surgical procedure were also analysed under physiological conditions for comparison purposes.

The results of the current analysis include: a) the deformation state of the arterial wall cross section at the patched region under physiological loading conditions, b) the maximum circumferential stress $\sigma_{\theta\theta}$ developed at the connecting region of the patch with the arterial wall, and c) maximum von Mises stress developed in the whole structure. The von Mises equivalent stress (a measure of shear stresses) is defined as $\sigma_{eq} = \sqrt{\frac{3}{2}\mathbf{s}:\mathbf{s}}$ where $\mathbf{s} = \boldsymbol{\sigma} - p\boldsymbol{\delta}$ is the stress deviator and p is the hydrostatic stress.

In all cases examined the maximum von Mises equivalent stress appears in the carotid bifurcation (Figure 4). The magnitude of stress is not influenced significantly either by the surgical procedure or the placement of the patch.

In the case of CCA internal diameter equal to 4.6 mm then, $\sigma_{eq}{}^{\max} \simeq 0.4$ MPa under normal physiological loading, $\sigma_{eq}{}^{\max} \simeq 0.04$ MPa after unloading, and $\sigma_{eq}{}^{\max} \simeq 0.8$ MPa under a twofold increase of systolic pressure loading. The maximum circumferential stress $\sigma_{\theta\theta}$ developed at the connecting region of the patch with the arterial wall is not influenced significantly by the stiffness of the patch (Figure 5). If the elastic modulus of patch is $E = 10$ MPa then $\sigma_{\theta\theta}{}^{\max} \simeq 0.15$ MPa under normal physiological loading, $\sigma_{\theta\theta}{}^{\max} \simeq 0.013$ MPa after unloading, and $\sigma_{\theta\theta}{}^{\max} \simeq 0.34$ MPa under a twofold increase of systolic pressure loading. If the elastic modulus of patch is $E = 40$ MPa then $\sigma_{\theta\theta}{}^{\max} \simeq 0.16$ MPa under normal physiological loading, $\sigma_{\theta\theta}{}^{\max} \simeq 0.014$ MPa after unloading, and $\sigma_{\theta\theta}{}^{\max} \simeq 0.35$ MPa under a twofold increase of systolic pressure loading. Finally, if the artery is not subjected to a surgical procedure then $\sigma_{\theta\theta}{}^{\max} \simeq 0.06$ MPa under normal physiological loading, $\sigma_{\theta\theta}{}^{\max} \simeq 0.0055$ MPa after unloading, and $\sigma_{\theta\theta}{}^{\max} \simeq 0.1$ MPa under a twofold increase of systolic pressure loading.

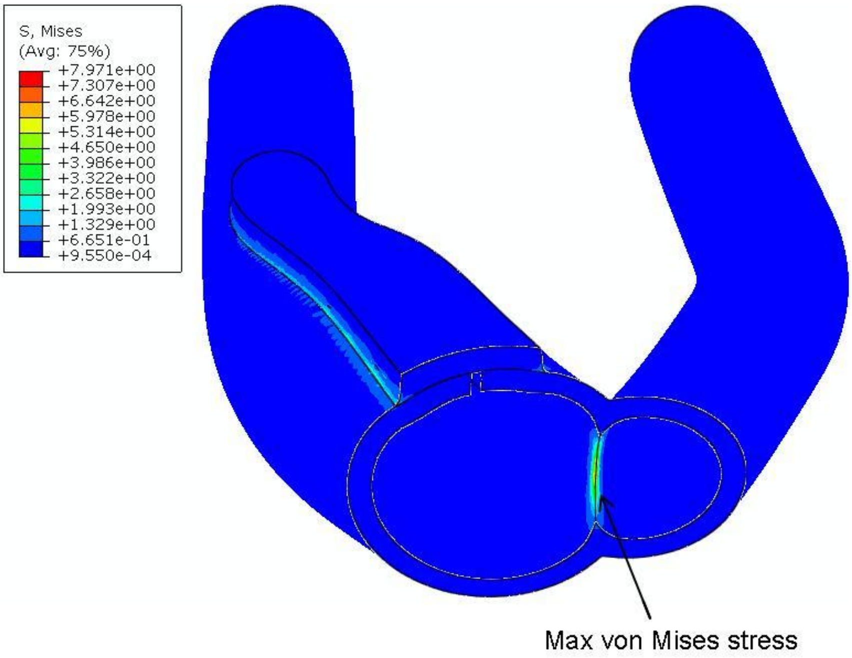

Figure 4: Carotid artery with CCA internal diameter 6mm after unloading and reloading with a twofold increase of systolic pressure. Maximum von Mises equivalent stress appears in the bifurcation.

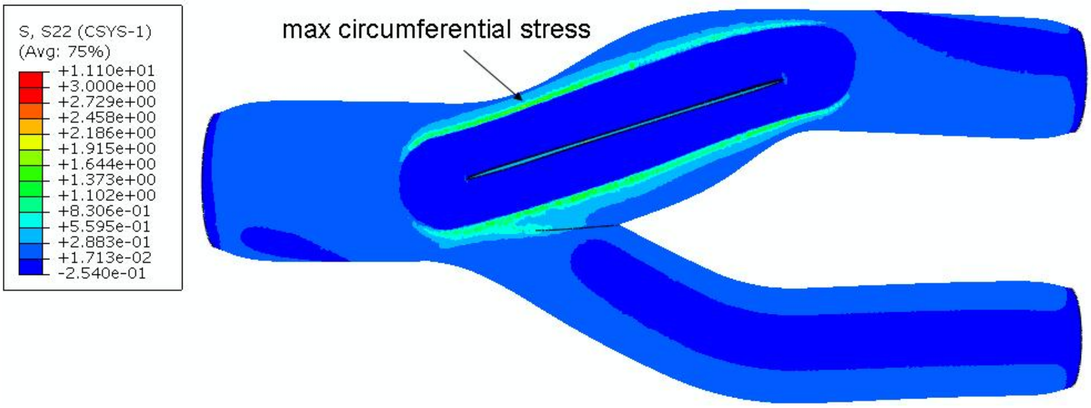

Figure 5: Carotid artery with CCA internal diameter 6mm after unloading and reloading with a twofold increase of systolic pressure. Maximum circumferential stress appears in the connecting region of the patch with the arterial wall.

In the case of CCA internal diameter equal to 6.0 mm then, $\sigma_{eq}^{\max} \simeq 0.7$ MPa under normal physiological loading, $\sigma_{eq}^{\max} \simeq 0.07$ MPa after unloading, and $\sigma_{eq}^{\max} \simeq 1.3$ MPa under a twofold increase of systolic pressure loading. The maximum circumferential stress $\sigma_{\theta\theta}$ developed at the connecting region of the patch with the arterial wall is not influenced significantly by the stiffness of the patch (Figure 5). If the elastic modulus of patch is $E = 10$ MPa then $\sigma_{\theta\theta}^{\max} \simeq 0.22$ MPa under normal physiological loading, $\sigma_{\theta\theta}^{\max} \simeq 0.018$ MPa after unloading, and $\sigma_{\theta\theta}^{\max} \simeq 0.47$ MPa under a twofold increase of systolic pressure loading. If the elastic modulus of patch is $E = 40$ MPa then $\sigma_{\theta\theta}^{\max} \simeq 0.23$ MPa under normal physiological loading, $\sigma_{\theta\theta}^{\max} \simeq 0.018$ MPa after unloading, and $\sigma_{\theta\theta}^{\max} \simeq 0.5$ MPa under a twofold increase of systolic pressure loading. Finally, if the artery is not subjected to a surgical procedure then $\sigma_{\theta\theta}^{\max} \simeq 0.085$ MPa under normal physiological loading, $\sigma_{\theta\theta}^{\max} \simeq 0.006$ MPa after unloading, and $\sigma_{\theta\theta}^{\max} \simeq 0.15$ MPa under a twofold increase of systolic pressure loading.

The stress results of the analyses regarding maximum von Mises stress and maximum circumferential stress developed on the arterial wall are presented in Tables 2 and 3 respectively.

CCA internal diameter	Loading	Max von Mises stress (MPa)
4.6 mm (patched)	Under normal systolic pressure	0.4
	Unloading	0.04
	Under x2 normal systolic pressure	0.8
4.6 mm (healthy)	Under normal systolic pressure	0.42
	Unloading	0.03
	Under x2 normal systolic pressure	0.9
6.0 mm (patched)	Under normal systolic pressure	0.7
	Unloading	0.07
	Under x2 normal systolic pressure	1.3
6.0 mm (healthy)	Under normal systolic pressure	0.8
	Unloading	0.04
	Under x2 normal systolic pressure	1.45

Table 2: Magnitude of maximum von Mises equivalent stress around the bifurcation.

CCA internal diameter	Loading	Patch Young’s Modulus (MPa)	Max circumferential stress (MPa)
4.6 mm (patched)	Under normal systolic pressure	10	0.15
		40	0.16
	Unloading	10	0.013
		40	0.014
	Under x2 normal systolic pressure	10	0.34
		40	0.35
6.0 mm (patched)	Under normal systolic pressure	10	0.22
		40	0.23
	Unloading	10	0.018
		40	0.018
	Under x2 normal systolic pressure	10	0.47
		40	0.5
4.6 mm (healthy)	Under normal systolic pressure	-	0.06
	Unloading	-	0.0055
	Under x2 normal systolic pressure	-	0.1
6.0 mm (healthy)	Under normal systolic pressure	-	0.085
	Unloading	-	0.006
	Under x2 normal systolic pressure	-	0.15

Table 3: Magnitude of maximum circumferential stress around the patch borders.

Also, the analysis revealed that the patch causes non-uniform expansion of the arterial wall and stretching of the arterial wall around the patch borders under physiological loading conditions (Figure 6). The non-uniform expansion of the arterial wall may have implications on the blood flow in the artery whereas the stretching of the arterial wall around the patch borders may have implications on the blood-proofness of the arterial wall.

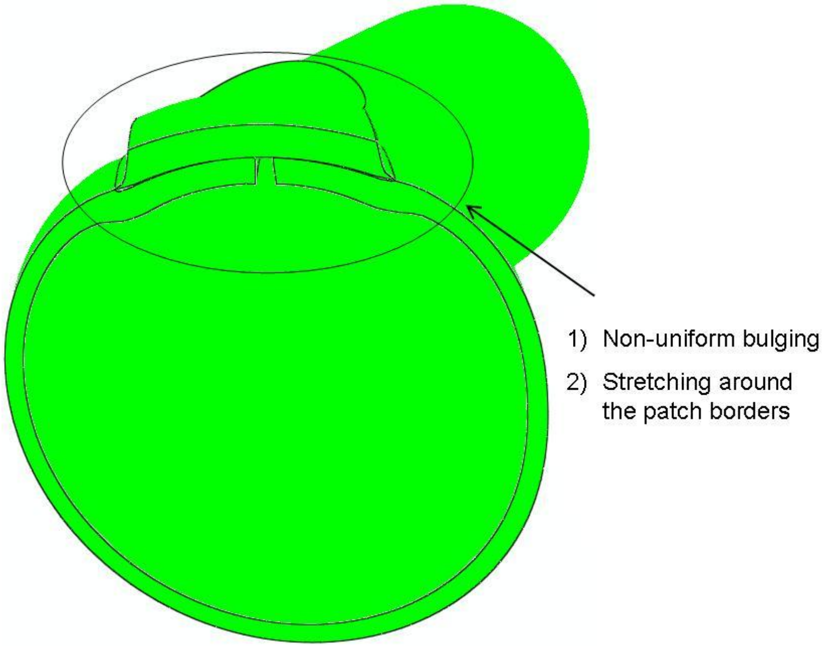

Figure 6: Deformed configuration $(2\times)$ of the carotid artery with CCA internal diameter 6mm and patch elastic modulus $E = 40$ MPa after unloading and reloading with a twofold increase of systolic pressure.

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