

IMPERIAL

BIOMECHANICS OF SOFT TISSUE-DEVICE
INTERACTIONS

AN INVESTIGATION INTO SKIN AND ARTERIAL TISSUE

Author

LULI LI

Supervised by

DR MARC MASEN

A Thesis submitted in fulfillment of requirements for the degree of
Doctor of Philosophy

Department of Mechanical Engineering
Imperial College London
2024

Abstract

This thesis adopts a multidisciplinary approach, integrating mathematical modelling, experimental testing, and numerical simulations, to investigate the complex interactions between soft tissues and medical devices. The work focuses on two main areas: characterising the contact behaviour of skin and mouse arterial tissue during indentation, and analysing artery–stent interactions using finite element simulations.

In Chapter [3](#), a novel method is developed to determine the Ogden hyperelastic parameters of soft tissues using indentation experiments on artificial skin. By incrementally applying Hayes’ model under the assumption that soft tissues behave approximately linearly at strain increments 1%, an equivalent shear modulus is calculated for each increment. Principal stresses are derived using Hooke’s law, and combined with principal strains to obtain compound stress–strain curves. The Ogden parameters are then extracted via curve fitting. Total contact force is computed by summing incremental Hayes forces.

Chapter [4](#) applies this methodology to mouse arterial tissue and introduces an approach to determine the equivalent thickness for Hayes’ model. The optimal thickness is defined as the one that minimises the variance of the calculated equivalent shear moduli. Based on 26 indentation tests from five mouse artery samples, the estimated thickness ranges from 0.05 to 0.5 mm, with an average shear modulus of 1.22 kPa.

Chapter [5](#) employs finite element simulations to study artery–stent interactions. The arterial model includes intima, media, adventitia, and a lipid pool, each assigned anisotropic hyperelastic properties. Contact is defined using hard contact with friction coefficients from 0.01 to 0.1. Simulations investigate different stent surface geometries. Results show stress and strain concentration in three critical areas: the contact interface, the cap’s right shoulder, and around the lipid pool. Larger friction and high-amplitude stent surfaces increase stresses, while small-amplitude convex surfaces reduce average shear stress, suggesting their potential to reduce arterial injury during stenting.

Declaration of Originality

I hereby declare that the work presented in this thesis is my own unless otherwise stated. To the best of my knowledge the work is original and ideas developed in collaboration with others have been appropriately referenced.

Copyright Declaration

The copyright of this thesis rests with the author. Unless otherwise indicated, its contents are licensed under a Creative Commons Attribution-Non Commercial-No Derivatives 4.0 International Licence (CC BY-NC-ND).

Under this licence, you may copy and redistribute the material in any medium or format on the condition that; you credit the author, do not use it for commercial purposes and do not distribute modified versions of the work.

When reusing or sharing this work, ensure you make the licence terms clear to others by naming the licence and linking to the licence text.

Please seek permission from the copyright holder for uses of this work that are not included in this licence or permitted under UK Copyright Law.

Acknowledgments

I would like to express my heartfelt gratitude to my supervisor, Dr Marc Arthur Masen, for his invaluable guidance, unwavering support, and constant encouragement throughout my research journey. His expertise and insightful feedback have been instrumental in shaping this thesis and enriching my understanding of the subject. I am equally indebted to Dr Alkystis Phinikaridou and Miss Ling Gao from King's College London for providing the mouse arterial specimens, which were crucial to the success of my experiments. Their dedication and collaboration have greatly contributed to my academic growth.

I am deeply thankful to my friends and colleagues, including Dr Eliane D. Gendreau, Dr Kian Kun Yap, Dr Tian Yuan, Dr Alexander MacLaren, Dr Ben Collard, Dr Jack Hayes, and Dr Rikeen D. Jobanputra, for their invaluable feedback on my papers and thesis, as well as their help with experiments. A special mention goes to my dear friends—Dr Marie-Louise Schlichting, Miss Janet Zhu, Dr Jie Zhang, Mr Roland Jones, Dr Thomas Kirkby, Mr Yun Zhao, and Mr Tom Briggs from the Tribology group—as well as Dr Yichang Shen, Dr Jingzi Huang, Dr Jessica Tjandra, Dr Tristan François Calvet, Mr Xianfeng Ren, and Dr Bo Yuan from Imperial College London. Your encouragement, support, and shared moments of laughter have brought joy and meaning to this journey, making the challenges more manageable and the successes more rewarding. Over the past decade, I have been privileged to be surrounded by an extraordinary group of friends and colleagues whose unwavering support and camaraderie have been a constant source of strength. I am profoundly grateful to each of you for helping me grow and persevere through this journey.

Finally, I extend my deepest gratitude to my family. To my mother and father, your unwavering support, love, and belief in me have been my guiding light and the foundation of all my achievements. To my boyfriend, Hang Su, your care and encouragement—both in life and in my career—have been an invaluable source of strength. You have all been my anchor, and for that, I am eternally grateful. This thesis is dedicated to you, with all my love and appreciation.

Contents

Abstract	i
Declaration of Originality	iii
Copyright Declaration	v
Acknowledgments	vii
Nomenclature	xi
List of Figures	xv
List of Tables	xxi
1 Introduction	1
1.1 Background	1
1.2 The Objective of the Research	2
1.3 This Thesis	3
2 Background	5
2.1 Histology and Mechanical Behaviour of Skin and Arterial Tissue	5
2.1.1 Fundamental Structural and Material Properties of Skin	6
2.1.2 Fundamental Structural and Material Properties of Arterial Tissue	7
2.2 Constitutive Models for Skin and Arterial Tissue	8
2.2.1 Basic Continuum Mechanics for Isotropic and Anisotropic Hyperelastic Tissue	8
2.2.2 Constitutive Equations for Hyperelasticity	10
2.3 Contact Modelling	16
2.3.1 Analytical Contact Model	16
2.3.2 Indentation of Tissue	20
2.3.3 Friction in Contact Problem	21
2.3.4 Friction of Tissue	24
2.4 Finite Element Method for Contact Problem	28
2.4.1 Fundamentals of Finite Element Method	29
2.4.2 Finite Element Modelling in ABAQUS	32
2.4.3 Contact Modelling in ABAQUS	36
2.4.4 Indentation Simulation on Tissue	39
3 Nonlinear Contact	41
3.1 Introduction	41
3.2 Modelling Non-linear Contact	43
3.3 Application of the Method	47
3.3.1 Indentation Experiments	47
3.3.2 Numerical Simulation	48
3.3.3 Determining the Ogden Parameters from the Assembled Curves	50
3.4 Verification	51
3.5 Conclusion	52
4 Nonlinear Contact on Mouse Arterial Tissue	55
4.1 Introduction	55
4.2 Method	57

4.2.1	Determining the Ogden Material Parameters	57
4.2.2	Application Window of the Method	61
4.2.3	Determining Ogden Parameters using Simulation Curves	64
4.3	Application of the Method on Mouse Arterial Tissue	67
4.4	Discussion	71
4.5	Conclusion	74
4.6	Supplementary	75
5	Finite Element Simulation of Artery-Stent Interaction	77
5.1	Introduction	77
5.2	Method	81
5.2.1	Finite Element Model of Artery-Stent Interaction	81
5.2.2	Surface Shape Modelling	87
5.3	Result	90
5.3.1	Verification of the Finite Element model	90
5.3.2	Verification of the Non-uniform Method	91
5.3.3	Artery-Stent Interaction	92
5.4	Discussion	102
5.5	Conclusion	107
6	Discussion and Recommendation	109
6.1	Discussion	109
6.1.1	Contact Modelling	109
6.1.2	Artery-Stent Interaction	114
6.2	Recommendations	117
6.2.1	Recommendations for Contact Modeling	117
6.2.2	Recommendations for Artery-Stent Interaction	120
	Conclusions	123
	Bibliography	125

Nomenclature

a	contact radius (m)
$\mathbf{a}_0$	mean/principal direction of fibres (-)
A	amplitude (m)
A_c	surface area (m ²)
A_{ijkl}	material parameters (-)
A_{real}	real contact area (m ²)
b	parameter of fibres distribution (-)
B	material parameter (-)
$\mathbf{B}$	left Cauchy-Green tensors (-)
$\mathfrak{B}_0^i$	structure tensor (-)
c	correction coefficient (-)
c_d	material wave speed (m/s ²)
$c, c_1, c_2, c_3, c_4, c_5$	material parameters (-)
C_1, C_2, C_{mn}, C_j	material parameters (-)
C_{ijkl}	stiffness matrix (-)
$\mathbf{C}$	right Cauchy-Green tensors (-)
D_{lumen}	smallest diameter of lumen (m)
$\mathbf{e}$	Cartesian coordinate axis (-)
E	elastic modulus (Pa)
E_{eff}	effective elastic modulus (Pa)
F	contact force (N)
f_i^b, f_i^s, f_i^α	force (N)
F_i	incremental contact force (N)
$F_{\text{hayes}}, F_{\text{sim}}$	contact force (N)
$\mathbf{F}$	deformation gradient tensor (-)
F_{spring}	spring force (N)
F_μ	friction force (N)
$F_{\mu, \text{adh}}$	adhesion friction (N)
$F_{\mu, \text{def}}$	deformation friction (N)
g	gravitational acceleration (m/s ²)
g_n	gap function (m)
g_τ	tangential gap (m)
$\dot{g}_\tau$	slip velocity (m/s)

G, G_i	shear modulus (Pa)
G^*	equivalent shear modulus (Pa)
$\bar{G}$	mean of shear moduli (Pa)
h, h_i	thickness (m)
h_{hayes}	equivalent thickness (m)
H	conical indenter parameter (°)
$\mathbf{H}, \mathbf{H}_i$	homogenised structure tensor (-)
I_j	strain invariants (-)
$\mathbf{I}$	identity tensor (-)
$\mathbf{I}_f$	internal element force (N)
J	Jacobian determinant of $\mathbf{F}$ (-)
J_m	material constants (-)
k_{spring}	elastic constant (N/m)
k_{i1}, k_{i2}	parameters for embedded fibres (-)
K_k	fibre stiffness (N/m)
$K_{iakb}, \mathbf{K}$	stiffness matrix (-)
l_{st}	wavelength of stent cell (m)
L_e	dimension of the mesh element (-)
m	mass (kg)
M_n	node number (-)
$\mathbf{n}, n_j$	normal direction (-)
n	number of fibres (-)
N	order of the Ogden model (-)
N_α	shape functions (-)
$N_{\alpha,j}, N_{b,l}$	partial derivative of the shape functions (-)
p	contact pressure (Pa)
p_n	contact pressure on nodes (Pa)
P	hydrostatic pressure (Pa)
$\mathbf{P}$	external applied force (N)
r	radius of stent (m)
R	radius of spherical indenter (m)
R_k	fibre orientation probability density function (-)
s_t	width of strut (m)
S	von Mises stress (-)
S^2, S_c^2, S_p^2	variance (-)
S_1, S_2, S_3, S_4, S_5	surface (-)
t	time (s)

t_{stable}	stable time increment (s)
$u_i, u_n, u_{\text{non}}, u_r, \mathbf{u}, u_1, u_2, u_3$	displacement (m)
$u_{k,l}, u_{l,k}$	displacement deviation (-)
u_i^s	displacement on the boundary (m)
$\ddot{u}$	acceleration (m/s ²)
U^2	unit sphere (-)
$\mathbf{U}, U_k^b$	nodal displacement (m)
v_{spring}	velocity of spring (-)
$\mathbf{v}_0$	direction of fibre (-)
V	domain (-)
V_c, V_e, V_k	volume fraction (-)
$\mathbf{x}$	material point in the Eulerian or current configuration (m)
$\mathbf{X}$	material point in the Lagrangian or reference configuration (m)
z_i	baseline (m)
α, α_p	material parameters for Ogden model (-)
β	viscoelastic loss fraction (-)
$\Gamma^C, \Gamma^{C,st}, \Gamma^{C,sl}$	contact interface (-)
$\Gamma, \Gamma_t, \Gamma_u, \Gamma_1, \Gamma_2$	boundary (-)
δu_i	virtual displacement (m)
δU_i^α	virtual nodal displacements (m)
Δt	time increment (s)
ΔF	force increment (N)
Δz	distance (m)
η	pressure coefficient (-)
ϵ_n	contact stiffness (N/m)
ϵ_j	principal strain (-)
$\epsilon^*, \epsilon_{kl}$	strain (-)
θ	semi-included angle of the indenter (°)
Θ	fibre angle of the indenter (°)
κ	parameters for Hayes' model (-)
κ_i	dispersion parameter (-)
λ_j	stretch ratio (-)
λ_l	Lamé coefficients (-)
λ_m	chain extensibility coefficient (-)
μ	friction coefficient (-)
μ_l	Lamé coefficients (-)
μ_0	initial coefficient of friction (-)

μ_p, μ	Ogden parameter (-)
ν	Poisson's ratio (-)
π	mathematical constant (-)
Ξ	nonlinear function (-)
ξ	node coordinate (-)
ρ	mass density (kg/m^3)
$\rho^{\text{a0}}(\Theta)$	density function (-)
$\sigma_{ij}, \sigma_{ij}^*, \sigma^*$	stress (Pa)
σ_j	principal Cauchy stress (Pa)
$\sigma_{ij,j}$	stress deviation (-)
$\tau, \tau_{xy}, \tau_{yz}, \tau_{xz}$	shear stress (Pa)
τ_0	yield stress at zero external pressure (Pa)
χ	Hayes' model parameter (-)
$\psi, \psi_{\text{bundle}}, \psi_{\text{fibre}}, \psi_g$	strain-energy function (Pa)
ω	indentation depth (m)
ω_i	incremental indentation depth (m)
$\omega(\tau)$	function for Fredholm integral equation
Ω_0	reference configuration of the material (-)

List of Figures

2.1	Skin tissue: (A) H & E stained human skin biopsy. The dermal-epidermal junction is denoted by a red dotted line [12]. (B) Schematic representation of the multi-layer nature of the skin featuring the epidermis (constituted of the stratum corneum and viable epidermis), dermis, and hypodermis [13].	6
2.2	Arterial tissue: (A) A healthy elastic artery composed of three layers: intima, media, and adventitia. The intima is the innermost layer consisting of a single layer of endothelial cells that rests on a thin basal membrane and a subendothelial layer whose thickness varies with topography, age, and disease. The media is composed of smooth muscle cells, a network of elastic and collagen fibrils, and elastic laminae which separate media into several fibre-reinforced layers. The primary constituents of adventitia are thick bundles of collagen fibrils arranged in helical structures. The adventitia is the outermost layer surrounded by loose connective tissue [8]. (B) Orientations of fibre families of an artery (without intima) and a strip for tension test [28].	7
2.3	Hyperelastic properties of soft tissue: (A) Hyperelastic properties of pig muscular tissue shown by different hyperelastic models [40]. (B) Schematic diagram of a typical (tensile) stress-strain curve for skin showing the associated collagen fibre morphology [8].	9
2.4	(A) Three-dimensional graphical representation of the orientation of the collagen fibres based on the transversely isotropic density function [33]. (B) Two-dimensional graphical representation of the (transversely isotropic) von Mises distribution of the collagen fibres [33].	15
2.5	Contact scenarios: (A) A spherical indenter pressing into a linear elastic material with an infinite half-space, as described by models developed by Hertz and Sneddon. (B) Scenarios accounting for finite-thickness effects representing using models such as those developed by Hayes and Dimitriadis, which incorporate corrections for boundary effects in finite layers. (C) Scenarios incorporating hyperelastic properties but assuming an infinite thickness, as described by Lin's model. (D) This research investigates a scenario that incorporates both finite-thickness and hyperelastic properties	17
2.6	Indentation test: (A) The schematic of the indentation test conducted for mouse arterial tissue in Chapter 5, where a spherical indenter with a radius of $R = 150$ μm is used. The indentation setup consists of several key components: the load transducer, the connections, the indenter itself, the specimen holder, and the specimen (arterial tissue) to be tested. This configuration ensures accurate measurement of the contact force and displacement depth during the test. (B) Biomomentum Mach-1 mechanical tester (Biomomentum, Laval, Canada).	21

2.7	Friction test: (A) Tribometer consists of a top solid substrate, referred to as the slider, placed on a bottom substrate known as the track. The slider and track may be separated by a thin lubricant film, and they are compressed together by a loading force, F , which is typically due to the mass m of the top block by $F = mg$, where g is the acceleration due to gravity. The slider is connected to the base, which moves at a constant velocity v_{spring} through a spring with an elastic constant k_{spring} . In practical applications, the elasticity of the top block itself may serve the role of the spring. The primary output parameter measured in this setup is the spring force, F_{spring} [163]. (B) Schematic of microtribometer setup for cell friction experiments. This setup includes key components: DMEM (Dulbecco's Modified Eagle Medium), FBS (Fetal Bovine Serum), and SMC (Smooth Muscle Cell). The piezoelectric-controlled stage linearly reciprocates the tissue culture plate containing the SMC monolayer against a glass pin, which is held under a constant normal load throughout a series of 10 cycles. Friction loop data is captured using the Anton Paar NTR3. Following the completion of tribological testing, the SMCs are immediately returned to 37 °C to ensure optimal cell viability [172].	25
2.8	Schematic of tribometers based on various contact configurations [160]: (A) Linear contact: this configuration involves two surfaces sliding against each other in a straight line. (B) Perpendicular rotating contact: in this setup, one surface rotates perpendicular to the other, allowing for a circular motion that generates friction. (C) Parallel rotating contact: this configuration features two surfaces rotating parallel to each other, which can affect the contact area and frictional behaviour. (D) User-controlled motion: this flexible setup allows the user to dictate the motion of the surfaces, which can include various trajectories and speeds to study their effects on friction	26
2.9	Coefficient of friction of the index finger as a function of normal load [177], [180], [182], [189], [196]–[201]. Image adapted from Kuilenburg et al [174].	28
2.10	FE discretisation [207]: the domain V is discretised into small quadrilateral elements with eight nodes per element, which are used to define the element geometry and perform the interpolation function across the element.	30
2.11	The contact algorithm in ABAQUS checks all contact interactions at the start of each increment, identifying each slave node as either open or closed. For closed nodes, it determines whether they are sliding or sticking and applies or removes constraints accordingly. ABAQUS then updates the model configuration and re-evaluates the contact conditions after each iteration. Nodes where clearance becomes non-positive switch to closed, while nodes with negative contact pressure switch to open. If contact changes are detected, ABAQUS marks it as a severe discontinuity iteration. Iterations continue until severe discontinuities are minimal (or absent) and equilibrium tolerances are met. Alternatively, ABAQUS can be set to iterate until no severe discontinuities remain before checking equilibrium [208].	36
3.1	Illustration of linearisation and segmentation: (A) $\Xi(\varepsilon_j, \alpha_p)$ for $\alpha_p = 0.5, 1, 2, 3, 4$ within the range $\varepsilon_{ji} \in [0, -0.3]$, showing that $\Xi(\varepsilon_j, \alpha_p)$ is independent of the value of α_p for strains smaller than 5%. (B) $\Xi(\varepsilon_1, 2)$ can be approximated by three adjoining linear segments of $k_i \varepsilon_{ji} + b_i$ ($i = 1, 2, 3$).	45
3.2	The contact behavior of hyperelastic materials can be modeled as linear materials, characterized by the equivalent shear modulus G_i^* in the i^{th} increment.	46
3.3	The procedure of the proposed method: the black block gives the information collected from the indentation test, while the purple block gives the data collected by the indentation simulation. The whole procedure was conducted in Matlab (Matlab 2021Ra).	47
3.4	The indentation tests: (A) Schematic of the indentation setup. (B) 8 measured force-displacement curves on different locations on the same specimen, where I, II, ..., VIII represents the test number.	48

3.5	The FE model and the calculated principal strains: (A) The FE model for indentation test. (B) Calculated principal strain field along the Z-axis at different indentation depths for the three simulated materials, the red numbers represent the indentation depth in mm.	49
3.6	Relationship between the Hill strain ε_j and the Green strain ε_j^G . For small strains, $\varepsilon_j \approx \varepsilon_j^G$	50
3.7	Determining the Ogden parameters: (A) The discretised force curve and (B) obtained shear moduli G_j^* . (C) strain fields ε_3 obtained by the numerical simulation. (D) stress fields σ_3 using Hooke's law.	51
3.8	Shear moduli obtained from the various indentation experiments: (A) the shear moduli along the Z-axis for one indentation result. (B) the average shear modulus for the 8 indentations performed on the material, as shown in Figure 3.4.	52
3.9	Comparison of the experimental data from 8 tests with the simulated data using the parameters obtained by Hayes' model and new developed model.	53
4.1	Identification of the equivalent thickness h_{hayes} : (A) Diagram of indentation setup. (B) (a) Arterial tissue exhibits a rough surface, which complicates the determination of thickness and prevents its direct application in Hayes' model; (b) A range of possible thickness values h_i are applied to assess the equivalent thickness h_{hayes} for the application of Hayes' model; (c) The equivalent thickness h_{hayes} is determined as the value for h_i that provides the smallest variation of material properties for each strain increment.	59
4.2	Hayes' model in thickness determination: (A) $\kappa\sqrt{\chi}$ versus indentation depth ω . (B) The calculated shear moduli G_i versus indentation depth ω for different thicknesses: red curve gives the G_i for $h_{\text{hayes}} = 100 \mu\text{m}$, purple curves represent the values given by other thicknesses.	60
4.3	FE model for indentation: (A) The FE model illustrates a $100 \mu\text{m}$ -thick arterial layer. The red line marks the path used to assess the principal strain ε_j . (B) The contour of strain distribution for material $G = 4 \text{ kPa}$, $\alpha = 1$, with the indentation depth $\omega = 20 \mu\text{m}$. (C) ε_j is evaluated along three paths for two materials with the same shear modulus $G = 4 \text{ kPa}$ and different Ogden parameter $\alpha = 1$ (purple line), $\alpha = 10$ (grey dotted line). For Path 3, the curves for $\alpha = 1$ and $\alpha = 10$ coincide. For Path 2, where the strains are higher, there is a small but noticeable difference. For path 1, the curves deviate considerably, particularly at higher strain levels. . .	62
4.4	(Two key aspects investigated through numerical simulations: (A) The hyperelastic material is linearized in small indentation increments, with the G_i^* characterizing the material property at each incremental step. (B) Validity window of the newly developed method: (a) To what indentation depths can the model be applied? (b) Is it possible to obtain an accurate value for h_{hayes} ?	62
4.5	Validity of Hayes' model in case of large relative deformations: (A) Simulation forces F_{sim} (purple lines) forces calculated using Hayes' model (grey lines). (B) The relative indentation depth where the difference between F_{sim} and F_{hayes} equals 5% for different specimen thicknesses. (C) The variance of shear moduli for different possible thicknesses is used to identify the equivalent thickness of the artery layer, with a true thickness of $100 \mu\text{m}$. The smallest variance is achieved at $h_i = 100 \mu\text{m}$, confirming the equivalent thickness. (D) The equivalent thickness h_{hayes} (purple dot) is identified by the smallest variances for different thickness values.	64
4.6	The procedure to obtain Ogden parameters from an indentation test. The black blocks outline the procedure for processing experimental curves, whilst the yellow blocks outline the method used to extract information from numerical simulations. For each indentation curve, the equivalent thickness h_{hayes} is identified from the possible thickness values first. Subsequently, the shear moduli and strains corresponding to h_{hayes} are used to fit the Ogden parameters.	65

4.7	Method verification: (A-B) Six indentation curves. (C) variances of the shear modulus for different tissue thickness values. The equivalent thickness h_{hayes} can be identified as the thickness for which G has the smallest variance, in this case $h_{\text{hayes}} = 100 \mu\text{m}$. (D) values for G_i^* for the six different tissues at $h_{\text{hayes}} = 100 \mu\text{m}$. (E) Average values of G were obtained from different nodes along Path 3 for the various simulated tissues. (F) Calculated values of α for the various simulated tissues.	66
4.8	Specimen and indentation test: (A) The image presents the arterial tissue opened longitudinally, fully prepared for indentation tests. It should be noted that the arterial tissue consists of an aortic arch and a descending part. In this research, we focus on the mechanical properties of arterial tissue, without distinguishing between the arch and descending segments. (B) The indentation equipment and setup.	67
4.9	The contact force F versus indentation depth ω collected from the indentation tests, where 1, 2, ..., 29 represents the test number.	69
4.10	Experimental results and obtained shear modulus: (A) Contact force versus indentation depth for various tissues with a thickness between 100 and 150 μm . Numbers represent the test ID (see table 4.2). The shear modulus G of Ogden material for 26 measurements: (B) Obtained shear modulus for the six thickness ranges.	70
4.11	Comparison of experimental results conducted on specimens with thicknesses between 100 and 150 μm , with simulated curves calculated using the obtained Ogden parameters α and μ and, for reference, curves calculated using the elastic modulus E from Hayes' model.	72
4.12	Equivalent shear moduli from Hertz's model and Hayes' model: (A) Hertz's model; (B) Hayes' model and newly developed method.	74
4.13	Ogden parameters fitting	75
5.1	The stent deployment and ISR procedure: (A) Stent expansion by balloon as described by Shimizu et al. [266]. (B) A schematic representation of the restenosis process [263]: (a) platelet aggregation: immediate result of stent placement with endothelial denudation and deposition of platelets and fibrinogen; (b) inflammatory phase: a variety of white blood cells will gather at the injury site; (c) proliferation phase: smooth muscle cells migrate and proliferate, creating the neointima in an attempt to repair the wound; (d) late remodelling phase: the neointima is transitions from predominantly cellular to a less cellular and more extracellular matrix-rich plaque.	78
5.2	Sections of the analyzed external iliac artery. Section B-B is the region with the smallest lumen diameter of 1.4 mm. The tissue components are adventitia (A), non-diseased media (M-nos), non-diseased intima (I-nos), fibrous cap (I-fc), lipid pool (I-lp), calcification (I-c), fibrotic intima at the medial border (I-fm) and diseased media (M-f). Image adapted from Holzapfel et al. [98].	80
5.3	The model utilized in Gasser's research [99] includes: (A) a segmented macroscopic view of a type V lesion according to the histological classification of Stary (adopted from Holzapfel et al. [98]). The borderlines represent different tissue components: A = adventitia, M-nos = non-diseased media, I-nos = non-diseased intima, I-fc = fibrous cap, I-fm = fibrous intima at the medial border, I-lp = lipid pool, and M-f = fibrous (diseased) media. (B) A 3D discretisation of a slice of the atherosclerotic lesion composed of four distinct tissue components (labeled as adventitia, media, intima, and lipid pool) using tetrahedral finite elements. The individual tissues are meshed separately and linked together employing the interface element method.	82
5.4	The geometric model of the artery: the artery consists of four components: adventitia, media, intima, and lipid pool from outer to inner. The thickness of the adventitia and media is 0.32 and 0.34 mm, respectively. The thickness of the intima varies from 0.1 to 0.5 mm, with an elliptical lumen and an arbitrarily shaped lipid pool. The elliptical lumen has a smallest diameter of $D_{\text{lumen}} = 0.14 \text{ mm}$.	83
5.5	The local coordinate system used in the simulation of the artery and stent is as follows: (A) <i>ORTZ</i> coordinate system applied to the FE model of the artery. (B) <i>ortz</i> coordinate system used for the stent.	84

5.6	The FE model for stent: (A) The 2D mesh part used for wrapping into a cylindrical stent. (B) The stent model used for FE simulation, with a diameter of 2 mm and a width of 1 mm	85
5.7	Artery-stent interaction: S_1 , S_2 and S_3 are the outer surfaces of the FE arterial model. Γ^C is the interface, consisting of S_4 (artery inner surface) of artery and S_5 (stent outer surface) of stent. Three coordinate systems are employed: the global coordinate system, $OXYZ$, defines the overall geometry and external loading of the model. Two local systems cylindrical coordinate system, $OTRZ$ is used for the artery, capturing its radial, circumferential, and longitudinal directions, while $otrz$ is applied for the stent, facilitating precise alignment and interaction modelling with the arterial structure	86
5.8	FE model for a different approach: (A) The realistic model, which shows the actual shape of the contact material. (B) The idealized model does not depict the actual shape but uses a displacement set for the nodes to represent the shape effect. . . .	88
5.9	Non-uniform displacement was applied to the upper surface nodes: (A) A uniform displacement (colored line) was applied to the idealized model over 10 steps, with the black dotted line indicating the baseline. The grey line represents the surface shape, while the black solid line shows the initial position of the surface. (B) A non-uniform displacement was applied to the idealized model over 10 steps. When considering the black dotted line as the starting point, the non-uniform displacement aligns precisely with the surface shape above the baseline.	89
5.10	The surface shapes of the stent: (A) Three surface shapes of the stent are considered in this study. (B) The surface shapes were given for the stent in $z = 0$. (C) Apply the non-uniform displacement to the stent.	91
5.11	Model verification: (A) The comparison between different mesh density, where D_{lumen} is the lumen diameter and d is the lumen center. (B) Quasi-static analysis	92
5.12	The maximum principal stress distribution of intima, media, and adventitia, respectively	92
5.13	The distribution of von Mises stress S for the idealised model and the realistic model during the simulation, with S plotted at each 10 increments.	93
5.14	The stress distribution of the artery: (A) The von Mises stress S distribution. (B) Maximum principal stress σ . (C) The shear stress τ_{xy} . (D) The strain energy density SED . (E) The maximum principal strain ε	94
5.15	The stress distribution at interface Γ^C : (A) The von Mises stress with a COF $\mu = 0.01$, where the cross-section $z = 0.1$ and $y = -3.35$ are highlighted by the red and yellow frames, respectively. (B) The von Mises stress of the cross-section $z = 0.1$ and the nodes used to obtain detailed information	95
5.16	The stress distribution at interface Γ^C : (A) The von Mises stress S . (B) The maximum principal stress σ . (C) The maximum principal strain ε . (D-F) The shear stresses τ_{xy} , τ_{xz} , and τ_{yz} , respectively.	96
5.17	The stress distribution obtained from the nodes: (A-B) 3D plots of the von Mises stress and the shear stress τ_{xy} , with the x and y coordinates aligned along the nodes, respectively. (C-D) Von Mises stress, S , along the x - and y -directions, respectively. (E-F) The shear stress τ_{xy} along the x - and y -directions, respectively.	98
5.18	Shear stress along the nodes: (A-B) Shear stress τ_{xz} along the x - and y -directions, respectively. (C-D) show the shear stress τ_{yz} along the x - and y -directions, respectively.	99
5.19	The stress distribution in struts: (A) The von Mises stress S . (B-D) The shear stress τ_{xy} , τ_{xz} and τ_{yz} , respectively.	100
5.20	The maximum von Mises stress S : (A) S decreases with increasing COF. (B) S increases with increasing the lumen diameter D_{lumen}	100

5.21	The average values of maximum and minimum shear stress, along with the SD, given by nodes at interface Γ between $z = 0$ and $z = 0.1$, with COF of 0.01 (dark) and 0.1 (light): (A-B) The results of shear stress τ_{xu} , where (A) represents the average values of maximum shear stresses and (B) represents the average values of minimum shear stresses. (C-D) The shear stress τ_{xz} . (E-F) The shear stress τ_{yz}	101
5.22	The average values of maximum von Mises stresses presented for Convex-1 (grey), Convex-2 (yellow), and Concave-1 (green) at COF of 0.01 and 0.03, using a flat surface (red) as the reference: (A-B) Convex-1 with the flat surface, where (A) represents the results at $\mu = 0.01$ and (B) represents the results at $\mu = 0.03$. (C-D) Convex-2 with flat surface. (E-F) : Concave-1 with flat surface.	103
5.23	The average values of maximum and minimum shear stresses for Convex-1 (grey), Convex-2 (yellow) and Concave-1 (green) at COF of 0.01, using a flat surface (red) as the reference: (A-B) Results of Convex-1 with the flat surface, where (A) represents the average maximum values and (B) represents the average minimum values. (C-D) Convex-2 with flat surface. (E-F) : Concave-1 with flat surface.	104
5.24	The average values of maximum and minimum shear stresses for Convex-1 (grey), Convex-2 (yellow), and Concave-1 (green) at COF of 0.03, using a flat surface (red) as the reference: (A-B) Results of Convex-1 with the flat surface, where (A) represents the average maximum values and (B) represents the average minimum values. (C-D) Convex-2 with flat surface. (E-F) : Concave-1 with flat surface.	105

List of Tables

2.1	Principal isotropic hyperelastic constitutive equations	13
2.2	Anisotropic hyperelastic models	15
2.3	The indentation test setup used for skin and artery	22
2.4	Influencing factors and their effect on friction	27
2.5	Arterial friction tests	29
3.1	Three Ogden materials used to assess differences in calculated stress and strain fields	48
4.1	Values of κ and χ for the spherical indenter used in this research ($\nu = 0.5$) [127]	59
4.2	The material parameters for mouse arterial tissue obtained from 26 measurements	71
4.3	Shear moduli and elastic moduli reported in literature	73
4.4	Fitting procedure for Ogden model parameters	76
5.1	Material properties of artery components	83
5.2	Material properties of SYNERGY TM BP Stent [292]	85
6.1	The indentation test on skin and arterial tissue	111
6.2	Shear moduli and elastic moduli reported in literature	112

1

Introduction

Contents

1.1 Background	1
1.2 The Objective of the Research	2
1.3 This Thesis	3

1.1 Background

Medical devices are crucial in preventing, diagnosing, and treating various diseases and medical conditions. However, despite their significant benefits, it is important to recognize that these devices can sometimes cause unintended tissue injuries during deployment and use. For example, during the COVID-19 pandemic, prolonged use of personal protective equipment by healthcare workers led to skin injuries ranging from minor irritations to pressure ulcers [1]. Similarly, in cardiology, in-stent restenosis remains a significant concern, occurring at rates of approximately 30% at six months for bare-metal stents and 10% at four years for drug-eluting stents [2, 3].

Biological tissues maintain their integrity through a delicate balance, known as homeostasis, which is achieved via ongoing processes of growth and remodelling. When this balance is disrupted, tissues can become injured. The introduction of a medical device into the human body can alter both mechanical and biochemical environments, significantly increasing the likelihood of disrupting homeostasis. Such disruptions can lead to a range of complications, from localised inflammatory responses to broader impacts on tissue function and integrity.

Considering the interaction between arterial tissues and medical devices, the arterial layers expe-

rience tension as they expand with the stent, while the intima comes into contact with the stent struts. Any material modelling efforts should simultaneously account for both tension and contact. That material model can then be integrated into finite element simulations to better understand the response of the artery to the use of stents. However, there are multiple challenges in such a procedure, both in terms of material modelling and performing the numerical simulations. Therefore, this research begins by modelling fundamental contact behaviour before progressing to full interaction simulations, with a focus on the hyperelastic effects of soft tissue. The material parameters for the artery in the numerical simulations are defined based on values from Holzapfel's studies, while the modelling and numerical investigations considering both contact-related and tension-related parameters will be further explored in future research.

To achieve this, a multidisciplinary approach is employed, integrating mathematical modelling, experimental testing, and numerical simulations. This approach addresses challenges such as nonlinear material responses and large deformations during complex contact and friction interactions. By leveraging insights from various disciplines, the research aims to develop more robust and accurate models for studying soft tissue–medical device interactions.

1.2 The Objective of the Research

The objective of this research focuses on modelling the nonlinear contact behaviour of soft tissue, including skin and arterial tissue, and conducting finite element simulations to mimic the artery-stent interaction. The consideration to begin with skin tissue arises from its ease of access, measurement, and characterisation, whereas arterial tissue presents greater challenges in terms of accessibility and modelling. An additional benefit of focusing on skin in the initial stages of this project is its relevance to ongoing work in our research group on the interaction of skin with respiratory devices [1], [4–7].

Chapter 3 explores the nonlinear contact mechanics of artificial skin tissue and proposes a newly developed method to account for the hyperelastic properties of thin-layer tissue by combining indentation testing with numerical simulations. Following validation of this method using artificial skin tissue, Chapter 4 proceeds to the study of mouse arterial tissue, further developing the proposed method to account for the material's roughness and surface irregularities.

Chapter 5 applies computational methods to predict the outcomes of patient-specific angioplasty treatments. The proposed approach provides insights into the development of mechanical stresses during stent-artery interactions and offers a computational assessment of the stent surface shape.

The research objectives are summarized below:

1. Develop a contact model incorporating the hyperelastic properties of thin-layer tissue (Chapter 3);
2. Application of the developed model to mouse arterial tissue (Chapter 4);
3. The finite element simulation for artery-stent interaction, considering the friction and the surface shape effect of the stent (Chapter 5).

1.3 This Thesis

This thesis begins with a background in Chapter 2, outlining the mathematical framework for soft tissue interactions. It covers the material properties of soft tissues, constitutive models for skin and arterial tissue, contact models with and without friction, and the fundamentals of the finite element method. Chapters 3 and 4 focus on the contact modelling of soft tissue. Specifically, Chapter 3 develops a method for assessing the hyperelastic properties of thin tissues using artificial skin tissue. Chapter 4 applies this method to mouse arterial tissue, addressing challenges posed by the tissue's rough surface and its effect on thickness determination for model application. Chapter 5 establishes the finite element model for artery-stent interactions, investigating the effects of friction and the stent's surface shape during these interactions. Chapter 6 presents the research findings, compares them with existing literature, and offers recommendations, followed by a concise conclusion in Chapter Conclusions to wrap up the thesis.

2

Background

Contents

2.1 Histology and Mechanical Behaviour of Skin and Arterial Tissue	5
2.1.1 Fundamental Structural and Material Properties of Skin	6
2.1.2 Fundamental Structural and Material Properties of Arterial Tissue	7
2.2 Constitutive Models for Skin and Arterial Tissue	8
2.2.1 Basic Continuum Mechanics for Isotropic and Anisotropic Hyperelastic Tissue	8
2.2.2 Constitutive Equations for Hyperelasticity	10
2.3 Contact Modelling	16
2.3.1 Analytical Contact Model	16
2.3.2 Indentation of Tissue	20
2.3.3 Friction in Contact Problem	21
2.3.4 Friction of Tissue	24
2.4 Finite Element Method for Contact Problem	28
2.4.1 Fundamentals of Finite Element Method	29
2.4.2 Finite Element Modelling in ABAQUS	32
2.4.3 Contact Modelling in ABAQUS	36
2.4.4 Indentation Simulation on Tissue	39

This chapter provides an overview of two key topics: the contact behaviour of skin and arterial tissue, and numerical simulations of contact interactions, along with a review of relevant research in these areas.

2.1 Histology and Mechanical Behaviour of Skin and Arterial Tissue

Skin and arterial tissue, as biological materials, share similar mechanical properties, including hyperelasticity, viscoelasticity, anisotropy, and a multi-layered structure [8], [9]. These properties

significantly influence their interactions with medical devices [10], [11]. This section provides a comprehensive description of the histology and mechanical properties of both skin and arterial tissue.

2.1.1 Fundamental Structural and Material Properties of Skin

Skin tissue has a complex structure that functions as a protective barrier against harmful microbes and the external environment.

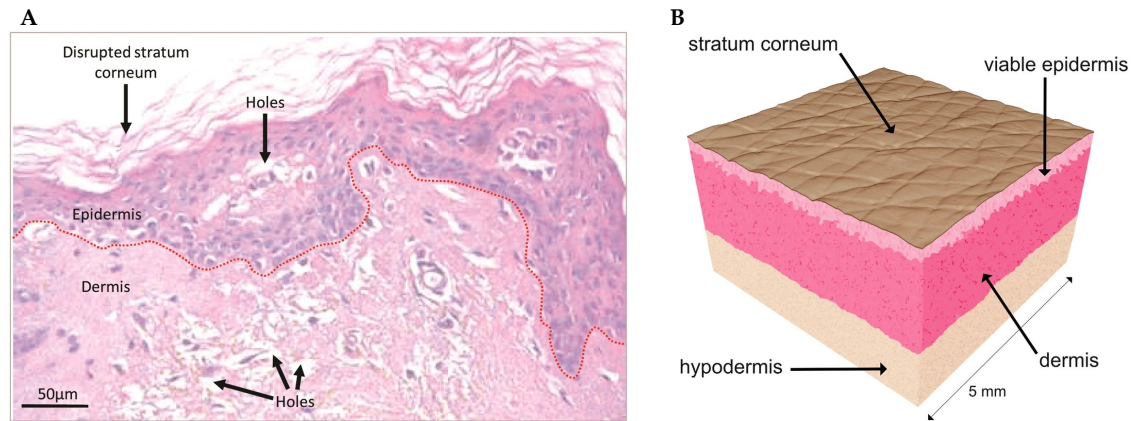


Figure 2.1: Skin tissue: (A) H & E stained human skin biopsy. The dermal-epidermal junction is denoted by a red dotted line [12]. (B) Schematic representation of the multi-layer nature of the skin featuring the epidermis (constituted of the stratum corneum and viable epidermis), dermis, and hypodermis [13].

Skin tissue consists of two distinct layers: the epidermis and the dermis, as illustrated in Figure 2.1 [13]. The epidermis, the outermost layer of the skin, is approximately 200 µm thick and comprises the stratum corneum and the living epidermis. The stratum corneum is a tightly bound layer of corneocytes (dead, flattened keratinocytes) that protects against mechanical stress from abrasion and penetration. The living epidermis contains other cell types, such as melanocytes [13–15] and living keratinocytes that migrate outwards and eventually replace the cells in the stratum corneum. Beneath the epidermis lies the dermis, which is roughly 15 to 20 times thicker and is characterised by a rich extracellular matrix (ECM) composed of collagen, elastin, and fibrillin proteins [13], [16]. These components are intertwined within a continuous, amorphous network, that provides the epidermis with structural support and elasticity [16]. The interface between the epidermis and the dermis, known as the dermal-epidermal junction (DEJ), forms a wavy, finger-like boundary.

The complex arrangement of collagen fibres and their mechanical interactions with elastin allow the dermis to primarily support skin and exhibit hyperelastic and anisotropic behaviour. Nu-

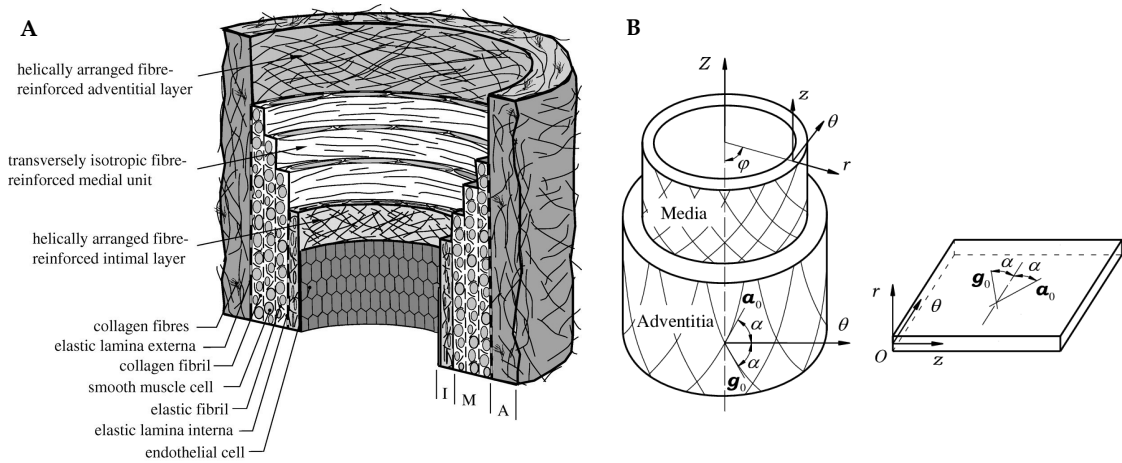


Figure 2.2: Arterial tissue: (A) A healthy elastic artery composed of three layers: intima, media, and adventitia. The intima is the innermost layer consisting of a single layer of endothelial cells that rests on a thin basal membrane and a subendothelial layer whose thickness varies with topography, age, and disease. The media is composed of smooth muscle cells, a network of elastic and collagen fibrils, and elastic laminae which separate media into several fibre-reinforced layers. The primary constituents of adventitia are thick bundles of collagen fibrils arranged in helical structures. The adventitia is the outermost layer surrounded by loose connective tissue [8]. (B) Orientations of fibre families of an artery (without intima) and a strip for tension test [28].

merous studies have assessed the hyperelastic and anisotropic properties of skin through various experiments, including tension [17–21], indentation [22], compression [23], and suction [13]. Additionally, the skin’s hierarchical structure and fluid phase contribute to its viscoelastic properties, leading to behaviours such as creep, stress relaxation, hysteresis, and strain rate dependency [13], [23–26]. This research focuses exclusively on the influence of hyperelastic properties on the interactions between biological tissue and medical devices.

2.1.2 Fundamental Structural and Material Properties of Arterial Tissue

Arterial tissue is primarily classified into two categories: elastic arteries and muscular arteries. Elastic arteries, such as the aorta, carotid arteries, and iliac arteries, are situated closer to the heart and have larger diameters. In contrast, muscular arteries, including the femoral, celiac, and cerebral arteries, are located at the periphery of the circulatory system and have relatively smaller diameters [27]. In this research, the two types of arteries are not differentiated; rather, the focus is on their shared morphological and general characteristics.

A healthy artery is structured with three distinct layers: intima, media, and adventitia, arranged from innermost to outermost (Figure 2.2 [8]). The intima consists of a single layer of endothelial cells, which may be accompanied by a subendothelial layer. Since the endothelial layer contributes minimally to load-bearing, the intima has a limited effect on the overall mechanical properties of

a healthy artery [29–32]. The media is composed of a complex, three-dimensional network of smooth muscle cells, elastin, and collagen fibrils [33]. These components work together to form a continuous fibrous helix that provides structural integrity to the media layer [31]. The adventitia is primarily composed of fibroblasts and fibrocytes, along with thick bundles of collagen fibres embedded in a histological ground matrix and surrounded by loose connective tissue. The collagen fibres within the adventitia are organised into two helical families, showing significant deviations from their average orientations [30, 31, 34].

Compared to the thin intima, the media and the adventitia contribute far more to load-bearing in both the longitudinal and circumferential directions due to their embedded fibre structures. These fibre structures provide considerable strength and resilience under load, while their dispersion contributes to the anisotropic hyperelastic properties of the arterial wall. Given that arteries are cylindrical, their anisotropic behaviour can be modelled as cylindrically orthotropic, as proposed by Patel et al. [35]. This modelling approach has been widely adopted in subsequent research [8, 31, 33, 36].

However, with ageing and pathological changes, clinical conditions such as atherosclerosis can alter the structure of arteries. In an atherosclerotic artery, the intima becomes thicker and stiffer [37, 38], significantly affecting its mechanical properties and transforming it into a more mechanically resistant layer. These changes reduce the artery’s functional capacity and resilience, impair its ability to accommodate blood flow effectively and increase the risk of cardiovascular events. Additionally, while arterial tissue demonstrates viscoelasticity and heterogeneity [39]; these factors are not addressed in this research.

2.2 Constitutive Models for Skin and Arterial Tissue

While this research focuses on the hyperelastic properties of skin and arterial tissue in modelling their contact behaviour, this section reviews the constitutive models used to characterise these properties in both tissue.

2.2.1 Basic Continuum Mechanics for Isotropic and Anisotropic Hyperelastic Tissue

Hyperelasticity refers to the near-perfect elasticity exhibited by materials under large deformations. Unlike linear elasticity, which follows a linear stress-strain relationship defined by Hooke’s Law (Eq. 2.1), hyperelasticity involves strain hardening at large deformations, as illustrated in Figure 2.3.

$$\sigma_{ij} = C_{ijkl} \varepsilon_{kl} \quad (2.1)$$

where σ_{ij} is the stress, ε_{kl} is the strain, and C_{ijkl} is the stiffness matrix.

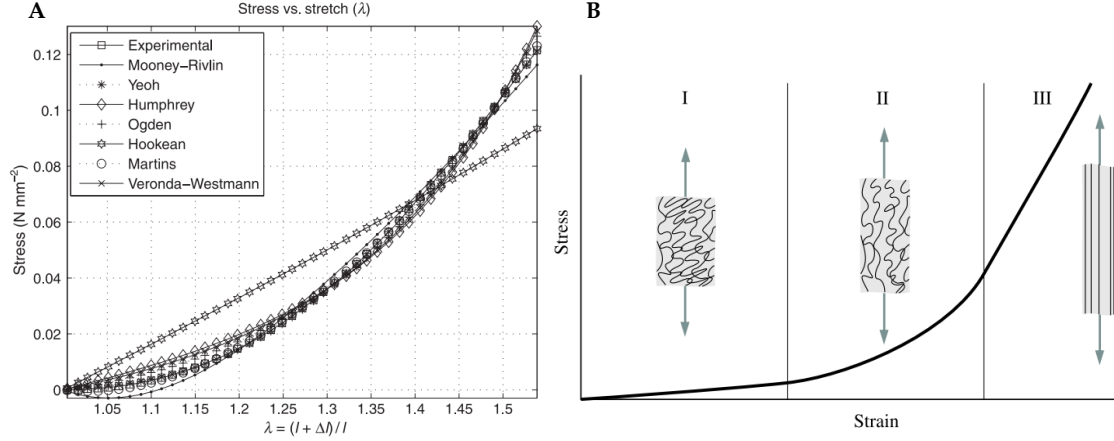


Figure 2.3: Hyperelastic properties of soft tissue: (A) Hyperelastic properties of pig muscular tissue shown by different hyperelastic models [40]. (B) Schematic diagram of a typical (tensile) stress-strain curve for skin showing the associated collagen fibre morphology [8].

Instead of directly defining the stress-strain relationship, hyperelastic constitutive models are typically formulated using a strain-energy function, denoted as ψ . This function describes the material's stored energy per unit volume as a function of deformation, $\psi(\mathbf{C})$, and is expressed in terms of principal stretches λ_j ($j = 1, 2, 3$) or strain invariants I_j ($j = 1, 2, 3$, see Eq. 2.2) [8], [13].

$$\begin{aligned} I_1 &= \text{tr} \mathbf{C} = \lambda_1^2 + \lambda_2^2 + \lambda_3^2, \\ I_2 &= \frac{1}{2}[(\text{tr} \mathbf{C})^2 - \text{tr} \mathbf{C}^2], \\ I_3 &= \det(\mathbf{C}), \end{aligned} \quad (2.2)$$

where 'tr' is the trace operator, and 'det' is the determinant operator. The right Cauchy-Green tensors $\mathbf{C}$ is given by $\mathbf{C} = \mathbf{F}^T \mathbf{F}$, where $\mathbf{F}$ is the deformation gradient tensor, defined as $\mathbf{F}(\mathbf{X}) = \partial \chi(\mathbf{X}) / \partial \mathbf{X}$. The function χ describes the deformation that maps a material point $\mathbf{X} \in \Omega_0$ to its new position $\mathbf{x} = \chi(\mathbf{X})$.

Soft tissue is often treated as an incompressible material [33], [41], [42], which implies the constraint $I_3 = \det(\mathbf{C}) = 1$ is maintained. As a result, the strain-energy function ψ depends only on invariants I_1 and I_2 .

2.2.1.1 Modelling of Fibre Dispersion

To characterise the mechanical contributions of the embedded collagen fibres, a widely accepted

assumption—consistent with physical observations—is that these fibres are distributed around a principal orientation, denoted as $\mathbf{a}_0$, with a certain probability of deviating within a specified range of angles from this direction [13]. Two prominent approaches for modelling this distribution are ‘angular integration’ (Eq. 2.3, 43) and ‘generalised structure tensor’ (Eq. 2.4, 31).

- *Angular integration*

$$\psi_{\text{bundle}} = n \int_{U^2} \rho(\mathbf{v}_0) \psi_{\text{fibre}}(\lambda) dA_c \quad (2.3)$$

where $\psi_{\text{fibre}}(\lambda)$ is the strain energy function of a fibre in terms of the fibre stretch λ ; ψ_{bundle} is the strain energy function of a bundle of these fibres per unit sphere U^2 ; dA_c is the elemental unit sphere surface area; n is the number of fibres; $\mathbf{v}_0$ is the direction of an arbitrary fibre, defined with respect to $\mathbf{a}_0$, and $\rho(\mathbf{v}_0)$ is the density function describing the relative angular distribution of fibres around the mean direction. The normalisation condition $\frac{1}{4\pi} \int_{U^2} \rho(\mathbf{v}_0) dA_c = 1$ must be satisfied.

- *Generalised structure tensor*

$$\psi_{\text{bundle}} = \psi_{\text{bundle}}(\mathbf{C}, \mathbf{H}) \quad (2.4)$$

where $\mathbf{H}$ is a homogenised structure tensor defined by integrating the structure tensor $\mathbf{v}_0 \otimes \mathbf{v}_0$, associated with an arbitrary fibre direction $\mathbf{v}_0$, over the unit sphere U^2 (Eq. 2.5). The normalisation condition $\mathbf{H} : \mathbf{I} = 1$ must be satisfied.

$$\mathbf{H} = \frac{1}{4\pi} \int_{U^2} \rho^{\mathbf{a}_0}(\mathbf{v}_0) \mathbf{v}_0 \otimes \mathbf{v}_0 dA_c \quad (2.5)$$

where ρ is the density function for the fibre family.

By incorporating fibre dispersion, the anisotropic hyperelastic model is expressed as the sum of the isotropic potential energy ψ_g for the non-collagenous ground matrix and the anisotropic potential energy ψ_{bundle} for the embedded collagen fibre families (Eq. 2.6).

$$\psi = \psi_g(\mathbf{C}) + \psi_{\text{bundle}}(\mathbf{C}, \mathbf{H}) \quad (2.6)$$

2.2.2 Constitutive Equations for Hyperelasticity

Numerous constitutive equations have been developed to describe the isotropic and anisotropic hyperelastic properties of soft tissue, including skin and arterial tissue. These models can be

broadly categorised into three main types: phenomenological, structural, and structurally-based phenomenological models [13], [44]. Phenomenological models, such as Neo-Hookean model (Eq. 2.7, [45]), Mooney–Rivlin model (Eq. 2.8, [46], [47]), Ogden model (Eq. 2.11, [48]) and Yeoh model (Eq. 2.10, [49]–[51]), capture the macroscopic behaviour of the tissue without accounting for the specific behaviour of its microstructural constituents and their interactions [13], [52]. However, the parameters of phenomenological models often lack direct physical interpretation, limiting their flexibility for studying mechanistic structural effects.

Structural models represent tissue as a composite material composed of defined microstructural elements, with interactions specified through appropriate equations. Unlike purely phenomenological models, structural models often use parameters that have direct physical meanings and can be experimentally measured. For instance, in models involving collagen fibres, parameters such as elastic modulus, waviness, and fibre dispersion are explicitly defined and physically interpretable. A key challenge of these models lies in obtaining precise information about the geometric and material properties of microstructural elements, as well as their spatial and interfacial arrangements, which poses significant experimental difficulties. Advances in technology and new techniques are gradually addressing these challenges. Additionally, finite element (FE) simulations of structural models spanning multiple length scales require extremely fine mesh densities to accurately capture component geometries. Despite improvements in computational power, these simulations can still be computationally demanding.

A middle ground between these approaches is represented by structurally-based phenomenological models, which integrate characteristics of both phenomenological and structural models, allowing for the incorporation of tissue structure into mechanical descriptions. Notable examples include models developed by Holzapfel et al. [8], Humphrey [53], and Lanir [43], [54].

2.2.2.1 Constitutive Equations for Isotropic Hyperelasticity

- *Neo-Hookean model*

$$\psi = \frac{G}{2}(I_1 - 3) \quad (2.7)$$

where G is the shear modulus in the ground state.

- *Mooney-Rivlin model*

$$\psi = C_1(I_1 - 3) + C_2(I_2 - 3) \quad (2.8)$$

where C_1 and C_2 are the material constants, which can be determined experimentally or by fitting to experimental data.

A strain energy function, as defined in Eq. 2.9 generalises the Mooney-Rivlin model.

$$\psi = \sum_{m,n=0}^{\infty} C_{mn} (I_1 - 3)^m (I_2 - 3)^n \quad (C_{00} = 0) \quad (2.9)$$

where C_{mn} are the material constants. When $m = 0$, the strain energy function reduces to the Yeoh model.

- *Yeoh model*

$$\psi = \sum_{j=1}^3 C_j (I_j - 3)^j \quad (2.10)$$

where C_j ($j = 1, 2, 3$) are the material constants.

- *Ogden model*

$$\psi = \sum_{p=1}^N \frac{\mu_p}{\alpha_p} (\lambda_1^{\alpha_p} + \lambda_2^{\alpha_p} + \lambda_3^{\alpha_p} - 3) \quad (2.11)$$

where N is the order of the Ogden model, which depends on the material properties and the required level of accuracy for the analysis. p is the term number, and μ_p and α_p are material parameters. The relationship of the shear modulus G with Ogden parameters μ_p and α_p is defined as $2G = \sum_{p=1}^N \mu_p \alpha_p$.

Among isotropic hyperelastic models, the Neo-Hookean model is notable for its simplicity and effectiveness in capturing hyperelastic behaviour under small to moderate deformations. It has been widely employed to characterise the material properties of biological tissue, including skin (see, e.g., [55]–[57]) and arterial tissue (see, e.g., [56], [58]–[60]). However, some studies such as [61]–[63] note that the Neo-Hookean model is not effective for biological tissue under large deformations in tensile tests. In contrast, the Mooney-Rivlin model provides a better fit for experimental data under relatively larger deformations, making it more appropriate for modelling skin [61], [62], [64], [65] and arterial tissue [66]–[70]. Additionally, the Yeoh model is widely utilised in hyperelastic modelling of both skin and arterial tissue (e.g., [40], [62], [71]–[73]) due to its simplicity relative to the more complex polynomial equation (Eq. 2.9).

Unlike Neo-Hookean, Mooney-Rivlin, and Yeoh models, the Ogden model does not rely on invari-

ants I_j of the deformation tensor but instead utilises the principal stretch ratios λ_j to characterise hyperelasticity. This alternative formulation enhances the Ogden model's versatility, enabling it to accurately represent nonlinear hyperelastic behaviour, even under quite large deformations. By adjusting its parameters μ_p and α_p , or by incorporating additional terms, the Ogden model can capture a wide range of nonlinear properties. It has been effectively applied to both skin (e.g. [74]–[76]) and arterial tissue (e.g., [77]–[81]).

Table 2.1 lists additional widely used hyperelastic models. The Fung-Demiray model [82], [83] and the Veronda-Westmann model [52], commonly known as exponential models, are discussed in studies [53], [84]–[87]. The Arruda-Boyce and the Gent models are both based on non-Gaussian statistical descriptions of the molecular structure of material. A detailed comparison of these two models is provided in [88], with further applications in [89]. Bischoff et al. successfully applied the Arruda-Boyce model to describe the mechanical properties of skin tissue [90], [91]. For more comprehensive reviews of these and other hyperelastic models, see [13], [92], [93].

Table 2.1: Principal isotropic hyperelastic constitutive equations

Model	Equation
Fung-Demiray model [82], [83]	$\psi = \frac{c_1}{2c_2} (\exp(c_2(I_1 - 3)) - 1)^{[1]}$
Veronda-Westmann model [52]	$\psi = c_1 (\exp(c_3(I_1 - 3)) - 1) - c_2(I_2 - 3)$
Arruda-Boyce model [94]	$\psi = G \sum_{i=1}^N \frac{c_i}{\lambda_m^{2c_i-2}} (I_1^i - 3^i)^{[2]}$
Gent model [95]	$\psi = -\frac{GJ_m}{2} \ln \left(1 - \frac{I_1 - 3}{J_m} \right), \quad I_1 < J_m + 3^{[3]}$

[1] $c_1 > 0$ is the infinitesimal shear modulus; $c_2 > 0$ is the stiffening parameter;

[2] λ_m is a limiting chain extensibility, there are $c_1 = \frac{1}{2}$, $c_2 = \frac{1}{20}$, $c_3 = \frac{11}{1050}$, $c_4 = \frac{19}{7000}$, $c_5 = \frac{519}{673750}$;

[3] $J_m > 0$ are material constants.

2.2.2.2 Constitutive Equations for Anisotropic Hyperelasticity

- *Gasser-Ogden-Holzapfel model*

The Gasser-Ogden-Holzapfel (GOH) model [8], [31], [33], [96] is a widely used constitutive model for modelling the anisotropic hyperelasticity of soft tissue, especially for arterial tissue. It combines an isotropic term representing the ground matrix with an anisotropic term that captures the contribution of the collagen fibre network. This formulation allows for an accurate representation of the complex mechanical behaviour of arterial tissue, particularly in response to large deformations, by incorporating both the matrix's elasticity and the directional stiffness of the embedded fibres.

$$\psi = \frac{G}{2} (I_1 - 3) + \sum_{i=1}^2 \frac{k_{i1}}{k_{i2}} (\exp k_{i2} [\mathbf{H}_i : \mathbf{C} - 1]^2 - 1) \quad (2.12)$$

where G is a shear modulus defining a neo-Hookean model for the ground matrix, and k_{i1} and

k_{i2} ($i = 1, 2$ represent two fibres) are respectively stress-like and unit-less scaling parameters for embedded fibres. The structure tensor $\mathbf{H}_i$, given by Eq. 2.5, accounts for fibre dispersion with i represents fibre family.

The GOH model assumes that the fibres in each fibre family are distributed with rotational symmetry around a mean referential (preferred) direction, denoted as $\mathbf{a}_0^i$ ($i = 1, 2$), as illustrated in Figure 2.2B. This assumption leads to a transversely isotropic behaviour of arterial tissue, where the material exhibits different mechanical properties in the direction of the fibres and the transverse directions. Consequently, the structure tensor $\mathbf{H}_i$, which describes the distribution of the fibres, can be simplified, as shown in Eq. 2.13.

$$\begin{aligned}\mathbf{H}_i &= \kappa_i \mathbf{I} + (1 - 3\kappa_i) \mathbf{a}_0^i \otimes \mathbf{a}_0^i = \kappa_i \mathbf{I} + (1 - 3\kappa_i) \mathfrak{B}_0^i \\ \kappa_1 &= \frac{1}{4} \int_0^\pi \rho(\Theta) \sin^3 \Theta d\Theta\end{aligned}\quad (2.13)$$

where κ_i is the dispersion parameter representing the distribution of the fibres within the tissue, $\mathfrak{B}_0^i = \mathbf{a}_0^i \otimes \mathbf{a}_0^i$, and $\rho(\Theta)$ is the density function for a specific distribution of fibre orientations. Here, Θ denotes the angle between $\mathbf{a}_0^1$ and the Cartesian coordinate axis $\mathbf{e}_1$.

As outlined in [33], the fibre distribution follows a von Mises distribution, described by the density function as Eq. 2.14

$$\rho(\Theta) = 4\sqrt{\frac{b}{2\pi}} \frac{\exp[b(\cos(2\Theta) + 1)]}{\text{erfi}(\sqrt{2b})} \quad (2.14)$$

where Θ is the angular orientation of the fibres, and b is a parameter controlling the concentration of the fibres around the mean direction. The von Mises distribution is periodic with period π , and it characterises how fibres are distributed concerning the preferred orientation, as illustrated in Figure 2.4A and B.

When $\kappa_i = 0$, the structure tensor $\mathbf{H}_i$ simplifies to the classical structure tensor $\mathfrak{B}_0^i$, which represents the case where all fibres are perfectly aligned with the preferred direction, i.e., no fibre dispersion. On the other hand, when κ_i reaches its maximum value of $1/3$, the fibre network is isotropically distributed, reflecting a symmetric, non-preferred fibre orientation throughout the tissue. This flexibility allows the GOH model to capture a wide range of fibre distributions, from perfectly aligned to fully isotropic.

Holzäpfel et al. provided constitutive parameters for the GOH model by fitting tension tests on each layer of an iliac artery [97]. This established material model was subsequently used for FE

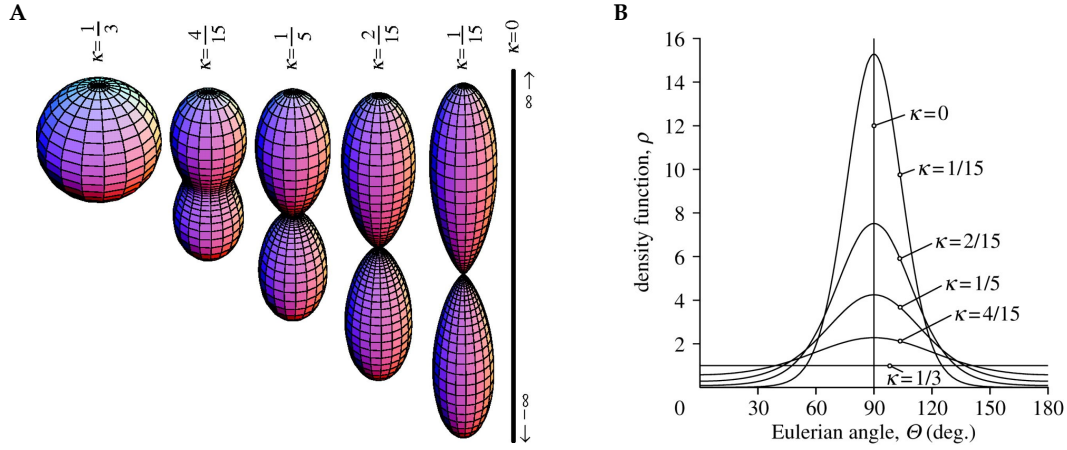


Figure 2.4: (A) Three-dimensional graphical representation of the orientation of the collagen fibres based on the transversely isotropic density function [33]. (B) Two-dimensional graphical representation of the (transversely isotropic) von Mises distribution of the collagen fibres [33].

investigations of artery-stent interactions in their later research [31], [96], [98], [99], where the geometric model of the artery was derived from high-resolution magnetic resonance images and reconstructed using non-uniform rational B-splines. The FE model of the artery is also referenced in Chapter 5 of this research to model the artery-stent interaction. Additionally, the GOH model has been applied to characterise human skin in studies by Annaidh [17], [100] and Tonge [101], [102].

For the other anisotropic models listed in Table 2.2, the Fung and Tong model provides a general form of hyperelastic modelling based on Green–Lagrange strain tensor components, with variations developed by researchers such as Humphrey [27], [53], Takamizawa [103], and Chuong and Fung [84]. The applications of these models on skin and arterial tissue are discussed extensively in [33], [53], [104]–[107]. Flynn et al. [75] highlighted that the Fung and Tong model more accurately captures the mechanical behaviour of skin compared to the two-term Ogden model.

Table 2.2: Anisotropic hyperelastic models

Model	Equation
Fung-Tong model	$\psi = c(\exp(A_{ijkl}E_{ij}E_{kl}) - 1)$ [1]
Lanir's model [43]	$\psi = (1 - V_c - V_e)\frac{G}{2}(I_1 - 3) + \sum_k V_k \sum_k R_k(\theta)\frac{k}{2}(\lambda - 1)^2$ [2]
Limbert-Middleton [108]	$\psi = \frac{1}{4} \sum_{k=1}^n \mu_k \left[\frac{1}{\alpha_k} \left(e^{(I_{[k]} - 1)^{\alpha_k}} \right) + \frac{1}{\beta_k} \left(J_{[k]}^{\beta_k} - 1 \right) + \frac{1}{\chi_k} \left(I_3^{-\chi_k} - 1 \right) \right]$ [3] $I_{[k]} = \sum_{i=1}^3 \omega_k^i (\mathfrak{B}_{0[k]}^i : \mathbf{C})$, $J_{[k]} = I_3 \sum_{i=1}^3 \omega_k^i \text{tr}(\mathbf{C}^{-1} \mathfrak{B}_{0[k]}^i)$

[1] where c and A_{ijkl} are material parameters.

[2] where G is the shear modulus; V_k , K_k , and R_k represent the volume fraction, fibre stiffness, and fibre orientation probability density function of collagen fibres ($k = c$) and elastin fibres ($k = e$), respectively.

[3] where μ_k , α_k , β_k , and χ_k are the material parameters.

Lanir's model is based on the 'angular integration' approach. Although Lanir did not validate

it with experimental data, subsequent studies have utilised it, such as those by [109], [110]. The Bischoff-Arruda-Grosh (BAG) model, an extension of the Arruda-Boyce model, incorporates different characteristic dimensions of the original cuboidal microscopic unit cell. Flynn et al. further extended the BAG model to explore various skin behaviours, including wrinkling [111], scar tissue contraction [112], and skin wrinkling/aging [113].

2.3 Contact Modelling

This section discusses contact problems, which describe the interactions between two bodies when they come into contact. The focus is on analytical modelling methods and experimental approaches for collecting data to establish a contact model. FE solutions for contact problems will be addressed in the subsequent section. In this research, viscoelasticity and friction were ignored to simplify the modelling process. Excluding viscoelastic effects allows us to concentrate on the instantaneous mechanical response—the hyperelastic behaviour of arterial tissue during short-term interactions with medical devices. Furthermore, incorporating external forces, such as friction between the arterial layers and stent struts, introduces significant modelling challenges. Given that existing approaches also neglect these forces, we opted to simplify our model accordingly. The expected effects of these assumptions are small, e.g., literature [114] noted that while the human abdominal aorta exhibits viscoelasticity, its fractional viscoelastic weighting is relatively low, making its behaviour closer to that of an elastic material. This simplification facilitates a more straightforward evaluation of deformation and stress distribution, ultimately providing valuable insights into the mechanical interactions involved.

2.3.1 Analytical Contact Model

There are numerous contact models tailored to various contact scenarios, such as those proposed by Hertz [115], [116], Lin [117], and Sneddon [118], each based on specific assumptions. Contact scenarios involving spherical indenters are illustrated in Figure 2.5, with the corresponding contact models discussed in the following sections.

These contact models offer valuable insights into the relationship between the contact force F and material parameters such as the elastic modulus E and Poisson's ratio ν , while also describing the geometric relationships among the contact radius a , the indenter radius R , and the indentation depth ω . As a result, these models are extensively used to determine material properties, analyse contact stresses, and evaluate performance in engineering design. Their applications span fields such as medical device development and clinical diagnostics, making them essential tools for both

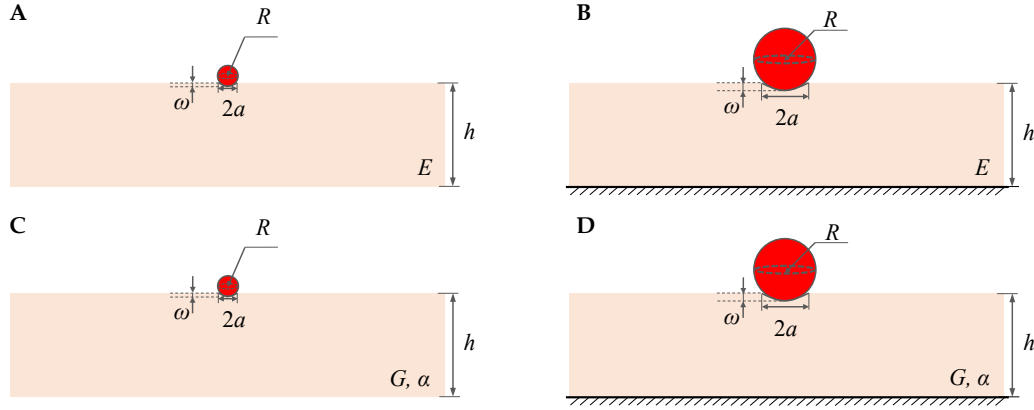


Figure 2.5: Contact scenarios: (A) A spherical indenter pressing into a linear elastic material with an infinite half-space, as described by models developed by Hertz and Sneddon. (B) Scenarios accounting for finite-thickness effects representing using models such as those developed by Hayes and Dimitriadis, which incorporate corrections for boundary effects in finite layers. (C) Scenarios incorporating hyperelastic properties but assuming an infinite thickness, as described by Lin's model. (D) This research investigates a scenario that incorporates both finite-thickness and hyperelastic properties.

research and practical applications.

- *Hertz's model*

Hertz's model (Eq. 2.15 [115]) describes the elastic deformation of two bodies in contact under the assumptions of a continuous frictionless surface, small deformation, and elastic half-space. The relationship between contact radius a , the indenter radius R , and the indentation depth ω is given by Eq. 2.16.

$$F = \frac{4ER^{\frac{1}{2}}\omega^{\frac{3}{2}}}{3(1-\nu^2)} \quad (2.15)$$

where ν is the Poisson's ratio.

$$a = \sqrt{R\omega} \quad (2.16)$$

- *Sneddon's model*

Sneddon developed comprehensive contact models for indenters with various profiles, including cones, cylindrical punches, and ellipsoidal shapes [118]. These models are based on the assumptions of small deformations, elastic half-spaces, and frictionless surfaces. For a spherical indenter, the contact model is given by Eq. 2.17.

$$F = \frac{G}{1-\nu} \left\{ (a^2 + R^2) \log \frac{R+a}{R-a} - aR \right\} \quad (2.17)$$

where G is the shear modulus of the indented material.

Hertz's model and Sneddon's model are both developed for the scenario illustrated in Figure 2.5A, which applies to elastic, isotropic materials occupying an infinite half-space. In this scenario, a rigid indenter is vertically loaded and brought into contact with the upper surface of the material. Compared to Sneddon's model and other models, Hertz's model has been widely applied to obtain the elastic modulus E of skin tissue [24], [119]–[122] and arterial tissue [123]–[125] in conjunction with indentation experiments, due to its ease of application. While Sneddon's model has also been used to assess the local properties of skin tissue [126], it has not been applied as frequently.

However, the application of Hertz's and Sneddon's models requires stringent experimental setups to meet their assumptions. First, the thickness of the indented material should be significantly larger than the contact radius a to satisfy the half-space assumption. Second, the indentation depth must be much smaller than the indenter's size to ensure that the strains within the material remain infinitesimal. Third, the indented material is supposed to be elastic. Lastly, adhesion and friction between the indenter and the material surface can be negligible.

Considering that soft tissue such as skin and arteries have thin thicknesses, consist of multiple layers, exhibit nonlinear material properties, and often experience large deformations, meeting the aforementioned requirements can be challenging without introducing errors. Therefore, for accurate characterisation, it is essential to consider factors such as finite thickness, large deformation, and nonlinear material behaviour. However, providing analytical solutions that incorporate these factors in contact problems is not straightforward. This process involves solving nonlinear equations associated with the nonlinear material properties and large deformations, as well as handling varying mixed boundary conditions imposed by the finite-thickness layers.

Contact models for the finite-thickness layers correspond to the scenario illustrated in Figure 2.5B. Hayes [127] addressed this scenario, formulating a contact problem in which cartilage, bonded to the subchondral bone, is indented by a spherical indenter. The problem is formulated using the Navier-Cauchy equation and solved using Fredholm integral equations to handle the mixed boundary conditions. This approach has been widely accepted for contact problems involving finite-thickness layers, as demonstrated in works such as [118], [128]–[131]. This method represents a complex mathematical procedure that entails extensive calculations, highlighting the analytical challenges of such problems.

- *Hayes' model*

$$F = \frac{2\kappa\sqrt{\chi}ER^{\frac{1}{2}}\omega^{\frac{3}{2}}}{(1-\nu^2)} \quad (2.18)$$

where E is the elastic modulus, κ and χ are the fixing parameters for the boundary conditions, given by:

$$\kappa = \frac{F(1-\nu)}{4aG\omega_0} = \int_0^1 \omega(\tau)d\tau, \quad \chi = \frac{a^2}{\omega_0 R} \quad (2.19)$$

where ω_0 is the applied the displacement. $\int_0^1 \omega(\tau)d\tau$ is a Fredholm integral equation of a second kind with the symmetric kernel. The values of κ and χ are provided in Table 2 of Hayes [127], for $\nu = 0.30, 0.35, 0.40, 0.45$, and 0.50 . These values of κ and χ may be used to calculate the relationships between the load F and the displacement ω , and contact region radius a for an indenter of radius R on a layer of thickness h , shear modulus G , and Poisson's ratio ν .

Alternative approaches to solving the finite-layer contact problem include the model proposed by Dimitriadis [132], which employs Green's functions. Compared to the Hayes model, Dimitriadis' model provides an explicit expression applicable within the range $h/d \leq 0.1$. Nevertheless, most studies on finite-layer contact rely on correction functions derived from finite element (FE) simulation data [133], as these offer greater flexibility and accuracy across a broader range of conditions.

Compared to Hertz's model, contact models that consider the finite-thickness effect, such as Hayes' model and the Dimitriadis' model, provide a more flexible experimental setup. The half-space assumption is not necessary, meaning the indented material does not have to be significantly larger than the contact radius a . Consequently, the indenter radius R can be larger, allowing for greater indentation depths. This flexibility effectively reduces errors induced by the rough surfaces of biological tissue. A more detailed demonstration of contact modelling for arterial tissue will be provided in Chapter 5.

- *Lin's model*

The effect of hyperelastic properties is addressed in the model established by Lin et al. [117], which describes the scenario illustrated in Figure 2.5C. This model assumes that the mean stress σ^* and strain ε^* during the indentation follow the stress and strain relationships defined by uniaxial compression. The contact model for the material, defined by a first-order Ogden model, is represented by Eq. 2.20.

$$F = \frac{B\pi a^2}{\alpha} \left[\left(1 - 0.2\frac{a}{R}\right)^{-\alpha/2-1} - \left(1 - 0.2\frac{a}{R}\right)^{\alpha-1} \right] \quad (2.20)$$

where α is the Ogden parameter, and R is the indenter radius. B is the material parameter given by Eq. 2.21.

$$B = \frac{40E}{9\pi(1 - \nu^2)} \quad (2.21)$$

Lin's model provides valuable insights into incorporating hyperelastic properties. However, it shares the same limitations as Hertz's model, requiring the use of small indenters with shallow indentation depths in experimental setups. Consequently, it does not address the errors introduced by the surface roughness of biological tissue.

Among the contact models discussed, except for Hertz's model, the others are not widely applied to skin and arterial tissue. Furthermore, numerous models exist for indenters of various shapes, which will be introduced in the next subsection. In this research, we focus exclusively on the contact model for spherical indenters and propose a novel method for the scenario illustrated in Figure 2.5D in Chapter 3. This method incorporates hyperelastic properties defined by the Ogden model, addresses the challenges posed by finite-space assumptions, accommodates relatively large deformations, and will be applied to the contact problem of mouse arteries in Chapter 4.

2.3.2 Indentation of Tissue

The indentation tests, illustrated in Figure 2.6, are normally used to collect data of contact force F versus displacement depth ω . A variety of instruments can be used for indentation tests, ranging from Atomic Force Microscopes, capable of applying piconewton (pN) loads, to nanoindentation devices that resolve nanonewton (nN) loads, and larger industrial indenters capable of applying micro-(μ N) to meganewton (mN) loads. In addition to spherical indenters, other types of indenters such as conical indenters, cylindrical flat punches, and diamond indenters are commonly used in indentation tests. Detailed information on these indenters can be found in the work of Ebenstein and Pruitt [134], as well as Gouldstone et al. [135]. The specific research objective and contact model will determine the most appropriate instrument and indenter. Table 2.3 lists the indenters used on skin and arterial tissue.

Among the various indenters and indentation tests conducted on skin and arteries, the spherical indenter and the cylindrical flat punch are the most widely used, as demonstrated in Table 2.3. In [132], it was reported that the estimated radius of curvature of the pyramidal tip can induce significant errors in estimating the elastic modulus, E , particularly for thin samples. A similar observation was made in [136], where they compared the elastic modulus obtained using a pyramidal indenter with that from a spherical indenter with radii $R = 1 \mu\text{m}$ and $2.5 \mu\text{m}$. The study found

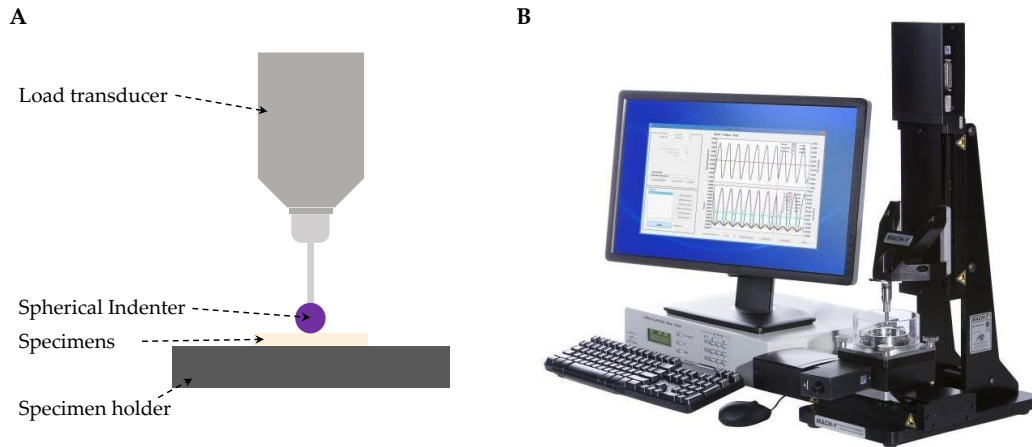


Figure 2.6: Indentation test: (A) The schematic of the indentation test conducted for mouse arterial tissue in Chapter 5, where a spherical indenter with a radius of $R = 150 \mu\text{m}$ is used. The indentation setup consists of several key components: the load transducer, the connections, the indenter itself, the specimen holder, and the specimen (arterial tissue) to be tested. This configuration ensures accurate measurement of the contact force and displacement depth during the test. (B) Biomomentum Mach-1 mechanical tester (Biomomentum, Laval, Canada).

that the pyramidal indenter tends to overestimate the Young's modulus of pulmonary vessels.

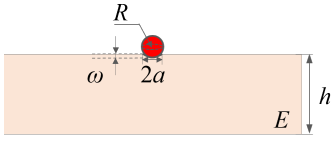
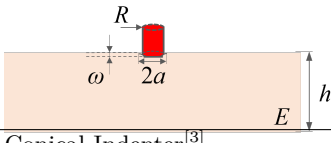
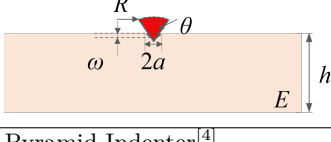
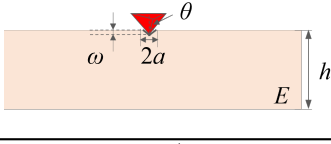
Among all the experiments conducted with spherical indenters, in skin tissue, the indenter radius R is typically larger than that used for arteries. Since the skin has larger dimensions, a larger indenter can facilitate deeper indentations without resulting in large strain.

In this research, the indentation tests on the skin (Chapter 3) and arterial tissue (Chapter 4) were all conducted using a Biomomentum Mach-1 mechanical tester (Biomomentum, Laval, Canada), as illustrated in Figure 2.6B.

2.3.3 Friction in Contact Problem

In the previous subsection, we discussed the contact problem without considering external forces such as friction f_i^b . However, in real-world scenarios, especially for biological tissue, external forces such as friction and adhesion significantly influence their contact behaviour. Friction forces arise from the tangential reaction between two contact surfaces and are affected by several factors, including contact geometry, surface topology, material properties of both the bulk and surface, physical and chemical conditions of the contact materials, displacement, relative velocity, and the presence of lubrication, among others [92], [158]–[160]. Incorporating friction force into the contact problem, commonly referred to as the frictional contact problem, adds non-linearity and introduces significant complexity to the contact problem, rendering analytical solutions impractical. Consequently, solving such problems necessitates the use of numerical methods. These methods often

Table 2.3: The indentation test setup used for skin and artery

The Experimental Setup	Skin	Artery
Spherical Indenter ^[1] 	$R = 5.9 \text{ mm}$ [137] $R = 6.22 \text{ mm}$ [138] $R = 2.25, 4.75, 6.35 \text{ mm}$ [139] $R = 4 \text{ mm}$ [140] $R = 7.8 \mu\text{m}$ [141] $R = 50 \mu\text{m}$ [143] $R = 250 \mu\text{m}$ [145] $R = 496 \mu\text{m}, 986 \mu\text{m}$ [146]	$R = 1 \mu\text{m}, 2.5 \mu\text{m}$ [136] $R = 4.5 \mu\text{m}$ [125] $R = 1 \text{ mm}$ [59] $R = 2.5 \text{ mm}$ [134] $R = 1 \text{ mm}$ [142] $R = 0.5 \text{ mm}$ [58], [144]
Cylindrical Flat Punch ^[2] 	$R = 9 \text{ mm}$ [147] $R = 2 \text{ mm}$ [149] $R = 2 \text{ mm}$ [149]–[151] $R = 4.5 \text{ mm}, 90 \text{ mm}$ [153] $R = 31 \mu\text{m}$ [146]	$R = 2.5 \text{ mm}$ [148] $R = 3.5 \text{ mm}$ [60] $R = 2.5 \text{ mm}$ [152]
Conical Indenter ^[3] 	$H = 10 \text{ mm}, \theta = 45^\circ$ [154]	$\theta = 20^\circ$ [155]
Pyramid Indenter ^[4] 	$\theta = 57.4^\circ$ (Vickers), 29.2° (Berkovich) [156] $\theta = 70.3^\circ$ (Berkovich) [146]	$\theta = 18^\circ$ [136]

[1] $F = \frac{4ER\omega^{3/2}}{3(1-\nu^2)}$, where R is the indenter radius, ω is the indentation depth, and a is the contact radius.

[2] $F = \frac{2ER\omega}{1-\nu^2}$ [118], where R is the indenter radius.

[3] $F = \frac{2E \tan \theta}{\pi(1-\nu^2)} \omega^2$ [118], where θ is the semi-included angle of the indenter.

[4] $F = \frac{2E \tan \theta}{\pi(1-\nu^2)} \omega^2$ [118], where θ is the semi-included angle of the indenter. The Berkovich and Vickers indenters are the most widely used pyramidal indenters with different shapes; Berkovich has three edges, while Vickers has four. A detailed comparison between them can be found in [157].

employ advanced mathematical frameworks, such as variational inequality and equality models [161], or rely on FE simulations [123], [162]. By applying these numerical approaches, it becomes possible to approximate solutions to the frictional contact problem, which is crucial for achieving a more realistic analysis of contact behaviours in biological tissue.

In this subsection, we will introduce friction laws that govern the interactions between contact surfaces, followed by a review of existing research on frictional contact as applied to skin and arteries.

2.3.3.1 Amontons' Law

The friction behaviour can broadly be categorized into static, quasi-static, and dynamic problems [163]. Since the friction in the contact problem addressed in this research is typically treated as either static or quasi-static [164], we will focus on static friction behaviour. Regardless of the

specific type of friction, Amontons' two well-known laws [163] generally apply.

- *First Amontons' Law*

The friction force, F_μ , is directly proportional to the normal load, F , as described by Eq. 2.22. In some research, this is referred to as Coulomb's law of friction.

$$\mu = \frac{F_\mu}{F} \quad (2.22)$$

- *Second Amontons' law*

The friction coefficient μ is independent of the contact area and the loading force, i.e., μ is approximately a constant ≤ 1 .

These foundational principles of friction provide a basis for understanding friction behaviour in various scenarios, including those involving biological tissue. However, Derler et al. pointed out that Amontons' empirical laws—stating that the friction force is directly proportional to the load and independent of the contact area and loading force—do not hold for skin friction due to the nonlinear, viscoelastic properties of skin [92]. In such cases, Bowden and Tabor theory [165], also known as the two-term friction model, is more appropriate [92].

2.3.3.2 Two-term Friction Model

The two-term friction model expresses the total friction force F_μ as the sum of adhesion friction, $F_{\mu,\text{adh}}$, and deformation friction, $F_{\mu,\text{def}}$, as shown in Eq. 6.8. While adhesion friction $F_{\mu,\text{adh}}$ corresponds to the breaking of adhesive bonds between the two surfaces (Eq. 2.24), the deformation friction $F_{\mu,\text{def}}$, accounts for the energy lost due to viscoelastic hysteresis per unit sliding distance (Eq. 6.10).

$$F_\mu = F_{\mu,\text{adh}} + F_{\mu,\text{def}} \quad (2.23)$$

$$F_{\mu,\text{adh}} = \tau A_{\text{real}} \quad (2.24)$$

where τ is the shear strength of the interface and A_{real} is the real contact area.

$$F_{\mu,\text{def}} = \frac{3}{16} \beta \frac{a}{R} F \quad (2.25)$$

where β is a viscoelastic loss fraction.

It has been demonstrated that the interface shear strength is a linear function of the contact pressure, p , as shown in Eq. 2.26 [165], [166].

$$\tau = \tau_0 + \eta p \quad (2.26)$$

where τ_0 is the yield stress at zero external pressure, η is the pressure coefficient, and p is the mean contact pressure [163].

Thus, the adhesion force can be rewritten as Eq. 2.27.

$$F_{\mu, \text{adh}} = \tau_0 A_{\text{real}} + \eta F \quad (2.27)$$

Adhesion friction has been widely recognized as the primary contributor to the friction of human skin, while deformation mechanisms are generally considered to play a minor role [92], [167].

2.3.4 Friction of Tissue

The experimental friction tests are conducted to study the friction characteristics of materials. These tests are highly dependent on various factors, including material properties, surface finish, and tissue conditions such as hydration, age, surface roughness, and chemical environment. Additionally, the conditions of the friction tests, including the normal force load and the sliding speed between the interacting surfaces, significantly influence the results.

The experimental apparatus commonly used for testing friction forces is called a tribometer. Depending on specific functionalities, examples include the scanning tunneling microscope (STM) [168], the atomic-force microscope (AFM) [169], and the friction-force microscope (FFM) [170]. The STM is especially effective for studying conducting surfaces, as it utilizes quantum tunneling to precisely measure surface features at the atomic scale. In contrast, the AFM is ideal for dielectric surfaces, providing high-resolution topographic measurements through interactions between a probe and the sample surface. Expanding on the AFM's capabilities, the FFM not only captures surface topography but also detects transverse forces, making it particularly well-suited for detailed friction force analysis.

The setup of the tribometer is schematically illustrated in Figure 2.7 [163], [171]. The measured spring force, F_{spring} , which is equal to F_{μ} , and normal force F are used to calculate the coefficient of friction (COF) according to Coulomb's law (Eq. 2.22).

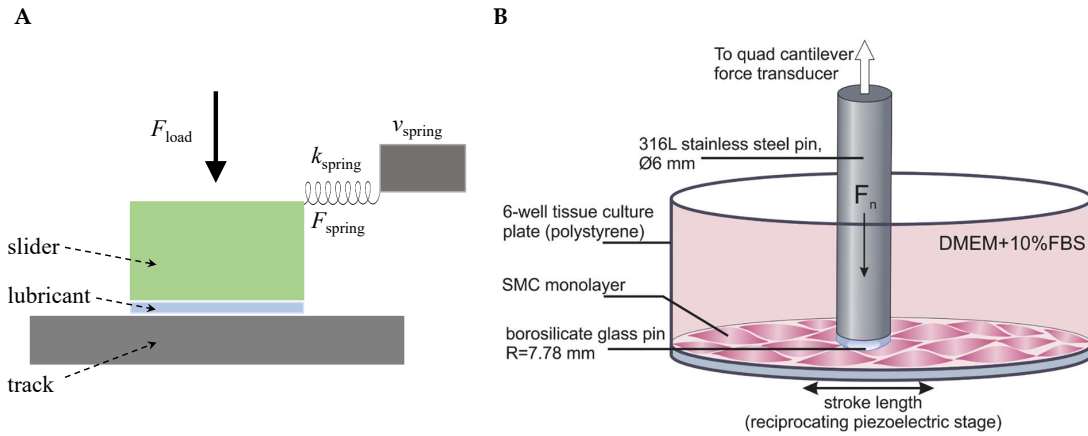


Figure 2.7: Friction test: (A) Tribometer consists of a top solid substrate, referred to as the slider, placed on a bottom substrate known as the track. The slider and track may be separated by a thin lubricant film, and they are compressed together by a loading force, F , which is typically due to the mass m of the top block by $F = mg$, where g is the acceleration due to gravity. The slider is connected to the base, which moves at a constant velocity v_{spring} through a spring with an elastic constant k_{spring} . In practical applications, the elasticity of the top block itself may serve the role of the spring. The primary output parameter measured in this setup is the spring force, F_{spring} [163]. (B) Schematic of microtribometer setup for cell friction experiments. This setup includes key components: DMEM (Dulbecco's Modified Eagle Medium), FBS (Fetal Bovine Serum), and SMC (Smooth Muscle Cell). The piezoelectric-controlled stage linearly reciprocates the tissue culture plate containing the SMC monolayer against a glass pin, which is held under a constant normal load throughout a series of 10 cycles. Friction loop data is captured using the Anton Paar NTR3. Following the completion of tribological testing, the SMCs are immediately returned to 37°C to ensure optimal cell viability [172].

Specific tribometers—including linear sliding or linearly reciprocating tribometers and rotating tribometers, as illustrated in Figure 2.8 [160]—have been extensively utilized for tribological studies of *in-vivo* skin. Compared to tribometers used for skin friction studies, micro-tribometers for arterial cell friction are generally simpler, focusing solely on linear sliding friction. Figure 2.7 illustrates micro-tribometers are employed for investigating arterial cell friction [172], [173]. The micro-tribometer features a cylindrical punch that applies both the normal force and the sliding velocity.

2.3.4.1 Skin Friction

Skin friction has been extensively studied in numerous studies, with comprehensive literature reviews provided by Derler et al. [92] and Masen et al. [160], [174]. Derler highlighted several key factors influencing skin friction, including hydration, age, anatomical region, and the type and direction of loading on the skin layers, with hydration being particularly emphasized. The COFs for both wet and dry skin have been reported in various studies, indicating the COF for dry skin is approximately 0.57 ± 0.32 for dry skin, while generally higher COFs are observed for wet skin, around 1.5 ± 0.6 . Masen also underscored the importance of hydration in their reviews [160], [174],

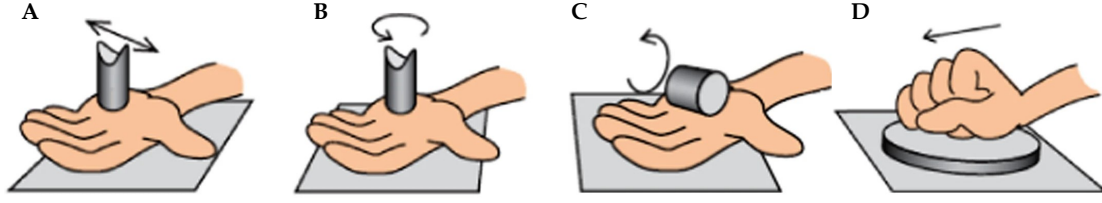


Figure 2.8: Schematic of tribometers based on various contact configurations [160]: (A) Linear contact: this configuration involves two surfaces sliding against each other in a straight line. (B) Perpendicular rotating contact: in this setup, one surface rotates perpendicular to the other, allowing for a circular motion that generates friction. (C) Parallel rotating contact: this configuration features two surfaces rotating parallel to each other, which can affect the contact area and frictional behaviour. (D) User-controlled motion: this flexible setup allows the user to dictate the motion of the surfaces, which can include various trajectories and speeds to study their effects on friction.

consistently showing that the COF increases significantly with higher hydration levels. The key factors identified in these studies are summarized in Table 2.4

In skin friction studies, a more detailed quantitative analysis of the friction effect has been provided by numerous researchers. The accepted relationships between the real contact area A_{real} and the normal force F , as well as the COF versus normal load, are expressed in given as Eq. 2.28 and 2.29, respectively. Eq. 2.28 is derived by considering the contact radius according to various contact models. Eq. 2.29 is derived from Eq. 2.28, where the friction force is considered to be proportional to the real contact area A_{real} , which corresponds to the adhesion friction, $F_{\mu,\text{adh}}$.

$$A_{\text{real}} \propto F^m \quad (2.28)$$

where m is the exponent coefficient.

$$\mu \propto F^{m-1} \quad (2.29)$$

The exponential coefficient m for the skin friction has been reported to range between 0.15 and 1.33, with a specific value of 0.66 associated with Hertz's contact theory. Several authors have noted that the COFs decrease with increasing normal load, following a power law characterised by exponents between -0.68 and -0.24. A more comprehensive quantitative relationship between contact area, COF, and surface texture parameters can be found in [194], [195].

Furthermore, if adhesion friction is considered separately from the applied normal load, the COF can be expressed as shown in Eq. 2.30. This relationship has been illustrated in various studies, including those referenced in [174] (see Figure 2.9).

Table 2.4: Influencing factors and their effect on friction

Factors	Effect on friction
Hydration	Linear relationship (increase) [175], [176] Power-law relationship (increase) [177] Exponential relationship (increase) [120] Bell-shaped relationship [178], [179] Increase [180]–[182]
Temperature	Negative correlation [183]
Ambient temperature	Positive correlation [183]
Amount of sebum	Increase [184], [185] Weak or no relation [185]
Skin-surface roughness	No significant correlation [186] R_a improves predictability of COF [186] Correlated with the density of primary lines ($> 20 \mu\text{m}$) [187]
Counter-surface roughness	Low roughness: decrease [177], [182], [188], [189] High roughness: increase [190], [191] Decrease then increase [92], [160]
Age	No effect [175], [186], [192] Has effect, higher in premenopausal females [193] More sensitive to moisture changes (female skin & textiles) [176]
Body region	Forehead, postauricular $>$ volar forearm, upper back $>$ dorsal forearm $>$ ankle, palm $>$ lower back $>$ thigh $>$ abdomen [92] Finger $>$ edge of hand; volar forearm $>$ cheek; vulva $>$ volar forearm [92] Finger pad higher than others, temple lower [183]

$$\mu = \mu_0 \left(1 + \frac{F_{\mu, \text{adh}}}{F} \right) \quad (2.30)$$

where μ_0 is the initial coefficient of friction.

2.3.4.2 Artery Friction

The friction between arteries and various materials is also widely explored in studies [123], [202]–[206]. Table 2.5 lists the friction experiments conducted on arterial tissue.

In these studies, the COFs between arteries and medical devices or arterial cells were measured using customized tribometers. Figure 2.8B illustrates the microtribometer setup for cell friction experiments, as used in Clark et al.’s study [172]. While the two-term friction model is commonly applied to the skin, Coulomb’s law remains the primary method for calculating arterial COFs. Among the studies on arterial friction, only Dunn et al. [173] accounted for adhesion friction in arteries. Braun [163] highlights that Bowden and Tabor’s theory provides a qualitative explanation for Amontons’ laws, with Amontons’ law serving as a simplified version of Bowden and Tabor’s theory, excluding adhesive and viscoelastic properties of the contact material. COFs for skin tissue

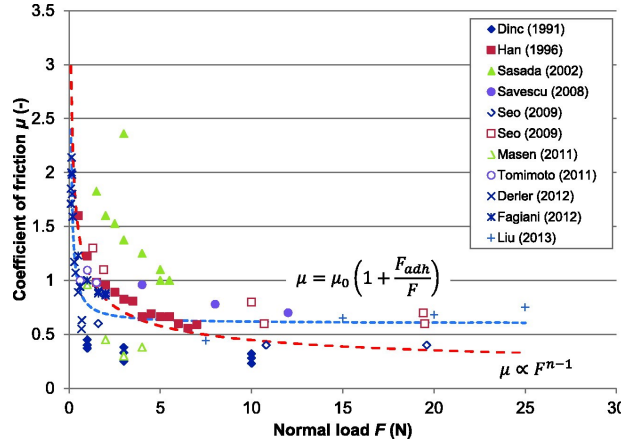


Figure 2.9: Coefficient of friction of the index finger as a function of normal load [177], [180], [182], [189], [196]–[201]. Image adapted from Kuilenburg et al [174].

typically range from 0.1 to 1 for dry skin and exceed 1 for wet skin, whereas arterial COFs are generally ≤ 0.1 .

The impact of friction in artery-stent interactions is further explored in Chapter 5, where the friction contact problem is addressed through FE simulations. Coulomb’s law is used to define frictional behaviour, aligning with the study’s focus on the hyperelastic properties of biological tissue. While some studies report higher COFs for arteries, the interactions in Chapter 5 are modeled using COFs ≤ 0.1 for artery-stent interactions, with a more detailed discussion provided later.

2.4 Finite Element Method for Contact Problem

Contact problems, especially those involving friction, pose significant challenges for analytical solutions. These challenges arise from various factors, including the nonlinearity introduced by external forces, the complexity of mixed boundary conditions, and the nonlinear behaviour of materials. Consequently, numerical methods, particularly the FE method, are frequently employed to address these frictional contact issues.

The FE method achieves approximate solutions by discretising the entire domain V into small, finite elements and applying the governing equations to these elements. Figure 2.10 illustrates the discretisation of the domain V into small quadrilateral elements, each defined by nodes located at their vertices and midpoints. These nodes serve as reference points for constructing the elements and defining the interpolation functions within each element, enabling an approximation of the solution across the domain. The FE method is highly valued for its ability to handle complex

Table 2.5: Arterial friction tests

Research	Material	COF
Takashima et al. [123]	a catheter and porcine aorta	0.015-0.13
Dunn et al. [173]	bovine aortic endothelial cells	0.03-0.06
Caldwell et al. [202]	untreated LDPE-porcine aorta ^[1] polyethylene LDPE-porcine aorta	0.0384 ± 0.01 0.0285 ± 0.0051
Vad et al. [205]	PET stent-artificial vascular ^[2] ePTFE stent-artificial vascular ^[3]	0.22-0.46 0.012-0.018
Chen et al. [203]	Zenith stent-porcine aorta	0.0328 ± 0.0019 (axial) 0.04426 ± 0.0036 (circumferential)
	Endurant stent-porcine aorta	0.05406 ± 0.0034 (axial) 0.05376 ± 0.0015 (circumferential)
	Anaconda stent-porcine aorta	0.04666 ± 0.0036 (axial) 0.04556 ± 0.0022 (circumferential)
Clark et al. [172]	SMCs from ND patients ^[4]	0.107 ± 0.03
	SMCs from T2DM patients ^[5]	0.22 ± 0.01
Kwak et al. [206]	stent retrievers-artificial vascular ^[6]	≈ 0.037 ^[7]

^[1] LDPE: low-density polyethylene

^[2] PET: polyethylene terephthalate

^[3] ePTFE: expanded polytetrafluoroethylene

^[4] SMC: smooth muscle cells; ND: non-diabetic

^[5] T2DM: Type 2 diabetic.

^[6] artificial vascular made by polydimethylsiloxane

^[7] $F_\mu = \mu F + \mu F_{\mu,adh}$, where F_μ and F are the friction and normal force, respectively. $F_{\mu,adh}$ is the adhesion force.

geometries, nonlinear material properties, and mixed boundary conditions. Numerous FE software packages, such as ABAQUS, ANSYS, and COMSOL, have been developed to streamline these analyses.

2.4.1 Fundamentals of Finite Element Method

In solid mechanics, the FE method is employed to determine the unknown displacement field $\mathbf{u}(\mathbf{x})$ governed by the momentum equation as described in Eq. 2.31.

$$\begin{aligned}
 \sigma_{ij,j} + \rho f_i^b - \rho \ddot{u} &= 0 \quad \text{on } V \\
 \sigma_{ij} n_j &= f_i^s \quad \text{on } \Gamma_t \\
 u_i &= u_i^s \quad \text{on } \Gamma_u
 \end{aligned} \tag{2.31}$$

where $\sigma_{ij,j}$ is the divergence of the stress tensor σ_{ij} , which represents the internal forces within the material; f_i^b is the body force; ρ is the density of the material. u_i^s defines the displacement on the boundary Γ_u , and f_i^s represents the traction on the boundary Γ_t ; $\ddot{u}$ is the acceleration.

Eq. 2.31 gives the strong form of the momentum equation, which requires strict continuity and

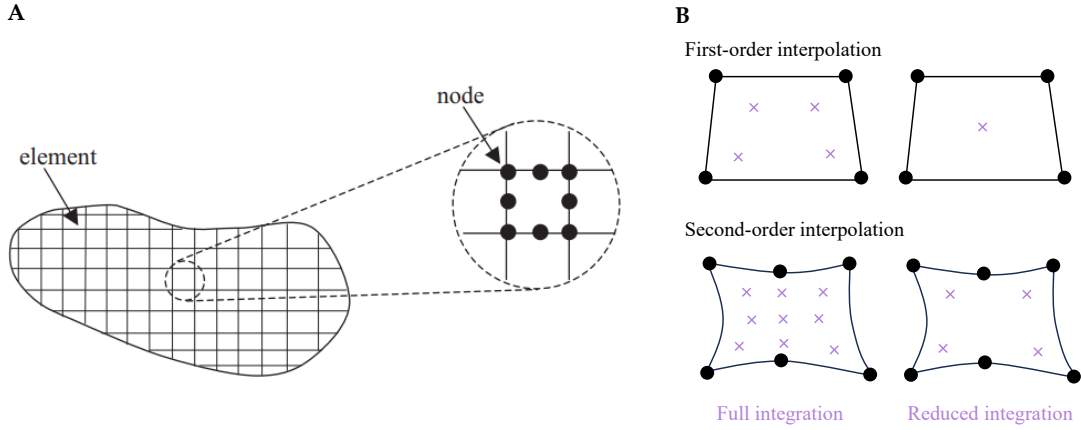


Figure 2.10: FE discretisation [207]: the domain V is discretised into small quadrilateral elements with eight nodes per element, which are used to define the element geometry and perform the interpolation function across the element.

differentiability of stresses and displacement fields across the domain V . However, these requirements are often difficult to meet in practical applications, particularly when the problem involves discontinuities in material properties (e.g., in composites or layered materials), irregular geometries, or nonlinear material behaviours. To overcome the limitations given by the strong form of the momentum equation, FE methods employ the weak form of the momentum equation, as shown in Eq. [2.32]. The weak form is derived from the principle of virtual work, which converts the strong form into an integral equation by integrating it over the domain V . The weak form provides increased flexibility by reducing the regularity requirements to piecewise differentiability, rather than continuous differentiability across the entire domain V . This flexibility allows each element to use independently defined functions, accommodating discontinuities between elements where necessary. Consequently, the weak form enables the FE method to handle complex material interfaces and geometries more effectively than the strong form, making it highly adaptable to a wide range of engineering problems.

$$\int_V \delta u_i (\sigma_{ij,j} + \rho f_i^b - \rho \ddot{u}_i) dV = 0 \quad (2.32)$$

where δu_i is a virtual displacement field.

By discretising the domain V into finite elements and assigning nodes to each element (see Figure. [2.10]), the virtual displacement $\delta u_i(\mathbf{x})$ can be interpolated using discrete virtual nodal displacements δU_i^α and shape functions $N_\alpha(\mathbf{x})$, as defined in Eq. [2.33].

$$\delta u_i(\mathbf{x}) = \sum_{\alpha=1}^{M_n} N_\alpha(\mathbf{x}) \delta U_i^\alpha \quad (2.33)$$

where M_n is the node number.

Substituting Eq. 2.33 into Eq. 2.32, the virtual work equation can be expressed as Eq. 2.34.

$$\int_V \sigma_{ij} N_{\alpha,j} dV - \int_{\Gamma_t} N_\alpha f_i^s d\Gamma = 0 \quad (2.34)$$

where i ranges from 1 to NDF (the number of spatial degrees of freedom, usually 2 or 3), and α ranges from 1 to M_n (M_n is the total number of nodes). $\Gamma = \Gamma_t + \Gamma_u$ is the boundary.

For linear elasticity with the stress-strain relationship given as $\sigma_{ij} = C_{ijkl} \varepsilon_{kl}$, and strain defined by displacement as $\varepsilon_{kl} = \frac{1}{2}(u_{k,l} + u_{l,k})$, the virtual work equation is reformulated in terms of displacement as Eq. 2.35.

$$\left(\int_V C_{ijkl} N_{b,l} N_{\alpha,j} dV \right) U_k^b = \int_{\Gamma_t} N_{\alpha,j} f_i^s d\Gamma \quad (2.35)$$

This leads to the final matrix form as Eq. 2.36.

$$\begin{aligned} K_{iakb} U_k^b &= f_i^\alpha \\ \mathbf{KU} &= \mathbf{f} \end{aligned} \quad (2.36)$$

where $\mathbf{U}$ (U_k^b) is the vector of unknown nodal displacement, $\mathbf{f}$ (f_i^α) is the nodal force vector, and $\mathbf{K}$ (K_{iakb}) is the FE stiffness matrix, which is often referred to as the ‘global’ stiffness matrix.

For dynamic problems, mass matrix $\mathbf{M}$ and acceleration vector $\mathbf{U}$ are included, leading to the governing equation as Eq. 2.37

$$\mathbf{MU} + \mathbf{KU} = \mathbf{f} \quad (2.37)$$

The main task in a FE analysis is the calculation and inversion of the global stiffness matrix $\mathbf{K}$. Each element contributes a ‘local’ stiffness matrix to $\mathbf{K}$, calculated independently using only information from the element itself. This independence is achieved through the use of the shape functions, N_α , whereby the value of the variables within an element depends only on the values at the nodes of that element.

Given the complexities involved in manually coding FE methods, this research utilizes the commer-

cial software ABAQUS (ABAQUS 2024) to conduct the indentation simulations and the simulations of artery-stent interactions. In this section, we outline the fundamentals of the FE method, with a particular emphasis on establishing and modelling contact interactions.

2.4.2 Finite Element Modelling in ABAQUS

To define an FE model in ABAQUS, the following steps are typically followed: create a part, define material properties, apply boundary conditions, specify interaction constraints, mesh the model, and run the simulation. Several important considerations are outlined below to enhance the modelling process.

2.4.2.1 ABAQUS Solver

ABAQUS provides two solvers: ABAQUS/Standard based on the implicit algorithm and ABAQUS/Explicit based on the explicit algorithm. In both ABAQUS/Standard and ABAQUS/Explicit, the governing equation is given by Eq. 2.37. The key difference between the two algorithms lies in how the nodal accelerations are computed. The implicit solver employs the Newton-Raphson method to iteratively solve for the correction, c , to the incremental displacements, Δu_j . This process seeks the solution to the dynamic equilibrium at the end of the time increment, $t + \Delta t$, while simultaneously computing the displacements (see Eq. 2.38). In contrast, the explicit algorithm utilizes an extrapolation approach, where the next state is directly calculated from the current state data (see Eq. 2.39).

$$\mathbf{K}c = \mathbf{P} - \mathbf{I}_f - \mathbf{M}\ddot{\mathbf{u}} \quad (2.38)$$

where $\mathbf{P}$ is the external applied forces, and $\mathbf{I}_f$ is the internal element force.

$$\begin{aligned} \ddot{\mathbf{u}}_t &= \mathbf{M}^{-1}(\mathbf{P}_{(t)} - \mathbf{I}_{f(t)}) \\ \dot{\mathbf{u}}_{(t+\frac{\Delta t}{2})} &= \dot{\mathbf{u}}_{(t-\frac{\Delta t}{2})} - \frac{\Delta t_{t+\Delta t} + \Delta t_{(t)}}{2} \\ \mathbf{u}_{(t+\Delta t)} &= \mathbf{u}_{(t)} + \Delta t_{(t+\Delta t)} \dot{\mathbf{u}}_{(t+\frac{\Delta t}{2})} \end{aligned} \quad (2.39)$$

The implicit algorithm is unconditionally stable, allowing for larger time increments, Δt , which are automatically determined by the solver. In contrast, the explicit solver's time increment is constrained by stability considerations. Excessively large time steps can lead to instability and significant errors in the solution. The maximum allowable time increment, known as the stable time increment, is governed by the Courant-Friedrichs-Lewy (CFL) condition (see Eq. 2.40). This

condition links the time-step size to the spatial discretisation and wave propagation characteristics of the system.

$$t_{\text{stable}} = \min\left(\frac{L_e}{c_d}\right) \quad c_d = \sqrt{\frac{\lambda_l + 2\mu_l}{\rho}} \quad (2.40)$$

where t_{stable} is the stable time increment; L_e is the dimension of the mesh element, c_d is the material wave speed, ρ is the material density, and λ_l and μ_l is the Lamé coefficients.

In an implicit solver, nonlinear problems typically require multiple iterations per increment to converge within prescribed tolerances, making the method computationally expensive compared to the explicit approach. As a result, the explicit solver is often better suited for problems involving significant nonlinearities. In this research, two types of simulations are conducted: the indentation simulation, representing a static contact problem, benefits from the implicit method due to its stability and accuracy (see Chapters 3 and 4). On the other hand, the artery-stent interaction in Chapter 5 involves substantial nonlinearities arising from friction, nonlinear material behaviour, complex geometry, and large deformations. Attempts to model these interactions using the implicit solver often resulted in premature termination due to convergence issues. In contrast, the explicit solver is more robust in handling strong nonlinearities and is widely used in quasi-static simulations of stent-artery interactions. Therefore, the explicit solver was chosen for the artery-stent interaction simulation to ensure both accuracy and stability.

2.4.2.2 Element Selection

In FE analysis, approximations primarily arise from the properties of the chosen elements, including their shape functions and the mesh distribution within the structure. Shape functions define the variation of displacement and strain across each element, directly influencing the accuracy of the results, particularly in regions with high stress or strain gradients. Therefore, both element selection and mesh distribution must be carefully tailored to the specific problem. In regions with stress concentrations or complex deformations, a finer mesh and higher-order elements may be required to accurately capture the behaviour. Conversely, areas with less variation may allow for the use of lower-order elements or a coarser mesh, optimising computational efficiency without sacrificing accuracy. Striking the right balance in element selection and mesh distribution is crucial for obtaining reliable results within practical computational limits.

- *Dimensionality*

ABAQUS offers a range of two-dimensional (2D) and three-dimensional (3D) elements tailored to

different modelling requirements, such as shell and surface elements for 2D analysis, and truss and solid elements for 3D. When possible, 2D elements with fewer degrees of freedom (DoF) and nodes are preferred over 3D elements to reduce computational costs.

Additionally, ABAQUS supports dimensional reduction through specific element types, such as axisymmetric, plane stress, and plane strain elements. For instance, axisymmetric elements are ideal when rotational symmetry is present in the model's geometry. This allows the simulation to focus on a cross-sectional slice rather than the entire geometry, ensuring computational efficiency while preserving essential boundary conditions and geometric features.

In this research, we used axisymmetric 2D solid elements for the indentation simulations, striking a balance between computational efficiency and model accuracy. However, for the artery-stent interaction simulations, we opted for 3D solid elements to accurately capture the complex contact and deformation characteristics inherent to the artery-stent system.

- *Element shape*

Common shapes for 2D elements include triangles and quadrilaterals. Triangular elements are advantageous for modelling complex geometries but may be less effective in accurately capturing bending behaviour. In contrast, quadrilaterals typically offer superior performance in plane stress and plane strain analyses due to their bilinear shape functions, which facilitate more precise stress distribution.

For 3D elements, tetrahedral elements are versatile and well-suited for complex geometries but can introduce inaccuracies in bending problems because of their reliance on linear shape functions. Hexahedral elements, on the other hand, are more efficient at capturing stress gradients and are ideal for structured meshes, making them the preferred choice for simulations that demand high accuracy in stress and deformation responses.

- *Degrees of freedom*

An element's DoF represents the independent displacements or rotations that each node within the element can undergo. In solid stress analysis, these movements typically consist of translations, with rotational DoF often incorporated for structural and shell elements.

For 2D elements, nodes can translate in the x and y directions, giving each node 2 DoF. In 3D elements, nodes can translate along the x , y , and z axes, providing 3 DoF per node. For example, linear rectangular elements have 4 nodes, each with 2 translational DoF in the x and y directions. Unlike Constant Strain Triangles, which assume constant strain across the element, rectangular

elements use a bilinear shape function, often referred to as an incomplete quadratic function, expressed as $C_1 + C_2x + C_3y + C_4xy$, where C_i are constants. This formulation allows for a linearly varying strain across the element, improving accuracy and performance, especially in higher-order analyses. However, rectangular elements may be less effective in situations involving acute angles, where their accuracy can diminish.

- *Number of nodes (Interpolation)*

ABAQUS offers two primary types of elements based on node configurations: first-order (linear) and second-order (quadratic) elements.

First-order elements have nodes only at the corners of the element, such as 3-node triangles and 4-node quadrilaterals (see first-order interpolation in Figure 2.10B). Linear triangular elements assume a uniform strain distribution across the element. This assumption limits their ability to capture complex deformations, particularly bending. These elements are suitable for problems with relatively simple deformations or when computational efficiency is prioritised over accuracy, though they may be less effective in representing detailed stress and strain variations.

In contrast, second-order elements have nodes at both the corners and midpoints of each edge, such as 6-node triangles or 8-node quadrilaterals (see second-order interpolation in Figure 2.10B). These elements use quadratic interpolation, enabling smoother transitions and more gradual changes in the element's behaviour. This allows them to more accurately capture nonlinear deformations, such as bending and stretching, although they come with higher computational costs.

- *Integration*

Integration within elements is vital for calculating material behaviour by assessing the element's response at specific internal points, known as integration points or Gauss points. At each integration point, ABAQUS computes the element's stiffness matrix, which helps approximate the overall response to applied loads and boundary conditions.

ABAQUS provides two primary integration options: full integration and reduced integration (Figure 2.10B). Full integration evaluates polynomial terms at all Gauss points within the element, offering higher accuracy but increasing computational costs. On the other hand, reduced integration uses fewer integration points, which speeds up calculations and reduces computational demand. However, this method can introduce numerical issues, such as 'hour-glassing' or element locking, especially in cases of low-strain or bending-dominated deformations.

In addition to defining the element section, specifying material properties is crucial in FE modelling.

ABAQUS offers a variety of material models, including the Ogden and the GOH model, enabling users to define and customize material behaviours according to their specific needs.

2.4.3 Contact Modelling in ABAQUS

Figure 2.11 illustrates the logical framework of the contact algorithm for both the implicit and explicit solvers. Compared to the implicit solver, the explicit solver offers an additional contact algorithm, the general contact algorithm, which is designed to handle multiple contact surfaces simultaneously. This algorithm is particularly useful for complex interactions involving many surfaces in a model.

In this research, however, both simulations involve only two contact surfaces: one for the indentation simulation and one for the artery-stent interaction. As a result, the contact pair algorithm, which is specifically tailored for modelling interactions between two surfaces, has been selected for both simulations.

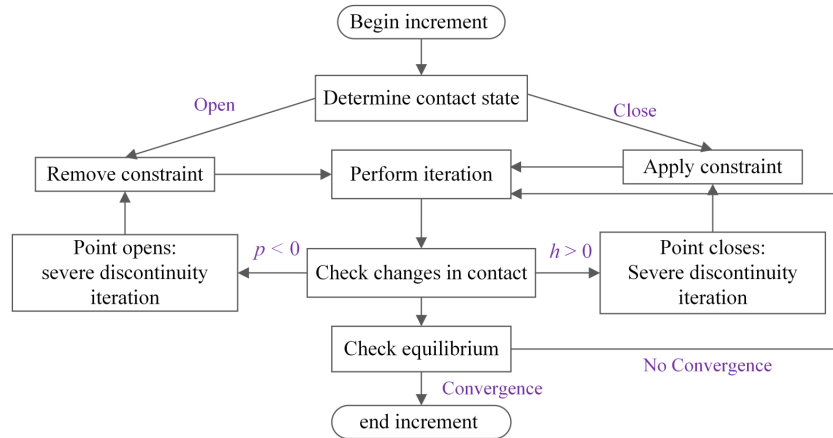


Figure 2.11: The contact algorithm in ABAQUS checks all contact interactions at the start of each increment, identifying each slave node as either open or closed. For closed nodes, it determines whether they are sliding or sticking and applies or removes constraints accordingly. ABAQUS then updates the model configuration and re-evaluates the contact conditions after each iteration. Nodes where clearance becomes non-positive switch to closed, while nodes with negative contact pressure switch to open. If contact changes are detected, ABAQUS marks it as a severe discontinuity iteration. Iterations continue until severe discontinuities are minimal (or absent) and equilibrium tolerances are met. Alternatively, ABAQUS can be set to iterate until no severe discontinuities remain before checking equilibrium [208].

2.4.3.1 Contact Properties

The contact properties definition includes the definition of both the normal behaviour, known as the contact pressure-overclosure relationship, and the friction behaviour during the interaction.

- *Contact pressure-overclosure relationship*

ABAQUS provides two methods for pressure-overclosure relationships: ‘hard contact’ and ‘softened contact’.

‘Hard contact’, represented by Eq. 5.3, minimises the penetration of the slave surface into the master surface at the constraint locations and prevents the transfer of tensile stress across the interface. In contrast, ‘softened contact’ defines the relationship between contact pressure and clearance more gradually, using linear, tabular piecewise-linear, or exponential laws. Although ‘softened contact’ enhances stability and convergence by mitigating potential slave surface penetration issues, it tends to be less accurate than ‘hard contact’.

Given the emphasis on accuracy in this research, ‘hard contact’ is chosen for both simulations to ensure a realistic representation of contact behaviour.

$$p_n = 0 \quad g_n > 0; \quad p_n > 0 \quad g_n = 0; \quad (2.41)$$

where p is the contact pressure on node n and g_n is the gap function.

- *Friction behaviour*

The Coulomb friction model is used to define the friction behaviour in ABAQUS, with specific equation provided in Eq. 2.42

$$\begin{aligned} g_\tau &= 0, \quad q_\tau \leq \mu |p_n| \quad \text{on } \Gamma^{C,st} \\ \dot{g}_\tau &\neq 0, \quad q_\tau = \mu |p_n| \frac{\dot{g}_\tau}{\|\dot{g}_\tau\|} \quad \text{on } \Gamma^{C,sl} \end{aligned} \quad (2.42)$$

where μ is the coefficient of friction; g_τ is the tangential gap; $\dot{g}_\tau$ is the sliding velocity; q_τ is the tangential reaction, and $\Gamma^{C,st}$, $\Gamma^{C,sl}$ are the stick and sliding region, respectively.

2.4.3.2 Contact Formulation

The contact formulation for modelling surface interactions includes four key components: the constraint enforcement method, contact surface weighting, tracking approach, and sliding formulation.

- *Constraint enforcement method*

In ABAQUS, two algorithms are available for achieving surface-to-surface contact: ‘kinematic contact’ and ‘penalty contact’. Kinematic contact ensures precise adherence to contact conditions through a predictor/corrector approach. Penalty contact models the contact force as the product of a penalty stiffness and the penetration distance, as described by Eq. 2.43. Although penalty contact can introduce artificial damping effects, it offers superior computational efficiency compared

to the more computationally demanding kinematic contact. Therefore, in this research, penalty contact was selected for both simulations, with the penalty stiffness carefully calibrated to minimise excessive damping and ensure the realistic behaviour of the contact interactions.

$$p_n = \epsilon_n g_n \quad (2.43)$$

where ϵ_n is the contact stiffness and g_n is the gap function. The penalty stiffness is automatically determined by the explicit solver.

- *Contact surface weighting*

In the implicit solver, only the pure master-slave contact algorithm is available, where one surface is the master and the other the slave, allowing only slave node penetration into master facets. The explicit solver offers an additional balanced master-slave method, which applies the pure master-slave technique twice with reversed surfaces. The balanced approach, invoked by a contact parameter f between 0 and 1.0, enhances stability but is more computationally demanding. It is typically used when mesh irregularities occur. However, since no such issues were encountered in this research, the pure master-slave contact was chosen to reduce computational cost.

- *Tracking approach*

In ABAQUS, the global and local search methods are key for efficient contact tracking in ABAQUS simulations. The global search method, applied at the start of each analysis step and every 100 increments, identifies the closest master facet for each slave node across the model. While precise, it can be computationally intensive. In contrast, the local search method is used between global search steps, focusing only on master facets associated with the previously tracked master node. This reduces the computational cost by limiting the search to fewer elements, improving efficiency.

- *Sliding approach*

ABAQUS offers small and finite sliding approaches. Small sliding assumes minimal relative displacements and a constant contact area, suitable for small deformations. Finite sliding, on the other hand, accommodates large displacements and varying contact areas, requiring more complex algorithms. Since the simulations in this research involve large deformations, the finite sliding approach was chosen to accurately capture the interaction dynamics.

2.4.4 Indentation Simulation on Tissue

Indentation simulations on skin and arterial tissue are widely used in biomechanical research for various applications. One key application is generating closed-form expressions or correction functions for contact models that involve hyperelastic properties and large deformations. For example, Zhang et al. [22] conducted finite element simulations to develop contact models corresponding to four hyperelastic materials characterised by the neo-Hookean, Mooney–Rivlin, Fung, and Arruda–Boyce constitutive models. Dimitriadis et al. [132] formulated correction functions for Hertz’s contact model, enabling more accurate predictions in cases involving large deformations during indentation.

Additionally, indentation simulations, when combined with experimental indentation tests, are commonly employed for inverse analyses to extract material properties of skin and arterial tissue. Notable studies include those by Tran et al. [55], Delalleau et al. [57], [138], Cox et al. [144], [209], Chai et al. [59], [60], as well as Fougerson [210] and Akbarzadeh [211]. In inverse analysis, material parameters are optimised by minimising the error between simulation results and experimental data, ensuring that the error remains within a predefined tolerance. We have also applied this technique in our work, with further details provided in Appendix A.

Indentation simulations are also valuable tools for validating theoretical contact models. For instance, Lin et al. [117] employed indentation simulations to validate their established contact model that incorporates hyperelastic material properties defined by the linear elastic, Mooney–Rivlin, and Ogden models. Additional examples can be found in the studies by Dai et al. [212], Dintwa [213], Wei et al. [214], and Guo et al. [215].

FE models in these studies are typically developed as 2D axisymmetric models due to the symmetric geometry of both the indenter and the indented material, with the material often represented as a thin layer. This approach is commonly applied in studies such as those by Zhang et al. [22], Delalleau et al. [57], [138], and Slaboch et al. [216]. However, some studies, such as those by Cox et al. [144], [209], Chai et al. [59], [60], and Valero et al. [217], use 3D axisymmetric FE models to simulate indentation on skin and arterial tissue. These models provide a more detailed and realistic representation but incur higher computational costs. Tran et al. [55] offer a realistic model based on MRI images of the human left forearm, which includes three skin layers (epidermis, dermis, and hypodermis), muscles, and bones. Similarly, [216] provides a three-layer skin model. While these offer valuable insights, they are computationally demanding. In contrast, for the purposes of this research, a 2D axisymmetric model was adopted to balance computational efficiency and model

accuracy, as detailed in Chapters [3](#) and [4](#).

Further details on the numerical simulations for artery-stent interactions are presented in Chapter [5](#).

3

Nonlinear Contact

Contents

3.1 Introduction	41
3.2 Modelling Non-linear Contact	43
3.3 Application of the Method	47
3.3.1 Indentation Experiments	47
3.3.2 Numerical Simulation	48
3.3.3 Determining the Ogden Parameters from the Assembled Curves	50
3.4 Verification	51
3.5 Conclusion	52

The content of this chapter has been published for publication in the Journal of the Mechanical Behaviour of Biomedical Materials [218].

3.1 Introduction

As previously discussed in Chapter 2, a comprehensive understanding of the hyperelastic characteristics of soft materials is essential for exploring their tribo-mechanical behaviour. One possible application is the interaction of soft tissue with medical devices, such as a respirator mask on the skin or a stent with arterial tissue. The hyperelastic material model proposed by Ogden [48] has been widely used and accurately describes the mechanical behaviour of soft materials, including skin [13], [75], [219] and arterial tissue [77]–[81]. Determining the Ogden parameters typically requires tensile experiments [220], [221], which may not be feasible for *in-vivo* tissue, such as human skin. Lin et al. [117] proposed a method for determining first-order Ogden parameters using indentation experiments, assuming the material behaves as an infinite half-space and that deformations are small. However, many studies assume linear elastic behaviour of soft tissue during the

indentation test [154], [214], [222], [223].

Reviewing the contact models discussed in Chapter 2, Hertz's model [115] for elastic contact is the most widely used and describes the contact behaviour between two spherical bodies by assuming small deformations, elastic infinite half-spaces, and friction-free surfaces. Although these assumptions do not hold for most contact cases, Hertz's model can still be used for many applications to determine the elastic modulus of the material, and the contact area for a given loading situation. Sneddon [118] established a general contact model for indenters with arbitrary profiles, such as cones, cylindrical punches, and ellipsoidal shapes. However, this model is still based on the assumptions of small deformations, elastic half-spaces, and friction-free surfaces. Hayes [127] developed a mathematical model to describe the contact behaviour of a finite elastic layer bonded to a rigid half-space. This model introduces two parameters, κ and χ , that depend on the Poisson's ratio as well as the ratio of contact radius to layer thickness to correct for the effects of the underlying substrate on the contact force F and the contact radius a , respectively. This model addresses the contact problem for thin, soft tissue covering a harder tissue, e.g., cartilage bonded to bone. However, Hayes' model assumes linear elastic material behaviour, and thus cannot be used for obtaining hyperelastic parameters. Other methods that are used in literature to determine Young's modulus for soft materials include the Bec/Tonck model [224] and the Song/Pharr model [225], but these are all based on linear elasticity. Lin's contact model [117] incorporates non-linear elasticity in the form of a first-order Ogden model, however, this model is still based on the assumptions of an infinite half-space and small deformations.

Inverse analysis is another widely used method for deriving Ogden parameters from indentation tests [57], [138], [226], [227]. It is an optimisation process where the unknown material parameters are iteratively optimized by minimizing the error between simulated and experimentally measured force. Unlike direct modelling, inverse analysis is heavily dependent on optimisation techniques, such as the Levenberg-Marquardt algorithm. Throughout each iteration, the simulation model is required to run at least once to obtain new parameters, rendering this iterative process notably time-consuming.

In this chapter, a method was proposed to determine the Ogden parameters of artificial soft tissue by combining indentation experiments with numerical simulations. The methodology involves performing indentation tests to obtain the curves of contact force F versus indentation depth ω curves, while numerical simulations are used to compute the principal strains ε_j during indentation. Assuming that the non-linear material behaviour can be linearised for small deformation increments, the contact problem during the indentation is discretised into multiple increments. In

each increment, the material is treated as linearly elastic, characterised by an ‘equivalent shear modulus’, G_i^* , derived from Hayes’ model. By applying the principal strains ε_j obtained from simulations, the principal stresses σ_j are subsequently calculated using Hooke’s law. By combining the principal stresses σ_j of all increments, non-linear stress σ_j versus strain ε_j curves are generated, from which the Ogden parameters, μ and α , are obtained through fitting.

3.2 Modelling Non-linear Contact

The constitutive model of soft materials can be described using an N -order Ogden model [48], which relates the strain energy potential ψ to the principal stretch ratios λ_j ($j = 1, 2, 3$) in three directions, as shown in Eq. [3.1].

$$\psi(\lambda) = \sum_{p=1}^N \frac{\mu_p}{\alpha_p} (\lambda_1^{\alpha_p} + \lambda_2^{\alpha_p} + \lambda_3^{\alpha_p} - 3) \quad (3.1)$$

where μ_p and α_p are material parameters with p is the term number.

The relationship of shear modulus G with Ogden parameters μ_p and α_p is defined as Eq. [3.2] [48].

$$2G = \sum_{p=1}^N \mu_p \alpha_p \quad (3.2)$$

Ogden’s model is formulated as a linear combination of invariants $\phi(\lambda)$, as given by Eq. [3.3] where the relationship between strain ε_j and stretch λ_j is described by Eq. [3.4] [48].

$$\phi(\lambda) = \begin{cases} (\lambda_1^{\alpha_p} + \lambda_2^{\alpha_p} + \lambda_3^{\alpha_p} - 3)/\alpha_p & \alpha_p \neq 0 \\ \ln \lambda_1 \lambda_2 \lambda_3 & \alpha_p = 0 \end{cases} \quad (3.3)$$

$$\varepsilon_j = \begin{cases} \frac{1}{\alpha_p} (\lambda_j^{\alpha_p} - 1) & \alpha_p \neq 0 \\ \ln \lambda_j & \alpha_p = 0 \end{cases} \quad (3.4)$$

In [48], the case of $\alpha_p = 0$ is excluded from consideration, and we follow the same approach here. Eq. [3.4] provides clear evidence that the principal strain depends only on λ_j and the Ogden parameter α_p , without being influenced by μ_p .

Considering the principal Cauchy stresses defined by the Ogden model, the principal Cauchy stresses can be given by Eq. [3.5] [48].

$$\sigma_j = \lambda_j \frac{\partial \psi}{\partial \lambda_j} - P = \sum_{p=1}^{\alpha_p} \mu_p \lambda_j^{\alpha_p} - P \quad (3.5)$$

where P is an arbitrary hydrostatic pressure introduced because of the incompressibility constraint.

To establish the relationship between stress and strain during the indentation process, we adopt the simplification that the stress in uniaxial compression can serve as an analogy for that in spherical indentation, as given in Eq. 3.6. Based on the works of Tabor [228], [229] and Johnson [230], Lin [117], [231] and Briscoe [232] showed that the stresses in a material during spherical indentation may be approximated by those occurring during uniaxial compression. Whilst formally this will introduce some error in the determination of the stress field, this is not explicitly accounted for.

$$\sigma_j = \sum_{p=1}^N \mu_p \left(\lambda_j^{\alpha_p} - \lambda_j^{-\frac{1}{2}\alpha_p} \right) \quad (3.6)$$

Substituting Eq. 3.4 into Eq. 3.6, the principal stress-strain relationship along the z -axis can be described by Eq. 3.7

$$\sigma_j = \sum_{p=1}^N \mu_p \alpha_p \left[\left(\varepsilon_j + \frac{1}{\alpha_p} \right) - \left(\frac{1}{\alpha_p} \right)^{\frac{3}{2}} \left(\varepsilon_j + \frac{1}{\alpha_p} \right)^{-\frac{1}{2}} \right] \quad (3.7)$$

Let $\Xi(\varepsilon_j, \alpha_p) = \left(\varepsilon_j + \frac{1}{\alpha_p} \right) - \left(\frac{1}{\alpha_p} \right)^{\frac{3}{2}} \left(\varepsilon_j + \frac{1}{\alpha_p} \right)^{-\frac{1}{2}}$. It is evident that $\Xi(\varepsilon_j, \alpha_p)$ is a nonlinear function of the principal strain ε_j and the Ogden parameter α_p . Expanding $\Xi(\varepsilon_j, \alpha_p)$ using a Taylor series yields the following expression:

$$\Xi(\varepsilon_j, \alpha_p) = \frac{3}{2}\varepsilon_j - \frac{3}{8}\alpha_p\varepsilon_j^2 + o(\varepsilon_j^2) \quad (3.8)$$

where $o(\varepsilon_j^2)$ higher-order terms beyond the quadratic term.

Since our goal is to determine the Ogden parameters μ_p and α_p , i.e. the shear modulus G (Eq. 3.2) from measured indentation data, we need to understand the stress field in the indented material. The strain field during indentation is obtained using a numerical simulation, and is independent of any underlying material model. The following paragraphs demonstrate how the stresses σ_j are determined by approximating the hyperelastic material as a linear material in small increments.

$\Xi(\varepsilon_j, \alpha_p)$ can be segmented into i adjoining segments as $\Xi_i(\varepsilon_{ji}, \alpha_p)$. For each segment, $\Xi(\varepsilon_{ji}, \alpha_p)$ can be approximated by the linear function, $k_i\varepsilon_{ji}$, as shown in Eq. 3.9, where $\Delta\varepsilon$ is the step size of each small increment.

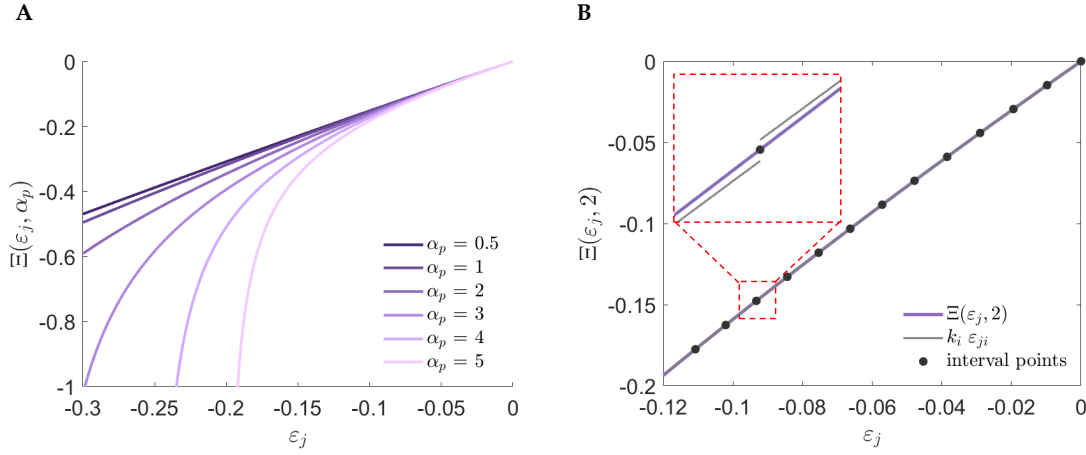


Figure 3.1: Illustration of linearisation and segmentation: **A**) $\Xi(\varepsilon_j, \alpha_p)$ for $\alpha_p = 0.5, 1, 2, 3, 4$ within the range $\varepsilon_{ji} \in [0, -0.3]$, showing that $\Xi(\varepsilon_j, \alpha_p)$ is independent of the value of α_p for strains smaller than 5%. **(B)** $\Xi(\varepsilon_1, 2)$ can be approximated by three adjoining linear segments of $k_i \varepsilon_{ji} + b_i$ ($i = 1, 2, 3$).

$$\Xi_i(\varepsilon_{ji}, \alpha_p) \approx \begin{cases} k_i \varepsilon_{ji} & i - 1 \leq \frac{\varepsilon_{ji}}{\Delta \varepsilon} \leq i \\ 0 & \text{otherwise} \end{cases} \quad (3.9)$$

Then, $\Xi(\varepsilon_j, \alpha_p)$ can be given as:

$$\Xi(\varepsilon_j, \alpha_p) \approx \sum_{i=1}^m \Xi_i(\varepsilon_{ji}, \alpha_p) = \sum_{i=1}^m k_i \varepsilon_{ji} \quad (3.10)$$

where m is the number of segments.

Figure 3.1A illustrates the $\Xi(\varepsilon_j, \alpha_p)$ for $\alpha_p = 0.5, 1, 2, 3, 4$ within the range $\varepsilon_{ji} \in [0, -0.3]$. The figure shows that at small strain $\varepsilon_{ji} \in [0, -0.2]$, $\Xi(\varepsilon_j, \alpha_p)$ behaves nearly linearly, especially for small α_p values, such as $\alpha_p = 0.5$ to 2. However, as ε_j increases ($\varepsilon_{ji} \in [-0.2, -0.3]$), the influence of α_p becomes more pronounced, leading to greater deviations from linearity, particularly for large α_p values, such as $\alpha_p = 3$ to 5.

The Cauchy stress can be written as:

$$\sigma_j \approx 2G \sum_{i=1}^m k_i \varepsilon_{ji} \quad (3.11)$$

where ε_{ji} ($j = 1, 2, 3, i = 1, 2, \dots, m$) represents the principal strains in small increment i .

As each increment introduces a different value for the tangential slope k_i , the equivalent shear modulus $G_i^* = 2Gk_i/(1 + \nu)$ can be used to approximate the hyperelastic material as a linear

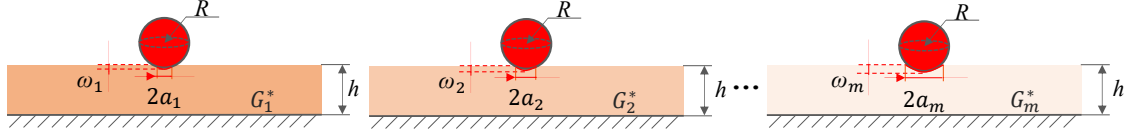


Figure 3.2: The contact behavior of hyperelastic materials can be modeled as linear materials, characterized by the equivalent shear modulus G_i^* in the i^{th} increment.

material provided the increments are small. The principal stress-strain relationship can then be expressed as:

$$\sigma_j \approx \sum_{i=1}^m 2(1+\nu)G_i^* \varepsilon_{ji} = \sum_{i=1}^m \sigma_{ji}^* \quad (3.12)$$

It is important to note that the principal Cauchy stresses σ_j , as defined by Ogden model, can be expressed as the sum of σ_{ji}^* over small increments, each corresponding to a linearised material with an equivalent shear modulus G_i^* .

Eq. 3.12 indicates that the principal Cauchy stresses σ_j can be obtained by assembling the incremental stresses σ_{ji}^* using Eq. 3.12. Using Hayes model, Gistar is then obtained from the measured force-displacement curve, according to y Eq. 3.13 [127].

$$G_i^* = \frac{F_i(1-\nu)}{\kappa\sqrt{\chi}R^{1/2}\omega_i^{3/2}} \quad (3.13)$$

where F_i is the incremental contact force, ω_i is the incremental indentation depth, R is the radius of the spherical indenter, κ and χ are the numerical coefficients as defined by Hayes' model, which depend on the ratio of contact radius with the thickness h , and Poisson's ratio ν .

Summarising, the overall contact behavior of hyperelastic materials can be described as the combination of multiple Hayes contact models across different small deformation increments, as defined by Eq. 3.14 and illustrated in Figure 3.2.

$$F = \sum_{i=1}^n F_i = \sum_{i=1}^n 2\kappa\sqrt{\chi} \frac{G_i^*}{1-\nu} R^{1/2} \omega_i^{3/2} \quad (3.14)$$

where F is the contact force for hyperelastic materials.

The procedure is schematically shown in Figure 3.3

3.3 Application of the Method

The developed incremental model is applied to determine the Ogden parameters of a commercially available tissue mimic (SynTissue Adult Skin Plate 2N, Syndaver, Tampa, FL). The following procedure is schematically illustrated in Figure 3.3.

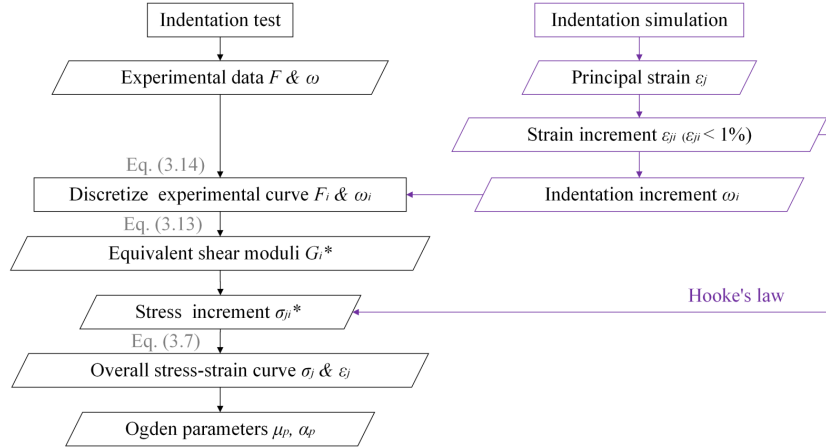


Figure 3.3: The procedure of the proposed method: the black block gives the information collected from the indentation test, while the purple block gives the data collected by the indentation simulation. The whole procedure was conducted in Matlab (Matlab 2021Ra).

A first-order Ogden model was used for these simulations, and the strain increment was set at an arbitrary value of 1%. For indentations up to 10 % of the material thickness, the numerically calculated strain fields are independent of the value of the Ogden parameters used in the simulation. This will also be discussed in more detail in Figure 3.5.

3.3.1 Indentation Experiments

Indentation tests on specimens of Syndaver tissue mimic were performed employing a Mach-1 mechanical tester (Biomomentum, Laval, Canada) using a spherical indenter with a radius R of 9.5 mm, as depicted in Figure 3.4A. Measurements were performed on 8 random locations on the specimen, at least 10 mm from the edge. The overall thickness of the tissue mimic is around 1 mm, but before performing the indentation a more accurate assessment of the local thickness h at each test point was determined by measuring the height difference between two surfaces of the specimen. After contact detection a downward displacement ω of 10 % of the local thickness h was applied. Each indentation took 10 s to complete. The measured data was processed to correct for variations in the detection of the point of initial contact, i.e., the zero-displacement location (233) thus ensuring that all curves commence at a similar starting point.

The eight resulting contact force F versus indentation depth ω curves are depicted in Figure. 3.4B. The curves show some variations, with the maximum indentation force ranging between 0.44 N and 0.67 N, which is expected for inhomogeneous materials such as artificial or biological tissue.

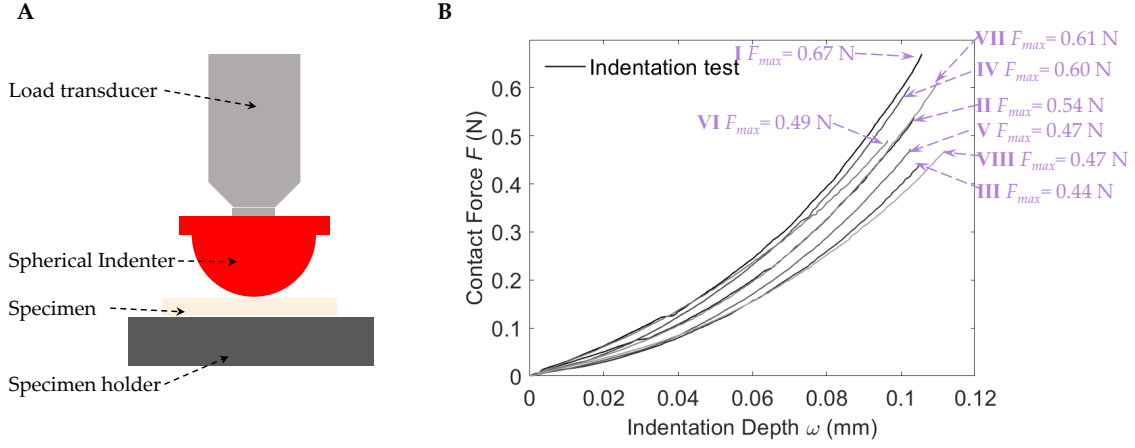


Figure 3.4: The indentation tests: (A) Schematic of the indentation setup. (B) 8 measured force-displacement curves on different locations on the same specimen, where I, II, ..., VIII represents the test number.

3.3.2 Numerical Simulation

A finite element model of the indentation experiments was established (ABAQUS/Standard 2024) as shown in Figure 3.5A. Similar to the experimental setup, the model comprises a spherical indenter with radius $R = 9.5$ mm that indents a specimen with a thickness $h = 1$ mm. The steel indenter was treated as a rigid body to simplify the simulation. A total indentation depth $\omega = 0.1$ mm was modelled, with contact properties set as ‘hard contact’ and ‘frictionless’. The bottom surface of the indented material was constrained in all directions. The indented material was modelled using a first-order Ogden model.

To assess the effects of the material parameters on the calculated principal strains and stresses, three Ogden materials, listed in Table 3.1, were defined and assigned to the indented material in subsequent simulations.

Table 3.1: Three Ogden materials used to assess differences in calculated stress and strain fields

Ogden Materials	Shear Modulus G (MPa)	Ogden Parameter α_p (-)
Material 1 (G_1, α_1)	0.256	1.2
Material 2 (G_2, α_2)	1.000	1.2
Material 3 (G_3, α_3)	0.256	10.0

Figure 3.5B shows the numerical strain at the centre of the contact ($x = y = 0$) as a function of depth z into the material, as calculated for the three simulated materials defined in Table 3.1. The

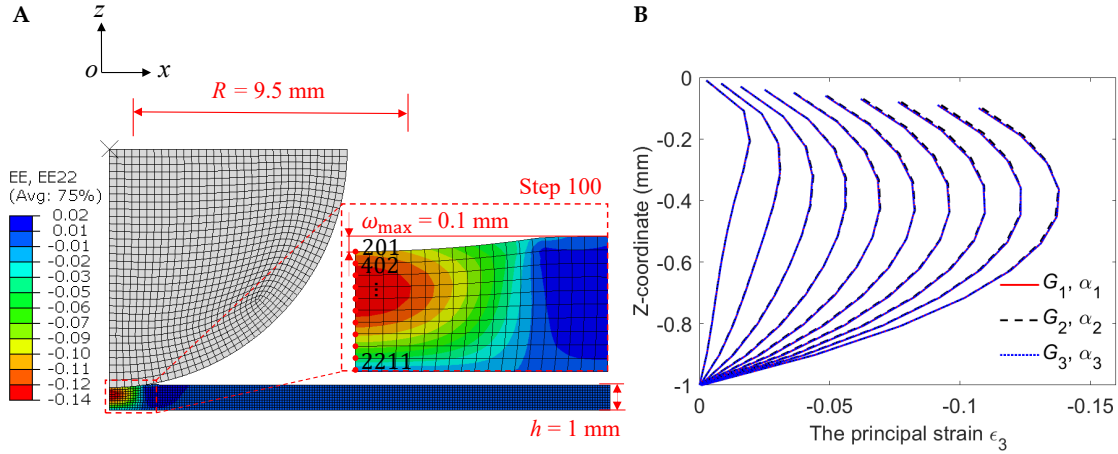


Figure 3.5: The FE model and the calculated principal strains: (A) The FE model for indentation test. (B) Calculated principal strain field along the Z-axis at different indentation depths for the three simulated materials, the red numbers represent the indentation depth in mm.

various curves refer to indentation depths between 0.01 and 0.12 mm. The curves for the three materials coincide, meaning that for these indentations the numerical strains are independent of the value of the shear modulus G and the Ogden parameter α_p .

The numerical strain calculated by ABAQUS is Green's strain, denoted as ϵ_j^G , which is equal to Hill strain (Eq. 3.4 [234], [235]) when $\alpha = 2$. The relationship between the general Hill strain used in Ogden's model ϵ_j and Green strain ϵ_j^G is given by Eq. 3.15

$$\epsilon_j = \frac{1}{\alpha} \left[(2\epsilon_j^G + 1)^{\alpha/2} - 1 \right] \quad (3.15)$$

Using Taylor's expansion, this can be expressed as Eq. 3.16

$$\epsilon_j = \epsilon_j^G + \frac{\alpha - 2}{2} \epsilon_j^{G^2} + o(\epsilon_j^{G^2}) \quad (3.16)$$

where $o(\epsilon_j^{G^2})$ represents terms of order higher than the second power of ϵ_j^G . For $\alpha_p = 2$, the higher order terms equal to 0 and $\epsilon_j = \epsilon_j^G$. The relationship between the FE calculated Green strain ϵ_j^G and the Hill strain ϵ_j is illustrated in Figure 3.6, with the curve for $\alpha_p = 2$ shown in black.

In this chapter, since the true principal strain ϵ_j is initially unknown, the numerical strain ϵ_j^G is used to compute the principal stresses via Hooke's law following the procedure outlined in Figure 3.3. Once the parameters α_p and μ_p are identified, the value of α_p is applied to the numerical strain field to estimate ϵ_j using Eq. 3.15. By repeating this procedure iteratively, the solution converges to a consistent value of α_p , which can then be used in the Ogden model.

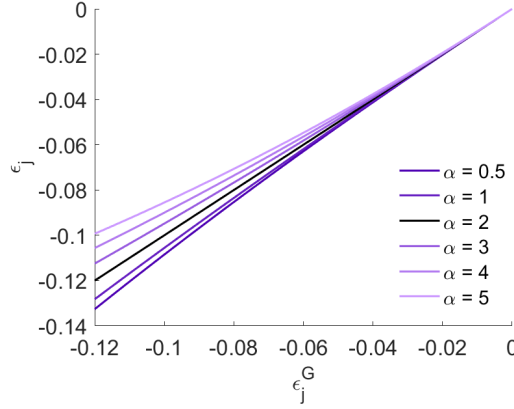


Figure 3.6: Relationship between the Hill strain ε_j and the Green strain ε_j^G . For small strains, $\varepsilon_j \approx \varepsilon_j^G$.

For larger indentation depths, i.e., when the indentation surpasses 12 to 15% of the specimen thickness, the value of α_p will affect the principal strains and the optimisation problem becomes more complex. For these conditions, the relationship between indentation depth, principal strains, and material properties requires further numerical investigations.

3.3.3 Determining the Ogden Parameters from the Assembled Curves

The measured force-displacement curves shown in Figure 3.4B are discretised based on the calculated principal strains and the maximum strain step of 1%. Figure 3.7A illustrates the discretisation, whilst Figure 3.8B shows the equivalent shear moduli G_i^* determined for each discretisation step using Hayes model (Eq. 3.14). It is important to note that for a linear elastic indentation, the contact force F scales with indentation depth ω^n , with $n = 3/2$, as shown in Eq. 3.14. This means that the contact stiffness increases with increasing indentation depth. The measured indentation curves in Figure 3.4B exhibit a slope $< 3/2$, which may be attributed to a reduction of the equivalent shear modulus of the material as the indentation depth increases, as illustrated in Figure 3.7B. The stress σ_3 in each increment can then be obtained by multiplying the strains (Figure 3.7C) with the obtained elastic modulus for each increment, as shown in Figure 3.7D.

To obtain the Ogden material parameters, the relationship between stress and strain is determined on a grid internally to the indented material. For each point on the grid, local values for μ_p and α_p can be determined by fitting Eq. 3.12 to this curve. The fitting of the first-order, second-order, and third-order Ogden models is conducted using a least-squares curve fit. The resulting shear modulus G for an N^{th} -order Ogden model, defined as Eq. 3.2 is then shown for every point inside the material, see Figure 3.8A. The obtained shear modulus is fairly constant with depth,

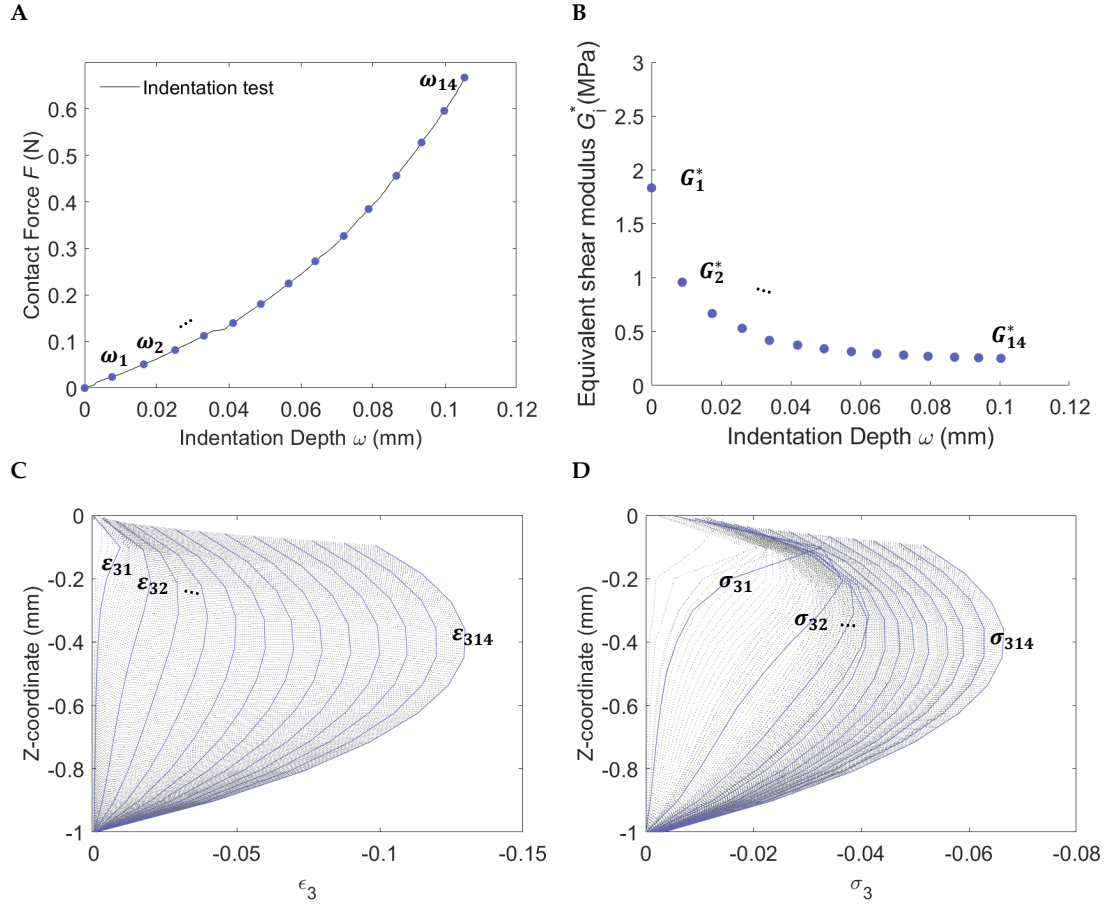


Figure 3.7: Determining the Ogden parameters: (A) The discretised force curve and (B) obtained shear moduli G_i^* . (C) strain fields ϵ_3 obtained by the numerical simulation. (D) stress fields σ_3 using Hooke's law.

at approximately 0.2 MPa, except for the interface of the indented material with the rigid sample holder. In the finite element model, this point was fully constrained and therefore exhibits an 'artificial' stiffness. Consequently, the calculated shear modulus for the interface will be ignored. There appears to be not much difference between the values obtained for the first, second, and third-order Ogden models, suggesting that under the prescribed strain conditions, the material behaviour is sufficiently accurately described using a first-order Ogden model. Figure 3.8B shows the average shear moduli obtained for the 8 indentation experiments originally shown in Figure 3.4. The shear moduli range between 0.19 and 0.13, indicative of the previously observed local variations in indentation behaviour.

3.4 Verification

To verify the obtained values, the determined Ogden material parameters were inserted into the finite element simulation described in Figure 3.5. The resulting calculated force-displacement curve

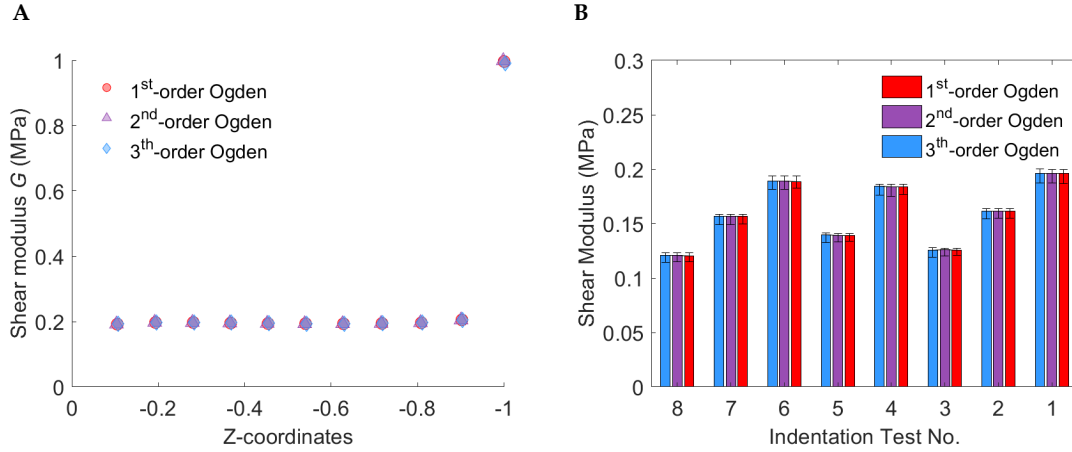


Figure 3.8: Shear moduli obtained from the various indentation experiments: (A) the shear moduli along the Z-axis for one indentation result. (B) the average shear modulus for the 8 indentations performed on the material, as shown in Figure 3.4.

from the simulation (solid black line) can be compared to the experimentally determined curves (black dots). For reference, results obtained using Hayes' model, i.e., assuming linear elasticity whilst taking into account the indented material being thin and placed on a rigid surface, are also included (gray solid line). The results for all eight indentation experiments are shown in Figures 3.9.

All graphs in Figure 3.9 show that Hayes' model does not fully capture the obtained experimental data, showing a smaller force at the start of the curve and a larger force at the end. The gradient of the force displacement curve obtained with Hayes' model differs significantly from the experimental results, particularly for larger indentation depths. This implies that modulus values obtained using Hayes' model may be too stiff at larger deformations. The experimentally obtained curves can be approximated more accurately using Ogden's hyperelastic material parameters, and in this investigation, it is suggested that a 1st order Ogden model is sufficiently accurate.

3.5 Conclusion

This chapter presents a newly developed method that enables determining the Ogden parameters from experimental indentation tests. The method uses the strain fields ε captured using a numerical simulation of the indentation process to enable establishing a relationship between deformation and strain, to replace any analytical equations that do not accurately capture the complex strain field underneath an indenter (40). The method is based on the assumption that the hyperelastic material behaviour can be linearised in small incremental deformations and that Hayes' model for thin, linear elastic contact can be applied within separate increments. It was found that the

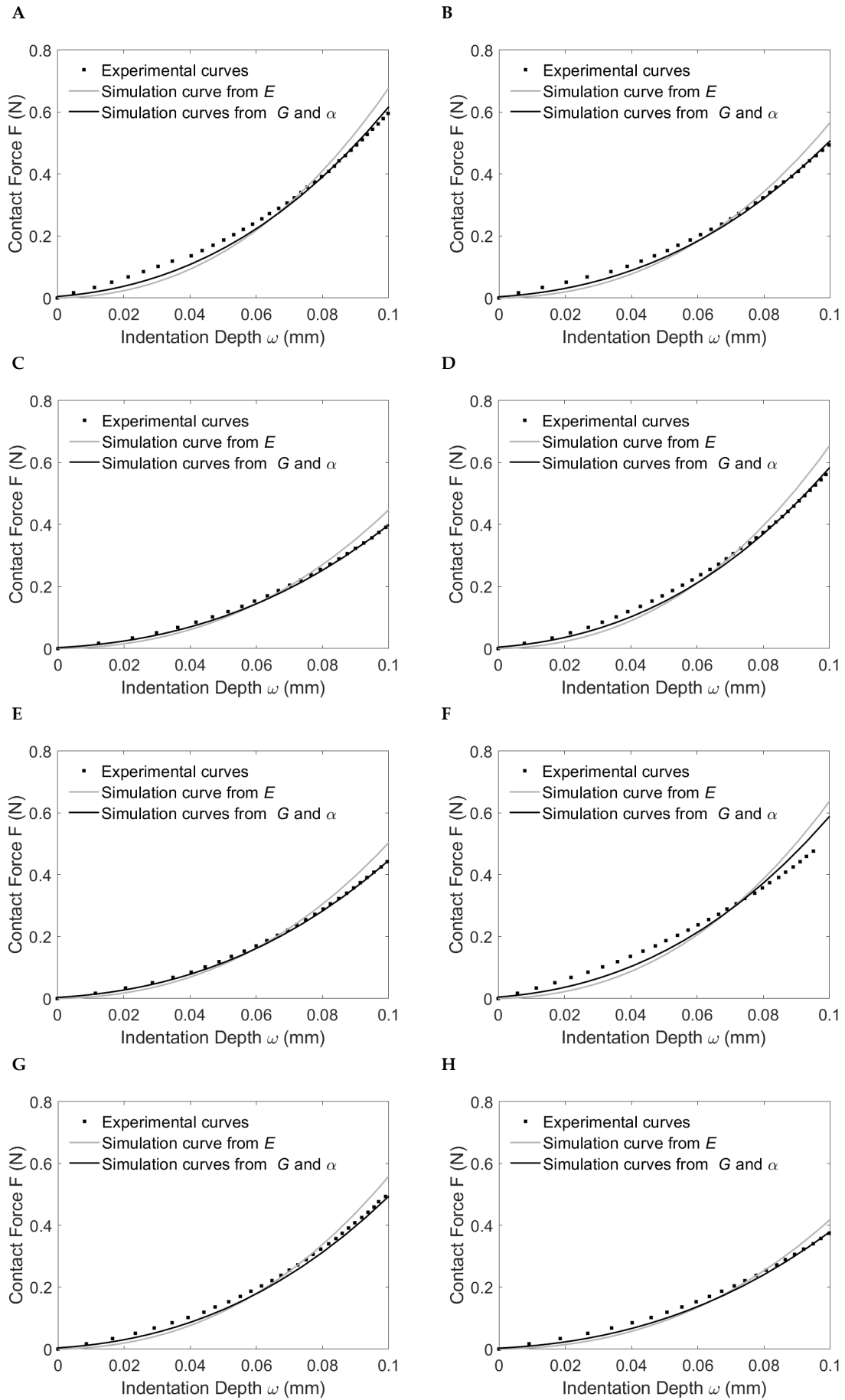


Figure 3.9: Comparison of the experimental data from 8 tests with the simulated data using the parameters obtained by Hayes' model and new developed model.

relationship between contact force and indentation depth as recorded in indentation tests can be accurately described using the obtained first-order Ogden parameters.

Nonlinear Contact on Mouse Arterial Tissue

Contents

4.1 Introduction	55
4.2 Method	57
4.2.1 Determining the Ogden Material Parameters	57
4.2.2 Application Window of the Method	61
4.2.3 Determining Ogden Parameters using Simulation Curves	64
4.3 Application of the Method on Mouse Arterial Tissue	67
4.4 Discussion	71
4.5 Conclusion	74
4.6 Supplementary	75

The content of this chapter has been published for publication in the Journal of the Mechanical Behaviour of Biomedical Materials [236]. As a result, there may be some repeat content from previous chapters.

4.1 Introduction

Arterial tissue plays a crucial role in the cardiovascular system and is characterised by unique structural and mechanical properties that facilitate blood flow. Typically, arterial tissue is thin, small, and has a narrow cross-section. The hyperelastic properties of arterial tissue allow it to undergo significant deformation in response to internal pressures and external forces. Characterising these hyperelastic properties is essential for analysing cardiovascular disease. However, the combination

of the tissue's small size, thin and narrow structure, and hyperelastic behaviour present substantial challenges for mechanical testing.

Indentation tests on arterial tissue have been explored in several studies (e.g., [125], [155], [237], [238]), typically employing Hertz's model to characterise the contact behaviour of the tissue. Hertz's model [116], [230] describes contact behaviour of materials under assumptions of linear elasticity, small deformations, and an infinite half-space. [239], [240] report that Hertz's model can accurately capture the elastic behaviour of arterial tissue only at very shallow indentations and when using a relatively large indenter, with $a/R \leq 0.1$ and $h/R \geq 20$, where R is the radius of the indenter, a is the radius of the contact patch between the indenter and the indented tissue, and h is the thickness of the indented tissue. However, arterial tissue has a rough surface, and this introduces significant noise, particularly at shallow indentation depths. This has the potential to compromise the accuracy of the measurements and, therefore, the extracted material properties [240].

To analyse indentation data that is obtained at relatively large depths, a contact model that accounts for the finite thickness of the indented tissue would be more suitable than Hertz's model. Both analytical and numerical approaches have been employed to address the effect of finite specimen thickness on contact mechanics [127], [132], [241]. Hayes introduced two numerical parameters, κ and χ , to correct for the effects of finite thickness on the contact force F and the contact radius a , respectively. However, Hayes' model is formulated for materials with smooth surfaces and requires precise knowledge of the thickness of the indented material. The rough surface of arterial tissue complicates the thickness measurement required for the application of the model, see e.g. [58], [233], [242], [243], whilst Chandran et al. [244] noted that surface roughness also can obscure the point of initial contact.

Contact problems involving materials that behave hyperelastically are more challenging to address than those for linearly elastic materials [245], especially using analytical methods. Assuming small deformations and an infinite half-space, Lin [117] proposed a method to determine the Ogden parameters from indentation experiments. However, their model results in significant errors when applied to thin biological tissue such as arteries. To obtain the hyperelastic properties of a material, an inverse FE approach may be employed, see e.g. [138], [210], [211]. In such an approach, the hyperelastic parameters are determined by changing the input material parameters in an FE model until the error between the calculated force and the measured experimental force falls within a specified tolerance. Compared to fitting an analytical expression, the inverse FE method entails significantly higher computational cost [246] and, therefore, research efforts continue to focus on

interpreting force-displacement curves using approximate formulations, see for instance [241], [247]–[250].

In this chapter, we use indentation testing to characterise the hyperelastic properties of arterial tissue. The hyperelastic behaviour of arterial tissues is assumed to be isotropic and modelled using the Ogden model, with the Ogden parameters μ and α determined to facilitate their use in numerical modelling efforts. The method is based on our previous work (Chapter 3 and [218]), which is applicable to thin and smooth tissue, i.e. tissue for which the thickness is unambiguously defined and known. In this chapter, we aim to quantify the thickness of thin tissue specimens with substantial surface irregularities. We propose defining an equivalent thickness h_{hayes} as the thickness value that yields the smallest variance of shear modulus for a range of indentation increments. The method is demonstrated by applying it to numerically simulated force-indentation curves and, subsequently, applied to 26 experimental contact force versus indentation depth curves that were obtained on mouse tissue.

4.2 Method

4.2.1 Determining the Ogden Material Parameters

The hyperelastic behaviour of mouse arterial tissue can be characterized using a first-order Ogden model, as described by Eq. 4.1 [48].

$$\psi = \frac{\mu}{\alpha}(\lambda_1^\alpha + \lambda_2^\alpha + \lambda_3^\alpha) \quad (4.1)$$

where ψ is the strain-energy function, and λ_j ($j = 1, 2, 3$) is the stretch ratio. It should be noted that, since the first-order Ogden model is used, the Ogden parameters μ_p and α_p used in Chapter 3 are simplified as μ and α here. The shear modulus G is defined in terms of the Ogden parameters μ and α by Eq. 4.2

$$G = \frac{\mu\alpha}{2} \quad (4.2)$$

Following the assumption made in Chapter 3 and Chapter 4, the overall contact force for a hyperelastic material can be described as the sum of multiple Hayes' contacts on linear material with the equivalent shear modulus G_i^* over different small deformation increments, as defined by Eq. 4.3. The principal Cauchy stresses can be obtained by assembling the principal Cauchy stresses

from these linear materials, described as Eq. 4.4

$$F = \sum_{i=1}^m F_i \approx \frac{R^{1/2}}{1+\nu} \sum_{i=1}^m \kappa \sqrt{\chi} G_i^* \omega_i^{3/2} \quad (4.3)$$

where F_i is the incremental contact force, ω_i is the incremental indentation depth, R is the radius of the spherical indenter, and ν is Poisson's ratio. κ and χ are the numerical coefficients as defined by Hayes [127], which depend on the ratio of contact radius over the thickness a/h and therefore have different values for each increment.

$$\sigma_j \approx \sum_{i=1}^m 2(1+\nu) G_i^* \varepsilon_{ji} = \sum_{i=1}^m \sigma_{ij}^* \quad (4.4)$$

where m is the number of increments and ε_{ji} represent the principal strain, σ_{ij}^* is the principal Cauchy stresses given by linear material with an equivalent shear modulus G_i^* .

Finite element model

A finite element (FE) model simulating the indentation of a thin tissue specimen with a 150 μm -radius spherical indenter was established in ABAQUS/Standard (ABAQUS 2024). The radius $R = 150 \mu\text{m}$ was chosen to match the experimental setup used in this study, as discussed later. The indenter is treated as a rigid body to simplify the simulation. The contact properties are defined as 'hard contact' and 'frictionless', and the bottom surface of the indented tissue is constrained in all degrees of freedom. The FE simulations were conducted to determine the principal strain for discretising the indentation curve and calculating the principal stress, as well as to verify the proposed method and define the equivalent thickness and application windows of the approach. Therefore, the thicknesses and the material properties of the arterial layer may be varied, based on the requirements of the simulation. Figure 4.3A illustrates the model, where the arterial tissue is represented as a 100 μm -thick layer.

Equivalent thickness h_{hayes}

The application of Hayes' model requires the thickness of the indented material to be known. However, arterial tissue has a rough surface, as schematically illustrated in Figure 4.1A. This complicates both performing the experiments as well as the determination of material parameters from the experimental results. To enable the use of Hayes' model for these rough specimens, we propose establishing a representative or equivalent thickness h_{hayes} (Figure 4.1C) that can be inserted into Hayes' model to analyse the indentation results obtained on rough specimens.

This thickness is established by applying Hayes' model to the tissue, assuming a range of thickness

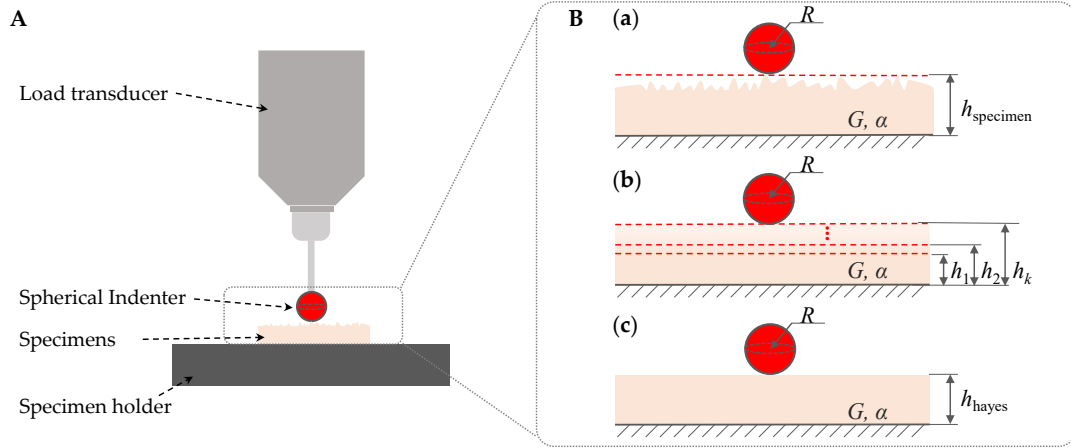


Figure 4.1: Identification of the equivalent thickness h_{hayes} : (A) Diagram of indentation setup. (B) (a) Arterial tissue exhibits a rough surface, which complicates the determination of thickness and prevents its direct application in Hayes' model; (b) A range of possible thickness values h_i are applied to assess the equivalent thickness h_{hayes} for the application of Hayes' model; (c) The equivalent thickness h_{hayes} is determined as the value for h_i that provides the smallest variation of material properties for each strain increment.

values h_i (Figure 4.1B). In Hayes' model, the effect of thickness is corrected by $\kappa\sqrt{\chi}$. The specific values of κ and χ are derived from Fredholm integral equations [127] and fully defined by the ratio of the radius of contact a and layer thickness h , with the contact radius a given by $a = \sqrt{R\omega}$.

$$\kappa = \frac{F(1 - \nu)}{4aG\omega} \quad (4.5)$$

$$\chi = \frac{a^2}{\omega R} \quad (4.6)$$

The specific equations are given in [127] and the values for κ and χ is given in Table 4.1.

Table 4.1: Values of κ and χ for the spherical indenter used in this research ($\nu = 0.5$) [127]

κ	0.6855	0.6963	0.7073	0.7187	0.7810	0.8530	0.9355	1.029	1.135	1.252	1.381	1.522	1.674	2.102	2.597	3.161	3.797	4.507	5.296	6.169	7.130
χ	1.031	1.047	1.063	1.080	1.167	1.258	1.348	1.435	1.516	1.592	1.662	1.725	1.784	1.911	2.017	2.109	2.189	2.260	2.325	2.384	2.438

The established procedure is illustrated by applying it to a simulated indentation curve, for a linear elastic material with a shear modulus $G = 4$ kPa and a thickness of $h = 100$ μm , indented by a sphere with radius $R = 150$ μm . The procedure to obtain the thickness and material properties is applied to this data, for specimens with thicknesses ranging from 50 to 150 μm . This should result in output values similar to those inserted into the simulation: Figure 4.2A illustrates the relationship between $\kappa\sqrt{\chi}$ and the indentation depth ω . The red curve represents the results obtained for the specimen when the thickness is set to 100 μm , while the purple curves depict the

values for the other thicknesses, with increments of 10 μm . It can be seen that the values for $\kappa\sqrt{\chi}$ are similar for small indentations, and gradually disperse with increasing indentation depth.

Figure 4.2B shows the various values for the shear modulus determined using Hayes' model, for specimen thicknesses ranging between 50 and 150 μm . For relatively small indentation depths, the results approximate to the correct input value of 4 kPa independent of the layer thicknesses, but with increasing indentation depth the values start deviating quite considerably. For small thickness values, the obtained shear modulus decreases with increasing indentation depth, whilst for large thickness values the obtained shear modulus increases. For the 'correct' thickness of $h = 100 \mu\text{m}$, the shear modulus is approximately constant with increasing indentation depth. The equivalent thickness h_{hayes} will be defined as the thickness for which the various shear modulus values obtained for different increments show the smallest variance (Eq. 4.7).

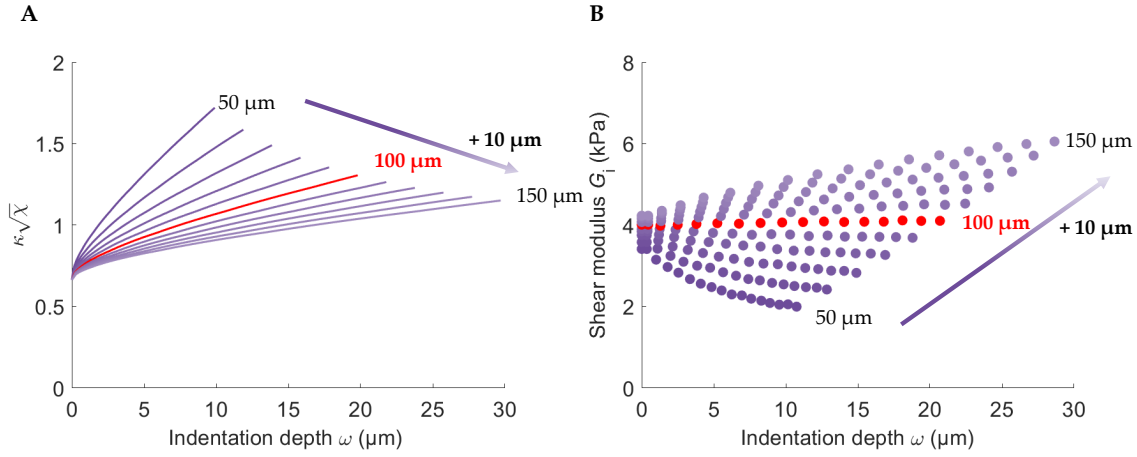


Figure 4.2: Hayes' model in thickness determination: (A) $\kappa\sqrt{\chi}$ versus indentation depth ω . (B) The calculated shear moduli G_i versus indentation depth ω for different thicknesses: red curve gives the G_i for $h_{\text{hayes}} = 100 \mu\text{m}$, purple curves represent the values given by other thicknesses.

$$S^2 = \frac{1}{m-1} \sum_{i=1}^m (G_i - \bar{G})^2 \quad (4.7)$$

where $\bar{G}$ is the mean of shear moduli for all deformation increments and m is the number of deformation increments.

Small principal strains

The procedure involves determining the principal strain during indentation through simulation of the indentation process. The simulated principal strains are then segmented into small strain increments, and the corresponding principal stress within each increment is calculated using Hooke's law.

The result is a compound stress-strain curve that comprises the linearised segments for each increment, from which the Ogden parameters μ and α are derived through the fitting. Since the material properties of the FE arterial layer are homogeneous, the derived values for μ and α should be the same throughout the tissue. It is therefore important that the segmented stress-strain curves are independent of the Ogden parameters, meaning the relationship between strain and displacement should be determined at a location in the tissue where the strains are sufficiently small for α not to have an influence. However, the use of a small indenter introduces large strains, particularly along the z -axis. We evaluate the principal strain at the nodes in the FE model along three selected paths during the indentation for two materials with the same shear modulus $G = 4$ kPa but different Ogden parameters $\alpha = 1$ and $\alpha = 10$. Path 1 is along the axis of indentation, Path 2 is on the edge of the indented zone, and Path 3 is further away from the indenter. Figure 4.3C shows the principal strain in the nodes for the three paths selected, for the two materials with $\alpha = 1$ and $\alpha = 10$. Only for Path 3, the two curves overlap. For Paths 1 and 2, there is a difference in the curves for the two materials. To ensure that there is no effect of α on strains, the strain values calculated along Path 3 were selected to calculate the stresses using Hooke's law. The strain-stress relationship described by Eq. 4.8 combined with Eq. 4.9 are then used to calculate μ and α .

$$\sigma_1 - \sigma_3 = \mu(\lambda_1^\alpha - \lambda_3^\alpha) \quad (4.8)$$

$$\varepsilon_j = \begin{cases} \frac{1}{\alpha}(\lambda_j^\alpha - 1) & \alpha \neq 0 \\ \ln \lambda_j & \alpha = 0 \end{cases} \quad (4.9)$$

4.2.2 Application Window of the Method

The method described above is based on Hayes' model. The shear moduli for all increments are derived using Hayes' model, and the equivalent thickness is subsequently identified based on these shear moduli. However, the small deformation condition $a = \sqrt{R\omega} < 0.1R$ may not be met when using a small indenter in combination with a relatively large indentation depth.

Thus, two key aspects were investigated through numerical simulations: the window in which Hayes' model can be used to determine both the shear moduli and the equivalent thickness (see Figure 4.4B).

Simulations were conducted by assigning arterial layer thicknesses ranging from 50 to 500 μm , with increments of 50 μm . The material of the arterial layer was defined as a linear elastic material

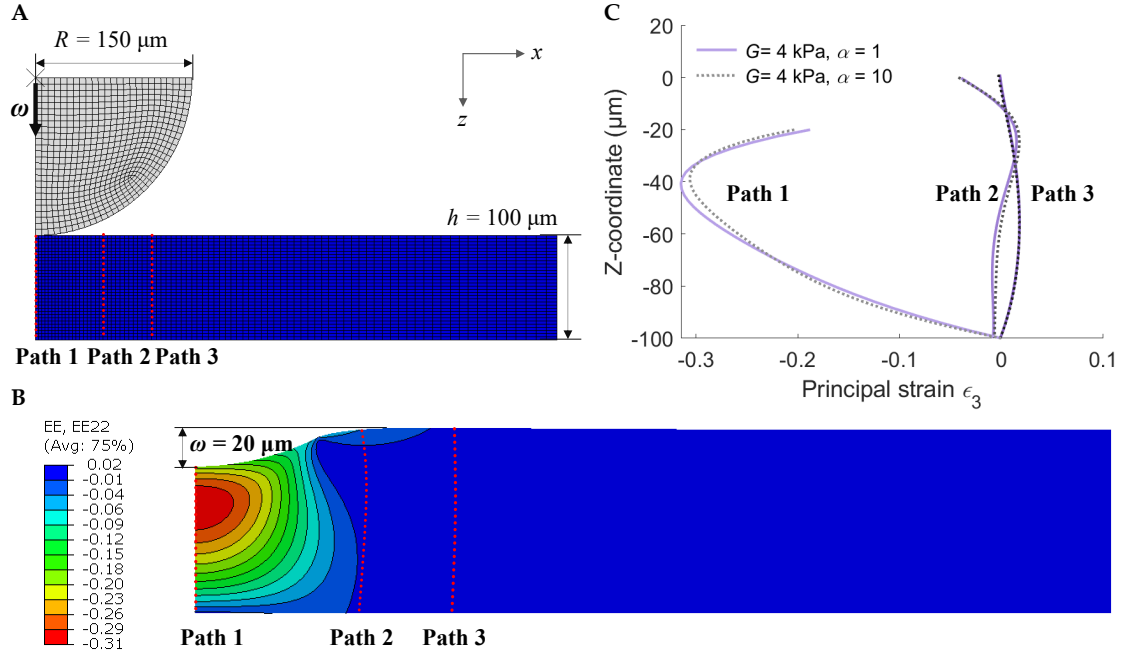


Figure 4.3: FE model for indentation: **(A)** The FE model illustrates a $100 \mu\text{m}$ -thick arterial layer. The red line marks the path used to assess the principal strain ε_j . **(B)** The contour of strain distribution for material $G = 4 \text{ kPa}$, $\alpha = 1$, with the indentation depth $\omega = 20 \mu\text{m}$. **(C)** ε_j is evaluated along three paths for two materials with the same shear modulus $G = 4 \text{ kPa}$ and different Ogden parameter $\alpha = 1$ (purple line), $\alpha = 10$ (grey dotted line). For Path 3, the curves for $\alpha = 1$ and $\alpha = 10$ coincide. For Path 2, where the strains are higher, there is a small but noticeable difference. For path 1, the curves deviate considerably, particularly at higher strain levels.

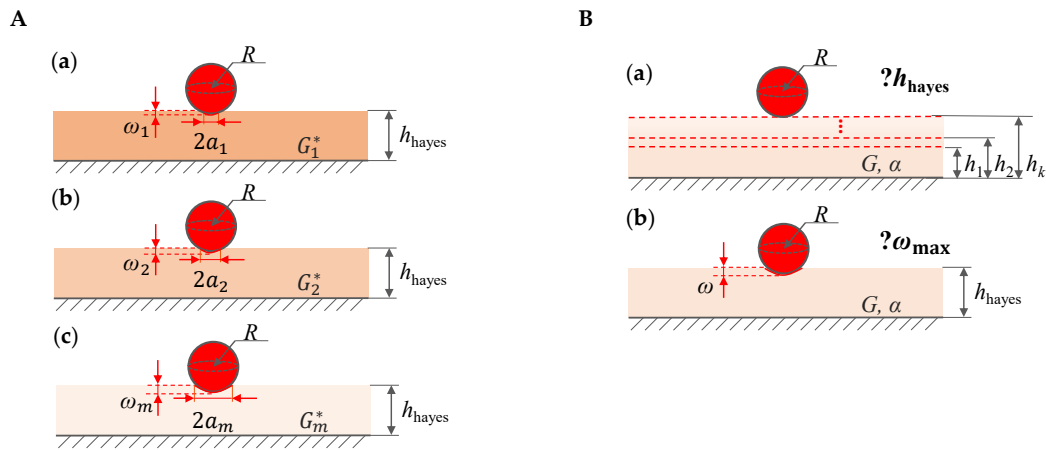


Figure 4.4: (Two key aspects investigated through numerical simulations: **(A)** The hyperelastic material is linearized in small indentation increments, with the G_i^* characterizing the material property at each incremental step. **(B)** Validity window of the newly developed method: (a) To what indentation depths can the model be applied? (b) Is it possible to obtain an accurate value for h_{hayes} ?

with a shear modulus G of 4 kPa and a Poisson's ratio ν of 0.499, meaning an elastic modulus $E = 2G(1 + \nu) = 12$ kPa. A large relative indentation depth, defined as the indentation depth over thickness, of 20 % was applied. As a result, 11 simulated curves of force versus indentation depth corresponding to the 11 thicknesses were collected.

Application windows of Hayes' model

The forces F_{sim} obtained from the numerical simulations were compared with the contact forces F_{hayes} calculated using Hayes' model. The difference between F_{sim} and F_{hayes} is calculated using Eq. 4.10.

$$\Delta F = \left| \frac{F_{\text{hayes}} - F_{\text{sim}}}{F_{\text{sim}}} \right| \quad (4.10)$$

Figure 4.5A shows the forces from the simulations F_{sim} (solid lines) and the contact forces calculated using Hayes' model F_{hayes} (dotted lines) for the 11 simulated thickness values, as a function of the relative indentation depth, i.e., the indentation depth over the thickness of the material. The red dots indicate where the difference between the simulated and Hayes' curves for each thickness reaches 5%, which was defined as the maximum allowable difference.

Figure 4.5B shows the relative indentation depth at which the difference between the input force (F_{sim}) and the output force F_{hayes} equals 5%.

Identification of the equivalent thickness

Figure 4.5C illustrates the variance of shear moduli for the simulation curve obtained with a specimen thickness of 100 μm . 11 possible thickness values, h_i , are assigned to define the FE artery layer for simulations. The calculated strains from each simulation are used to discretise the simulation curve and calculate the shear moduli and the variance. The smallest variance is achieved at $h_i = 100$ μm , which matches the input value.

Figure 4.5D presents the equivalent thickness obtained for 11 thickness (purple dots) values using the same procedure. It shows that the method accurately identifies the actual thickness (black dotted line) using the smallest variance of shear modulus when the thickness is below 300 μm . However, for thicknesses exceeding 300 μm , errors are introduced. Correcting this error, whether through numerical or analytical approaches, remains a significant challenge. Nonetheless, since the arterial thicknesses analysed in this study are predominantly below 300 μm , we address this error manually during the thickness determination process.

From Figures 4.5A and B it can be concluded that the method may be applied for indentations up

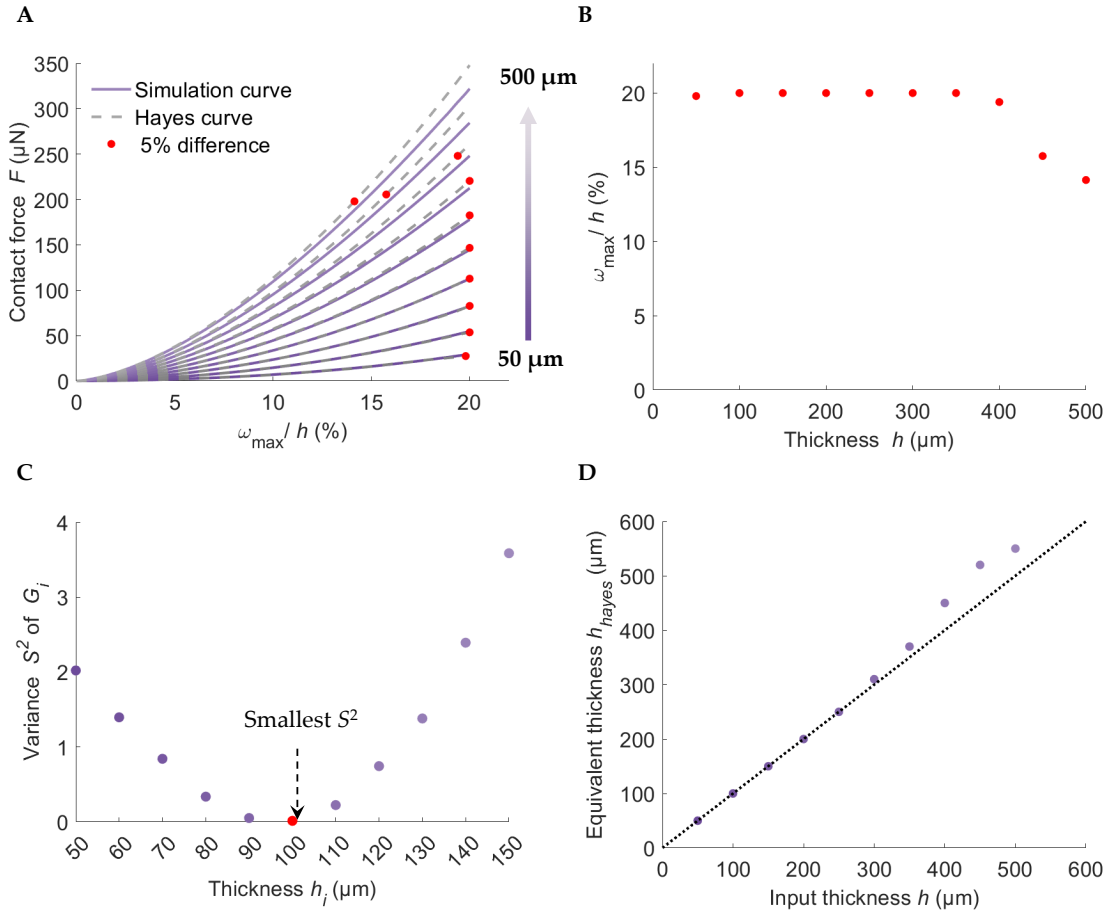


Figure 4.5: Validity of Hayes' model in case of large relative deformations: **(A)** Simulation forces F_{sim} (purple lines) forces calculated using Hayes' model (grey lines). **(B)** The relative indentation depth where the difference between F_{sim} and F_{hayes} equals 5% for different specimen thicknesses. **(C)** The variance of shear moduli for different possible thicknesses is used to identify the equivalent thickness of the artery layer, with a true thickness of 100 μm . The smallest variance is achieved at $h_i = 100$ μm , confirming the equivalent thickness. **(D)** The equivalent thickness h_{hayes} (purple dot) is identified by the smallest variances for different thickness values.

to 20 % of the thickness for specimens with a thickness up to 300 μm , whilst for tissue specimens with a larger thickness the small size of the indenter limits the maximum indentation depth. Figure 4.5D confirms this finding, showing that for specimen thickness above 300 μm the obtained equivalent thickness is larger than the input thickness.

A flowchart of the method is illustrated in Figure 4.6.

4.2.3 Determining Ogden Parameters using Simulation Curves

The functionality of the procedure is verified by evaluating a set of six curves (Figure 4.7A-B) generated from the indentation simulations of arterial tissue with a thickness h of 100 μm . The tissue is modelled as a first-order Ogden material, all with the same shear modulus $G = 4$ kPa,

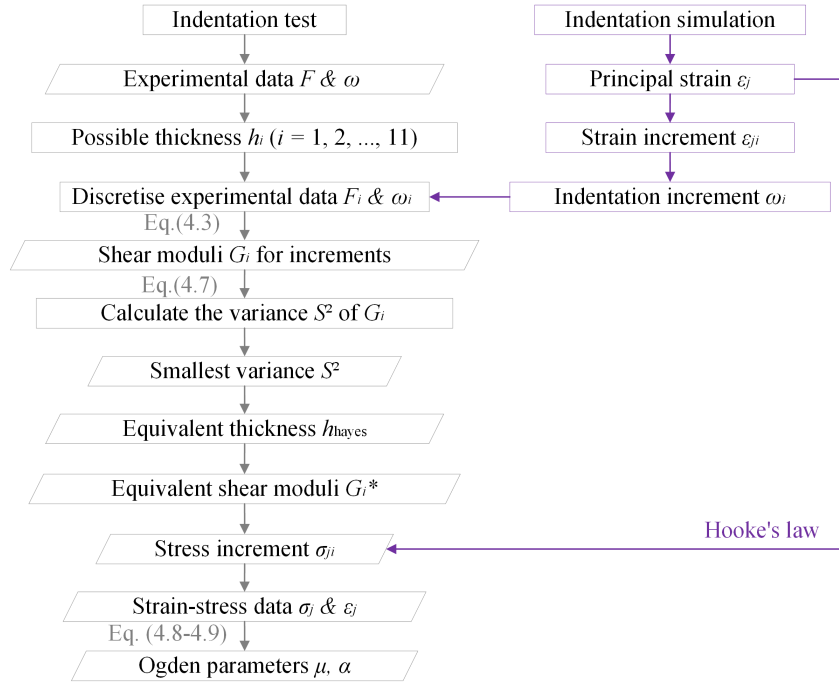


Figure 4.6: The procedure to obtain Ogden parameters from an indentation test. The black blocks outline the procedure for processing experimental curves, whilst the yellow blocks outline the method used to extract information from numerical simulations. For each indentation curve, the equivalent thickness h_{hayes} is identified from the possible thickness values first. Subsequently, the shear moduli and strains corresponding to h_{hayes} are used to fit the Ogden parameters.

but with different values of the Ogden parameters μ and α . Values for α were chosen as 0.5, 1, 2, 3, 4, and 5, which, with Eq. 4.2, give $\mu = 16, 8, 4, 2.667, 2$ and 1.6. Assuming that the material properties G , μ , α , and tissue thickness h are not known *a priori*, the proposed method is expected to accurately recover these values from the six simulated curves.

The equivalent thickness $h_{\text{hayes}} = 100 \mu\text{m}$ is identified from the FE artery model shown in Figure 4.7C, into which a range of possible thicknesses, h_i , from 50 to 150 μm with increments of 10 μm was inserted. The resulting equivalent shear moduli for a thickness of $h_{\text{hayes}} = 100 \mu\text{m}$ are shown in Figure 4.7D. The principal strain ϵ_j given by Path 3 (in Figure 4.3) was used to calculate the principal stresses σ_j with G_i^* using Hooke's law.

After combining σ_{ji} for all increments, the first-order Ogden parameters μ and α were determined by iteratively fitting overall stress-strain curves using Eq. 4.8 and 4.9. As detailed in the supplementary information, initial values for $\mu = 0.1 \text{ kPa}$ and $\alpha = 10$ were subsequently iteratively adjusted to minimize the variances of μ and α across all nodes, until the variance reached a threshold of 0.001. The fitted shear modulus G (Eq. 4.2) and α for six simulation curves are plotted in Figure 4.7E and F, respectively, showing a maximum error of less than 5% for all simulated curves.

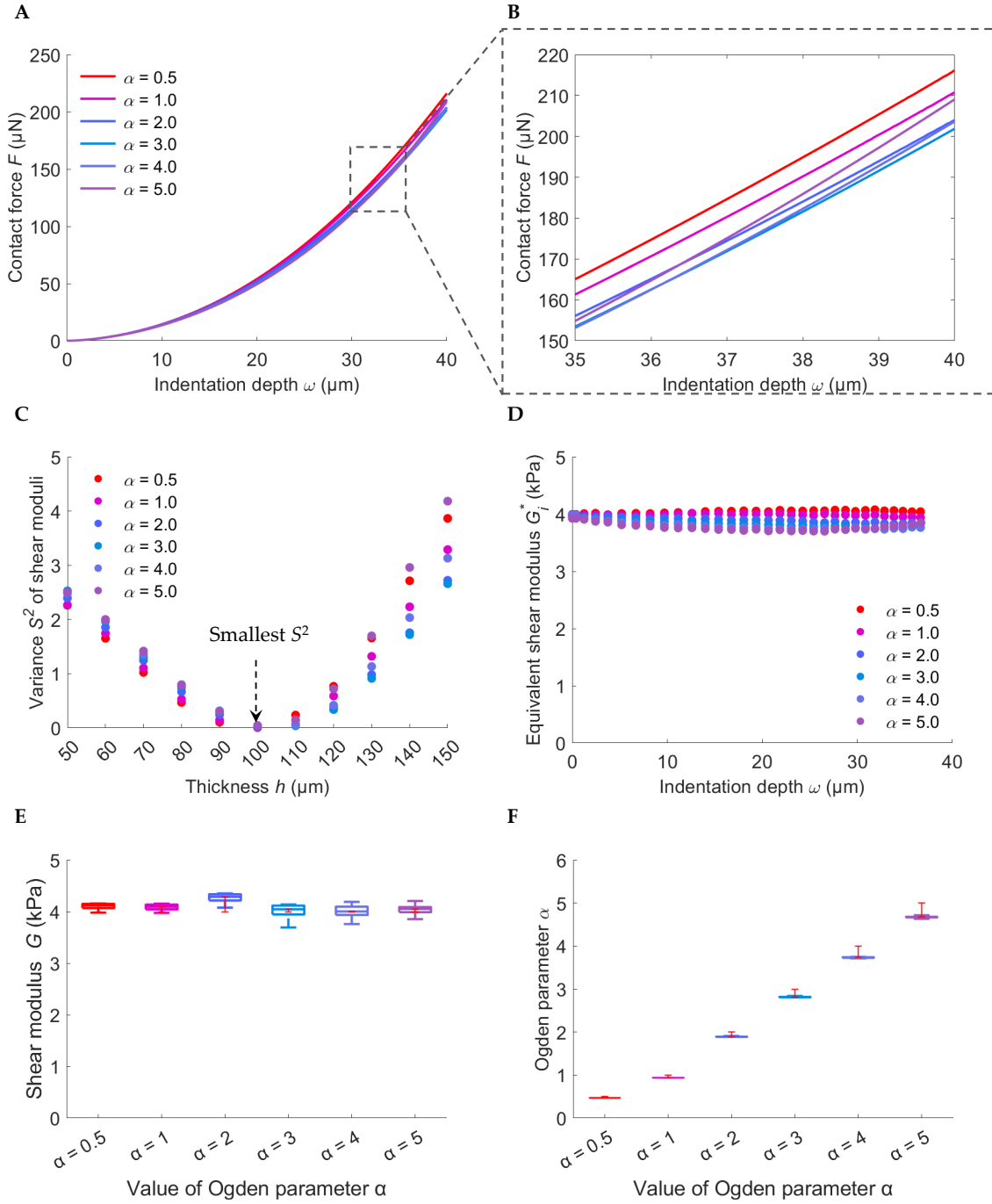


Figure 4.7: Method verification: (A-B) Six indentation curves. (C) variances of the shear modulus for different tissue thickness values. The equivalent thickness h_{hayes} can be identified as the thickness for which G has the smallest variance, in this case $h_{\text{hayes}} = 100$ μm . (D) values for G_i^* for the six different tissues at $h_{\text{hayes}} = 100$ μm . (E) Average values of G were obtained from different nodes along Path 3 for the various simulated tissues. (F) Calculated values of α for the various simulated tissues.

4.3 Application of the Method on Mouse Arterial Tissue

Tissue Preparation

Female apolipoprotein E gene-deficient (ApoE^{-/-}) mice purchased from Charles Rivers Laboratories (Edinburgh, UK) were bred in the Biological Service Facility at King's College London. All animal procedures were performed in accordance with the Animals (Scientific Procedures) Act 1986 and Amendment Regulations 2012, Guidance of Operations of the Animals (Scientific Procedures) Act 1986 and carried out under licenses from United Kingdom of Great Britain and Northern Ireland Home Office. Animals were housed in accordance with Code of Practice for the Housing and Care of Animals Bred, Supplied or Used for Scientific Purposes. Briefly, all animals were maintained in groups of 2 to 4 individuals, or in rare incidents of home-cage aggression, housed separately in individually ventilated cages under controlled environmental conditions (12 hours of light/dark cycle, at approximately 21°C and humidity $55 \pm 10\%$).

Five fresh aortic specimens were collected from 30-33 weeks old ApoE^{-/-} mice. The aorta, extending from the aortic root to the iliac bifurcation was excised en block. Any adhering connective tissue was removed from the arteries, after which specimens were pinned onto a cork base, kept in refrigerated phosphate-buffered saline (PBS) solution, and transferred to the laboratory the same day. Six indentation experiments were performed for each tissue specimen, approximately 1 hour after tissue collection. Before indentation, the arteries were opened longitudinally to expose the intima layer. The specimen is illustrated in Figure 4.8A.

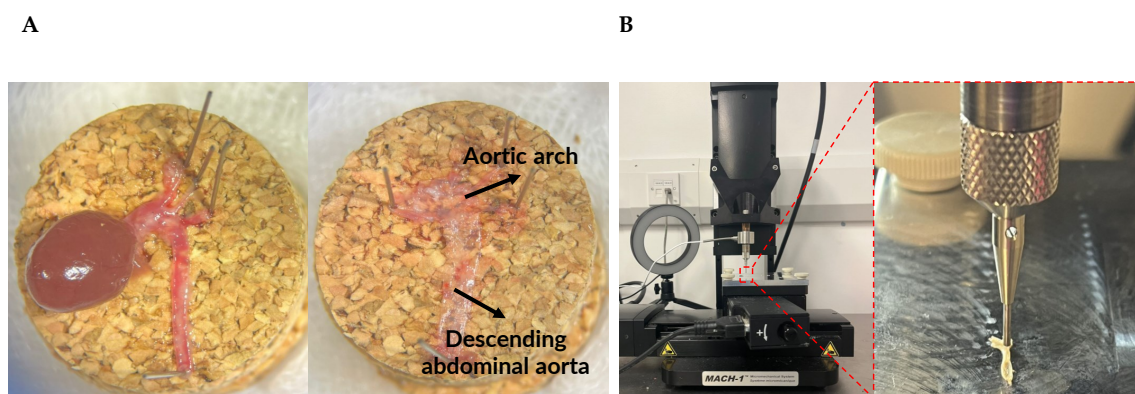


Figure 4.8: Specimen and indentation test: (A) The image presents the arterial tissue opened longitudinally, fully prepared for indentation tests. It should be noted that the arterial tissue consists of an aortic arch and a descending part. In this research, we focus on the mechanical properties of arterial tissue, without distinguishing between the arch and descending segments. (B) The indentation equipment and setup.

Indentation experiments

The indentation experiments were performed on a Mach-1 mechanical tester (Biomomentum, Laval, Canada, see Figure 4.8B). A force transducer (Biomomentum MA999) with a capacity of 1.5 N was used to collect contact force F versus indentation depth ω data. A spherical indenter with a radius R of 150 μm was used in these measurements, to minimise the influence of lateral boundaries, since the width of arterial specimens averages $\leq 1000 \mu\text{m}$. A downward displacement ω was applied at a velocity of 5 $\mu\text{m/s}$ until the contact force reached 1 mN. The resulting curves of contact force F versus indentation depth ω were recorded at a rate of 100 Hz.

Data analysis

Data analysis for the 30 experimentally obtained curves was conducted using MATLAB by employing the ‘smooth’ function with a window size of 1,000 points. The initial contact point was identified as the point from which the data exhibited a continuously increasing slope, and data before this point were excluded from the analysis. Some experimental data exhibited abrupt rises in force towards the end of the indentation, caused by the substrate’s contribution. These data points were also removed. A total of 4 measurements were excluded from the dataset, as they contained an abrupt rise in force at the beginning of the curve. The approximate thickness of the specimen was estimated from the distance between initial contact during indentation and a light contact between indenter and specimen holder when the tissue specimen was not present. The 26 measurements were then categorized into six brackets: 50-100 μm , 100-150 μm , 150-200 μm , 200-250 μm , 250-350 μm , and $\geq 350 \mu\text{m}$. Figure 4.10A illustrates the measured load-displacement diagrams, after noise removal, for tissue in the 100-150 μm bracket. The measurements for the five other thickness brackets can be found in Figure 4.9. It is observed that some noise remains even after noise removal, this can be attributed to the rough surface of the tissue, the use of an indenter with a small radius, and the relatively small thickness of the specimens. The remaining noise is inherent to the tissue and the measurement method used. In addition to noise, the various curves demonstrate significant variation, which is typical for heterogeneous materials with differing thicknesses.

Indentation tests

Following the method illustrated in Figure 4.6, the equivalent thickness h_{hayes} and the Ogden parameters α and μ were obtained, see Table 4.2. From α and μ the shear modulus G can be calculated (Figure 4.10B). The results illustrate that the shear modulus exhibits significant variance, even within each thickness bracket. This is attributed to the heterogeneous nature of the

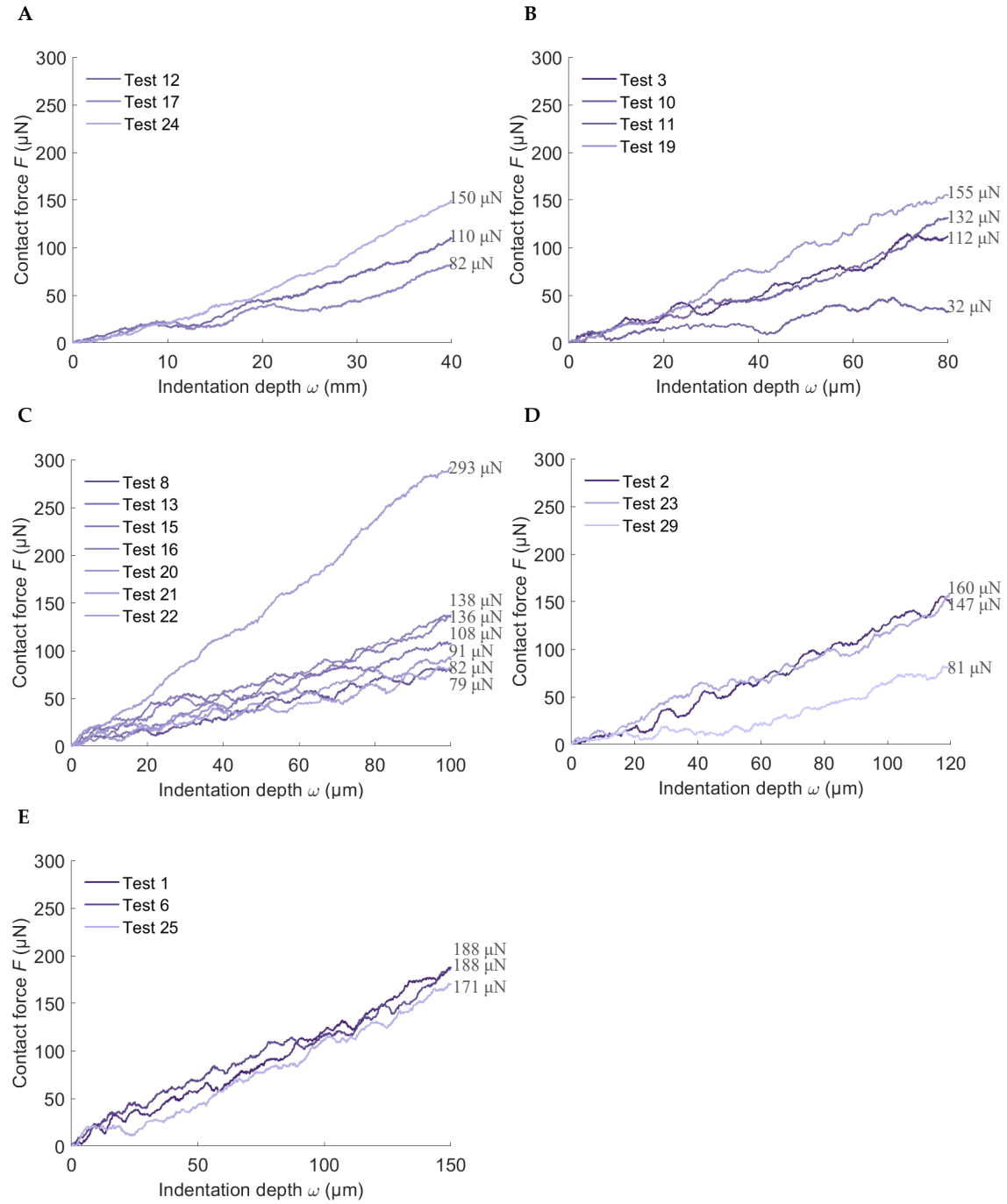


Figure 4.9: The contact force F versus indentation depth ω collected from the indentation tests, where 1, 2, ..., 29 represents the test number.

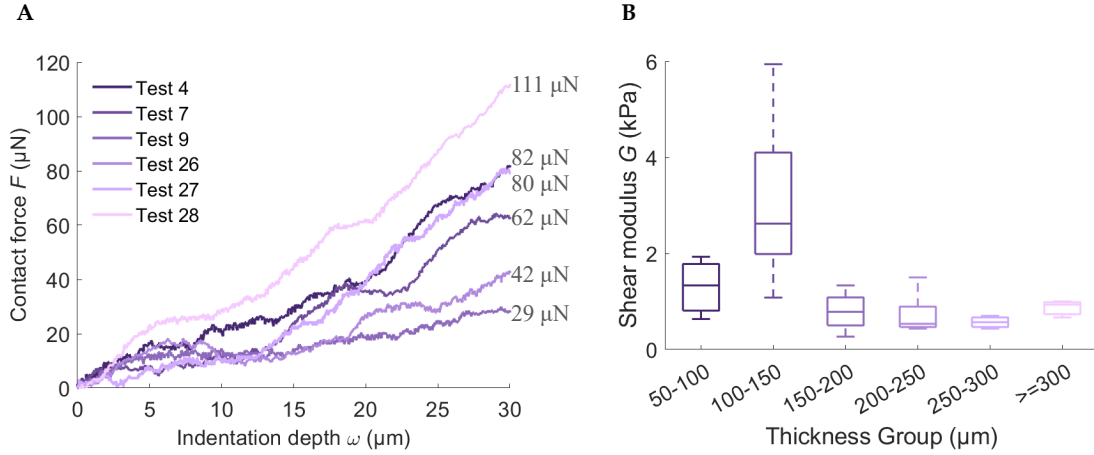


Figure 4.10: Experimental results and obtained shear modulus: (A) Contact force versus indentation depth for various tissues with a thickness between 100 and 150 μm . Numbers represent the test ID (see table 4.2). The shear modulus G of Ogden material for 26 measurements: (B) Obtained shear modulus for the six thickness ranges.

arterial tissue. The average shear modulus G for all conducted tests was found to be 1.22 kPa, with values ranging between 0.27 kPa and 5.02 kPa.

Simulations to verify the results

Numerical simulations were conducted to verify the parameters h_{hayes} , μ , and α obtained from the proposed method. Since $G = \mu\alpha/2$ provides a clearer material's modulus values, we have chosen to present G in these results. The numerically generated curves were then compared to the experimental curve to assess the accuracy of the fitted material parameters.

Figure 4.11 shows the comparison, with the dotted curve the experimental data and the solid black curve the simulated data using the extracted material properties. The calculated material parameters for G and α (black) are displayed at the top left corner of each sub-figure. In grey, the value for E is displayed, as obtained by applying Hayes' method directly to the measured data. As a reference, the indentation curve for that value is also shown in the graphs, in grey colour.

Test 27 at the top left of the figure is an example of where the procedure works well. The experimental data lies on a smooth curve that follows $F \sim \omega^n$, and the simulated result based on the obtained values for G and α aligns with the experimental data. In contrast, the force-displacement curve calculated using the elastic modulus E derived from Hayes' model underestimates the force at a given indentation depth, particularly at large indentation depths. For experiment 7 (top right), which exhibits some fluctuations, the simulation given by G and α effectively captures the overall indentation trends. Tests 4, 26, and 28, show fluctuations at the initial 0 to 25 μm of indentation and the simulation result aligns well with the experimental data, but only at larger indentation

Table 4.2: The material parameters for mouse arterial tissue obtained from 26 measurements

Thickness bracket	Test ID	Material Parameters			
		h_{hayes} (μm)	G (kPa)	μ (-)	α (-)
50-100 μm	12	50	1.9	4.8	0.8
	17	90	0.6	0.8	1.2
	24	100	1.3	0.9	1.5
100-150 μm	4	110	2.8	5	0.8
	7	110	2.4	4.4	1.1
	9	110	1.1	2	0.9
	26	150	2.0	1.75	1.6
	27	150	4.1	4.32	1.9
	28	150	5.9	3.8	2.3
150-200 μm	3	200	2.0	4.4	0.9
	10	160	2.4	5.3	0.9
	11	150	0.9	1.125	1.6
	19	200	1.4	1.22	2.3
200-250 μm	8	240	0.5	0.59	1.7
	13	250	0.5	0.43	2.3
	15	250	0.9	2.57	0.7
	16	250	0.8	1.14	1.4
	20	250	0.5	0.48	2.1
	21	250	1.5	3.75	0.8
	22	230	0.4	0.67	1.2
250-300 μm	2	250	0.7	2	0.7
	23	260	0.6	1.09	1.1
	29	260	0.4	0.44	1.8
≥ 300 μm	1	600	0.1	0.07	2.8
	6	350	0.7	0.67	2.1
	25	480	0.9	1.38	1.3

depths. The experimental result of Test 9 does not follow a 'normal' indentation curve $F \sim \omega^n$ and indeed this reduces the accuracy of the obtained Ogden parameters, and the simulated curve is not representative.

Overall, the results obtained using the Ogden model are significantly closer to the experimental data than those using the elastic modulus obtained through the use of Hayes' model. The simulated data based on Hayes' model consistently underestimates the experimental data, particularly at larger indentation depths.

4.4 Discussion

The shear modulus G of mouse arterial tissue obtained in this study averages 1.22 kPa, with values ranging from 0.27 to 5.02 kPa. The obtained equivalent thicknesses of mouse arterial tissue ranges from 50 to 500 μm , with the majority of obtained values below 300 μm .

Table 4.3 lists various studies that reported the elastic modulus $E = 2G(1 + \nu)$ of arterial tissue. The elastic modulus values obtained in this study are consistent with those reported by [125], [136], [237], [251], whereas values at least an order of magnitude higher are observed in [58–60], [252].

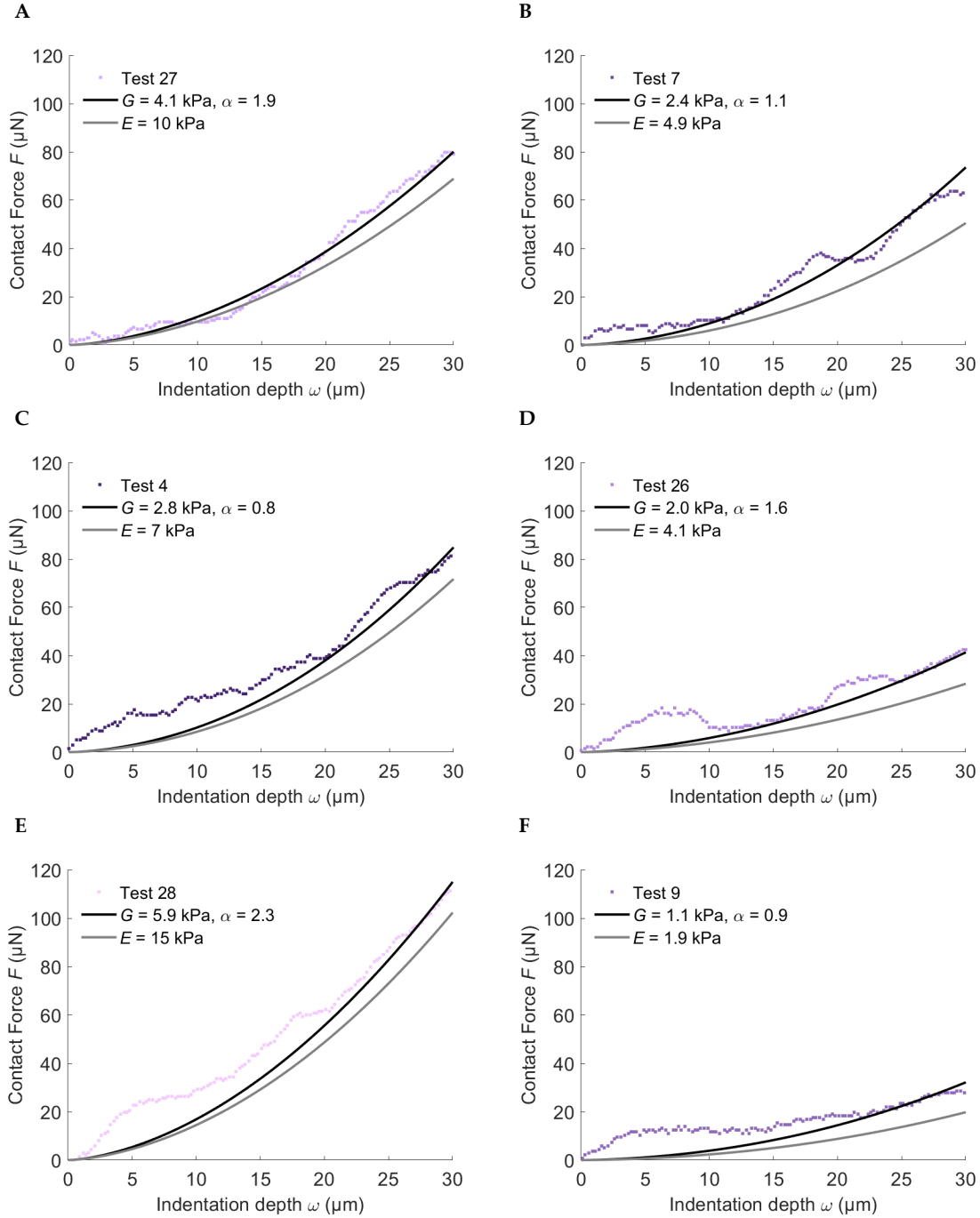


Figure 4.11: Comparison of experimental results conducted on specimens with thicknesses between 100 and 150 μm , with simulated curves calculated using the obtained Ogden parameters α and μ and, for reference, curves calculated using the elastic modulus E from Hayes' model.

Table 4.3: Shear moduli and elastic moduli reported in literature

Reference	Tissue Type	Thick- ness (μm)	Material model	Modulus (kPa)	Method
Meng et al. [251]	Mouse pulmonary vessels	250	Linear elastic	3.29 ± 1.2	Hertz's model
Liu et al. [237]	Rat arterial tissue	50	Linear elastic	1.66 ± 0.49	Hertz's model
	Human pulmonary vessel	10		5.44 ± 2.96	
Peloquin et al. [125]	Endothelium (bovine)	10-1000	Linear elastic	2.5 ± 1.9	Hertz's model
	Subendothelium (bovine)			2.77 ± 1.1	
Sicard et al. [136]	Bovine arteries	10-50	Linear elastic	1.74 ± 0.96	Hertz's model
Melo et al. [252]	Human tunica intima	10	Linear elastic	79.3 ± 7.2	Hertz's model
	Human tunica adventitia	-		41.1 ± 8.0	Hertz's model
Engler et al. [253]	Arterial medial layer	-	Linear elastic	5-8	Dimitriadis's model
Barrett et al. [58]	Atherothrombotic plaque	-	Neo-Hookean	$G = 7-100$	Inverse analysis
Chai et al. [59]	Human carotid artery	200	Neo-Hookean	6-891	Inverse analysis

The elastic modulus values reported by [125], [136], [237], [251] were measured by micro- or nano-indentation, with indenter radii ranging from nanometers to tens of micrometers. These studies limited the maximum indentation depth to 10% of the indenter radius, facilitating the application of Hertz's model for elastic modulus calculation. While these works neglected the effects of finite specimen thickness, Engler et al. [253] accounted for this factor by employing Dimitriadis' model [132] to estimate the elastic modulus of the arterial medial layer. The hyperelastic or Neo-Hookean properties for plaques, were reported by Barrett et al. [58] and Chai [59], [60] using the FE inverse method.

Sicard et al. [136] emphasized the significant influence of indenter radius and shape on the measured elastic modulus. They compared results obtained with pyramidal tips with a radius of 20 nm and spherical tips with radii of 1 and 2.5 μm , attributing the discrepancies to the large strains induced by sharp tips. Sicard et al. recommend the use of micro-sized spherical indenters for more accurate measurements. Similar observations of localized strains exceeding the linear elastic regime with sharp tips were reported in other studies [132], [242].

Harris et al. [242] reported that the rough surfaces of arterial tissue introduced significant errors in measuring indentation depth and estimating the contact area between the cell and the indenter. To address these issues, they combined atomic force microscopy measurements with confocal microscopy on epithelial cells expressing a green fluorescent protein-tagged membrane marker, allowing for direct characterization of the indentation geometry and precise measurement of the indentation depth. Surface roughness challenges in mouse joint cartilage were addressed in Chandran et al. [244] by introducing a virtual contact point as a fitting parameter in Hertz's model. By optimizing this parameter for the initial contact point and correcting for finite thickness, they improved the accuracy of the analysis. Similarly, Garcia et al. [233] proposed a method to eliminate uncertainties regarding the initial contact point by smoothing the measured curve, and fitting the

first derivative of the applied normal force to the indenter's position. This approach was later applied by Johnson et al. [243] to determine the elastic modulus of gradient-density polyacrylamide hydrogels. Unlike these methods, the method proposed in this study does not depend on a specific measurement apparatus or assumptions about the initial contact point, offering a more generalized approach.

In the current research, the equivalent shear modulus G_i^* for each increment was derived using Hayes' model to address the effect of finite thickness of the specimen on the contact behavior. While Hayes' model is used in this study, the equivalent shear modulus G_i^* can also be derived from other contact models, for example, Hertz's model, which may be more appropriate for materials with larger thickness. Figure 4.12 illustrates the equivalent shear moduli G_i^* versus a/h for specimens with a finite thickness as predicted by Hertz's model and Hayes' model, respectively. In contrast to the values derived from Hayes' model, where the equivalent shear modulus is fairly stable with increasing length scale, the equivalent shear modulus predicted by Hertz's model consistently increases with increasing indentation depth.

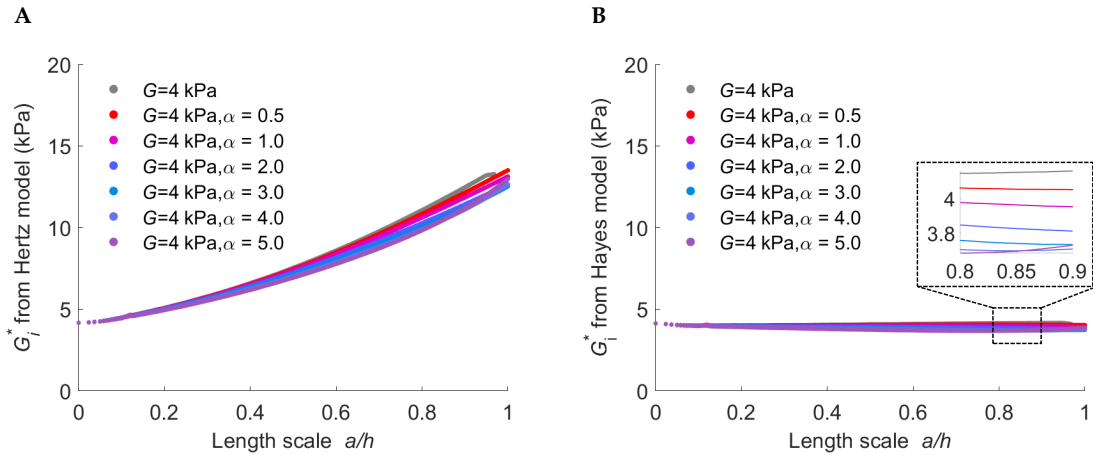


Figure 4.12: Equivalent shear moduli from Hertz's model and Hayes' model: (A) Hertz's model; (B) Hayes' model and newly developed method.

4.5 Conclusion

In this research, we developed a procedure for determining the Ogden parameters μ and α for arterial tissue, expanding upon previous work. The roughness of the specimen surface plays a significant role and must be considered when modelling the contact behaviour of arterial tissue during indentation. This can be addressed by introducing an equivalent thickness, h_{hayes} , which accounts for surface roughness. Analysis of experimental results shows that the thickness of mouse arterial tissue ranges from 50 to 500 μm , with the majority of specimens having a thickness

between 100 and 300 μm . The Ogden parameters, derived from the stress-strain data using the equivalent shear moduli G_i^* and the equivalent thickness h_{hayes} , provide a reliable representation of the tissue's hyperelastic behaviour. The average shear modulus G calculated from 26 measurements is 1.22 kPa. This approach ensures an accurate characterization of arterial tissue's mechanical properties, particularly concerning specimen thickness and material non-linearity. It allows for precise determination of Ogden parameters from indentation experiments, contributing to a better understanding of arterial tissue mechanics.

4.6 Supplementary

Ogden parameters fitting

The fitting process is conducted in Matlab. Since the fitting process is sensitive to initial values, the procedure would be conducted iteratively by adjusting the initial values to minimize the variance of Ogden parameter μ and α across all nodes until it is below 0.005. The algorithm is illustrated in Figure 4.13 with by 'Lsqcurvefit' fitting the stress-strain curve. Since the fitting process is sensitive to initial values, the procedure would be conducted iteratively by adjusting the initial values to minimize the variance of Ogden parameter μ and α across all nodes until it is below 0.005.

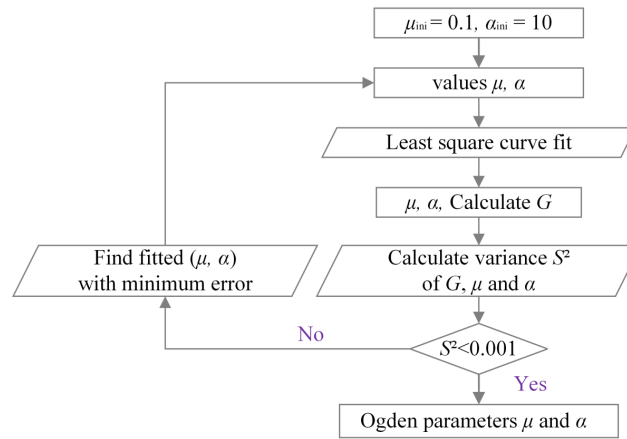


Figure 4.13: Ogden parameters fitting

An example

Taking the simulation curve from $G = 4$ kPa, $\alpha = 2$ as an example, the fitting procedure is started at the initial values (0.1, 10) for Ogden parameters (μ, α) . The shear modulus $G = \mu\alpha/2$. The obtained values of (μ, α) is listed in in Table, with the variance for standard. Initially, with a shear modulus $G = 4$ kPa provided in the first step with a small variance of 3.6×10^{-9} , the Ogden parameter

α exhibits a large variance of 1.3 across nodes. Finding the minimum fitting error is given by node 100 with the values of (0.0013, 6.1112), put these values as the initial values, giving the variance $S_c^2 = 0.97$, which is smaller than the previous one S_p^2 . the fitting procedure demonstrates that α converges to similar values around 1.9 for most nodes, with only slight fluctuations observed for the bottom nodes.

Table 4.4: Fitting procedure for Ogden model parameters

Step	Initial Values (μ, α)	$S^2(G)$	$S^2(\mu)$	$S^2(\alpha)$	Best Fit (μ, α)
1	(0.1, 10)	3.59×10^{-9}	8.80×10^{-7}	1.34	(0.0013, 6.1112)
2	(0.0013, 6.1112)	7.10×10^{-9}	2.18×10^{-7}	0.9727	(0.0014, 5.8427)
3	(0.0014, 5.8427)	7.06×10^{-9}	2.32×10^{-7}	0.9551	(0.0014, 5.6182)
4	(0.0014, 5.6182)	6.52×10^{-9}	2.33×10^{-7}	0.7625	(0.0015, 5.4152)
5	(0.0015, 5.2442)	6.19×10^{-9}	2.73×10^{-7}	0.8534	(0.0015, 5.2442)
6	(0.0015, 5.2442)	6.16×10^{-9}	2.64×10^{-7}	0.6782	(0.0017, 4.6203)
7	(0.0017, 4.6203)	5.10×10^{-9}	3.31×10^{-7}	0.5505	(0.0019, 4.2198)
8	(0.0019, 4.2198)	1.07×10^{-8}	2.17×10^{-7}	0.2957	(0.002, 4.1227)
9	(0.002, 4.1227)	1.20×10^{-8}	6.21×10^{-8}	0.1061	(0.0021, 3.7161)
10	(0.0021, 3.7161)	3.92×10^{-9}	4.85×10^{-7}	0.3659	(0.0023, 3.5254)
11	(0.0023, 3.5254)	9.80×10^{-9}	1.51×10^{-8}	0.0133	(0.0028, 2.8748)
12	(0.0028, 2.8748)	6.77×10^{-9}	1.52×10^{-8}	0.0048	(0.0028, 2.8336)
13	(0.0028, 2.8336)	7.29×10^{-9}	1.68×10^{-8}	0.0032	(0.0031, 2.6067)
14	(0.0031, 2.6067)	5.11×10^{-9}	1.61×10^{-8}	0.0042	(0.0034, 2.3651)
15	(0.0034, 2.3651)	4.27×10^{-9}	1.34×10^{-8}	0.0023	(0.0035, 2.3152)
16	(0.0035, 2.3152)	2.78×10^{-9}	1.34×10^{-8}	0.0027	(0.0039, 2.0694)
17	(0.0039, 2.0694)	2.78×10^{-9}	1.34×10^{-8}	0.0016	(0.004, 2.0239)
18	(0.004, 2.0239)	2.9066×10^{-9}	4.9711×10^{-9}	1.3121×10^{-9}	(0.004, 2.009)

Finite Element Simulation of Artery-Stent Interaction

Contents

5.1 Introduction	77
5.2 Method	81
5.2.1 Finite Element Model of Artery-Stent Interaction	81
5.2.2 Surface Shape Modelling	87
5.3 Result	90
5.3.1 Verification of the Finite Element model	90
5.3.2 Verification of the Non-uniform Method	91
5.3.3 Artery-Stent Interaction	92
5.4 Discussion	102
5.5 Conclusion	107

5.1 Introduction

Atherosclerotic stenosis refers to the narrowing of an artery caused by the buildup of plaque along the arterial walls. The plaque, composed of fatty substances, cholesterol, calcium, and other debris, causes artery blockages and reduces the supply of oxygen and nutrient to tissues, increasing the risk of serious cardiovascular diseases such as heart attacks and strokes.

Percutaneous Coronary Intervention (PCI) is a widely adopted procedure to treat atherosclerotic stenosis by placing a tiny tubular device called a stent at the site of stenosis to open the artery and restore blood flow. The stent deployment process is illustrated in Figure 5.1A. While PCI effectively

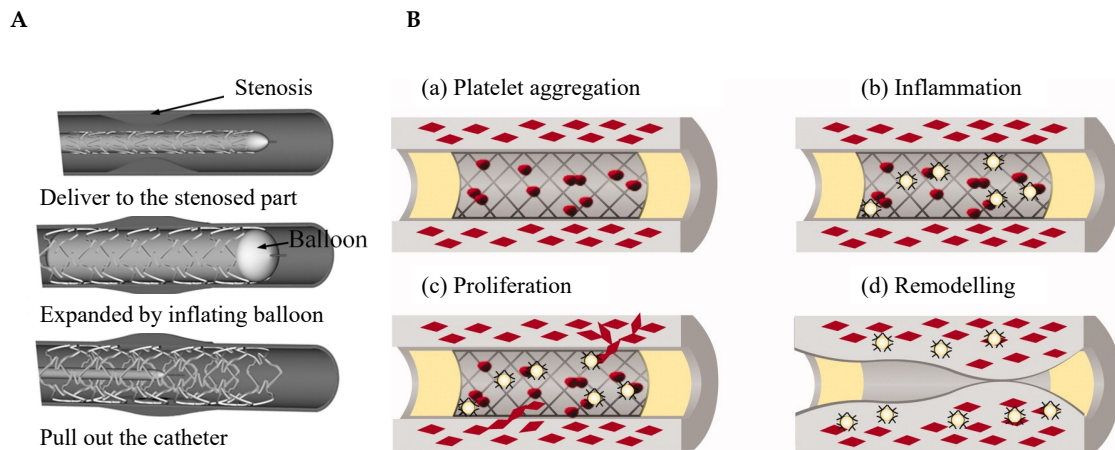


Figure 5.1: The stent deployment and ISR procedure: (A) Stent expansion by balloon as described by Shimizu et al. [266]. (B) A schematic representation of the restenosis process [263]: (a) platelet aggregation: immediate result of stent placement with endothelial denudation and deposition of platelets and fibrinogen; (b) inflammatory phase: a variety of white blood cells will gather at the injury site; (c) proliferation phase: smooth muscle cells migrate and proliferate, creating the neointima in an attempt to repair the wound; (d) late remodelling phase: the neointima is transitions from predominantly cellular to a less cellular and more extracellular matrix-rich plaque.

alleviates arterial blockages, complications can arise, with in-stent restenosis (ISR) being the most common. ISR refers to the re-narrowing of the artery after stent deployment (see Figure 5.1B), typically caused by excessive tissue growth during the artery's healing process. Studies [2, 254, 255] show that within one year of placing a bare-metal stent, the risk of restenosis ranges from 11% to 60%, leading to approximately 10% of patients requiring a repeat PCI to manage recurrent stenosis. To reduce ISR rates and improve long-term PCI outcomes, significant efforts are focused on enhancing stent design [8, 256–261] and improving deployment techniques [262–265].

Drug-eluting stents (DES) were introduced to the clinical in the early 2000s as an advancement over traditional bare-metal stents. DES are bare-metal stents coated with biocompatible materials and antiproliferative drugs. By releasing drugs that inhibit excessive tissue growth, DES significantly reduce the restenosis rate to approximately 8-15% within a year after stenting [3], marking a substantial improvement compared to bare-metal stents. Following the introduction of drug-eluting stents, biodegradable stents have also been introduced to the clinical market. These stents are made from materials that can gradually dissolve or be absorbed by the body within a few months after implantation. After the early stages of healing, the artery has stabilized and regained normal function. The breakdown and elimination of the stent material by the body can significantly reduce long-term risks and complications associated with a permanent stent.

Computational simulations based on the finite element (FE) method have become an adopted approach for simulating the stenting procedure and analyse the stent-induced stress on arterial walls.

Numerous studies have evaluated the stent performance by examining key parameters such as radial and longitudinal recoil (the stent's tendency to revert to its original shape), foreshortening (the reduction in stent length after expansion), dog-boning (the uneven expansion of the stent ends relative to the center) and the metal-to-artery ratio during stent expansion [267, 268] (see details in Appendix B). In these studies, stent expansion is typically achieved by applying displacement or pressure to the stent surface, or by inflating an expanding balloon within the stent. Since the recent introduction of bioresorbable scaffolds (BRS), most studies have focused on investigating their mechanical performance by developing new constitutive models for their materials, such as poly (L-lactic acid) (PLLA), to characterise the complex time-dependent mechanical behavior of these devices and to improve scaffold designs [258, 269, 270]. These studies provide valuable insights into stent design by exploring different geometries, material properties, and deployment techniques.

Regarding the artery model used in FE simulations, earlier studies typically relied on idealised models that represent the artery as a cylindrical vessel with a parabola-shaped plaque, simplifying both the artery and plaque as isotropic, homogeneous, and incompressible materials [271, 272]. Cheng et al. [273] introduced one of the earliest patient-specific FE models based on histological images of coronary artery lesions. A more detailed patient-specific model was developed by Holzapfel and Gasser et al. [96, 274, 275] for a diseased iliac artery from a 65-year-old patient. With clear geometric information of the artery provided by the high-resolution magnetic resonance images, a three-dimensional (3D) model of the artery was reconstructed, consisting of eight components: the fibrotic part, fibrotic intima, calcification, lipid pool, non-diseased media, diseased fibrotic media, and adventitia, as illustrated in Figure 5.2. An anisotropic constitutive model, the Gasser-Ogden-Holzapfel model, was employed to define the material properties of these components, using data collected from quasi-static stress-stretch tests. Three stents—Multi-Link, NIR, and InFlow Gold (Boston Scientific)—were investigated by estimating key indicators: intimal pressure, stress, and luminal change (the ratio of the smallest inner cross-section before and after stenting) in relation to stent-artery mismatch (the difference between the smallest lumen diameter and the diameter of the expanded stent), strut thickness, and cell geometry (details see Appendix B). Subsequently, more studies have utilized patient-specific models based on real arteries, as seen in [73, 276]. Meanwhile, many numerical models still rely on idealised representations [81, 267, 277].

In all FE studies, maximum/minimum principal stresses [73, 96, 98, 99, 267, 277], von Mises stress [78, 79, 81, 267, 276, 278], and strain field [80, 267, 276, 277, 279, 280] are used

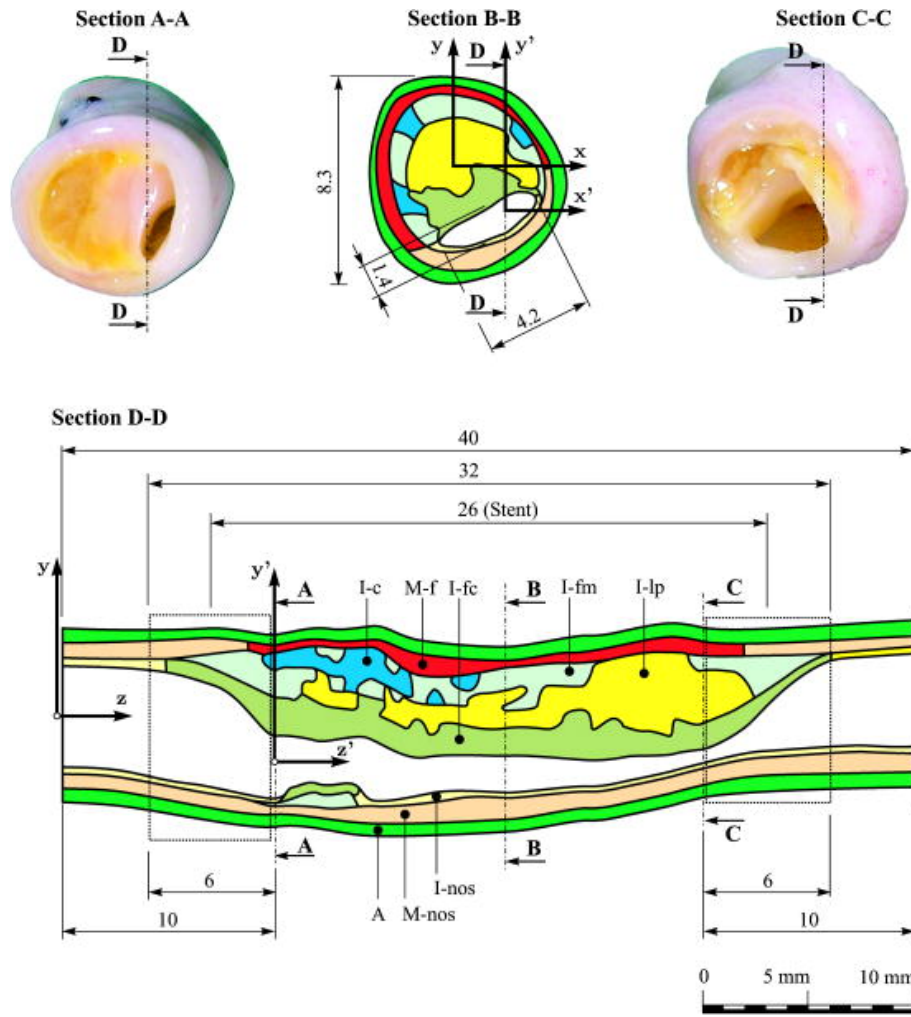


Figure 5.2: Sections of the analyzed external iliac artery. Section B-B is the region with the smallest lumen diameter of 1.4 mm. The tissue components are adventitia (A), non-diseased media (M-nos), non-diseased intima (I-nos), fibrous cap (I-fc), lipid pool (I-lp), calcification (I-c), fibrotic intima at the medial border (I-fm) and diseased media (M-f). Image adapted from Holzapfel et al [98]

to measure the artery fracture and plaque rupture related to restenosis, with smaller values of these variables generally being preferable. For instance, [281] reports that a stress of 1720 ± 890 kPa can cause artery failure, while [282] provides stress thresholds of 600 kPa for non-ulcerated plaques and 200 kPa for ulcerated plaques. The strain at fracture reported in [282] is $47 \pm 23\%$ for non-ulcerated plaques and 25% for ulcerated plaques. Cheng et al. reported a plaque rupture stress threshold of 300 kPa [273], which has been widely cited in subsequent studies [72], [283]–[285]. Although the relevance of stress and strain to physiological injury mechanisms remains unclear, these values can still provide insights that allow us to judge the stent performance and artery states.

In computational simulations of artery-stent interactions, the friction effects are less reported, with most studies either considering the coefficient of friction (COF) as a single, constant value [73],

[81], [123], [276], [277] or assuming a 'frictionless' interface [78], [79], [98], [267], [278], [286]. In [287], the effect of friction was examined by setting the COF to 0.05, 0.1, and 0.15, with findings indicating that variations in COF between the Nitinol stent and artery did not result in significant changes in both stress and strain. In [288], the friction effect was investigated by setting the COF between 0.03 and 0.06, revealing that friction between the artery and stent contributed to larger stress alterations only for the plaque. Although less reported in numerical studies, experimentally measured COF values between the artery and stent have been widely reported, with most studies suggesting that the COF should fall within the range of 0.01 to 0.05 [162], [173], [202], [203], [206].

In recent years, surface modification has become an active area of research aimed at enhancing stent performance. Efforts have focused on developing non-thrombogenic surface modifications that promote vascular wall healing by altering surface roughness and feature size [289], [290]. This presents a promising topic for computational modelling, as the effects of stent topography have not yet been extensively investigated in computational models.

In this study, FE simulations of artery-stent interaction are conducted using ABAQUS/Explicit, accounting for the effects of friction and stent surface shape. The interaction between the artery and stent is modelled as 'hard contact' with a COF ranging from 0.01 to 0.1. Non-uniform displacements are applied at the nodal level to reconstruct the concave and convex surface shapes of the stent. The results indicate elevated von Mises stress S , maximum principal stress σ , and maximum principal strain ε in the intima, with peak values occurring at the right shoulder of the fibrous cap. The higher strain is also observed around the lipid pool, indicating a high likelihood of plaque rupture. An increase in the COF leads to a moderate rise in stress (von Mises stress S and shear stress τ) at the interface. The surface shape of the stent significantly affects the maximum stress in the interface, with increased stress observed in the convex surface with large amplitudes and concave surface. Conversely, convex surfaces with small amplitude result in lower average shear stress, which suggests that surface modifications with these features may be beneficial.

5.2 Method

5.2.1 Finite Element Model of Artery-Stent Interaction

The FE simulation for artery-stent interaction is conducted in ABAQUS/Explicit (ABAQUS 2024). The FE model involves a 1 mm-wide segment of an artery interacting with a stent made of platinum-chromium alloy. The FE model of the artery is based on Gasser and Holzapfel et al.'s research [96], [275] and consists of four components: the intima, media, adventitia, and pool. The simulation is

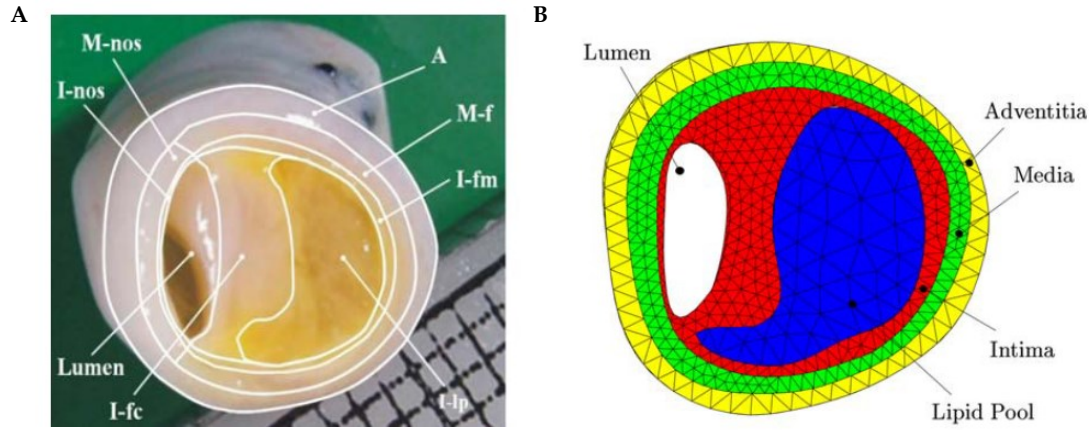


Figure 5.3: The model utilized in Gasser’s research [99] includes: (A) a segmented macroscopic view of a type V lesion according to the histological classification of Stary (adopted from Holzapfel et al. [98]). The borderlines represent different tissue components: A = adventitia, M-nos = non-diseased media, I-nos = non-diseased intima, I-fc = fibrous cap, I-fm = fibrous intima at the medial border, I-lp = lipid pool, and M-f = fibrous (diseased) media. (B) A 3D discretisation of a slice of the atherosclerotic lesion composed of four distinct tissue components (labeled as adventitia, media, intima, and lipid pool) using tetrahedral finite elements. The individual tissues are meshed separately and linked together employing the interface element method.

carried out as a quasi-static analysis, ensuring that the ratio of kinetic energy to the total strain energy remains below 5%, as recommended by Kim et al [164]. The details are provided as follows.

5.2.1.1 Artery Model

In the research by Gasser and Holzapfel et al. [96], [98], the geometric and material model of the artery is based on a diseased iliac artery from a 65-year-old female. The geometric model, depicted in Figure 5.2, was derived from high-resolution magnetic resonance images (hrMRI) and reconstructed using non-uniform rational B-splines (NURBS) to represent eight components. The material properties of these components were defined using the Gasser-Ogden-Holzapfel (GOH) model, with the material parameters obtained from the tensile experiments [97]. As a simplification, Gasser [96], [99] combined some of these components, resulting in an artery model with three main layers: intima, media, adventitia along with a lipid pool, as illustrated in Figure 5.3.

In this research, a geometric model of the artery, as shown in Figure 5.4 is established based on the FE model provided by Gasser [96]. It consists of three main layers, which are adventitia, media, intima, and an arbitrarily shaped lipid pool.

The material models for the adventitia, the media, and the intima are defined by the GOH model, as described in Eq. 5.1. The lipid pool is modelled as an isotropic matrix defined by a linear elastic. As discussed in Chapter 2, the GOH model characterises the arterial layers as an isotropic ground matrix embedded with two families of collagen fibres. It assumes that the collagen fibres exhibit a

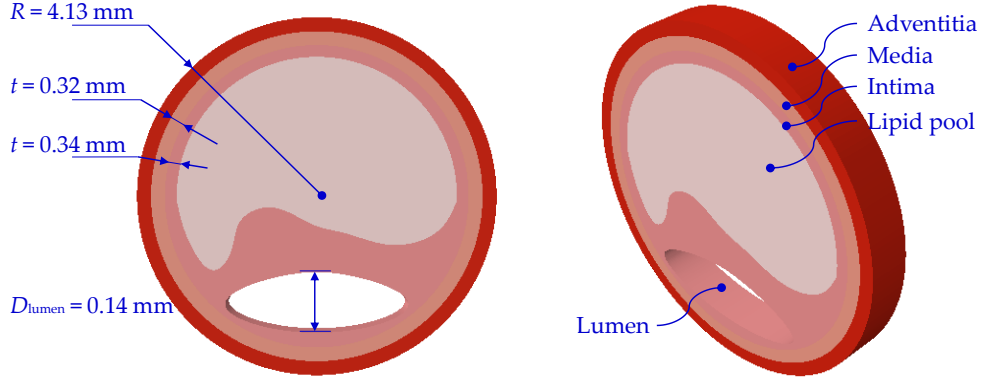


Figure 5.4: The geometric model of the artery: the artery consists of four components: adventitia, media, intima, and lipid pool from outer to inner. The thickness of the adventitia and media is 0.32 and 0.34 mm, respectively. The thickness of the intima varies from 0.1 to 0.5 mm, with an elliptical lumen and an arbitrarily shaped lipid pool. The elliptical lumen has a smallest diameter of $D_{\text{lumen}} = 0.14$ mm

transversely isotropic distribution along a preferred direction a_{0i} , with dispersion characterised by the parameter κ . The material parameters for each layer are listed in Table. 5.1, which is based on the tensile test data reported by Holzapfel et al. [98].

$$\psi = \frac{G}{2}(I_1 - 3) + \sum_{i=1}^2 \frac{k_{i1}}{k_{i2}} (\exp k_{i2} [\mathbf{H}_i : \mathbf{C} - 1]^2 - 1) \quad (5.1)$$

$$\kappa = \frac{1}{4} \int_0^\pi \rho(\Theta) \sin^3 \Theta d\Theta \quad (5.2)$$

where ψ is the strain energy density; C_{10} , D , k_1 and k_2 are material parameters κ , $\rho(\Theta)$, and Θ describe the fibre distribution: κ characterises fibre dispersion, where $\kappa = 1/3$ indicates an isotropic distribution of collagen fibres; the function $\rho(\Theta)$ represents the orientation density function associated with the Eulerian angle Θ which denotes the Eulerian angles of the collagen fibres of the collagen fibres.

Table 5.1: Material properties of artery components

Tissue	G (kPa)	k_1 (kPa)	k_2 (-)	α (-)	κ (-)
Intima	78.9	23.7	26.3	0.0	0.0
Media	15.0	4.0	2.3	7.0	0.0
Adventitia	1.75	65.6	61.8	49.0	0.0
Lipid Pool	0.1	0.0	-	-	-

Note: G is the shear modulus, k_1 and k_2 are material stiffness constants, α is the fibre orientation angle, and κ is the fibre dispersion parameter.

To model the collagen direction $\mathbf{a}_0^i$, a cylindrical coordinate system $ORTZ$ is defined for the artery, as illustrated in Figure 5.5A. The local material orientation is specified along the T direction, with an angle α (indicated by the blue lines) relative to the T axis. This is achieved by modifying the input file.

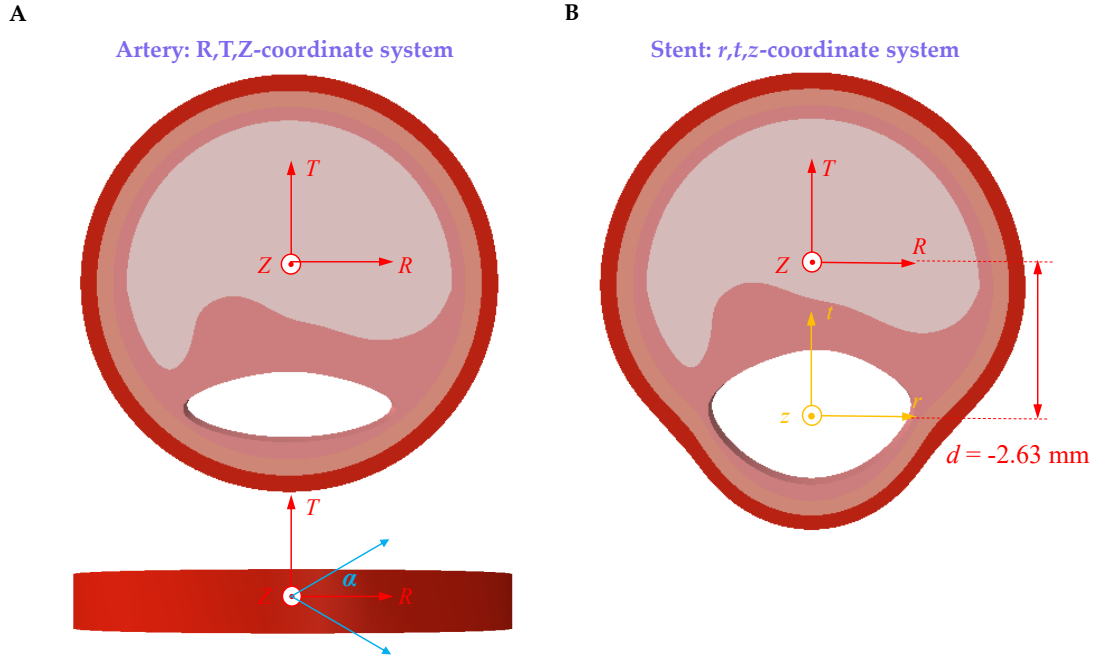


Figure 5.5: The local coordinate system used in the simulation of the artery and stent is as follows: (A) $ORTZ$ coordinate system applied to the FE model of the artery. (B) $ortz$ coordinate system used for the stent.

The artery layers are discretised using C3D8R elements, which widely employed in related studies [79, 267, 277, 288, 291]. Reduced integration elements were employed to save computation time while maintaining acceptable accuracy. Meshes of varying densities were generated, with coarser elements assigned to the media and the adventitia layers, and finer elements for the intima. As the slave surface in the interaction, the finest mesh is used for the inner surface of the intima. To balance computational accuracy and efficiency, six different discretisations were created, consisting of 229,680, 259,050, 441,850, 493,250, 1,616,000, and 1,747,400 elements. These correspond to combinations of the finest mesh sizes of 0.03 mm^3 , 0.02 mm^3 (used for the inner surface of intima), and 0.01 mm^3 with coarsest mesh sizes of 0.13 mm^3 and 0.1 mm^3 (used for the outer surface of adventitia), respectively. Mesh convergence was assessed by ensuring the variation in key results, such as stress and lumen diameter (defined by the minimum lumen diameter), between successive refinements was less than 3%.

To avoid rigid-body motions of the artery, the displacements in the T - and Z - directions for the

nodes on the outer surface of the adventitia are fixed, and all Z -displacements are constrained at both end faces of the artery. To simulate physiological loading conditions, a uniform pressure $p = 100$ mmHg (equivalent to $p = 13.3$ kPa) is applied to the entire inner surface of the artery. The lumen is expanded as depicted in Figure 5.5B, reflecting the *in-vivo* condition of an artery in the body. To maintain these stress conditions during the artery-stent interaction, a predefined stress field is applied to the artery model.

5.2.1.2 Stent Model

Due to patent restrictions, we cannot obtain detailed information about the geometric features of commercial stents. In this research, the stent model is developed based on the SYNERGYTM BP Everolimus-Eluting Platinum Chromium Coronary Stent, with the information provided in [292].

The FE model of the stent is directly generated in ABAQUS. The mesh part with 16 stent cells, shown in Figure 5.6A, is wrapped into a cylindrical shell with a radius of 1 mm, where the width of strut s_t is 0.09 mm and the wavelength l_{st} of the stent cells is 0.39 mm. A solid stent, as depicted in Figure 5.6B, is then formed by generating two solid layers from the cylindrical shell by sharing nodes and removing the base shell. Since the outer surface of the stent serves as the master surface in the artery-stent interaction, the mesh size of the stent depends on the mesh size of the slave surface, which uses three mesh scales: 0.03 mm³, 0.02 mm³, and 0.01 mm³.

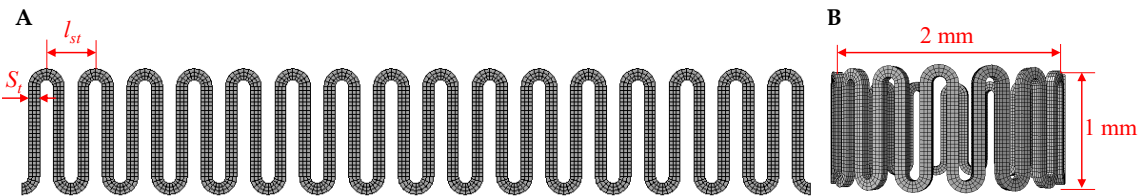


Figure 5.6: The FE model for stent: (A) The 2D mesh part used for wrapping into a cylindrical stent. (B) The stent model used for FE simulation, with a diameter of 2 mm and a width of 1 mm.

The material properties of the stent are given as a platinum-chromium alloy (Pt-Cr) [292], see Table. 5.2.

Table 5.2: Material properties of SYNERGYTM BP Stent [292]

Property Type	Property	Value
Elastic	Young's modulus (MPa)	203000
	Poisson's ratio	0.3
Plastic	Yield stress (MPa)	Plastic strain
	480.96	0.00
	1207.63	0.37

For assembling the stent into the artery, a cylindrical coordinate system ($ortz$) for the stent is defined with the origin o at the center of the artery lumen, as shown in Figure 5.5B. To prevent rigid-body motions of the stent, the t - and z -displacements are fixed at node **A**. Additionally, a tie constraint is applied to the end surfaces of the stent to prevent it from opening during expansion.

5.2.1.3 Artery-Stent Interaction

Figure 5.7 illustrates the interaction between the artery and stent, with a coordinate system ($oxyz$) provided for the artery-stent assembly.

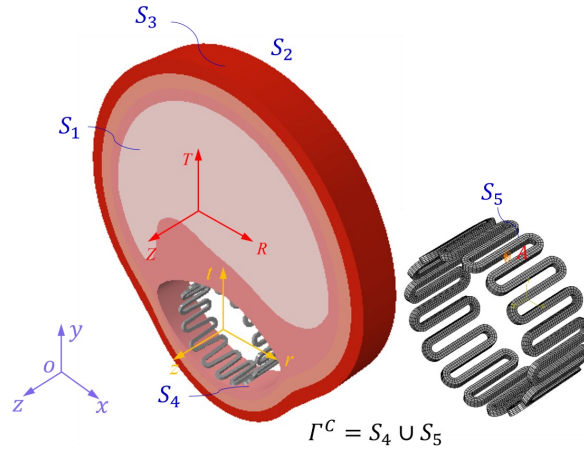


Figure 5.7: Artery-stent interaction: S_1 , S_2 and S_3 are the outer surfaces of the FE arterial model. Γ^C is the interface, consisting of S_4 (artery inner surface) of artery and S_5 (stent outer surface) of stent. Three coordinate systems are employed: the global coordinate system, $OXYZ$, defines the overall geometry and external loading of the model. Two local systems cylindrical coordinate system, $OTRZ$ is used for the artery, capturing its radial, circumferential, and longitudinal directions, while $ortz$ is applied for the stent, facilitating precise alignment and interaction modelling with the arterial structure.

The artery-stent interaction is modelled using surface-to-surface contact in ABAQUS/Explicit, by assigning the inner surface of the intima S_4 , and the outer surface of stent S_5 as the slave and master surface, respectively. The contact properties are defined as ‘hard contact’ with a COF set to vary between 0.01 and 0.1 in steps of 0.01. The ‘hard contact’ defines the pressure-overclosure relationships as Eq. 5.3. ‘Hard contact’ ensures that the contact forces are transmitted accurately without allowing significant penetration between the stent and artery, which is essential for capturing the true mechanical interaction.

$$p_n = 0 \quad g_n > 0; \quad p_n > 0 \quad g_n = 0; \quad (5.3)$$

where p_n is the contact pressure on node n , and g_n is the gap function.

Then, the governing functions of the interaction are described by Eq. 5.4 and the boundary conditions are specified in Eq. 5.6 293.

$$\begin{aligned}
 g_n &\geq 0, \quad p_n \leq 0 \quad g_n p_n = 0 \quad \text{on} \quad \Gamma^C \\
 g_n &\leq 0, \quad p_n = \epsilon_n g_n \quad \text{on} \quad \Gamma^C \\
 g_\tau &= 0, \quad \|q_\tau\| \leq \mu |p_n| \quad \text{on} \quad \Gamma^C \\
 \dot{g}_\tau &\neq 0, \quad \|q_\tau\| = \mu |p_n| \quad \frac{\dot{g}_\tau}{\|\dot{g}_\tau\|} \quad \text{on} \quad \Gamma^C
 \end{aligned} \tag{5.4}$$

where $\Gamma^C = S_4 \cup S_5$ represents the contact interface; g_n and $\dot{g}_\tau$ are the gap function and the slip velocity, defined as Eq. 5.5, respectively; p_n and q_τ represent the contact force and friction force, respectively.

$$\begin{aligned}
 g_n &= \mathbf{n} \cdot \Delta \mathbf{u} \\
 \dot{g}_\tau &= (\mathbf{I} - \mathbf{n} \otimes \mathbf{n}) \cdot (\dot{\mathbf{u}}_\tau)
 \end{aligned} \tag{5.5}$$

where $\Delta \mathbf{u}$ represents the increment of displacements on the contact surface, and $\mathbf{n}$ is the unit normal vector direction. Since the stent is established based on cylindrical coordinates, the n -direction corresponds to the r -direction. $\mathbf{I}$ is the identity tensor.

$$\begin{aligned}
 \mathbf{u} &= \mathbf{u} \quad \text{on} \quad S_5 \\
 u_1 &= u_2 = 0 \quad \text{on} \quad A \\
 u_1 &= u_2 = u_3 = 0 \quad \text{on} \quad S_3 \\
 u_3 &= 0 \quad \text{on} \quad S_1, S_2
 \end{aligned} \tag{5.6}$$

where u_1 , u_2 , and u_3 represent the displacement in r -, t -, and z - direction, respectively.

5.2.2 Surface Shape Modelling

5.2.2.1 Non-uniform Displacement Method

In addition to the friction effect, the surface shape of the stent is also considered. Incorporating the realistic geometry created by CAD software may result in poor mesh quality. Therefore, we do not represent the exact geometry of the stent. Instead, we use the same FE stent model, applying a non-uniform modelling method to account for the surface shape effects. This method, inspired by Paggi's approach 293, can also effectively model the surface topographies of the stent.

A simple example illustrated in Figure 5.8A was used to demonstrate the application of this

method. The realistic model involves contact between a flat surface characterized by $\Gamma_1(\xi) = z_1$ and a rough surface characterized by $\Gamma_2(\xi) = z_2 + h(\xi)$, where ξ is the node coordinate, z_i ($i = 1, 2$) is a constant representing the baseline and $h(\xi)$ represents the surface roughness. By applying a normal displacement u_n to the grey part to make it contact with the blue part, the gap function is given as Eq. 5.7.

$$\begin{aligned} g_n(\xi) &= |z_1 - z_2| - h(\xi) - u_n = \Delta z + h(\xi) - u_n \quad \Delta z = u_n + h(\xi) \\ g_n(\xi) &= 0 \quad \Delta z > u_n + h(\xi) \end{aligned} \quad (5.7)$$

where Δz is the distance between the two baselines.

By replacing the uniform u_n with a non-uniform displacement $u_{\text{non}} = u_n + h(\xi)$, the realistic model can be represented using two flat surfaces in contact, as illustrated in 5.8B. This is referred to as the idealised model, and the displacement is defined by Eq. 5.8.

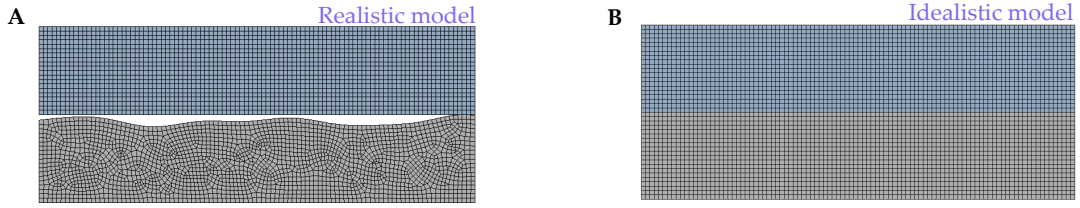


Figure 5.8: FE model for a different approach: (A) The realistic model, which shows the actual shape of the contact material. (B) The idealized model does not depict the actual shape but uses a displacement set for the nodes to represent the shape effect.

$$\begin{aligned} g_n(\xi) &> 0, \quad u_{\text{non}} = u_n + h(\xi) \\ g_n(\xi) &< 0, \quad u_n = 0 \end{aligned} \quad (5.8)$$

The method is verified by implementing the realistic and idealised models in ABAQUS/Explicit to do the simulation. The shape of a realistic model is defined by Eq. 5.9. Contact is established for both a realistic model and an idealised model by defining contact pairs with the top surface of the grey part as the master surface and the bottom surface of the blue part as the slave surface. The contact properties are specified as ‘hard contact’ with ‘friction-less’ behavior. The material of the grey part is given by an elastic modulus E of 206,000 MPa and Poisson’s ratio $\nu = 0.3$, and the material of the blue part is given by a Neo-Hookean model with an elastic modulus E of 0.1 MPa. All nodes on the top surface of the blue part are fixed. For the realistic model, a uniform displacement $u_n = 0.275$ mm is applied to the nodes on the top surface of the grey part, whereas a non-uniform displacement u_{non} , as illustrated in Figure 5.9 is given to the idealised model. The

simulation is completed in 10 steps.

$$\Gamma_1 = 0.1 \sin(1.5x) + 0.05 \sin(2x + \pi/4) + 0.03 \cos(3x + \pi/2) \quad (5.9)$$

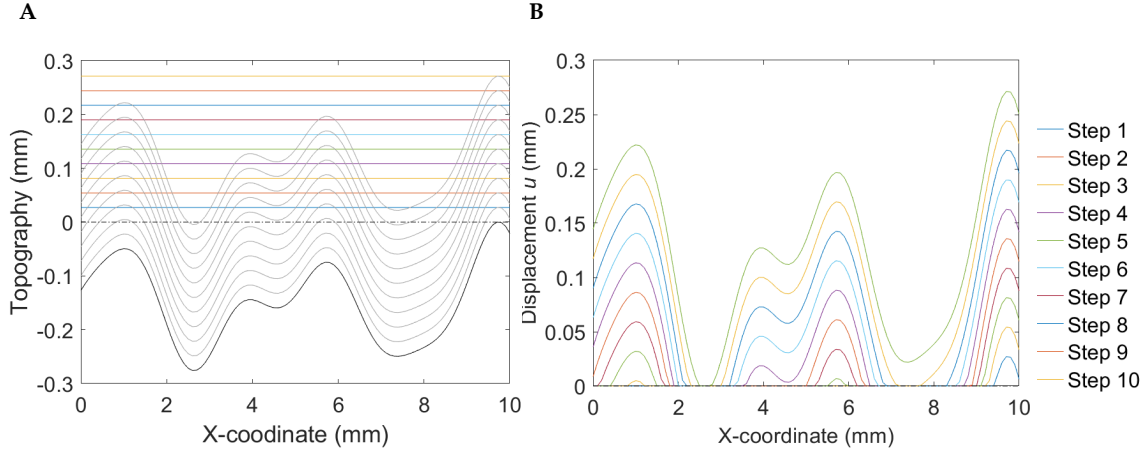


Figure 5.9: Non-uniform displacement was applied to the upper surface nodes: (A) A uniform displacement (colored line) was applied to the idealized model over 10 steps, with the black dotted line indicating the baseline. The grey line represents the surface shape, while the black solid line shows the initial position of the surface. (B) A non-uniform displacement was applied to the idealized model over 10 steps. When considering the black dotted line as the starting point, the non-uniform displacement aligns precisely with the surface shape above the baseline.

5.2.2.2 Surface Shape of the Stent

In stent design, the topographies and surface shape are used to enhance stent performance by improving the mechanical interaction between the stent and the arterial wall, potentially reducing complications such as restenosis. In this study, two surface shapes—convex (named Convex-1, Convex-2) and concave (Concave-1)—are investigated, as they represent the simplest and most commonly used designs in current commercial stents [290]. To accurately capture the surface shape of the stent, a finer mesh size of 0.01 mm^3 is applied to the stent surface. As the mesh size of master and slave surfaces should remain consistent, the same mesh size of 0.01 mm^3 is applied to the inner wall of the intima. This results in a total of 1,616,000 elements being used for the artery. A finer mesh size of 0.008 mm^3 for the stent and inner surface of the artery was attempted, however, the simulation terminated automatically, and reported a large mesh distortion problem. The reasons for this will be discussed further in section Discussion. Given that the width of the strut is only 0.09 mm , a mesh size of 0.01 mm^3 constrains the scale modelling of the more complex surface shapes and topographies of the stent.

With the outer flat surface defined by $R(\theta) = 1$, where θ represents the angular coordinates, the convex and concave surfaces are described by Eq. [5.10]. To illustrate the effect of the surface shape

of the stent, a non-uniform displacement u_r , as defined in Eq. 5.11, is applied to the nodes on the outer surface of the stent using a Python script. The COF for the contact surfaces was set to 0.01, 0.02, and 0.03. Due to the complex contact behavior introduced by the uniform surface shape, simulations with higher COF values led to automatic termination due to excessive mesh distortion. In this section, we focus primarily on the effects of the stent's surface shape; thus, a COF of 0.01 to 0.03 is considered acceptable.

$$R(\theta) = r + A \cdot \sin 64\theta \quad (5.10)$$

where A is the amplitude of the stent surface, with the positive value represented for convex and the negative represented for concave. 64 is the given since there are 16 stent cells, featuring 32 struts and 32 interval features.

$$\begin{aligned} u_r &= u_r + A \cdot \sin 64\theta, & g_r(\theta) &> 0, \\ u_r &= 0, & g_r(\theta) &< 0, \end{aligned} \quad (5.11)$$

5.3 Result

5.3.1 Verification of the Finite Element model

5.3.1.1 Mesh Convergence

The maximum von Mises stress S calculated, as well as the distance between the two origin points, d , and lumen diameter D_{lunem} , under pressure and inflation are shown in Figure 5.11A. As illustrated, the mesh discretisations with 441850 elements demonstrate an acceptable variance ($\leq 3\%$) in S , d , and D_{lunem} compared to the results from discretisations with 493,250, 1,616,000 and 1,747,400 elements. Consequently, the mesh discretisations with 441850 elements, combining the finest mesh of 0.02 mm^3 and the coarsest mesh size of 0.13 mm^3 , is selected for the numerical simulations.

5.3.1.2 Quasi-static Analysis

The kinetic energy and the total strain energy during the interaction are shown in Figure 5.11B. As illustrated, the ratio of the kinetic energy to the total strain energy is nearly 0 %, which satisfies the conditions for quasi-static analysis.

5.3.1.3 Finite Element Model Validation

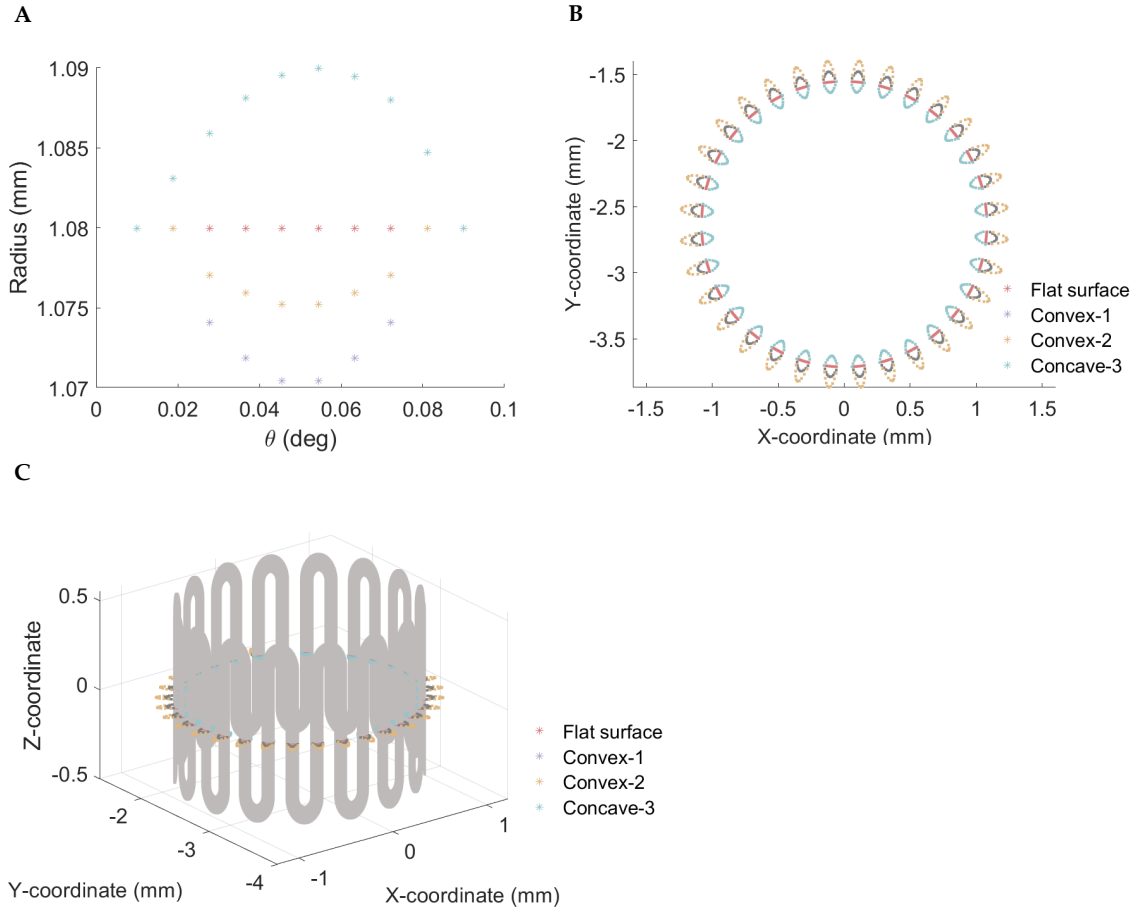


Figure 5.10: The surface shapes of the stent: (A) Three surface shapes of the stent are considered in this study. (B) The surface shapes were given for the stent in $z = 0$. (C) Apply the non-uniform displacement to the stent.

The results of maximum principal stresses, calculated under a lumen diameter D_{lumen} of 3.51 mm without friction, are illustrated in Figure 5.12. Compared to the results reported in [31], where the maximum principal stresses are around 500 kPa, 100 kPa, and 40 kPa for the intima, media, and adventitia, respectively, our findings of 540.4 kPa, 177.4 kPa, and 32.1 kPa fall within a same range of the stress values. The observed differences can be attributed to the intima rupture in [31], which leads to less stretching of the media and adventitia, resulting in lower stress values. Therefore, the finite element model established in this study is validated and accurate.

5.3.2 Verification of the Non-uniform Method

The stress distribution of nodes on the slave surface for each step is illustrated in Figure 5.13. Despite slight differences in values at each peak point, the stress distribution in the idealised model is similar to that in the realistic model. Therefore, we believe the method can effectively model topography effects.

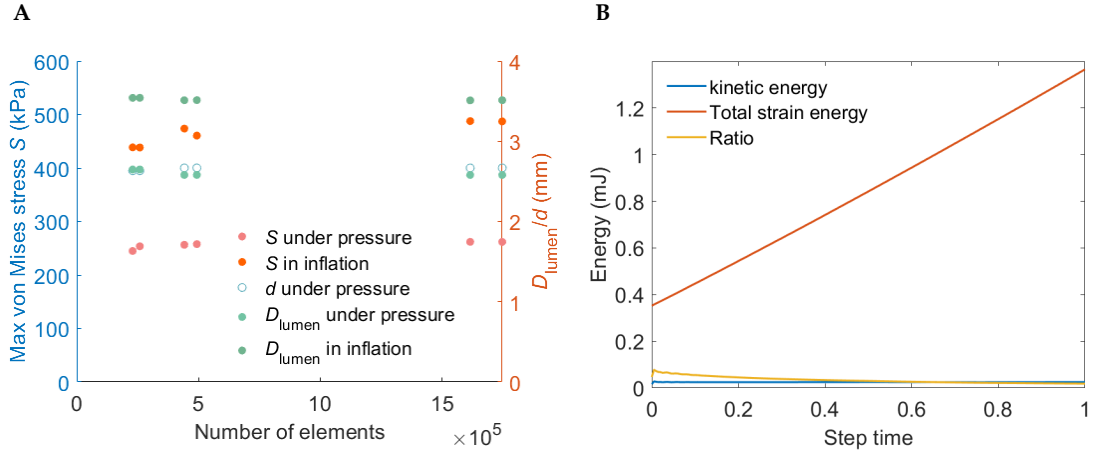


Figure 5.11: Model verification: (A) The comparison between different mesh density, where D_{lumen} is the lumen diameter and d is the lumen center. (B) Quasi-static analysis

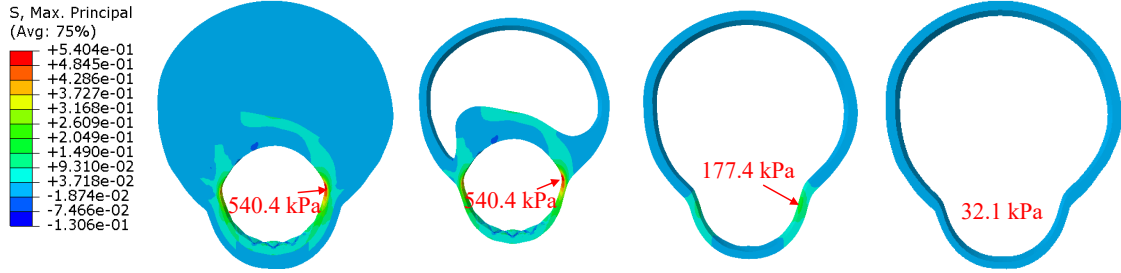


Figure 5.12: The maximum principal stress distribution of intima, media, and adventitia, respectively

5.3.3 Artery-Stent Interaction

In this research, five factors are considered to evaluate artery responses during artery-stent interactions: von Mises stress S , maximum principal stress σ , strain energy density SED , shear stress τ , and maximum principal strain LE . A particular focus is given to von Mises stress S , maximum principal stress σ , maximum principal strain ε , and shear stress τ at the artery-stent interface Γ^C to better understand the effects of friction and topography.

5.3.3.1 Artery Response

Figure 5.14 shows the results for von Mises stress S , maximum principal stress σ , strain energy density SED , shear stress τ , and maximum principal strain ε with $\mu = 0.01$. As illustrated in Figure 5.14A, the intima, with the highest shear modulus $\mu = 78.9$ kPa, exhibits greater von Mises stress S compared to the media and the adventitia. The maximum von Mises stress of 494.1 kPa is found at the right shoulder of the fibrous cap, which has been identified as the initial fracture point in Gasser's study [96]. The non-uniform stress distribution in the intima is attributed to the

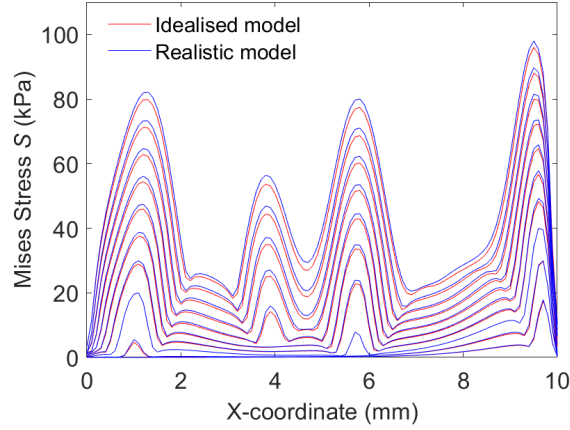


Figure 5.13: The distribution of von Mises stress S for the idealised model and the realistic model during the simulation, with S plotted at each 10 increments.

irregular shape of the lipid pool. A larger lipid pool on the left side reduces the material modulus of the intima, resulting in higher stress on the right side. Unlike the intima, the non-uniform stress distribution in the media and the adventitia is due to the orientation of the fibres in these layers. With the fibre oriented at a specific angle α ($< 90^\circ$) along the T - direction, the right side exhibits higher von Mises stress S . These observations are also supported by Figure 5.14B and C, where the maximum σ of 540.4 kPa and maximum SED of 164.5 kJ/mm³ occur at the same point. Since the maximum principal stress σ does not account for shear stress, unlike von Mises stress S , it shows that the stent induces maximum negative stress values, while the surrounding areas exhibit predominantly positive stress values, except for a small region (Figure 5.14B). A higher concentration of SED around the lumen is illustrated in Figure 5.14C.

In Figure 5.14D, the maximum shear stress τ_{xy} of 82.4 kPa is observed at the artery-stent interface Γ^C (τ_{xy} is denoted as S12 in Abaqus, where 12 refers to the xy -plane. Similarly, τ_{xz} and τ_{yz} are denoted as S13 and S23, with 13 and 23 referring to the xz - and yz -planes, respectively). Additionally, two regions, indicated by the red arrows, exhibit relatively higher shear stresses, exceeding 40 kPa. Similar to the von Mises stress S , the shear stress is higher on the right side of the media and adventitia, where the presence of fibers enhances the material's mechanical properties.

The maximum value of 71% for the maximum principal strain ε is observed at the right shoulder of the fibrous cap, where the maximum S , σ , and SED are also recorded, as illustrated in Figure 5.14E. However, unlike the other indicators, higher strain values are found in the lower half of the intima, media, and adventitia, as well as around the outline of the lipid pool. This observation aligns with reports suggesting that the lipid pool is likely to rupture during stent deployment [294].

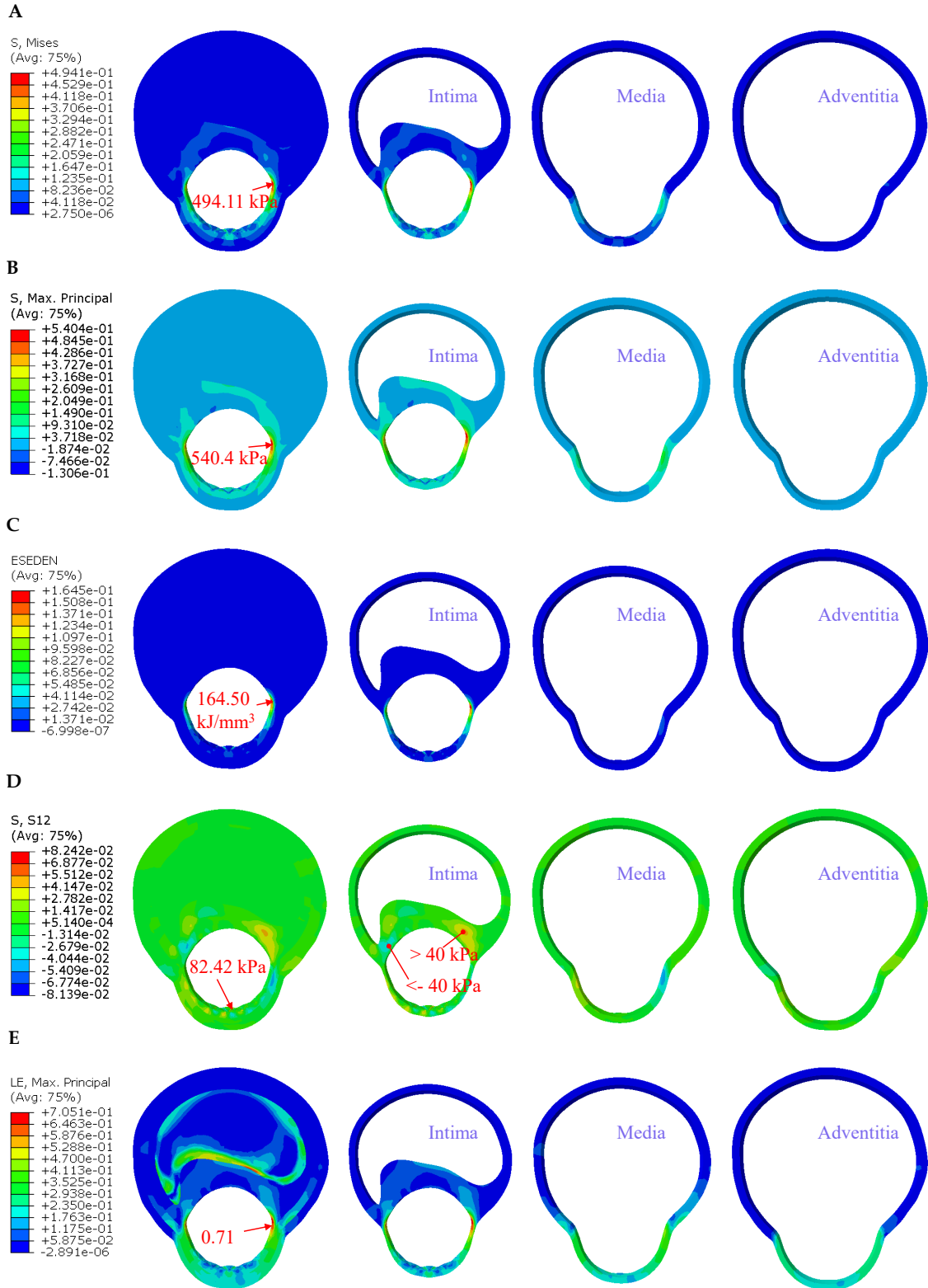


Figure 5.14: The stress distribution of the artery: (A) The von Mises stress S distribution. (B) Maximum principal stress σ . (C) The shear stress τ_{xy} . (D) The strain energy density SED . (E) The maximum principal strain ϵ .

5.3.3.2 Results of Friction Effects

Two cross sections: $z = 0.1$ (red frame) and $y = -3.35$ (yellow frame, depicted in Figure 5.15A), are examined to explore the stress distribution at the interface Γ^C . Detailed stress information for the nodes around the lumen from $z = 0.1$ (red frame) (Figure 5.15B) is accessed, with the results plotted in Figure 5.16.

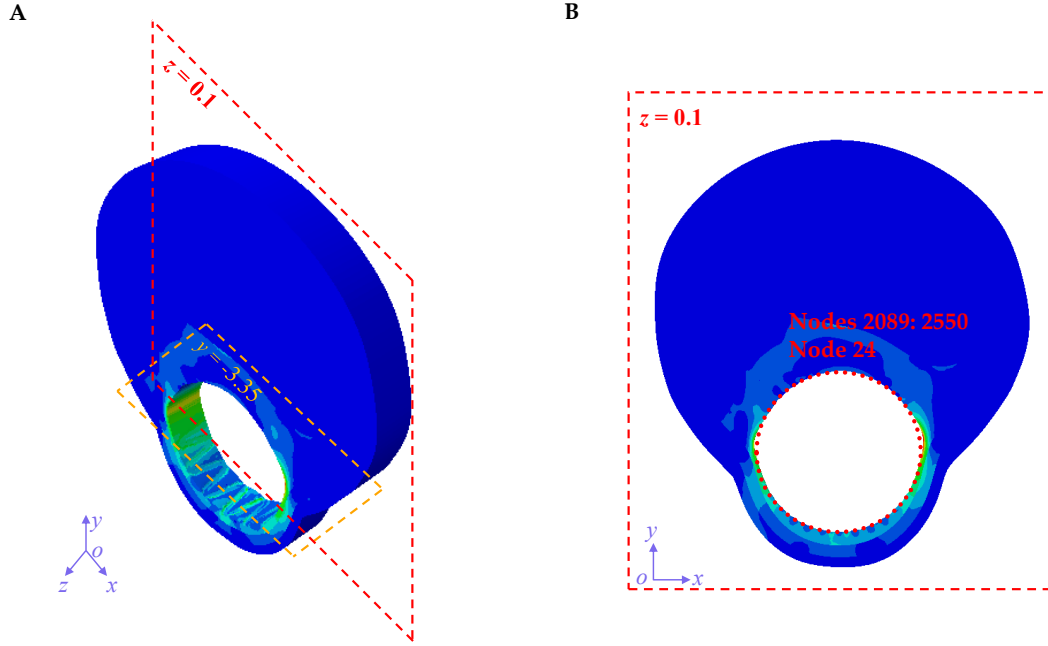


Figure 5.15: The stress distribution at interface Γ^C : **(A)** The von Mises stress with a COF $\mu = 0.01$, where the cross-section $z = 0.1$ and $y = -3.35$ are highlighted by the red and yellow frames, respectively. **(B)** The von Mises stress of the cross-section $z = 0.1$ and the nodes used to obtain detailed information.

The von Mises stresses S , maximum principal stress σ and maximum principal strain ε along with the shear stresses τ_{xy} , τ_{xz} , and τ_{yz} for the stent from $y = -3.35$, are shown in Figures 5.16. The stent cells are numbered from I to XVI, as shown in the upper right corner of Figure 5.14, with each stent cell containing two struts. Figure 5.14 illustrates the comparison of stress distribution between the contact and non-contact regions. Both the von Mises stress S and maximum principal strain ε are higher in the contact region, with noticeable stress concentrations along the stent's edges. This effect is even more pronounced for the maximum principal stress σ , where higher negative values are observed in the contact area, contrasting with the positive values in the surrounding non-contact area. The maximum absolute value of σ occurs at the cross-section $y = -3.35$.

The stress concentrations along the stent's edges are also evident in the shear stresses τ_{xy} , τ_{xz} , and τ_{yz} , as shown in Figures 5.14D–F, respectively. Additionally, there are substantial differences in von Mises stress S and maximum principal strain ε across the three arterial layers, as discussed

earlier. However, unlike the von Mises stress S , the variations in shear stresses, particularly τ_{xz} and τ_{yz} , across different arterial layers are less pronounced.

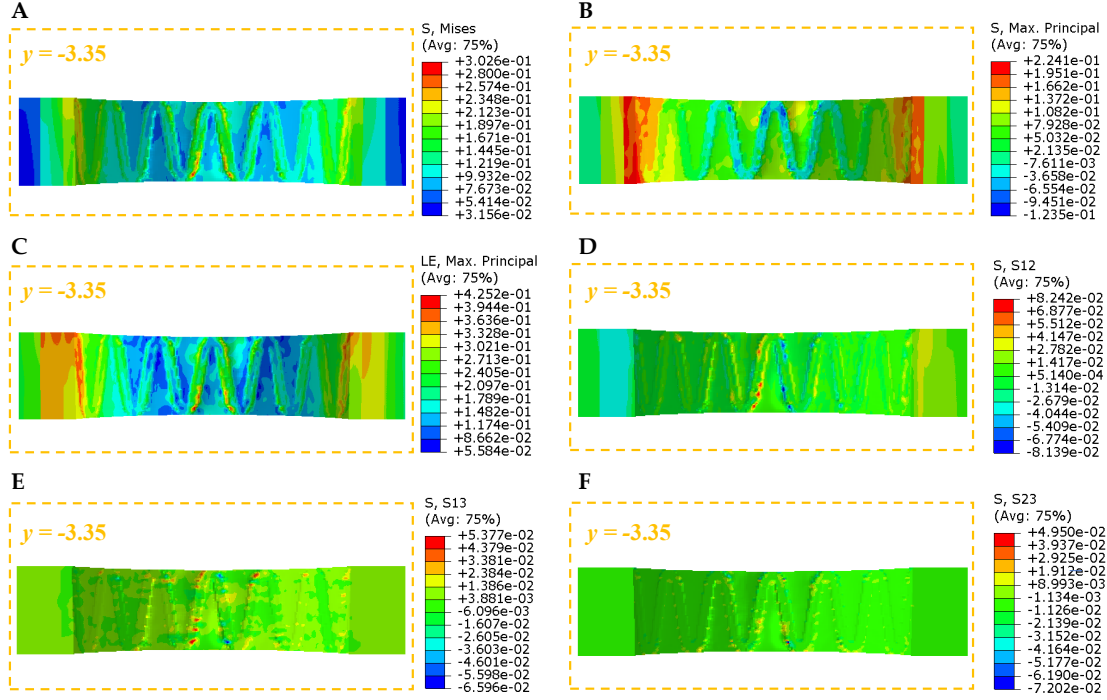


Figure 5.16: The stress distribution at interface Γ^C : (A) The von Mises stress S . (B) The maximum principal stress σ . (C) The maximum principal strain ε . (D-F) The shear stresses τ_{xy} , τ_{xz} , and τ_{yz} , respectively.

The stress distribution along the nodes is depicted in Figure 5.17. As previously mentioned, the contact regions exhibit higher stress than the surrounding non-contact regions. Except for stent cells V and XIII, which do not contact the artery wall, all other struts contribute to two peaks and one trough, with the two peaks observed at the ends and a trough in the middle. The peaks represent the stress concentrations caused by the edges of the struts.

For most stent cells, the von Mises stress S generally increases with increasing COF, as indicated by the upward grey arrow. However, a notable exception is observed in stent cell I and the left strut of stent cell II, where the von Mises stress decreases with increasing COF, as indicated by the downward grey arrow.

The shear stresses τ_{xy} along the x and y -directions are depicted in Figure 5.17E and F, respectively. Except for stent cells V and XIII, two distinct fluctuations with opposite peaks on either side are observed for each strut. It is evident that lower shear stress τ_{xy} is observed in the upper half of the artery where it is in contact with stent cells I, II, III, XIV, XV, and XVI. This can be attributed to the lipid pool weakening the material properties in the upper regions. Additionally, the shear stress τ_{xy} generally increases with increasing COF, as indicated by the upward red arrow. This

trend can also be observed in the shear stresses τ_{xz} and τ_{yz} as depicted in Figure 5.18

To illustrate the contribution of each stent cell, the maximum von Mises stresses S as well as the maximum and minimum shear stresses τ_{xy} , τ_{xz} , τ_{yz} for stent cells are provided in Figure 5.19. Each stent cell, with its two struts (or say contact areas), contributes two sets of data.

From Figure 5.19A, a decrease in von Mises stress with increasing COF is observed in the struts of stent cells I, II, IV, V, VII, VIII, and IX. In contrast, an increase in von Mises stress with increasing COF is noted in the struts of stent cells III, XIII, and XIV. The remaining struts show varying fluctuations in von Mises stress as COF values change. The maximum stresses are observed at the position of strut V, which corresponds to the right shoulder of the fibrous cap. Since this position is in a non-contact region, these maximum stresses cannot be attributed to friction effects.

It was found that the lumen diameter, D_{lumen} , is related to the maximum stress. The plot of maximum von Mises stress versus the lumen diameter, D_{lumen} , is shown in Figure 5.20. It is evident that the maximum von Mises stress, S , increases as D_{lumen} increases.

From Figure 5.19B-D, the maximum shear stresses τ_{xy} , τ_{xz} , and τ_{yz} show significant fluctuations, with τ_{xy} generally being higher than τ_{xz} and τ_{yz} . These fluctuations can likely be attributed to the inherent complexities of the contact simulations. Due to these fluctuations, there is no consistent pattern regarding the increase in τ_{xy} , τ_{xz} , and τ_{yz} with increasing COF values. Nevertheless, higher shear stresses are observed in contact regions compared to non-contact regions, with struts of stent cells V and XIII showing zero shear stresses τ_{xz} and τ_{yz} . Notably, a high shear stress τ_{xy} of 108 kPa is observed in stent cell XII, while shear stresses of -63 kPa and 54 kPa are observed in stent cell II, respectively (highlighted in red). With relatively small τ_{xz} and τ_{yz} , large values of τ_{xz} of -25 kPa, τ_{xz} of -30 kPa and 30 kPa are shown in stent cells I, II and X, respectively (highlighted in yellow). These stress levels indicate potential areas where the artery could sustain injury.

To better understand the effects of friction, additional data were extracted from the nodes at the interface Γ , ranging from cross-section $z = 0$ to cross-section $z = 0.1$. Figure 5.21 presents the average values of maximum and minimum shear stresses for the stent struts (indicated by dots) along with their standard deviations (SD) for $\mu = 0.01$ and $\mu = 0.1$. The deep blue color represents the results for $\mu = 0.01$, while the light blue corresponds to $\mu = 0.1$. The plot indicates that the average of the maximum and minimum shear stresses, τ_{xy} , τ_{xz} , and τ_{yz} , are considerably higher at $\mu = 0.1$ compared to a $\mu = 0.01$ except in non-contact stent cells V and XIII. The plot also illustrates that larger deviations occur at $\mu = 0.1$, indicating that higher COF leads to locally higher stresses. This suggests that a larger COF is more likely to induce stress concentration,

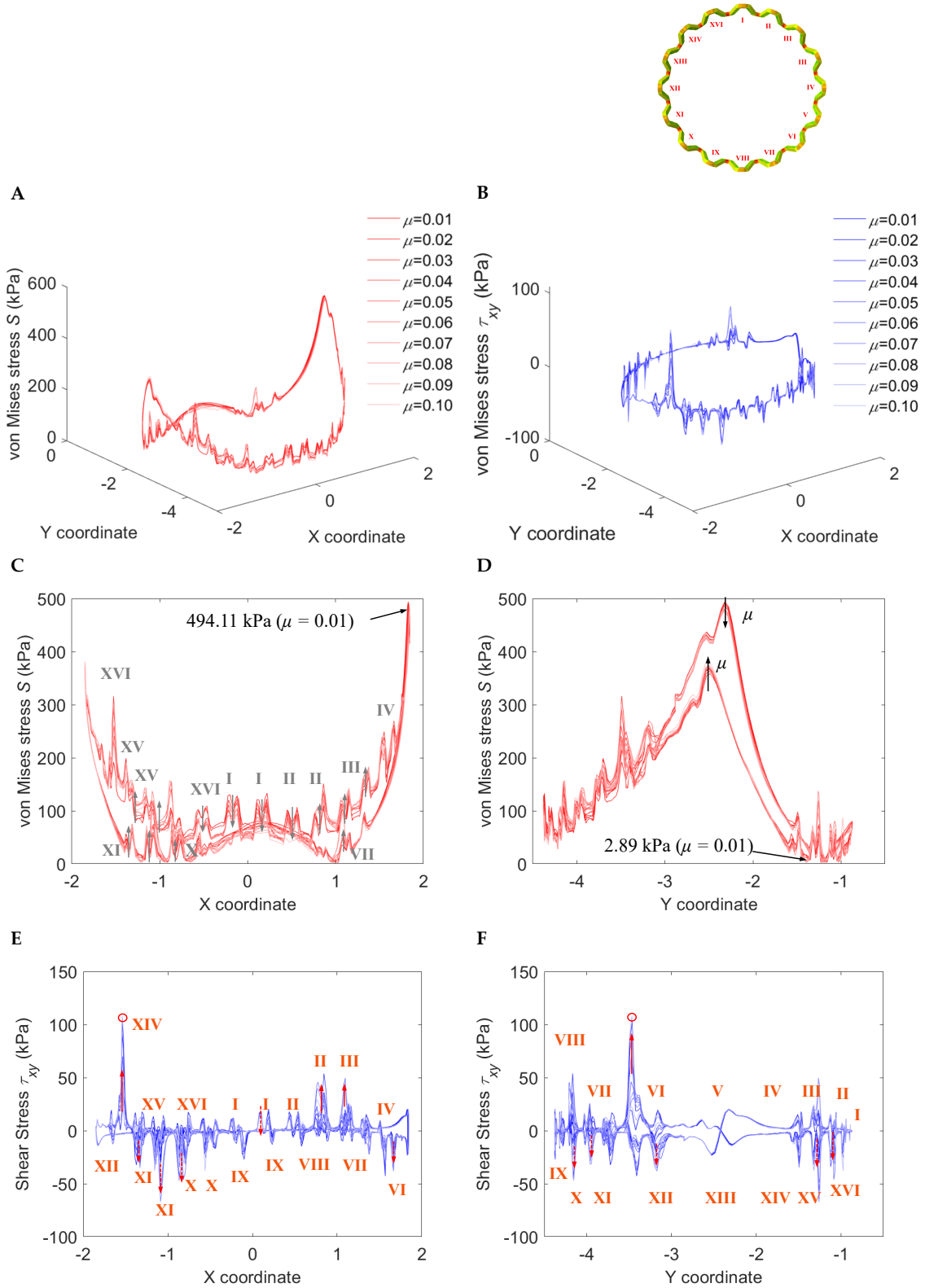


Figure 5.17: The stress distribution obtained from the nodes: (A-B) 3D plots of the von Mises stress and the shear stress τ_{xy} , with the x and y coordinates aligned along the nodes, respectively. (C-D) Von Mises stress, S , along the x - and y -directions, respectively. (E-F) The shear stress τ_{xy} along the x - and y -directions, respectively.

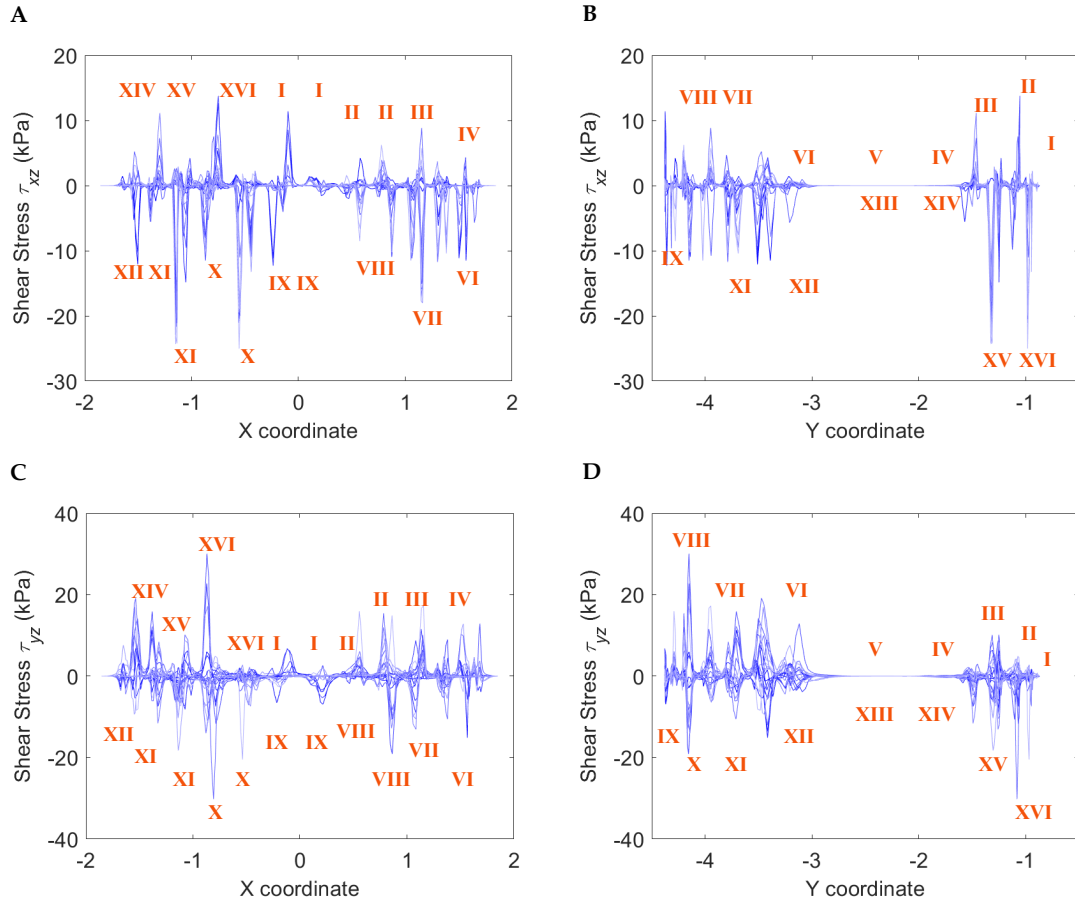


Figure 5.18: Shear stress along the nodes: (A-B) Shear stress τ_{xz} along the x - and y -directions, respectively. (C-D) show the shear stress τ_{yz} along the x - and y -directions, respectively.

contributing to increased instability in numerical simulations.

5.3.3.3 Results of Surface Shape Effects

Figure 5.22 presents the average values of the maximum von Mises stresses S at the nodes on the contact interface Γ^C between cross-section $z = 0$ and $z = 0.1$. The results are shown for Convex-1 (grey), Convex-2 (yellow), and Concave-1 (green), with a flat surface (red) as the reference. The left-side sub-figures (dark colors) correspond to $\mu = 0.01$, while the right-side sub-figures (light colors) represent $\mu = 0.03$.

Compared to a flat surface, Convex-1 (grey) exhibits lower von Mises stresses for most of the struts, except for the non-contact stent cell V for both $\mu = 0.01$ and $\mu = 0.03$. In contrast, Convex-2 (yellow) and Concave-1 (green) demonstrate higher von Mises stresses than the flat surface. However, the surface shapes improve the stability of the numerical simulations, as evidenced by the smaller SDs for the average von Mises stresses S .

The average values of maximum and minimum shear stresses τ_{xy} with their SD for three surfaces

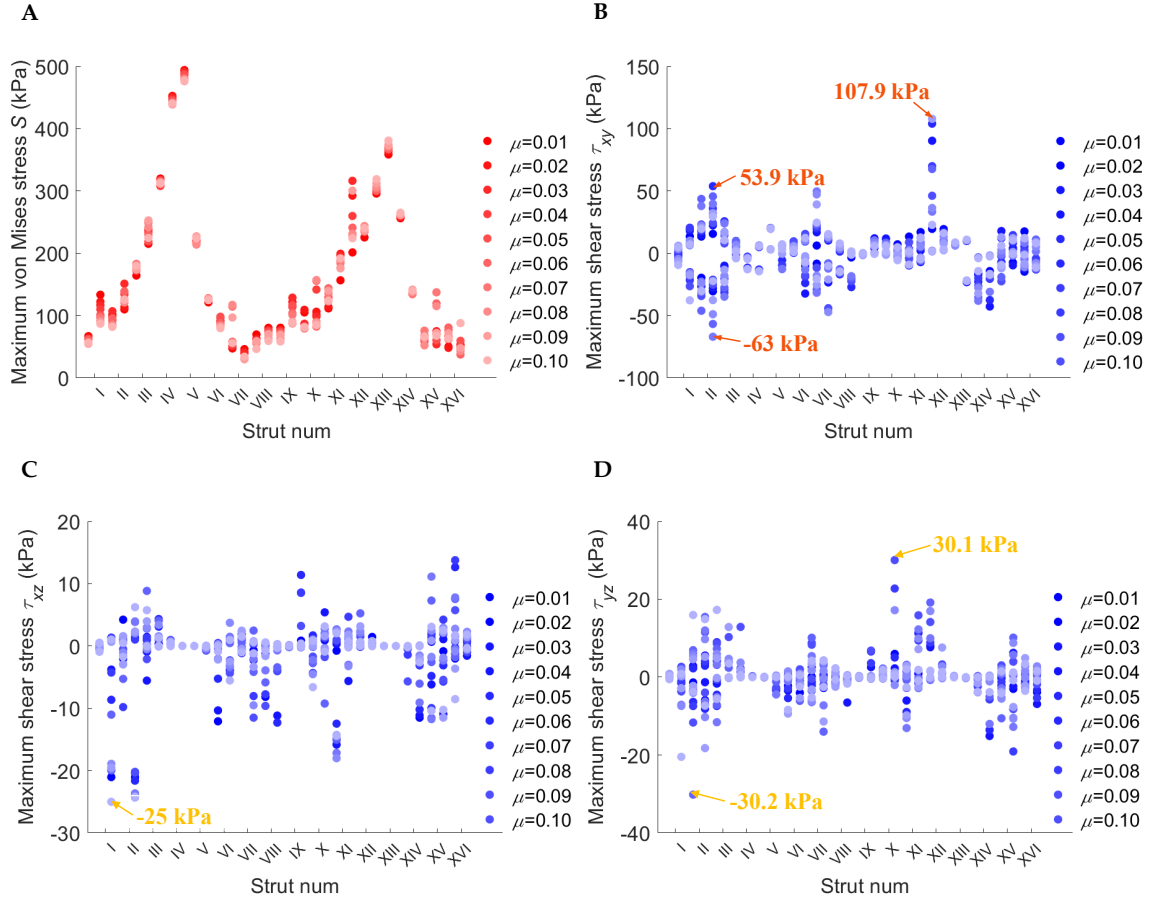


Figure 5.19: The stress distribution in struts: (A) The von Mises stress S . (B-D) The shear stress τ_{xy} , τ_{xz} and τ_{yz} , respectively.

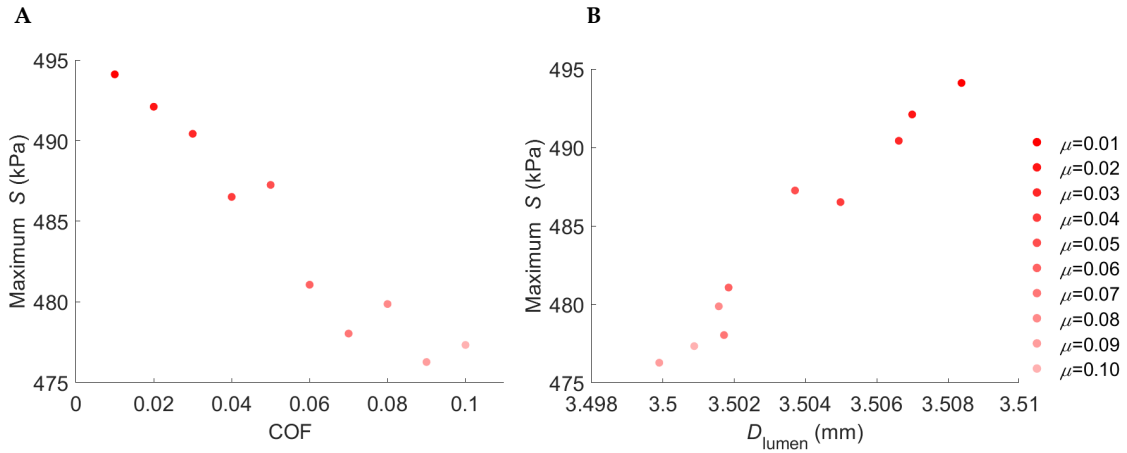


Figure 5.20: The maximum von Mises stress S : (A) S decreases with increasing COF. (B) S increases with increasing the lumen diameter D_{lumen} .

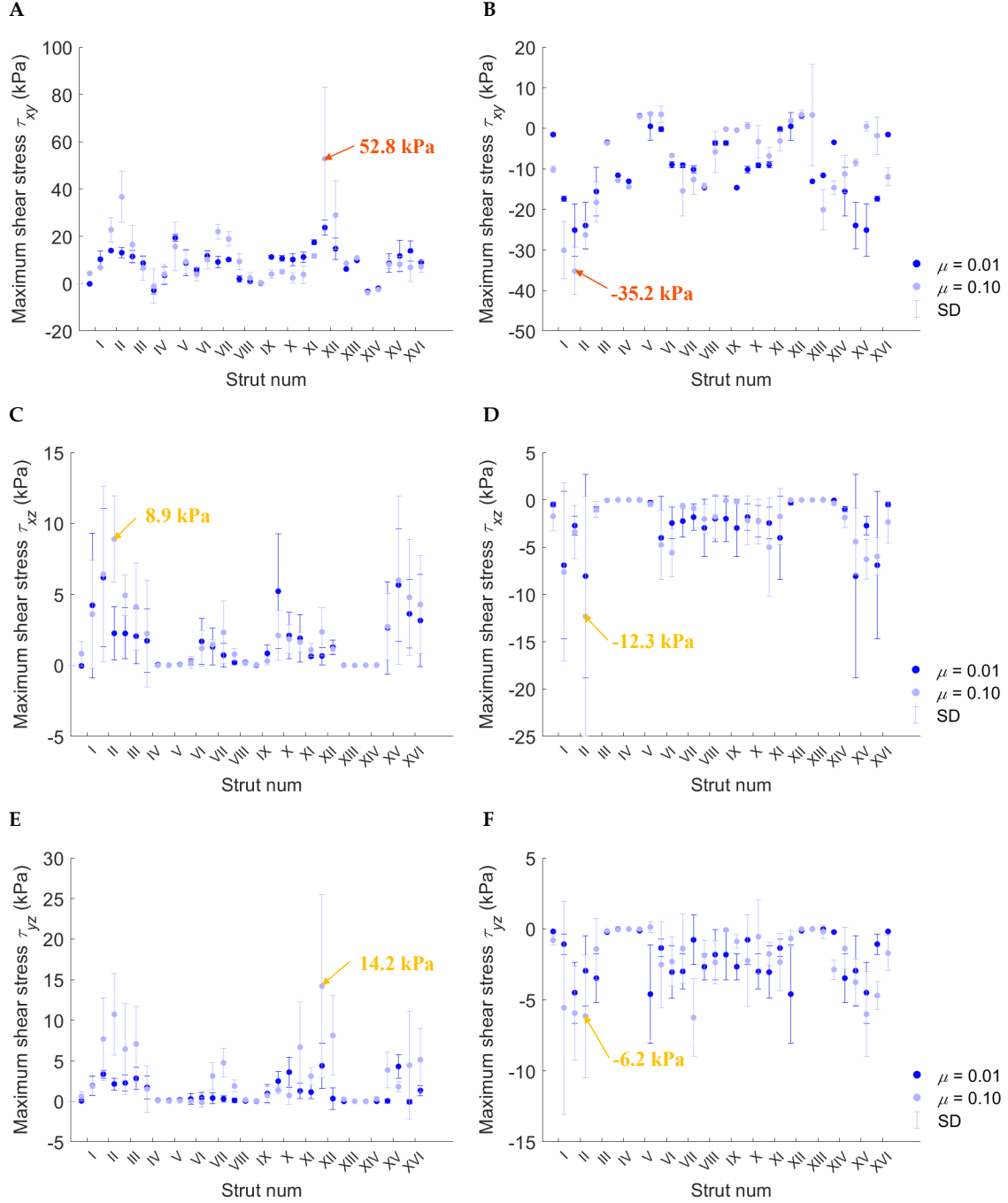


Figure 5.21: The average values of maximum and minimum shear stress, along with the SD, given by nodes at interface Γ between $z = 0$ and $z = 0.1$, with COF of 0.01 (dark) and 0.1 (light): **(A-B)** The results of shear stress τ_{xy} , where **(A)** represents the average values of maximum shear stresses and **(B)** represents the average values of minimum shear stresses. **(C-D)** The shear stress τ_{xz} . **(E-F)** The shear stress τ_{yz} .

and the flat surface are illustrated in Figure 5.23 for $\mu = 0.01$ and Figure 5.24 for $\mu = 0.03$. Convex-2 (yellow) and Concave-1 (green) exhibit relatively larger shear stresses compared to the flat surface (red) for both $\mu = 0.01$ and $\mu = 0.03$. Convex-1 (grey) shows slightly lower shear stress than the flat surface at $\mu = 0.01$; however, this trend is less clear when the COF is increased to 0.03.

Therefore, a convex surface with an appropriate amplitude A , such as $A = 0.01$ mm for Convex-1, results in lower stress values—including both von Mises stress S and shear stress τ_{xy} —at a lower COF ($\mu = 0.01$), which is more favorable for the artery-stent interface. In contrast, concave shapes, like Concave-1, or convex shapes with larger amplitude, like Convex-2, lead to higher von Mises and shear stresses. From a mechanical perspective, these shapes are not recommended for stent design.

5.4 Discussion

Compared to the media and adventitia, the intima exhibits higher values for five key indicators: von Mises stress S , maximum principal stress σ , strain energy density SED , and shear stress τ_{xy} due to its higher shear modulus, as well as maximum principal strain ε due to the largest stretch. Except for the shear stress τ_{xy} , which shows the peak values at the interface Γ^C , the maximum values of S (494.11 kPa), σ (540.4 kPa), SED (164.5 kJ/mm³), and ε (71 at $\mu = 0.01$) are all located at the right shoulder of the fibrous cap. These elevated values are caused by the stretching of arterial components in the circumferential (T –) direction as the artery expands during the artery-stent interaction. The right shoulder of the fibrous cap was identified as the initial fracture point in the research by Gasser et al. [99], with peak values reported around 500 kPa. Similarly, Cheng et al. [273] noted this location as the site of highest stress, with rupture stresses ranging from 300 to 705 kPa. In contrast, studies by Mulvihill et al. [72] and Li et al. [77] observed even stress distributions around a circular lumen, attributed to their use of circular lumen shapes expanded by circular stents in their models.

The artery-stent interaction induces higher stresses in the contact region than in the non-contact region (see Figure 5.16A–C). Similar observations have been widely reported in studies [79], [81], [267], [277], [288]. Due to differences in material models, geometric models, and boundary conditions, the reported maximum stress values in the contact region vary significantly, typically ranging from tens to hundreds of kilopascals. Schiavone et al. [81] reported maximum stress values between 290 and 580 kPa in contact areas with various stents. Zahedmanesh et al. [79] observed

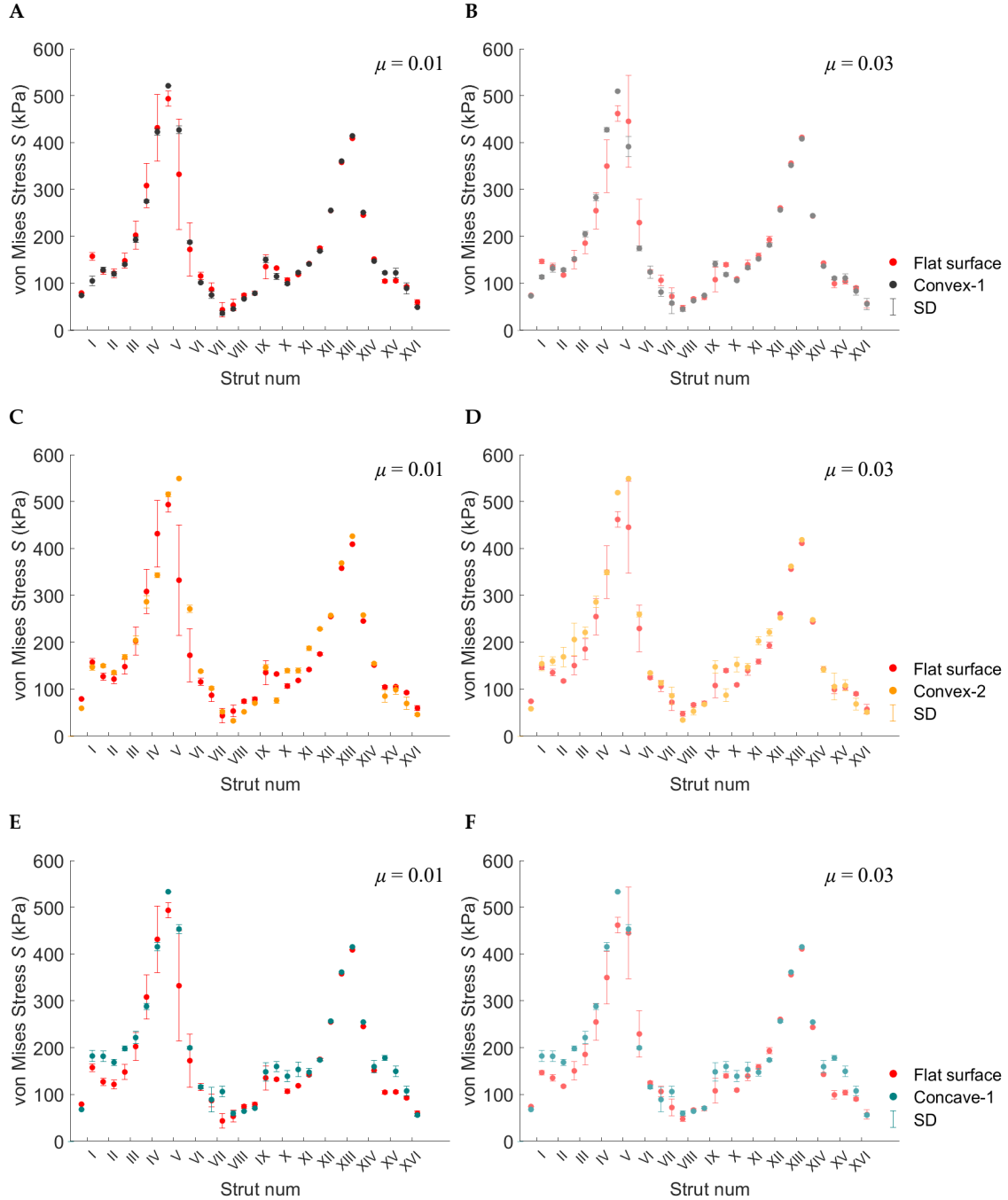


Figure 5.22: The average values of maximum von Mises stresses presented for Convex-1 (grey), Convex-2 (yellow), and Concave-1 (green) at COF of 0.01 and 0.03, using a flat surface (red) as the reference: **(A-B)** Convex-1 with the flat surface, where **(A)** represents the results at $\mu = 0.01$ and **(B)** represents the results at $\mu = 0.03$. **(C-D)** Convex-2 with flat surface. **(E-F)**: Concave-1 with flat surface.

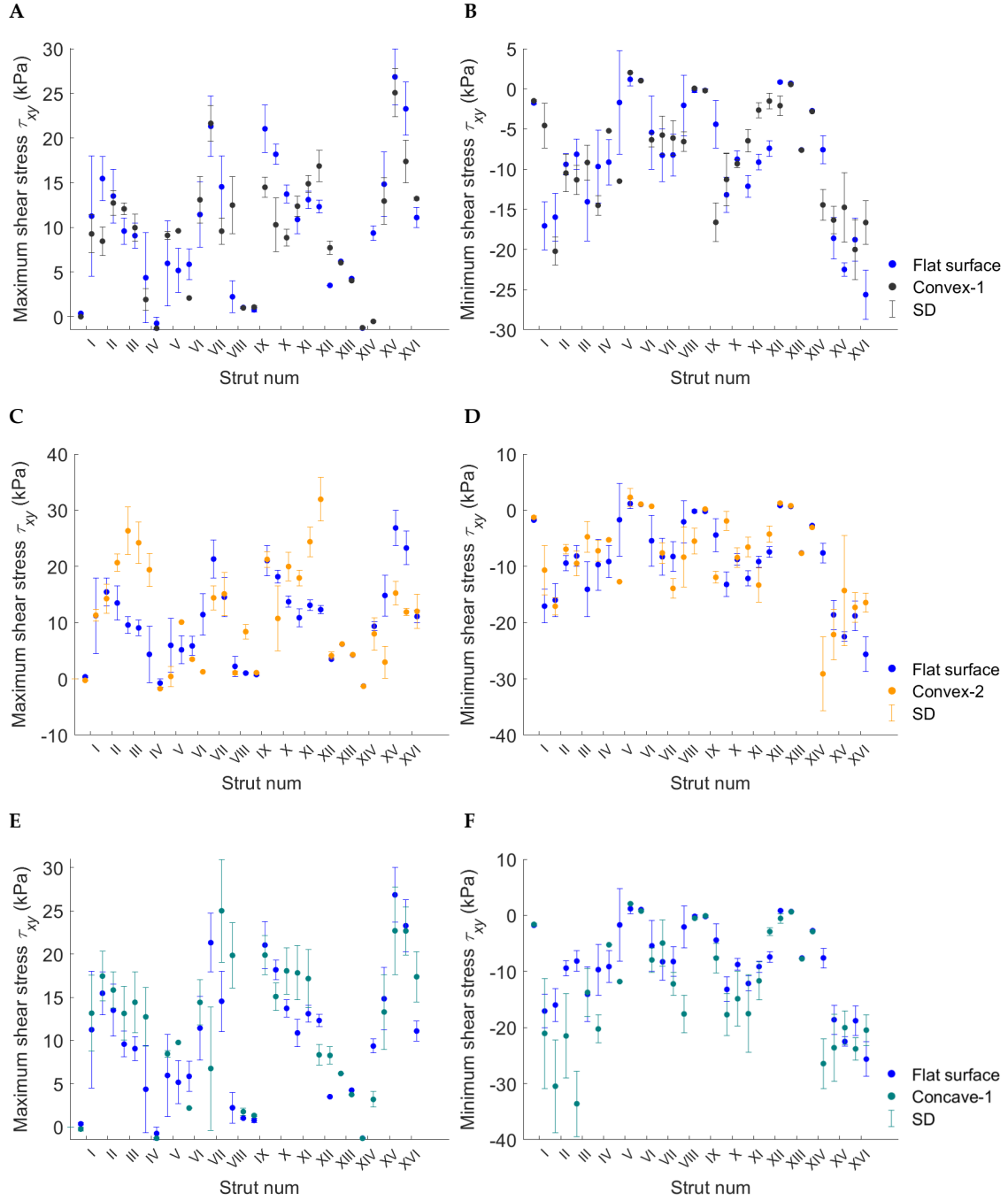


Figure 5.23: The average values of maximum and minimum shear stresses for Convex-1 (grey), Convex-2 (yellow) and Concave-1 (green) at COF of 0.01, using a flat surface (red) as the reference: (A-B) Results of Convex-1 with the flat surface, where (A) represents the average maximum values and (B) represents the average minimum values. (C-D) Convex-2 with flat surface. (E-F): Concave-1 with flat surface.

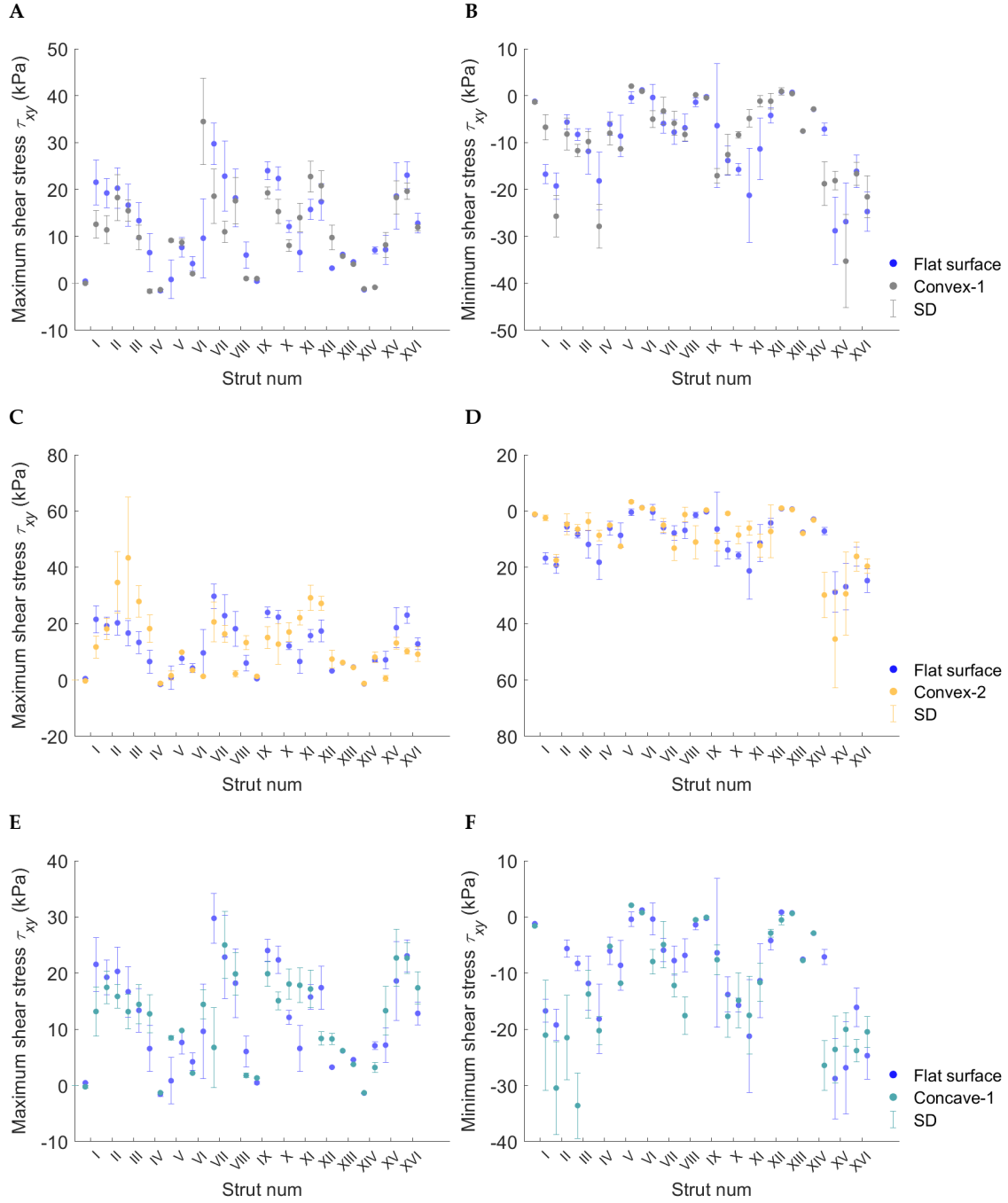


Figure 5.24: The average values of maximum and minimum shear stresses for Convex-1 (grey), Convex-2 (yellow), and Concave-1 (green) at COF of 0.03, using a flat surface (red) as the reference: (A-B) Results of Convex-1 with the flat surface, where (A) represents the average maximum values and (B) represents the average minimum values. (C-D) Convex-2 with flat surface. (E-F): Concave-1 with flat surface.

a maximum von Mises stress of 950 kPa in the plaque, while Conway et al. [267] reported much higher values, reaching up to 10,000 kPa—significantly exceeding those typically observed. In these studies, the attached plaque can be considered analogous to the fibrous cap examined in both our research and the work conducted by Gasser and Holzapfel’s group [96, 98, 99].

In this research, friction effects are most pronounced in the shear stresses at the contact area, with higher shear stresses observed as the COF increases (Figures 5.19 and 5.21), which aligns with Coulomb’s friction theory, as described by Eq. 5.4. Additionally, it is observed that the von Mises stress in some struts increases with higher COF (Figure 5.21A), although these changes are not substantial. A smaller COF, in this study, results in a larger lumen diameter D_{lumen} , and the resulting stretching effect mitigates the impact of friction in the overall artery FE model. The combined results demonstrate a notable decrease in von Mises stress, S , with increasing COFs, whereas S increases with larger lumen diameters, D_{lumen} . Nematzadeh et al. [287] also observed a decrease in maximum stress with increasing COF, though they did not discuss the underlying reasons. They further reported that increasing the COF between the Nitinol stent and the peripheral artery did not lead to significant changes in stress and strain across the entire artery model, which is consistent with our findings.

These higher stresses, either at the contact interface or at the right shoulder of the fibrous cap, serve as critical indicators of vulnerable regions during stent interactions, offering essential guidance for stent design and placement. Regarding the reported rupture stress of 600 kPa in [282], 1720 ± 890 kPa in [281] and 300 kPa in [273], as well as the rupture strain of $47 \pm 23\%$, the maximum principal stress of 540 kPa and the maximum principal strain ε of 70% observed in our studies suggests that rupture is likely. A smaller but effective lumen diameter is expected to mitigate this risk.

In this research, the topography effect is modelled at the nodal level using a non-uniform method, as described by Eq. 5.8. The method for modelling the topography effect has been widely reported in research [293, 295–298]. While Yastrebov et al. [295] provided detailed surface topography, more recent methods have shifted toward approaches that do not show detailed topographical data for FE models. These newer methods help to reduce the computational complexity associated with mesh generation and simulation. Our method is inspired by Paggi [293, 298]. They introduced the eMbedded Profile for Joint Roughness (MPJR) interface FE method, which allows for a nominally flat surface to incorporate arbitrary heights at the nodal level, effectively reconstructing the displacement field due to contact in the presence of surface roughness. However, implementing the MPJR interface FE method in ABAQUS/Explicit involves complex scripting to define the field

variables and contact properties for the interface layers. To simplify this process, we propose an alternative method by directly applying a non-uniform displacement to represent the surface topography instead of defining an interface layer. Given that contact simulations in ABAQUS/Explicit start with zero initial gaps, the non-uniform displacement is divided into two parts: the contact region receives a displacement equal to the baseline displacement u plus the topography profile $h(\xi)$, while the non-contact region is assigned a displacement of zero. This approach is even more straightforward in ABAQUS/Standard, which uses an implicit method and includes a gap element. The gap element can handle the non-contact region by setting its value to the maximum gap, and the displacement can then be directly applied as $u + h(\xi)$.

By investigating convex and concave surfaces, it was found that convex surfaces with an appropriate amplitude, such as $A = 0.01$ mm, result in a lower stress distribution, which is more favorable for the artery-stent interface. However, topographies featuring concave or large amplitude convex lead to higher von Mises stress and shear stress, making them less suitable for surface modification.

In this research, the minimum mesh size for the stent and the inner surface of the lumen of the artery is set to 0.01 mm^3 , which results in a total of 1,616,000 elements for the artery and 66780 elements for the stent. We attempted to refine the mesh further by using a mesh size of 0.008 mm^3 , however, the simulation terminated automatically, and reported a large mesh distorted problem. In ABAQUS/Explicit, particularly for large deformation problems, finer meshes create more contact points, potentially introducing localized stress concentrations and leading to instability in the contact algorithm. This is more significant in simulations with high friction coefficients. Therefore, the minimized mesh size is given as 0.01 mm^3 , which restricts the simulation of topographical features at the scale of 0.01 mm. Given the small dimensions of the stent, with a width of only 0.09 mm, there is limited space for detailed topography modelling. Additionally, due to the mesh size and the finite element algorithm built in ABAQUS/Explicit, higher COFs often result in significant mesh distortion, causing the simulation to terminate automatically. These issues currently remain unsolved. Furthermore, while blood flow-induced wall shear stress can significantly influence vascular healing and contribute to restenosis noted in [98], [299–301], this aspect is not considered in our studies.

5.5 Conclusion

In this chapter, the artery-stent interaction was investigated using finite element simulations conducted in ABAQUS/Explicit. The effects of friction were examined by setting the COF as 0.01 ,

0.02, ... 0.1. Additionally, the impact of shapes of stent surfaces, including convex and concave surfaces, was simulated using a non-uniform displacement.

Our findings reveal that the maximum von Mises stress, maximum strain energy density and maximum principal logarithmic strain are observed at the shoulders of the fibrous cap, with the right side exhibiting higher values than the left. These values are positively associated with the lumen diameter D_{lumen} , with large D_{lumen} contributing larger stress values. The shoulders of the fibrous cap are identified as the initial fracture point for an atherosclerotic artery. Additionally, a large principal logarithmic strain ε is also observed around the lipid pool, indicating a higher likelihood of plaque rupture.

The higher COF gives larger shear stress at the interface. Stent surface shape features different stress distribution at the interface. We recommend using a convex surface with an appropriate amplitude for stent surface modification, rather than concave or convex surfaces with a large amplitude, to achieve a more favorable stress distribution.

Discussion and Recommendation

Contents

6.1 Discussion	109
6.1.1 Contact Modelling	109
6.1.2 Artery-Stent Interaction	114
6.2 Recommendations	117
6.2.1 Recommendations for Contact Modeling	117
6.2.2 Recommendations for Artery-Stent Interaction	120

This chapter provides a discussion of the results for two key topics: contact modeling and the FE simulation of artery-stent interactions. It compares these results with relevant literature and offers recommendations for future work.

6.1 Discussion

6.1.1 Contact Modelling

Chapters 3 and 4 explore the contact behaviour of skin and mouse arterial tissue using indentation experiments, emphasizing the hyperelastic properties and the thin characteristics of both tissue types. Table 6.1 and 6.2 summarizes the studies on evaluating the material properties of skin and arterial tissue using indentations, respectively, with key points discussed below.

- *Method developed for material properties of skin and arterial tissue*

Among the studies investigating the linear elastic properties of skin tissue, as summarized in Table 6.1, the elastic modulus E is typically determined using contact models such as Hertz's model, Sneddon's model, Hayes' model, and Oliver-Pharr's model, with experiments conducted using a

small indenter (on the order of millimeters) and shallow indentation depths. For arterial tissue, E is often derived using Hertz's and Sneddon's models, with even smaller indenters on the order of micrometers. Reported values for E range from 0.07 kPa [119] to ≥ 10 kPa [302]. For arterial tissue, E is often derived using Hertz's model, with reported values spanning from 0.01 to 0.77 MPa [125, 136, 237, 251, 303].

The inverse FE method is widely adopted in current studies investigating the hyperelastic properties of both skin and arterial tissues through indentation experiments. Material parameters for skin tissue have been defined by various hyperelastic models, including the Neo-Hookean model [227, 304], the Mooney-Rivlin model [227, 304], and the Ogden model [55, 75, 304, 305], while the arterial tissue was modelled using the Neo-Hookean model [58–60]. To balance accuracy with computational efficiency, many researchers adopt simplified FE models for skin tissue, as demonstrated in [75, 138], which significantly reduce computation time and complexity. However, [55, 305, 306] use more realistic FE models to achieve greater precision in simulating tissue behavior. Additionally, studies by Tran et al. [55] and Delalleau [138] investigated initial parameter values for inverse analysis in the context of *in-vivo* forearm experiments. Their work provided close approximations for these values, thereby enhancing the accuracy of the inverse analysis. Similar studies addressing other soft materials can be found in [137, 307].

Lin's model was used in [24, 214] to characterize the hyperelastic properties of skin tissue. Compared with the inverse FE model, Lin's model is computationally efficient. However, since Lin's model is established for small deformations and infinite layers with shallow indentation depths, significant errors are introduced when applied to large deformations and finite layers, such as in our studies.

Compared to Hertz's, Lin's, and Sneddon's models, the approach combining FE simulation with Hayes' model proposed in this research offers a more accurate analysis of thin, deeper indentation data. Additionally, it is significantly more efficient than the inverse FE method. By demonstrating the equivalent shear moduli, G_i^* , throughout the indentation process, the proposed method proves well-suited for comprehensive material characterization.

- *Specimens choice*

This research primarily focuses on developing methods for exploring the hyperelastic properties of soft tissues. Therefore, instead of using human tissue, which has been chosen in previous studies such as [119, 139, 154, 302, 308], we selected commercially available artificial skin with a smooth surface, a relatively thick profile (approximately 1 mm), and homogeneous material prop-

Table 6.1: The indentation test on skin and arterial tissue

Research	Material	Modelling	Material model	Material parameters	Experimental Setup
Bader et al. [308]	Young male thigh Young male forearm Young female forearm Aged female forearm	Hayes Model	Linear elastic	$E = 1.99 \pm 0.43$ kPa $E = 1.51 \pm 0.32$ kPa $E = 1.09 \pm 0.54$ kPa $E = 1.11 \pm 0.17$ kPa	plane-end indenter $R = 20$ mm, $F = 11.7$ kN/m ²
Jachowicz et al. [119]	artificial skin <i>in-vivo</i> skin	Hertz's Model	Linear elastic	$E = 0.55 - 1.77$ kPa $E = 0.07 - 0.33$ kPa	$R = 4$ mm, $\omega = 0.6$ mm
Zahouani et al. [302]	<i>in-vivo</i> skin	Sneddon's Model	Linear elastic	$E = 8.3 \pm 2.1$ kPa	$R = 2.25$ mm, $\omega = 600$ μ m
Pailler et al. [139]	<i>in-vivo</i> skin	Hertz's Model	Linear elastic	$E = 9.5 \pm 2$ kPa	$R = 6.35$ mm, $\omega = 2$ mm
Pailler et al. [154]	<i>in-vivo</i> skin	Sneddon's model Two-layer model	Linear elastic Linear elastic	$E = 12.5 - 13.7$ kPa $E = 4.5 - 8$ kPa	conical indenter $H = 10$ mm, $\alpha = 45^\circ$ $\omega = 15$ mm
Delalleau et al. [138]	<i>in-vivo</i> forearm	Inverse FE Method	Linear elastic	$E = 5.67$ kPa	$R = 6.22$ mm, $\omega = 800$ μ m
Geerligts et al. [145]	<i>ex-vivo</i> abdominal skin	Oliver and Pharr's Model	Linear elastic	$E_{SC} = 2.6 \pm 0.6$ kPa $E_{VD} = 1.1 \pm 0.2$ kPa	$D = 500$ μ m, $\omega = 8-12$ μ m
Tran et al. [55]	<i>in-vivo</i> forearm	Inverse FE Method	Neo-Hookean	$C_{10E} = 0.12 \pm 0.06$ MPa $C_{10D} = 1.11 \pm 0.09$ MPa - $C_{10H} = 0.42 \pm 0.05$ MPa	
Flynn et al. [75]	<i>in-vivo</i> forearm	Inverse FE Method	Ogden	$\mu = 6.568$ kPa, $\alpha = 25.152$	
Iivarinen et al. [309]	<i>in-vivo</i> forearm	Inverse FE Method	Neo-Hookean	$E = 210-651$ kPa	$R = 2$ mm $F = 4$ N
Iivarinen et al. [310]	<i>in-vivo</i> forearm	Inverse FE Method	Neo-Hookean	$E = 130.2$ kPa	$R = 2$ mm $F = 4$ N
Crichton et al. [24]	mouse ear skin	Lin's Model	Linear elastic	$E_{SC} = 0.75 - 1.62$ MPa $E_{VD} = 0.49 - 1.51$ MPa $E_D = 7.33 - 13.48$ MPa	$D = 1.9$ μ m, $\omega = 10\%$
Wei et al. [214]	human skin <i>in-vivo</i> human skin	Lin's Model	Ogden	$E = 1.99 - 50.26$ kPa $E = 1.5 - 108.19$ kPa	$D = 0.18 - 3.15$ mm, $\omega = 10\%h$
Dai et al. [212]	<i>in-vivo</i> human skin	Lin's Model	Linear elastic	$E = 3.9 - 13.7$ kPa	$R = 6-10$ mm, $\omega = 2.5$ mm
Marinopoulos et al. [311]	<i>in-vivo</i> human skin	inverse analysis	Ogden	$E = 3.9 - 13.7$ kPa	$R = 6-10$ mm, $\omega = 2.5$ mm

[1] SC, VD, D represent the stratum corneum, viable epidermis, and dermis, respectively.

Table 6.2: Shear moduli and elastic moduli reported in literature

Research	Material	Modelling	Material model	Material parameters	Experimental Setup
Meng et al. [251]	Mouse pulmonary vessels	Hertz's model	Linear elastic	$E = 3.29 \pm 1.2$ kPa	$R = 50$ μm , $\omega = 500$ nm
Liu et al. [237]	Rat arterial tissue	Hertz's model	Linear elastic	$E = 1.66 \pm 0.49$ kPa	$D = 5$ μm , $F = 26$ nN
	Human pulmonary vessel			$E = 5.44 \pm 2.96$ kPa	
Peloquin et al. [125]	Bovine endothelium	Hertz's model	Linear elastic	$E = 2.5 \pm 1.9$ kPa	$R = 4.5$ μm , $F = 1$ nN
	Bovine subendothelium			$E = 2.77 \pm 1.1$ kPa	
Sicard et al. [136]	Bovine arteries	Hertz's model	Linear elastic	$E = 167.66 \pm 122.26$ kPa	pyramidal tips $R = 20$ nm, $\omega = 2.26 \pm 1.07$ μm
				$E = 8.36 \pm 7.55$ kPa	$R = 1$ μm , $\omega = 2.04 \pm 1.02$ μm
				$E = 9.41 \pm 11.21$ kPa	$R = 2.5$ μm , $\omega = 1.74 \pm 0.96$ μm
Melo et al. [252]	Human tunica intima	Hertz's model	Linear elastic	$E = 79.3 \pm 7.2$ kPa	pyramidal tip $\theta = 20^\circ$, $\omega = 0.5$ μm
	Human tunica adventitia			$E = 41.1 \pm 8.0$ kPa	
Engler et al. [253]	Arterial medial layer	Dimitriadis' model	Linear elastic	$E = 5.8$ kPa	$R = 2.5$ μm , $\omega = 2$ μm
Barrett et al. [58]	Atherothrombotic plaque	Inverse analysis	Neo-Hookean	$G = 7\text{--}100$ kPa	$R = 0.5$ mm, $F = 0.2$ N
Chai et al. [59]	Human carotid artery	Inverse analysis	Neo-Hookean	$E = 6\text{--}891$ kPa	$D = 2$ μm , $\omega = 30\%h$
Zhang et al. [303]	Porcine arteries	Sneddon's model	Linear elastic	$E = 0.77 \pm 0.11$ MPa	$R = 0.75$ mm, $\omega \leq 10\%h$
	Human arteries			$E = 0.75 \pm 0.15$ MPa	
Lundkvist [312]	Human femoral artery	Sneddon's model	Linear elastic	$E = 34.3$ kPa	-
	Fresh aorta wall			$E = 63.4 \pm 4.1$ kPa	conical indenter $R = 10$ μm , $\theta = 10^\circ$, $\omega \leq 40$ μm
Hemmasizadeh et al. [148]	48h-cold storage aorta	Sneddon's model	Linear elastic	$E = 59.0 \pm 4.3$ kPa	
	48h-cold storage aorta			$E = 56.9 \pm 4.4$ kPa	

erties. This artificial skin serves as an ideal specimen for developing our experimental methodology.

The selection of mouse arterial tissue was mainly due to its structural and mechanical similarities to human arterial tissue and its ready availability through collaboration with King's College London. The mouse arterial model has been widely used as a substitute for human arterial tissue in various research studies, including [313], [314], particularly in studies focusing on the mechanical properties of atherosclerotic arteries or plaques, such as [315]–[317].

- *Thickness determination*

The rough surface of mouse arterial tissue introduces excessive noise during data collection and complicates the determination of tissue thickness, which is essential for contact model application, as discussed in Chapter 4. These challenges have also been addressed in previous studies on indentation of arterial tissue, such as those by [58], [145], as well as in various works related to other soft tissues, including [22], [233], [240], [242]–[244]

Geerligs et al. [145] used a large indenter with a diameter of 500 μm to create a larger contact area relative to the tissue cell size. This ensured that the deformation was more homogenized across the tissue, effectively averaging out surface irregularities and minimizing the influence of individual cellular features on the measurements. Similarly, Chandran et al. [244] noted that the initial contact point for applying contact models was located deeper than the physical (or slope-based) contact point. They identified the initial contact point by finding the optimal fit of Hertz's model. Chandran's approach is similar to the method proposed in our study, as both rely on the optimal fitting of a contact model. While Chandran et al. used the best fit for Hertz's model to locate the initial contact point, our method used the smallest variance (S^2) of the equivalent elastic moduli (E_i^*) to directly identify the equivalent thickness, h_{hayes} . This approach allows for thickness determination without the need for apparatus-based measurements, ensuring both accuracy and the applicability of the Hayes model.

Considering that the thickness of the tissue cannot be measured before the indentation test, the displacement-controlled approach typically used for indentation tests on skin tissue—where the indentation depth is set to 10% of the thickness—was not feasible for mouse arterial tissues. Therefore, a force-controlled approach, with a contact force set to $F = 0.001 \text{ N}$, was employed, as detailed in Chapter 4. This method simplifies the data collection process by eliminating the need to know the thickness of the initial contact point and allows for deeper indentation depths. However, this approach also introduces additional challenges in data analysis, particularly in identifying the starting point of contact.

6.1.2 Artery-Stent Interaction

Chapter 5 demonstrated the artery-stent interaction using FE simulations conducted in ABAQUS/Explicit.

6.1.2.1 Finite Element Model

The explicit solver is widely used for modeling artery-stent interactions, see e.g., [73], [78–81], [267], [271], [318], [319], where the stent deployment was treated as a quasi-static process with a common criterion being to maintain the kinetic energy below 5-10% of the internal energy, as suggested by [320]. While the implicit solver has also been used for artery-stent interaction in some studies [279], [280], [286], [321], it is generally applied when the stent undergoes relatively small displacements. Studies such as [80], [320] demonstrate that the explicit solver significantly reduces computational time and mitigates the convergence issues often encountered with implicit methods.

In this research, the FE model of the artery is developed based on models provided by Gasser and Holzapfel et al. [96], [275], incorporating essential parameters such as the maximum and minimum lumen diameters, outer artery diameter, component thicknesses, and the shape of the lipid pool. Holzapfel et al [98] reconstructed the arterial geometry using non-uniform rational B-splines derived from high-resolution magnetic resonance imaging, allowing for detailed characterisation of morphological features at various vascular length locations. The results in [275] were tailored to meet specific arterial and clinical requirements. In this research, we adopt Gasser's method [31], [96], [99] and focus on a 1 mm-wide artery segment. By reducing the representation of detailed morphological sections, the computational efficiency was improved. A similar realistic modelling approach for artery-stent interaction has been used in studies such as [73], [273], [276], [318]. Compared to idealised models, which represent the artery as a cylindrical vessel with a parabolically shaped plaque defined by isotropic, homogeneous, and incompressible materials in [81], [267], [271], [272], [277], a realistic FE model can offer a more accurate and reliable representation of artery-stent interactions.

In this research, the Gasser-Ogden-Holzapfel (GOH) model was used to define the material properties of different arterial components. In earlier studies, such as [273], [322], the material properties of arterial tissue were modeled as linear elastic. More recent FE models typically define the material properties of arterial tissues using hyperelastic models, including Ogden model [77–81], Yeoh model [72], [73], Mooney-Rivlin model [271], [278], [280], [287], [288], [323], [324], and GOH model [31], [98], [99], [318], [325]. Among these material models, the GOH model is particularly effective in capturing the anisotropic behavior of arterial tissues, and the material parameters provided

in [97], [98] have been widely accepted and used as a foundation for constructing other material models, see studies [78], [79].

In this research, ‘hard contact’ is defined as the contact property, with the penalty method employed to enforce it. ‘Hard contact’ offers a more realistic representation of contact behavior compared to softened contact. While ‘hard contact’ can be prone to introducing stress concentrations or mesh distortions—particularly around sharp stent edges, which could potentially lead to simulation interruptions—these issues did not occur in this study. ‘Hard contact’ is commonly used in similar studies (e.g., [81], [286], [321], [326]). The choice of contact formulation algorithm, whether the kinematic method [323] or the penalty method [96], [99], [125], [286], does not significantly impact the results of this study, but the penalty method significantly improves computational efficiency.

In a clinical scenario, a stent is compressed before insertion to navigate through the tortuous lumen and is then expanded once correctly positioned. The simulation of the stent compression process has been widely studied in the literature, e.g., [79], [270], [291], [327], [328]). However, this process incurs high computational costs, as noted by Filipovic et al. [292], and is therefore not included in our study, where we directly expand the stent. Several methods have been employed to simulate stent expansion, including the direct application of uniform pressure [98], [278] or radial displacement [80], [287], [288], [323] to the stent’s internal surface. Another common approach involves using a balloon to facilitate expansion [78], [79], [99], [123], [279], [324], [329]–[334]. The balloon expansion method can be implemented in various ways, such as using a folded balloon or applying pressure or displacement to the balloon’s interior, which has been detailed in studies by Gervaso et al. [328] and Martin et al. [258]. Given the focus of this research on the artery behaviour during the interaction, along with the high computational costs associated with balloon expansion, we directly apply radial displacement to the internal surface of the stent in our simulations.

6.1.2.2 Finite Element Results

The simulation results reveal that significantly higher stress and strain occur in the intima layer compared to the media and adventitia layers. The higher stress is attributed to the intima’s greater shear modulus ($G = 78.9$ kPa) compared to that of the media ($G = 15.0$ kPa) and the adventitia ($G = 1.75$ kPa). The higher strain in the intima is attributed to its greater stretch, as it is the innermost layer with a smaller diameter, leading to a more significant relative change in length compared to the media and adventitia. Additionally, the effects of fibre, defined by the GOH model, are evident in the asymmetric stress distribution within the media and adventitia, with higher stress showing along the fibre direction. Except for the shear stress τ_{xy} , which shows the peak values at the interface Γ^C , the maximum values of the von Mises stress S , strain energy

density, and maximum principal strain ε are all found at the right shoulder of the fibrous cap. It also found that the maximum stress is associated with the lumen diameter. These findings align with the research conducted by Gasser and Holzapfel group [31], [96], [98], [99] as well as the studies [72], [77], [273]. The highest principal stress of 540 kPa is given in this research with [99] reported the peak values around 500 kPa and Cheng et al. [273] reported an average stress of 545 ± 160 kPa for ruptured plaques. In contrast, studies by [72], [77] identified the highest stresses distributed along the circular lumen without localised stress concentrations. This difference can be attributed to the different shape of the lumen, with the elliptical lumen used in [31], [96], [98], [99], [273], [72], [77] feature a circular shape of the lumen.

However, for studies using idealised artery FE models with a parabola-shaped plaque attached to the intima, the highest stresses were observed in the contact region (see [79], [81], [277], [288]). The maximum stress values vary widely, ranging from hundreds of kilopascals, e.g., 290 kPa to 580 kPa in [81], to 950 kPa [79], and even reaching thousands of kilopascals, e.g., 10,000 kPa in [267], due to differences in material models, geometric configurations, and boundary conditions of the FE models. For the realistic models, higher stress is also observed in the contact region, but it is not the highest stress.

An interesting observation emerged: a smaller coefficient of friction (COF) led to a larger lumen diameter (LD), and larger LD corresponded to higher stress. Specifically, for $\mu = 0.01$, $LD = 3.508$ mm and $S = 494.11$ kPa, whereas for $\mu = 0.1$, $LD = 3.500$ mm and $S = 476.28$ kPa. This is also reported in [287]. Although the reason why a smaller COF results in a larger lumen diameter remains unclear, the relationship between a larger LD and higher stress is evident, as increased stretching correlates with elevated stress. This suggests that the stretching effect predominantly contributes to stress levels. Additionally, elevated maximum principal strains around the lipid pool indicate an increased likelihood of plaque rupture during stent deployment, a finding supported by previous studies [273], [281], [282]. At the interface Γ^C , shear stresses (τ_{xy}) are observed to increase with higher COF, consistent with Coulomb's friction theory.

Chapter 5 also introduces a method to model the effect of stent surface shape by applying a non-uniform displacement to represent the surface geometry. This approach is inspired by the eMbedded Profile for Joint Roughness (MPJR) interface finite element method proposed by Paggi [293], but it is easier to implement in ABAQUS/Explicit. However, a notable limitation of this method is its inability to account for the adhesion effects of normal and friction forces, which will be discussed further in the recommendations. Another limitation in simulating surface shape arises from the minimum mesh size for the stent, which is set to 0.01 mm^3 . This restricts the accurate

modeling of surface features at smaller scales. Attempts to use finer mesh sizes result in significantly higher computational costs and often lead to mesh distortion, causing the simulation to terminate prematurely. These challenges are likely tied to the finite element algorithm in ABAQUS/Explicit, which remains difficult to overcome with current capabilities.

6.2 Recommendations

Further research into contact modeling and tissue characterization is needed to improve understanding of tissue mechanics and overcome current limitations.

6.2.1 Recommendations for Contact Modeling

6.2.1.1 Multi-layer Model

In this study, skin and arterial tissue are modeled as single-layer homogeneous materials. However, real soft tissues are multi-layered structures, as discussed in Chapter 2. Investigating the mechanical properties of individual layers can offer deeper insights into the behavior of skin and arterial tissue during mechanical interactions.

The mechanical properties of distinct skin layers, including the stratum corneum, viable epidermis, and dermis, have been investigated in prior studies (e.g., [24], [214], [335]). Pailler et al. [154] introduced a two-layer elastic model to account for the impact of subcutaneous layers, such as muscles, on the mechanical behavior of human skin during indentation tests. Their findings revealed that traditional one-layer models fail to adequately capture variations in Young's modulus at shallow indentation depths. Expanding on this concept, Van Kuilenburg et al. [336] proposed a generalized multi-layer model. The effective elastic modulus, E_{eff} , for a multi-layer system is described by Eq. 6.1.

$$\frac{1}{E_{eff}} = 2 \sum_{i=1}^{n-1} \frac{t_i}{f_i(a)\pi a E_i} + \frac{1}{f_n(a)E_n} \quad (6.1)$$

where f_i is a correction function for each layer i , defined as:

$$\begin{aligned} f_1(a) &= 1 + \frac{2t_1}{\pi a} \\ f_i(a) &= \left(1 + \frac{2}{\pi a} \sum_{j=1}^{i-1} t_j \right) \left(1 + \frac{2t_1}{\pi a} \right) \\ f_n(a) &= 1 + \frac{2t_1}{\pi a} \end{aligned} \quad (6.2)$$

Further research aims to develop a multi-layer model for soft tissues, with each layer characterized by hyperelastic material properties. This endeavor will require advanced modelling techniques, supported by a specifically designed set of experiments, potentially combined with sophisticated computational methods. Additionally, investigating the application of multi-layer models in more complex geometries or under *in-vivo* conditions could provide valuable insights into the mechanical behavior of soft tissues.

6.2.1.2 Adhesion Modelling in Contact

Adhesion forces, while not considered in this study, play a significant role in contact modelling, particularly for soft tissues or small-scale interactions. These forces arise from attractive surface forces between contacting bodies and become increasingly relevant as the applied load decreases toward zero. Two principal theories for modeling adhesion in contact are the Johnson, Kendall, and Roberts (JKR) theory and the Derjaguin, Muller, and Toporov (DMT) theory. These theories define the extremes of adhesive contact behavior, depending on material and surface properties.

- *JKR theory*

The JKR theory applies to compliant, elastic samples with large tip radii and high surface energy, making it suitable for soft tissue modeling. This theory accounts for the effects of adhesion on the contact area and pressure distribution, as demonstrated in [337], [338]. The adhesive force F_{adh} is defined as:

$$F_{adh} = -\frac{3}{2}\pi\gamma R \quad (6.3)$$

where γ represents the energy per unit contact area, and R is the indenter radius. The contact radius a is given by:

$$a^3 = \frac{3R}{4E_r} \left[P + 3\pi R\gamma + \sqrt{6\pi RP\gamma + (3\pi R\gamma)^2} \right] \quad (6.4)$$

where E_r is the reduced modulus, discussed below. Studies, such as [139] and [339], have shown that adhesion significantly influences the contact area and pressure distribution. Dai et al. [339] demonstrated improved agreement with experimental data by incorporating a modified hyperelastic adhesion model to account for contact radius variations under different conditions.

- *DMT model*

The DMT theory applies to stiffer materials with small tip radii and low surface energy, representing

the opposite limit of adhesive contact behavior. The adhesive force F_{adh} in the DMT model is defined as:

$$F_{adh} = 4\pi\gamma R \quad (6.5)$$

The boundary between the JKR, DMT, and intermediate (transition) models is defined by the Tabor parameter μ (Eq. 6.6).

$$\mu = \left(\frac{R\gamma^2}{E_r^2 z_0^3} \right)^{1/3} \quad (6.6)$$

where z_0 is the equilibrium separation of the surfaces (typically taken to be on the order of 0.3 to 0.5 nm), and E_r is the reduced modulus, given by:

$$\frac{1}{E_r} = \frac{1 - \nu_i^2}{E_i} \frac{1 - \nu_s^2}{E_s} \quad (6.7)$$

Here, E_i and E_s are the elastic moduli of the indenter and sample, respectively, while ν_i and ν_s are their Poisson's ratios. The Tabor parameter determines the appropriate model:

- $\mu < 0.1$: DMT theory
- $\mu > 5$: JKR theory
- $0.1 < \mu < 5$: Transition model.

Further research aims to incorporate adhesive forces into the contact model to provide a more accurate explanation of the collected experimental data.

6.2.1.3 Two-term Friction Model

In this research, friction effects weren't a consideration in contact modeling of skin and arterial tissues. The two-term model is proposed by Johnson et al. and Adams et al. As described by Eq. 6.8, the friction force is given by the sum of an adhesion term and a deformation term.

$$F_\mu = F_{\mu,adh} + F_{\mu,def} \quad (6.8)$$

where $F_{\mu,adh}$ represents the adhesive friction, which is the force required to break the adhesive bonds between the two surfaces, see Eq. 6.9. $F_{\mu,def}$ refers to the deformation friction, which is the work lost due to viscoelastic hysteresis per unit sliding distance, see Eq. 6.10.

$$F_{\mu,\text{adh}} = \tau A_{\text{real}} \quad (6.9)$$

where τ is the shear strength of the interface and A_{real} is real contact area.

$$F_{\mu,\text{def}} = \frac{3}{16} \beta \frac{a}{R} F \quad (6.10)$$

where β is a viscoelastic loss friction.

Further research aims to conduct friction tests on soft tissue, establish a friction model for arterial tissue with different stent materials, and incorporate it into the FE simulation.

6

6.2.2 Recommendations for Artery-Stent Interaction

6.2.2.1 Adhesion Modeling in Finite Element Simulation

Paggi and Bonari et al. introduced a method to model adhesion effects for both normal (Eq. 6.11) and frictional forces (Eq. 6.12) in their proposed approach, known as the MPJR (eMbedded Profile for Joint Roughness) interface finite element method. The MPJR interface finite element method utilizes a nominally flat interface element capable of embedding arbitrary surface profiles at the nodal level, effectively reconstructing the displacement field arising from contact in the presence of surface roughness. A single layer of interface elements stores contact geometry information as a corrected gap, as discussed previously.

$$P_h = \frac{A_H}{6\pi g_0^3} \left[\left(\frac{g_0}{g_n} \right)^9 - \left(\frac{g_0}{g_n} \right)^3 \right] \quad (6.11)$$

where A_H is the Amaker's constant characterizing the strength of adhesion and g_0 represents the equilibrium distance between two approaching half-spaces. This equation is derived from a Lennard-Jones potential-like relationship define the normal direction.

$$\mathbf{q}_r = \mu(|p_n| - p_c) H(g_c - g_n) \frac{\dot{\mathbf{g}}_\tau}{\|\dot{\mathbf{g}}_\tau\|} \tanh \frac{\|\dot{\mathbf{g}}_\tau\|}{\dot{\epsilon}} \quad (6.12)$$

where p_c is the value of the normal traction corresponding to a specific cut-off normal gap g_c , g_p is the normal gap related to the pull-out normal traction, and $H(x)$ is the Heaviside step function.

While implementing this method in ABAQUS/Explicit would require extensive effort, particularly in defining the normal and tangential forces, it holds significant potential for accurately representing contact behavior under adhesion effects. Incorporating this approach could improve the fidelity of

simulations for applications where adhesion is critical.

Further research aims to incorporate adhesive effects into FE simulations to better capture the complexities in real artery-stent interaction.

6.2.2.2 Hemodynamic Effect

Computational fluid dynamics (CFD) is increasingly used to simulate and analyze hemodynamic flow patterns in arteries. Among the key parameters derived from CFD, wall shear stress (WSS) and wall shear stress gradient (WSSG), generated by blood flow, have been identified as critical contributors to the development of restenosis, as noted in numerous studies [340], [341]. These hemodynamic forces influence endothelial function, vascular remodeling, and the healing process following interventions.

Hemodynamic parameters such as blood flow rate, blood resistance, and blood inertia are also pivotal in shaping vascular responses. They interact dynamically with the arterial wall, contributing to changes in vessel structure and function over time.

Although this research has not addressed hemodynamic factors due to its defined scope, their significance in arterial health and disease progression makes them a valuable area for future research. We highly recommend exploring CFD-based hemodynamic simulations in future studies to gain deeper insights into vascular mechanics and improve predictive models for restenosis and vascular healing.

Conclusions

This thesis examines the contact behavior of artificial skin and mouse arterial tissue in Chapters 3 and 4 through a novel method that integrates indentation experiments with numerical simulations. This approach discretises the contact behavior of hyperelastic tissue into incremental steps, applying Hayes' model to approximate the linearised material response within each step. The incremental equivalent shear moduli, G_i^* , derived from Hayes' model, effectively represent the hyperelastic behavior of both artificial skin and mouse arterial tissue. In Chapter 4, the method is further developed to address uncertainties in the thickness of mouse arterial tissue. An 'equivalent thickness' parameter is introduced to improve the applicability of Hayes' model in accounting for these variations.

The findings reveal that the hyperelastic properties of both artificial skin and arterial tissue can be effectively characterized using a first-order Ogden model. For artificial skin, the shear modulus determined by the first-order Ogden model ranges from 0.13 MPa to 0.19 MPa. In the case of mouse arterial tissue, an average shear modulus of 1.22 kPa is derived from 26 measurements, with tissue thicknesses ranging from 50 to 500 μm .

Chapter 5 investigates the interaction between arteries and stents using finite element simulations, considering the effects of friction and stent surface shape. The friction effect is analysed by varying the coefficient of friction (COF) between 0.01 and 0.1. Additionally, the impact of surface shape is studied by simulating stents with small- and large-amplitude convex surfaces, as well as small-amplitude concave surfaces, using non-uniform displacement inputs. Key indicators including von Mises stress, maximum principal stress, maximum principal strain, strain energy density, and shear stress are used to evaluate the outcomes.

The results identify three key damage-prone areas: the contact interface, the right shoulder of the fibrous cap, and the lipid pool within the artery. Among these, the right shoulder of the fibrous cap exhibits the highest values for von Mises stress, maximum principal stress, maximum principal strain, and strain energy density, all of which are positively correlated with the lumen diameter. The friction effects are most significant at the contact interface, where shear stress is highly sensitive to changes in the coefficient of friction. An increase in the COF significantly amplifies the shear stress at this interface. Surface shape also influences the stress distributions. Stents with small-

amplitude convex surfaces (with $A = 0.01$ mm) lead to more favorable stress distributions, showing lower values of von Mises and shear stress. On the other hand, concave surfaces or large-amplitude convex surfaces result in higher von Mises stress and shear stress, making them less suitable for stent surface modifications. Based on these findings, it is recommended to ensure sufficient stent expansion to prevent arterial fractures and rupture of the lipid pool. Additionally, stent surfaces should be designed with appropriately shaped convex profiles to minimize stress concentrations. A low coefficient of friction should be maintained, and concave or excessively convex surfaces should be avoided to reduce the risk of arterial damage.

Bibliography

- [1] M. A. Masen, A. Chung, J. U. Dawczyk, *et al.*, “Evaluating lubricant performance to reduce covid-19 ppe-related skin injury,” *PLoS One*, vol. 15, no. 9, e0239363, 2020.
- [2] D. Antoniucci, R. Valenti, G. M. Santoro, *et al.*, “Restenosis after coronary stenting in current clinical practice,” *American Heart Journal*, vol. 135, no. 3, pp. 510–518, 1998.
- [3] G. D. Dangas, B. E. Claessen, A. Caixeta, E. A. Sanidas, G. S. Mintz, and R. Mehran, “In-stent restenosis in the drug-eluting stent era,” *Journal of the American College of Cardiology*, vol. 56, no. 23, pp. 1897–1907, 2010.
- [4] R. Jobanputra, C. Boyle, D. Dini, and M. Masen, “Modelling the effects of age-related morphological and mechanical skin changes on the stimulation of tactile mechanoreceptors,” *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 112, p. 104073, 2020.
- [5] R. D. Jobanputra, J. Hayes, S. Royyuru, and M. A. Masen, “A numerical analysis of skin–ppe interaction to prevent facial tissue injury,” *Scientific Reports*, vol. 11, no. 1, p. 16248, 2021.
- [6] K. K. Yap, M. Murali, Z. Tan, X. Zhou, L. Li, and M. A. Masen, “Wax-oil lubricants to reduce the shear between skin and ppe,” *Scientific Reports*, vol. 11, no. 1, p. 11537, 2021.
- [7] X. Zhou, M. A. Masen, Y. Y. Li, K. K. Yap, M. Murali, and Z. M. Jin, “Influence of different fluid environments on tactile perception and finger friction,” *Journal of the Royal Society Interface*, vol. 19, no. 188, p. 20210783, 2022.
- [8] G. Holzapfel, “Biomech preprint series biomechanics of soft tissue,” *Handbook of Materials Behavior Models*, 2000.
- [9] M. Freutel, H. Schmidt, L. Dürselen, A. Ignatius, and F. Galbusera, “Finite element modeling of soft tissues: Material models, tissue interaction and challenges,” *Clinical Biomechanics*, vol. 29, pp. 363–372, 4 2014.
- [10] C. T. McKee, J. A. Last, P. Russell, and C. J. Murphy, “Indentation versus tensile measurements of young’s modulus for soft biological tissues,” *Tissue Engineering Part B: Reviews*, vol. 17, pp. 155–164, 3 2011.

-
- [11] R. M. Wagner, R. Maiti, M. J. Carré, C. M. Perrault, P. C. Evans, and R. Lewis, “Biotribology of vascular devices: A review of tissue/device friction research,” *Biotribology*, vol. 25, 2021.
 - [12] L. A. Walton, R. S. Bradley, P. J. Withers, *et al.*, “Morphological characterisation of unstained and intact tissue micro-architecture by x-ray computed micro- and nano-tomography,” *Scientific Reports*, vol. 5, 2015.
 - [13] G. Limbert, “Constitutive modelling of skin mechanics,” *Skin Biophysics: From Experimental Characterisation to Advanced Modelling*, pp. 19–76, 2019.
 - [14] M. Venus, J. Waterman, and I. McNab, “Basic physiology of the skin,” *Surgery (Oxford)*, vol. 29, no. 10, pp. 471–474, 2011, Wound Management.
 - [15] E. C. Naylor, R. E. Watson, and M. J. Sherratt, “Molecular aspects of skin ageing,” *Maturitas*, vol. 69, pp. 249–256, 3 2011.
 - [16] W. L. Wong, T. J. Joyce, and K. L. Goh, “Resolving the viscoelasticity and anisotropy dependence of the mechanical properties of skin from a porcine model,” *Biomechanics and Modeling in Mechanobiology*, vol. 15, pp. 433–446, 2 2016.
 - [17] A. N. Annaidh, K. Bruyère, M. Destrade, M. D. Gilchrist, and M. Otténio, “Characterization of the anisotropic mechanical properties of excised human skin,” *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 5, no. 1, pp. 139–148, 2012.
 - [18] R. B. Groves, S. A. Coulman, J. C. Birchall, and S. L. Evans, “An anisotropic, hyperelastic model for skin: Experimental measurements, finite element modelling and identification of parameters for human and murine skin,” *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 18, pp. 167–180, 2013.
 - [19] C. Flynn, A. J. Taberner, P. M. Nielsen, and S. Fels, “Simulating the three-dimensional deformation of in vivo facial skin,” *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 28, pp. 484–494, 2013.
 - [20] C. Flynn, I. Stavness, J. Lloyd, and S. Fels, “A finite element model of the face including an orthotropic skin model under in vivo tension,” *Computer Methods in Biomechanics and Biomedical Engineering*, vol. 18, pp. 571–582, 6 2015.
 - [21] J. W. Jor, M. D. Parker, A. J. Taberner, M. P. Nash, and P. M. Nielsen, “Computational and experimental characterization of skin mechanics: Identifying current challenges and future directions,” *Wiley Interdisciplinary Reviews: Systems Biology and Medicine*, vol. 5, pp. 539–556, 5 2013.

-
- [22] M. G. Zhang, Y. P. Cao, G. Y. Li, and X. Q. Feng, "Spherical indentation method for determining the constitutive parameters of hyperelastic soft materials," *Biomechanics and Modeling in Mechanobiology*, vol. 13, pp. 1–11, 1 2014.
- [23] J. Z. Wu, R. G. Dong, W. P. Smutz, and A. W. Schopper, "Nonlinear and viscoelastic characteristics of skin under compression: Experiment and analysis," *Biomedical materials and engineering*, vol. 13, no. 4, pp. 373–385, 2003.
- [24] M. L. Crichton, B. C. Donose, X. Chen, A. P. Raphael, H. Huang, and M. A. Kendall, "The viscoelastic, hyperelastic and scale dependent behaviour of freshly excised individual skin layers," *Biomaterials*, vol. 32, no. 20, pp. 4670–4681, 2011.
- [25] S. Pan, D. Malhotra, and N. Germann, "Nonlinear viscoelastic properties of native male human skin and in vitro 3d reconstructed skin models under laos stress," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 96, pp. 310–323, 2019.
- [26] D. Cao, S. Malakooti, V. N. Kulkarni, Y. Ren, and H. Lu, "Nanoindentation measurement of core–skin interphase viscoelastic properties in a sandwich glass composite," *Mechanics of Time-Dependent Materials*, vol. 25, pp. 353–363, 3 2021.
- [27] J. D. Humphrey, "Mechanics of the arterial wall: Review and directions," *Critical ReviewsTM in Biomedical Engineering*, vol. 23, no. 1-2, 1995.
- [28] M. T. Duong, "Hyperelastic modeling and soft-tissue growth integrated with the smoothed finite element method-sfem," Ph.D. dissertation, Dissertation, RWTH Aachen University, 2014, 2015.
- [29] J. D. Hartman, "Structural changes within the media of coronary arteries related to intimal thickening," *The American Journal of Pathology*, vol. 89, no. 1, p. 13, 1977.
- [30] P. B. Canham, H. M. Finlay, J. G. Dixon, D. R. Boughner, and A. Chen, "Measurements from light and polarised light microscopy of human coronary arteries fixed at distending pressure," *Cardiovascular Research*, vol. 23, no. 11, pp. 973–982, 1989.
- [31] T. C. Gasser, R. W. Ogden, and G. A. Holzapfel, "Hyperelastic modelling of arterial layers with distributed collagen fibre orientations," *Journal of the Royal Society Interface*, vol. 3, no. 6, pp. 15–35, 2006.
- [32] G. A. Holzapfel, G. Sommer, M. Auer, P. Regitnig, and R. W. Ogden, "Layer-specific 3D residual deformations of human aortas with non-atherosclerotic intimal thickening," *Annals of Biomedical Engineering*, vol. 35, pp. 530–545, 2007.

-
- [33] G. A. Holzapfel, T. C. Gasser, and R. W. Ogden, "A new constitutive framework for arterial wall mechanics and a comparative study of material models," *Journal of Elasticity*, vol. 61, pp. 1–48, 2000.
- [34] H. M. Finlay, L. McCullough, and P. B. Canham, "Three-dimensional collagen organization of human brain arteries at different transmural pressures," *Journal of Vascular Research*, vol. 32, no. 5, pp. 301–312, 1995.
- [35] D. J. Patel, D. L. Fry, and J. S. Janicki, "The elastic symmetry of arterial segments in dogs," *Circulation Research*, vol. 24, no. 1, pp. 1–8, 1969.
- [36] M. R. Roach and S. Song, "Variations in strength of the porcine aorta as a function of location," *Clinical and Investigative Medicine*, vol. 17, no. 4, p. 308, 1994.
- [37] A. Milutinović, D. Šuput, and R. Zorc-Pleskovič, "Pathogenesis of atherosclerosis in the tunica intima, media, and adventitia of coronary arteries: An updated review," *Bosnian Journal of Basic Medical Sciences*, vol. 20, no. 1, p. 21, 2020.
- [38] J. C. Kohn, M. C. Lampi, and C. A. Reinhart-King, "Age-related vascular stiffening: Causes and consequences," *Frontiers in Genetics*, vol. 6, p. 112, 2015.
- [39] Y. C. Fung, *Biomechanics: mechanical properties of living tissues*. Springer Science & Business Media, 2013.
- [40] P. Martins, R. Natal Jorge, and A. Ferreira, "A comparative study of several material models for prediction of hyperelastic properties: Application to silicone-rubber and soft tissues," *Strain*, vol. 42, no. 3, pp. 135–147, 2006.
- [41] T. E. Carew, R. N. Vaishnav, and D. J. Patel, "Compressibility of the arterial wall," *Circulation Research*, vol. 23, no. 1, pp. 61–68, 1968.
- [42] J. Vossoughi, R. N. Vaishnav, and D. J. Patel, "Compressibility of the myocardial tissue," *Advances in Bioengineering*, pp. 45–48, 1980.
- [43] Y. Lanir, "Constitutive equations for fibrous connective tissues," *Journal of Biomechanics*, vol. 16, no. 1, pp. 1–12, 1983.
- [44] R. Fan and M. S. Sacks, "Simulation of planar soft tissues using a structural constitutive model: Finite element implementation and validation," *Journal of Biomechanics*, vol. 47, pp. 2043–2054, 9 2014.
- [45] L. R. G. Treloar, "The elasticity of a network of long-chain molecules.," *Transactions of the Faraday Society*, vol. 39, pp. 36–41, 1943.

-
- [46] M. Mooney, "A theory of large elastic deformation," *Journal of Applied Physics*, vol. 11, no. 9, pp. 582–592, 1940.
- [47] R. Rivlin, "Large elastic deformations of isotropic materials. v. the problem of flexure," *Proceedings of the Royal Society of London. Series A. Mathematical and Physical Sciences*, vol. 195, no. 1043, pp. 463–473, 1949.
- [48] R. W. Ogden, "Large deformation isotropic elasticity—on the correlation of theory and experiment for incompressible rubberlike solids," *A Mathematical, Physical and Engineering Sciences*, vol. 326.1567, pp. 565–584, 1972.
- [49] O. H. Yeoh, "Characterization of elastic properties of carbon-black-filled rubber vulcanizates," *Rubber Chemistry and Technology*, vol. 63, no. 5, pp. 792–805, 1990.
- [50] O. H. Yeoh, "Some forms of the strain energy function for rubber," *Rubber Chemistry and Technology*, vol. 66, no. 5, pp. 754–771, 1993.
- [51] O. H. Yeoh, "Hyperelastic material models for finite element analysis of rubber," *Journal of Natural Rubber Research*, vol. 12, no. 1997, pp. 142–153, 1997.
- [52] D. Veronda and R. Westmann, "Mechanical characterization of skin—finite deformations," *Journal of Biomechanics*, vol. 3, no. 1, pp. 111–124, 1970.
- [53] J. D. Humphrey and K. Rajagopal, "A constrained mixture model for growth and remodeling of soft tissues," *Mathematical Models and Methods in Applied Sciences*, vol. 12, no. 03, pp. 407–430, 2002.
- [54] Y. Lanir, "A structural theory for the homogeneous biaxial stress-strain relationships in flat collagenous tissues," *Journal of Biomechanics*, vol. 12, no. 6, pp. 423–436, 1979.
- [55] H. Tran, F. Charleux, M. Rachik, A. Ehrlicher, and M. Ho Ba Tho, "In vivo characterization of the mechanical properties of human skin derived from mri and indentation techniques," *Computer Methods in Biomechanics and Biomedical Engineering*, vol. 10, no. 6, pp. 401–407, 2007.
- [56] N. Gundiah, M. B. Ratcliffe, and L. A. Pruitt, "Determination of strain energy function for arterial elastin: Experiments using histology and mechanical tests," *Journal of Biomechanics*, vol. 40, no. 3, pp. 586–594, 2007.
- [57] A. Delalleau, G. Josse, J.-M. Lagarde, H. Zahouani, and J. Bergheau, "A nonlinear elastic behavior to identify the mechanical parameters of human skin in vivo," *Skin Research and Technology*, vol. 14, no. 2, pp. 152–164, 2008.

-
- [58] S. Barrett, M. Sutcliffe, S. Howarth, Z. Li, and J. Gillard, "Experimental measurement of the mechanical properties of carotid atherothrombotic plaque fibrous cap," *Journal of Biomechanics*, vol. 42, no. 11, pp. 1650–1655, 2009.
- [59] C.-K. Chai, A. C. Akyildiz, L. Speelman, *et al.*, "Local axial compressive mechanical properties of human carotid atherosclerotic plaques—characterisation by indentation test and inverse finite element analysis," *Journal of Biomechanics*, vol. 46, no. 10, pp. 1759–1766, 2013.
- [60] C.-K. Chai, L. Speelman, C. W. Oomens, and F. P. Baaijens, "Compressive mechanical properties of atherosclerotic plaques—indentation test to characterise the local anisotropic behaviour," *Journal of Biomechanics*, vol. 47, no. 4, pp. 784–792, 2014.
- [61] N. Elango, A. A. M. Faudzi, A. Hassan, and M. R. M. Rusydi, "Experimental investigations of skin-like material and computation of its material properties," *International Journal of Precision Engineering and Manufacturing*, vol. 15, pp. 1909–1914, 2014.
- [62] C. Wex, S. Arndt, A. Stoll, C. Bruns, and Y. Kupriyanova, "Isotropic incompressible hyperelastic models for modelling the mechanical behaviour of biological tissues: A review," *Biomedical Engineering/Biomedizinische Technik*, vol. 60, no. 6, pp. 577–592, 2015.
- [63] S. D. Łagan and A. Liber-Kneć, "Experimental testing and constitutive modeling of the mechanical properties of the swine skin tissue," *Acta of Bioengineering and Biomechanics*, vol. 19, pp. 93–102, 2 2017.
- [64] O. A. Shergold, N. A. Fleck, and D. Radford, "The uniaxial stress versus strain response of pig skin and silicone rubber at low and high strain rates," *International Journal of Impact Engineering*, vol. 32, no. 9, pp. 1384–1402, 2006.
- [65] A. Karimi, R. Faturechi, M. Navidbakhsh, and S. A. Hashemi, "A nonlinear hyperelastic behavior to identify the mechanical properties of rat skin under uniaxial loading," *Journal of Mechanics in Medicine and Biology*, vol. 14, no. 05, p. 1 450 075, 2014.
- [66] D. Tang, C. Yang, S. Kobayashi, J. Zheng, and R. P. Vito, "Effect of stenosis asymmetry on blood flow and artery compression: A three-dimensional fluid-structure interaction model," *Annals of Biomedical Engineering*, vol. 31, pp. 1182–1193, 2003.
- [67] C. Lally, A. Reid, and P. J. Prendergast, "Elastic behavior of porcine coronary artery tissue under uniaxial and equibiaxial tension," *Annals of Biomedical Engineering*, vol. 32, no. 10, pp. 1355–1364, 2004.

-
- [68] A. Valencia and F. Solis, "Blood flow dynamics and arterial wall interaction in a saccular aneurysm model of the basilar artery," *Computers & Structures*, vol. 84, no. 21, pp. 1326–1337, 2006.
- [69] M. H. Kural, M. Cai, D. Tang, T. Gwyther, J. Zheng, and K. L. Billiar, "Planar biaxial characterization of diseased human coronary and carotid arteries for computational modeling," *Journal of Biomechanics*, vol. 45, no. 5, pp. 790–798, 2012.
- [70] A. Karimi, M. Navidbakhsh, A. Shojaei, K. Hassani, and S. Faghihi, "Study of plaque vulnerability in coronary artery using mooney–rivlin model: A combination of finite element and experimental method," *Biomedical Engineering: Applications, Basis and Communications*, vol. 26, no. 01, p. 1450013, 2014.
- [71] M. Rackl, "Curve fitting for ogden, yeoh and polynomial models," in *ScilabTEC Conference 2015*, vol. 1, 2015, pp. 1–11.
- [72] J. J. Mulvihill and M. T. Walsh, "On the mechanical behaviour of carotid artery plaques: The influence of curve-fitting experimental data on numerical model results," *Biomechanics and Modeling in Mechanobiology*, vol. 12, pp. 975–985, 2013.
- [73] X. Kan, T. Ma, J. Lin, L. Wang, Z. Dong, and X. Y. Xu, "Patient-specific simulation of stent-graft deployment in type b aortic dissection: Model development and validation," *Biomechanics and Modeling in Mechanobiology*, vol. 20, pp. 2247–2258, 6 2021.
- [74] S. L. Evans and C. A. Holt, "Measuring the mechanical properties of human skin in vivo using digital image correlation and finite element modelling," *The Journal of Strain Analysis for Engineering Design*, vol. 44, no. 5, pp. 337–345, 2009.
- [75] C. Flynn, A. Taberner, and P. Nielsen, "Modelling the mechanical response of in vivo human skin under a rich set of deformations," *Annals of Biomedical Engineering*, vol. 39, pp. 1935–1946, 2011.
- [76] C. Flynn, A. J. Taberner, P. M. Nielsen, and S. Fels, "Simulating the three-dimensional deformation of in vivo facial skin," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 28, pp. 484–494, 2013.
- [77] Z.-Y. Li, T. Tang, M. Graves, M. Sutcliffe, J. H. Gillard, *et al.*, "Assessment of carotid plaque vulnerability using structural and geometrical determinants," *Circulation Journal*, vol. 72, no. 7, pp. 1092–1099, 2008.

-
- [78] H. Zahedmanesh and C. Lally, “Determination of the influence of stent strut thickness using the finite element method: Implications for vascular injury and in-stent restenosis,” *Medical and Biological Engineering and Computing*, vol. 47, pp. 385–393, 4 2009.
- [79] H. Zahedmanesh, D. J. Kelly, and C. Lally, “Simulation of a balloon expandable stent in a realistic coronary artery—determination of the optimum modelling strategy,” *Journal of Biomechanics*, vol. 43, pp. 2126–2132, 11 2010.
- [80] R. Araújo, T. A. Guimarães, and S. A. Oliveira, “An analysis of the contact between the stent and the artery using tube hydroforming simulation,” *International Journal for Numerical Methods in Biomedical Engineering*, vol. 29, pp. 1214–1222, 11 2013.
- [81] A. Schiavone and L. G. Zhao, “A study of balloon type, system constraint and artery constitutive model used in finite element simulation of stent deployment,” *Mechanics of Advanced Materials and Modern Processes*, vol. 1, 1 2015.
- [82] Y. Fung, “Elasticity of soft tissues in simple elongation,” *American Journal of Physiology-Legacy Content*, vol. 213, no. 6, pp. 1532–1544, 1967.
- [83] H. Demiray, “A note on the elasticity of soft biological tissues,” *Journal of Biomechanics*, vol. 5, no. 3, pp. 309–311, 1972.
- [84] C. Chuong and Y. Fung, “Three-dimensional stress distribution in arteries,” *Journal of Biomechanics*, vol. 105, no. 3, pp. 268–274, 1983.
- [85] C. O. Horgan and M. G. Smayda, “The importance of the second strain invariant in the constitutive modeling of elastomers and soft biomaterials,” *Mechanics of Materials*, vol. 51, pp. 43–52, 2012.
- [86] H. Demiray and R. P. Vito, “A layered cylindrical shell model for an aorta,” *International Journal of Engineering Science*, vol. 29, no. 1, pp. 47–54, 1991.
- [87] G. Y. Li, Q. He, R. Mangan, *et al.*, “Guided waves in pre-stressed hyperelastic plates and tubes: Application to the ultrasound elastography of thin-walled soft materials,” *Journal of the Mechanics and Physics of Solids*, vol. 102, pp. 67–79, 2017.
- [88] C. O. Horgan and G. Saccomandi, “Constitutive modelling of rubber-like and biological materials with limiting chain extensibility,” *Mathematics and Mechanics of Solids*, vol. 7, no. 4, pp. 353–371, 2002.
- [89] C. Horgan and G. Saccomandi, “A description of arterial wall mechanics using limiting chain extensibility constitutive models,” *Biomechanics and Modeling in Mechanobiology*, vol. 1, no. 4, pp. 251–266, 2003.

-
- [90] J. Bischoff, E. Arruda, and K. Grosh, "A microstructurally based orthotropic hyperelastic constitutive law," *Journal of Applied Mechanics*, vol. 69, no. 5, pp. 570–579, 2002.
- [91] J. E. Bischoff, E. M. Arruda, and K. Grosh, "A rheological network model for the continuum anisotropic and viscoelastic behavior of soft tissue," *Biomechanics and Modeling in Mechanobiology*, vol. 3, pp. 56–65, 2004.
- [92] S. Derler and L. Gerhardt, "Tribology of skin: Review and analysis of experimental results for the friction coefficient of human skin," *Tribology Letters*, vol. 45, pp. 1–27, 2012.
- [93] G. Limbert, "Mathematical and computational modelling of skin biophysics: A review," *Proceedings of the Royal Society A: Mathematical, Physical and Engineering Sciences*, vol. 473, no. 2203, p. 20170257, 2017.
- [94] E. M. Arruda and M. C. Boyce, "A three-dimensional constitutive model for the large stretch behavior of rubber elastic materials," *Journal of the Mechanics and Physics of Solids*, vol. 41, no. 2, pp. 389–412, 1993.
- [95] A. N. Gent, "A new constitutive relation for rubber," *Rubber Chemistry and Technology*, vol. 69, no. 1, pp. 59–61, 1996.
- [96] T. C. Gasser and G. A. Holzapfel, "Physical and numerical modeling of dissection propagation in arteries caused by balloon angioplasty," in *Proceedings of the 3rd IASTED International Conference on Biomechanics*, ACTA Press Anaheim, 2005.
- [97] G. A. Holzapfel, G. Sommer, C. T. Gasser, and P. Regitnig, "Determination of layer-specific mechanical properties of human coronary arteries with nonatherosclerotic intimal thickening and related constitutive modeling," *American Journal of Physiology-Heart and Circulatory Physiology*, vol. 289, pp. 2048–2058, 2005.
- [98] G. A. Holzapfel, M. Stadler, and T. C. Gasser, "Changes in the mechanical environment of stenotic arteries during interaction with stents: Computational assessment of parametric stent designs," *Journal of Biomechanical Engineering*, vol. 127, pp. 166–180, 1 2005.
- [99] T. C. Gasser and G. A. Holzapfel, "Modeling plaque fissuring and dissection during balloon angioplasty intervention," *Annals of Biomedical Engineering*, vol. 35, pp. 711–723, 2007.
- [100] A. Ní Annaidh, K. Bruyère, M. Destrade, *et al.*, "Automated estimation of collagen fibre dispersion in the dermis and its contribution to the anisotropic behaviour of skin," *Annals of Biomedical Engineering*, vol. 40, no. 8, pp. 1666–1678, 2012.

-
- [101] T. K. Tonge, L. S. Atlan, L. M. Voo, and T. D. Nguyen, "Full-field bulge test for planar anisotropic tissues: Part i-experimental methods applied to human skin tissue," *Acta Biomaterialia*, vol. 9, no. 4, pp. 5913–5925, 2013.
- [102] T. K. Tonge, L. M. Voo, and T. D. Nguyen, "Full-field bulge test for planar anisotropic tissues: Part II—a thin shell method for determining material parameters and comparison of two distributed fiber modeling approaches," *Acta Biomaterialia*, vol. 9, no. 4, pp. 5926–5942, 2013.
- [103] K. Takamizawa and K. Hayashi, "Strain energy density function and uniform strain hypothesis for arterial mechanics," *Journal of Biomechanics*, vol. 20, no. 1, pp. 7–17, 1987.
- [104] J. Humphrey and F. Yin, "A new constitutive formulation for characterizing the mechanical behavior of soft tissues," *Biophysical Journal*, vol. 52, no. 4, pp. 563–570, 1987.
- [105] Y. Kvistedal and P. Nielsen, "Estimating material parameters of human skin in vivo," *Biomechanics and Modeling in Mechanobiology*, vol. 8, no. 1, pp. 1–8, 2009.
- [106] S. Federico, A. Grillo, S. Imatani, G. Giaquinta, and W. Herzog, "An energetic approach to the analysis of anisotropic hyperelastic materials," *International Journal of Engineering Science*, vol. 46, no. 2, pp. 164–181, 2008.
- [107] J. P. Wilber and J. R. Walton, "The convexity properties of a class of constitutive models for biological soft issues," *Mathematics and Mechanics of Solids*, vol. 7, no. 3, pp. 217–235, 2002.
- [108] M. Itskov, A. Ehret, and D. Mavrilas, "A polyconvex anisotropic strain-energy function for soft collagenous tissues," *Biomechanics and Modeling in Mechanobiology*, vol. 5, no. 1, pp. 17–26, 2006.
- [109] R. Meijer, L. Douven, and C. Oomens, "Characterisation of anisotropic and non-linear behaviour of human skin in vivo," *Computer Methods in Biomechanics and Biomedical Engineering*, vol. 2, no. 1, pp. 13–27, 1999.
- [110] J. W. Jor, M. P. Nash, P. M. Nielsen, and P. J. Hunter, "Estimating material parameters of a structurally based constitutive relation for skin mechanics," *Biomechanics and Modeling in Mechanobiology*, vol. 10, pp. 767–778, 2011.
- [111] C. Flynn and B. A. McCormack, "Finite element modelling of forearm skin wrinkling," *Skin Research and Technology*, vol. 14, no. 3, pp. 261–269, 2008.
- [112] C. Flynn and B. A. McCormack, "A simplified model of scar contraction," *Journal of Biomechanics*, vol. 41, no. 7, pp. 1582–1589, 2008.

-
- [113] C. Flynn and B. A. McCormack, "Simulating the wrinkling and aging of skin with a multi-layer finite element model," *Journal of Biomechanics*, vol. 43, no. 3, pp. 442–448, 2010.
- [114] W. Zhang, G. Sommer, J. A. Niestrawska, G. A. Holzapfel, and D. Nordsletten, "The effects of viscoelasticity on residual strain in aortic soft tissues," *Acta Biomaterialia*, vol. 140, pp. 398–411, 2022, ISSN: 1742-7061.
- [115] H. Hertz, "The contact of elastic solids," *Journal for Pure and Applied Mathematics*, vol. 92, pp. 156–171, 1881.
- [116] H. Hertz, "Ueber die berührung fester elastischer körper.," *Journal für die reine und angewandte Mathematik (Crelles Journal)*, vol. 1882, pp. 156–171, 1882.
- [117] D. C. Lin, D. I. Shreiber, E. K. Dimitriadis, and F. Horkay, "Spherical indentation of soft matter beyond the hertzian regime: Numerical and experimental validation of hyperelastic models," *Biomechanics and Modeling in Mechanobiology*, vol. 8, pp. 345–358, 2009.
- [118] I. N. Sneddon, "The relation between load and penetration in the axisymmetric boussinesq problem for a punch of arbitrary profile," *International Journal of Engineering Science*, vol. 3, no. 1, pp. 47–57, 1965.
- [119] J. Jachowicz, R. McMullen, and D. Prettypaul, "Indentometric analysis of in vivo skin and comparison with artificial skin models," *Skin Research and Technology*, vol. 13, no. 3, pp. 299–309, 2007.
- [120] M. Kwiatkowska, S. Franklin, C. Hendriks, and K. Kwiatkowski, "Friction and deformation behaviour of human skin," *Wear*, vol. 267, no. 5-8, pp. 1264–1273, 2009.
- [121] S. Derler, G. Rotaru, W. Ke, *et al.*, "Microscopic contact area and friction between medical textiles and skin," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 38, pp. 114–125, 2014.
- [122] S. V. Kontomaris, "The hertz model in afm nanoindentation experiments: Applications in biological samples and biomaterials," *Micro and Nanosystems*, vol. 10, no. 1, pp. 11–22, 2018.
- [123] K. Takashima, R. Shimomura, T. Kitou, H. Terada, K. Yoshinaka, and K. Ikeuchi, "Contact and friction between catheter and blood vessel," *Tribology International*, vol. 40, pp. 319–328, 2 SPEC. ISS. 2007.
- [124] K. Hidaka, M. Nakamura, K. Osuga, H. Miyazaki, and S. Wada, "Elastic characteristics of microspherical embolic agents used for vascular interventional radiology," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 3, no. 7, pp. 497–503, 2010.

-
- [125] J. Peloquin, J. Huynh, R. M. Williams, and C. A. Reinhart-King, "Indentation measurements of the subendothelial matrix in bovine carotid arteries," *Journal of Biomechanics*, vol. 44, no. 5, pp. 815–821, 2011.
- [126] V. Castellano, R. Jackson, and M. Zabala, "Contact mechanics modeling of the semmes-weinstein monofilament on the plantar surface of the foot," *International Journal of Foot and Ankle*, vol. 5, no. 2, pp. 1–13, 2021.
- [127] W. C. Hayes, L. M. Keer, G. Herrmann, *et al.*, "A mathematical analysis for indentation tests of articular cartilage," *Journal of Biomechanics*, vol. 5, no. 5, pp. 541–551, 1972.
- [128] T. S. Wu and Y. P. Chiu, "On the contact problem of layered elastic bodies," *Quarterly of Applied Mathematics*, vol. 25, no. 3, pp. 233–242, 1967.
- [129] M. Abdou, "Fredholm–volterra integral equation of the first kind and contact problem," *Applied Mathematics and Computation*, vol. 125, no. 2-3, pp. 177–193, 2002.
- [130] M. Abdou, "On a symptic methods for fredholm–volterra integral equation of the second kind in contact problems," *Journal of Computational and Applied Mathematics*, vol. 154, no. 2, pp. 431–446, 2003.
- [131] M. Abdou, S. Raad, and S. Al-Hazmi, "Fundamental contact problem and singular mixed integral equation," *Life Sciences Journal*, vol. 11, no. 9, pp. 119–125, 2014.
- [132] E. K. Dimitriadis, F. Horkay, J. Maresca, *et al.*, "Determination of elastic moduli of thin layers of soft material using the atomic force microscope," *Biophysical Journal*, vol. 82, no. 5, pp. 2798–2810, 2002.
- [133] Y. Hu, E. P. Chan, J. J. Vlassak, and Z. Suo, "Poroelastic relaxation indentation of thin layers of gels," *Journal of Applied Physics*, vol. 110, no. 8, 2011.
- [134] D. Ebenstein and L. Pruitt, "Nanoindentation of soft hydrated materials for application to vascular tissues," *Journal of Biomedical Materials Research Part A: An Official Journal of The Society for Biomaterials, The Japanese Society for Biomaterials, and The Australian Society for Biomaterials and the Korean Society for Biomaterials*, vol. 69, no. 2, pp. 222–232, 2004.
- [135] A. Gouldstone, N. Chollacoop, M. Dao, J. Li, A. M. Minor, and Y. L. Shen, "Indentation across size scales and disciplines: Recent developments in experimentation and modeling," *Acta Materialia*, vol. 55, pp. 4015–4039, 12 2007.

-
- [136] D. Sicard, L. E. Fredenburgh, and D. J. Tschumperlin, "Measured pulmonary arterial tissue stiffness is highly sensitive to afm indenter dimensions," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 74, pp. 118–127, 2017.
- [137] C. Gamonpilas, M. N. Charalambides, J. G. Williams, P. J. Dooling, and S. R. Gibbon, "Predicting the mechanical behaviour of starch gels through inverse analysis of indentation data," *Applied Rheology*, vol. 20, pp. 1–9, 3 2010.
- [138] A. Delalleau, G. Josse, J.-M. Lagarde, H. Zahouani, and J.-M. Bergheau, "Characterization of the mechanical properties of skin by inverse analysis combined with the indentation test," *Journal of Biomechanics*, vol. 39, no. 9, pp. 1603–1610, 2006.
- [139] C. Pailler-Mattei and H. Zahouani, "Analysis of adhesive behaviour of human skin in vivo by an indentation test," *Tribology International*, vol. 39, no. 1, pp. 12–21, 2006.
- [140] K. Khaothong, "In vivo measurements of the mechanical properties of human skin and muscle by inverse finite element method combined with the indentation test," in *6th World Congress of Biomechanics (WCB 2010)*, In conjunction with 14th International Conference on Biomedical Engineering (ICBME) and 5th Asia Pacific Conference on Biomechanics (APBiomech), Singapore: Springer Berlin Heidelberg, 2010, pp. 1467–1470.
- [141] C. Pailler-Mattei, S. Pavan, R. Vargiolu, F. Pirot, F. Falson, and H. Zahouani, "Contribution of stratum corneum in determining bio-tribological properties of the human skin," *Wear*, vol. 263, no. 7-12, pp. 1038–1043, 2007.
- [142] C.-K. Chai, A. C. Akyildiz, L. Speelman, *et al.*, "Local anisotropic mechanical properties of human carotid atherosclerotic plaques—characterisation by micro-indentation and inverse finite element analysis," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 43, pp. 59–68, 2015.
- [143] Y. Yuan and R. Verma, "Measuring microelastic properties of stratum corneum," *Colloids and Surfaces B: Biointerfaces*, vol. 48, no. 1, pp. 6–12, 2006.
- [144] M. A. Cox, N. J. Driessen, C. V. Bouten, and F. P. Baaijens, "Mechanical characterization of anisotropic planar biological soft tissues using large indentation: A computational feasibility study," *Journal of Biomechanical Engineering*, vol. 128, no. 3, pp. 428–436, 2006.
- [145] M. Geerligs, L. Van Breemen, G. Peters, P. Ackermans, F. Baaijens, and C. Oomens, "In vitro indentation to determine the mechanical properties of epidermis," *Journal of Biomechanics*, vol. 44, no. 6, pp. 1176–1181, 2011.

-
- [146] S. Kucheyev, A. Hamza, J. Satcher Jr, and M. Worsley, "Depth-sensing indentation of low-density brittle nanoporous solids," *Acta Materialia*, vol. 57, no. 12, pp. 3472–3480, 2009.
- [147] Y. Zheng and A. F. Mak, "Effective elastic properties for lower limb soft tissues from manual indentation experiment," *IEEE Transactions on Rehabilitation Engineering*, vol. 7, no. 3, pp. 257–267, 1999.
- [148] A. Hemmasizadeh, K. Darvish, and M. Autieri, "Characterization of changes to the mechanical properties of arteries due to cold storage using nanoindentation tests," *Annals of biomedical engineering*, vol. 40, pp. 1434–1442, 2012.
- [149] G. Boyer, L. Laquière, A. Le Bot, S. Laquière, and H. Zahouani, "Dynamic indentation on human skin in vivo: Ageing effects," *Skin Research and Technology*, vol. 15, no. 1, pp. 55–67, 2009.
- [150] G. Boyer, H. Zahouani, A. Le Bot, and L. Laquiere, "In vivo characterization of viscoelastic properties of human skin using dynamic micro-indentation," in *2007 29th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, IEEE, 2007, pp. 4584–4587.
- [151] G. Boyer, C. P. Mattei, J. Molimard, M. Pericoi, S. Laquiere, and H. Zahouani, "Non contact method for in vivo assessment of skin mechanical properties for assessing effect of ageing," *Medical Engineering & Physics*, vol. 34, no. 2, pp. 172–178, 2012.
- [152] F. M. Weafer, S. Duffy, I. Machado, *et al.*, "Characterization of strut indentation during mechanical thrombectomy in acute ischemic stroke clot analogs," *Journal of neurointerventional surgery*, vol. 11, no. 9, pp. 891–897, 2019.
- [153] A. Choi and Y. P. Zheng, "Estimation of young's modulus and poisson's ratio of soft tissue from indentation using two different-sized indentors: Finite element analysis of the finite deformation effect," *Medical and Biological Engineering and Computing*, vol. 43, pp. 258–264, 2005.
- [154] C. Pailler-Mattei, S. Bec, and H. Zahouani, "In vivo measurements of the elastic mechanical properties of human skin by indentation tests," *Medical Engineering & Physics*, vol. 30, no. 5, pp. 599–606, 2008.
- [155] A. Rezvani-Sharif, M. Tafazzoli-Shadpour, and A. Avolio, "Progressive changes of elastic moduli of arterial wall and atherosclerotic plaque components during plaque development in human coronary arteries," *Medical & Biological Engineering & Computing*, vol. 57, pp. 731–740, 2019.

-
- [156] A. Giannakopoulos, “Elastic and viscoelastic indentation of flat surfaces by pyramid indentors,” *Journal of the Mechanics and Physics of Solids*, vol. 54, no. 7, pp. 1305–1332, 2006.
- [157] N. Sakharova, J. Fernandes, J. Antunes, and M. Oliveira, “Comparison between berkovich, vickers and conical indentation tests: A three-dimensional numerical simulation study,” *International Journal of Solids and Structures*, vol. 46, no. 5, pp. 1095–1104, 2009.
- [158] H. Olsson, K. J. Åström, C. C. D. Wit, M. Gäfvert, and P. Lischinsky, “Friction models and friction compensation,” *European Journal of Control*, vol. 4, no. 3, pp. 176–195, 1998.
- [159] V. L. Popov, *Contact Mechanics and Friction*. Berlin: Springer Berlin Heidelberg, 2010.
- [160] M. A. Masen, N. Veijgen, and M. Klaassen, “Experimental tribology of human skin,” *Skin Biophysics: From Experimental Characterisation to Advanced Modelling*, pp. 281–295, 2019.
- [161] A. R. Mijar and J. S. Arora, “Review of formulations for elastostatic frictional contact problems,” *Struct Multidisc Optim*, vol. 20, pp. 167–189, 2000.
- [162] S. N. Sanders, R. G. Lopata, L. C. van Breemen, F. N. van de Vosse, and M. C. Rutten, “A novel technique for the assessment of mechanical properties of vascular tissue,” *Biomechanics and Modeling in Mechanobiology*, vol. 19, pp. 1585–1594, 5 2020.
- [163] O. M. Braun and A. G. Naumovets, “Nanotribology: Microscopic mechanisms of friction,” *Surface Science Reports*, vol. 60, no. 6-7, pp. 79–158, 2006.
- [164] J. Kim, H.-H. Choi, S.-M. Hwang, Y. Kang, and B. Kang, “Comparison of implicit and explicit finite-element methods for the hydroforming process of an automobile lower arm,” *International Journal of Advanced Manufacturing Technology*, vol. 20, pp. 407–413, 2002.
- [165] F. P. Bowden and D. Tabor, *The Friction and Lubrication of Solids*. Oxford University Press, 2001.
- [166] B. Briscoe and D. Tabor, “The effect of pressure on the frictional properties of polymers,” *Wear*, vol. 34, no. 1, pp. 29–38, 1975.
- [167] M. J. Adams, B. J. Briscoe, and S. A. Johnson, “Friction and lubrication of human skin,” *Tribology Letters*, vol. 26, pp. 239–253, 2007.
- [168] G. Binnig and H. Rohrer, “Scanning tunneling microscopy—from birth to adolescence,” *Reviews of Modern Physics*, vol. 59, no. 3, p. 615, 1987.
- [169] D. Smith, G. Binnig, and C. Quate, “Atomic point-contact imaging,” *Applied Physics Letters*, vol. 49, no. 18, pp. 1166–1168, 1986.

-
- [170] C. M. Mate, G. M. McClelland, R. Erlandsson, and S. Chiang, "Atomic-scale friction of a tungsten tip on a graphite surface," *Scanning Tunneling Microscopy*, pp. 226–229, 1993.
- [171] F. Heslot, T. Baumberger, B. Perrin, B. Caroli, and C. Caroli, "Creep, stick-slip, and dry-friction dynamics: Experiments and a heuristic model," *Physical review E*, vol. 49, no. 6, p. 4973, 1994.
- [172] E. R. Clark, K. Hemmings, S. Greco, and et al., "Tribological characteristics of human vascular smooth muscle cells: The implication of disease state on friction," *Biotribology*, vol. 22, p. 100 122, 2020.
- [173] A. C. Dunn, T. D. Zaveri, B. G. Keselowsky, and W. G. Sawyer, "Macroscopic friction coefficient measurements on living endothelial cells," *Tribology Letters*, vol. 27, pp. 233–238, 2 2007.
- [174] J. van Kuilenburg, M. A. Masen, and E. van der Heide, "A review of fingerpad contact mechanics and friction and how this affects tactile perception," *Proceedings of the Institution of Mechanical Engineers, Part J: Journal of engineering tribology*, vol. 229, no. 3, pp. 243–258, 2015.
- [175] A. B. Cua, K.-P. Wilhelm, and H. Maibach, "Frictional properties of human skin: Relation to age, sex and anatomical region, stratum corneum hydration and transepidermal water loss," *British Journal of Dermatology*, vol. 123, no. 4, pp. 473–479, 1990.
- [176] L.-C. Gerhardt, V. Strässle, A. Lenz, N. D. Spencer, and S. Derler, "Influence of epidermal hydration on the friction of human skin against textiles," *Journal of the Royal Society Interface*, vol. 5, no. 28, pp. 1317–1328, 2008.
- [177] C. Hendriks and S. Franklin, "Influence of surface roughness, material and climate conditions on the friction of human skin," *Tribology Letters*, vol. 37, pp. 361–373, 2010.
- [178] T. André, P. Lefèvre, and J.-L. Thonnard, "A continuous measure of fingertip friction during precision grip," *Journal of Neuroscience Methods*, vol. 179, no. 2, pp. 224–229, 2009.
- [179] S. Tomlinson, R. Lewis, X. Liu, C. Texier, and M. Carré, "Understanding the friction mechanisms between the human finger and flat contacting surfaces in moist conditions," *Tribology Letters*, vol. 41, pp. 283–294, 2011.
- [180] O. Dinc, C. Ettles, S. Calabrese, and H. Scarton, "Some parameters affecting tactile friction," *ASME Journal of Tribology*, 1991.

-
- [181] S. Comaish and E. Bottoms, "The skin and friction: Deviations from amonton's laws, and the effects of hydration and lubrication," *British Journal of Dermatology*, vol. 84, no. 1, pp. 37–43, 1971.
- [182] S. Derler, L.-C. Gerhardt, A. Lenz, E. Bertaux, and M. Hadad, "Friction of human skin against smooth and rough glass as a function of the contact pressure," *Tribology International*, vol. 42, no. 11-12, pp. 1565–1574, 2009.
- [183] N. K. Veijgen, "Skin friction: A novel approach to measuring in vivo human skin," PhD Thesis, Ph.D. dissertation, Faculty of Engineering Technology, University of Twente, 2013.
- [184] C. Pailler-Mattei, S. Nicoli, F. Pirot, R. Vargiolu, and H. Zahouani, "A new approach to describe the skin surface physical properties in vivo," *Colloids and Surfaces B: Biointerfaces*, vol. 68, no. 2, pp. 200–206, 2009.
- [185] A. Gupta, B. Haldar, and M. Bhattacharya, "A simple device for measuring skin friction," *Indian Journal of Dermatology*, vol. 40, no. 3, pp. 116–121, 1995.
- [186] M. Egawa, M. Oguri, T. Hirao, M. Takahashi, and M. Miyakawa, "The evaluation of skin friction using afriictional feel analyzer," *Skin Research and Technology*, vol. 8, no. 1, pp. 41–51, 2002.
- [187] K. Nakajima and H. Narasaka, "Evaluation of skin surface associated with morphology and coefficient of friction," *International Journal of Cosmetic Science*, vol. 15, no. 4, pp. 135–151, 1993.
- [188] L. Skedung, K. Danerlöv, U. Olofsson, *et al.*, "Finger friction measurements on coated and uncoated printing papers," *Tribology Letters*, vol. 37, pp. 389–399, 2010.
- [189] M. A. Masen, "A systems based experimental approach to tactile friction," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 4, no. 8, pp. 1620–1626, 2011.
- [190] S. Tomlinson, R. Lewis, and M. Carré, "The effect of normal force and roughness on friction in human finger contact," *Wear*, vol. 267, no. 5-8, pp. 1311–1318, 2009.
- [191] S. Derler, R. Huber, H.-P. Feuz, and M. Hadad, "Influence of surface microstructure on the sliding friction of plantar skin against hard substrates," *Wear*, vol. 267, no. 5-8, pp. 1281–1288, 2009.
- [192] R. K. Sivamani, G. C. Wu, N. V. Gitis, and H. I. Maibach, "Tribological testing of skin products: Gender, age, and ethnicity on the volar forearm," *Skin Research and Technology*, vol. 9, no. 4, pp. 299–305, 2003.

-
- [193] Y. Zhu, S. Song, W. Luo, P. Elias, and M. Man, "Characterization of skin friction coefficient, and relationship to stratum corneum hydration in a normal chinese population," *Skin Pharmacology and Physiology*, vol. 24, no. 2, pp. 81–86, 2011.
- [194] J. van Kuilenburg, M. A. Masen, M. Groenendijk, V. v. Bana, and E. van der Heide, "An experimental study on the relation between surface texture and tactile friction," *Tribology International*, vol. 48, pp. 15–21, 2012.
- [195] J. van Kuilenburg, M. A. Masen, and E. van der Heide, "The role of the skin microrelief in the contact behaviour of human skin: Contact between the human finger and regular surface textures," *Tribology International*, vol. 65, pp. 81–90, 2013.
- [196] R. Fagiani, F. Massi, E. Chatelet, J. P. Costes, and Y. Berthier, "Contact of a finger on rigid surfaces and textiles: Friction coefficient and induced vibrations," *Tribology Letters*, vol. 48, pp. 145–158, 2012.
- [197] T. Sasada, "The friction of human skin," in *Proceedings of the 21st Biotribology Symposium, JSME*, Fukuoka, 2000, pp. 46–52.
- [198] X. Liu, Z. Lu, R. Lewis, M. Carré, and S. Matcher, "Feasibility of using optical coherence tomography to study the influence of skin structure on finger friction," *Tribology International*, vol. 63, pp. 34–44, 2013.
- [199] A. V. Savescu, M. L. Latash, and V. M. Zatsiorsky, "A technique to determine friction at the fingertips," *Journal of Applied Biomechanics*, vol. 24, no. 1, pp. 43–50, 2008.
- [200] N. J. Seo and T. J. Armstrong, "Friction coefficients in a longitudinal direction between the finger pad and selected materials for different normal forces and curvatures," *Ergonomics*, vol. 52, no. 5, pp. 609–616, 2009.
- [201] H.-Y. Han, A. Shimada, and S. Kawamura, "Analysis of friction on human fingers and design of artificial fingers," in *Proceedings of IEEE International Conference on Robotics and Automation*, IEEE, vol. 4, 1996, pp. 3061–3066.
- [202] R. A. Caldwell, J. E. Woodell, S. P. Ho, and et al., "In vitro evaluation of phosphonylated low-density polyethylene for vascular applications," *Journal of Biomedical Materials Research: An Official Journal of The Society for Biomaterials, The Japanese Society for Biomaterials, and The Australian Society for Biomaterials and the Korean Society for Bio-* materials, vol. 62, no. 4, pp. 514–524, 2002.

-
- [203] T. Chen, M. Lancaster, D. S. Lin, M. G. Doyle, T. L. Forbes, and C. H. Amon, "Measurement of frictional properties of aortic stent grafts and their delivery systems," *Journal of Medical Devices, Transactions of the ASME*, vol. 13, 2 2019.
- [204] J. Nelson, D. Patel, H. J. Sant, J. Shea, B. K. Gale, and J. Agarwal, "Compression of the vascular wall to create a friction fit in a vascular anastomotic coupler," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 123, 2021.
- [205] S. Vad, A. Eskinazi, T. Corbett, T. McGloughlin, and J. P. V. Geest, "Determination of coefficient of friction for self-expanding stent-grafts," *Journal of Biomechanical Engineering*, vol. 132, 12 2010.
- [206] Y. Kwak, W. Son, B.-J. Kim, *et al.*, "Frictional force analysis of stent retriever devices using a realistic vascular model: Pilot study," *Frontiers in Neurology*, vol. 13, p. 964354, 2022.
- [207] D. Balint, *Computational continuum mechanics*, Department of Mechanical Engineering, Imperial College London, Course materials, 2019.
- [208] Dassault Systèmes, *Abaqus user's manual, version 2024*, 2021.
- [209] M. A. Cox, N. J. Driessen, R. A. Boerboom, C. V. Bouten, and F. P. Baaijens, "Mechanical characterization of anisotropic planar biological soft tissues using finite indentation: Experimental feasibility," *Journal of Biomechanics*, vol. 41, no. 2, pp. 422–429, 2008.
- [210] N. Fougerson, P. Y. Rohan, D. Haering, J. L. Rose, X. Bonnet, and H. Pillet, "Combining freehand ultrasound-based indentation and inverse finite element modeling for the identification of hyperelastic material properties of thigh soft tissues," *Journal of Biomechanical Engineering*, vol. 142, 9 2020.
- [211] M. Akbarzadeh Khorshidi, S. Bose, B. Watschke, E. Mareena, and C. Lally, "Characterisation of human penile tissue properties using experimental testing combined with multi-target inverse finite element modelling," *Acta Biomaterialia*, 2024.
- [212] A. Dai, S. Wang, L. Zhou, H. Wei, Z. Wang, and W. He, "In vivo mechanical characterization of human facial skin combining curved surface imaging and indentation techniques," *Skin Research and Technology*, vol. 25, no. 2, pp. 142–149, 2019.
- [213] E. Dintwa, E. Tijskens, and H. Ramon, "On the accuracy of the hertz model to describe the normal contact of soft elastic spheres," *Granular Matter*, vol. 10, pp. 209–221, 2008.

-
- [214] J. C. Wei, G. A. Edwards, D. J. Martin, H. Huang, M. L. Crichton, and M. A. Kendall, "Allometric scaling of skin thickness, elasticity, viscoelasticity to mass for micro-medical device translation: From mice, rats, rabbits, pigs to humans," *Scientific Reports*, vol. 7, no. 1, p. 15 885, 2017.
- [215] Z. Guo, Q. Lyu, L. Jiang, Y. Chen, L. Dong, and H. Zhang, "Contact model for incompressible neo-hookean materials under finite spherical indentation," *Journal of Applied Mechanics*, vol. 87, 5 2020.
- [216] C. L. Slaboch, M. S. Alber, E. D. Rosen, and T. C. Ovaert, "Mechano-rheological properties of the murine thrombus determined via nanoindentation and finite element modeling," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 10, pp. 75–86, 2012.
- [217] C. Valero, B. Navarro, D. Navajas, and J. García-Aznar, "Finite element simulation for the mechanical characterization of soft biological materials by atomic force microscopy," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 62, pp. 222–235, 2016.
- [218] L. Li and M. Masen, "A new method for determining the ogden parameters of soft materials using indentation experiments.," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 155, p. 106 574, 2024.
- [219] D. R. Nolan, A. L. Gower, M. Destrade, R. W. Ogden, and J. McGarry, "A robust anisotropic hyperelastic formulation for the modelling of soft tissue," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 39, pp. 48–60, 2014.
- [220] G. A. Holzapfel *et al.*, "Biomechanics of soft tissue," *The Handbook of Materials Behavior Models*, vol. 3, no. 1, pp. 1049–1063, 2001.
- [221] J. Lim, J. Hong, W. W. Chen, and T. Weerasooriya, "Mechanical response of pig skin under dynamic tensile loading," *International Journal of Impact Engineering*, vol. 38, no. 2-3, pp. 130–135, 2011.
- [222] A. Kalra, A. Lowe, and A. M. Al-Jumaily, "Mechanical behaviour of skin: A review," *Journal of Materials Science and Engineering*, vol. 5, no. 4, p. 1 000 254, 2016.
- [223] T. Virén, J. Iivarinen, J. Sarin, I. Harvima, and H. Mayrovitz, "Accuracy and reliability of a hand-held in vivo skin indentation device to assess skin elasticity," *International Journal of Cosmetic Science*, vol. 40, no. 2, pp. 134–140, 2018.
- [224] S. Bec, A. Tonck, J.-M. Georges, E. Georges, and J.-L. Loubet, "Improvements in the indentation method with a surface force apparatus," *Philosophical Magazine A*, vol. 74, no. 5, pp. 1061–1072, 1996.

-
- [225] A. Rar, H. Song, and G. Pharr, "Assessment of new relation for the elastic compliance of a film-substrate system," *MRS Online Proceedings Library (OPL)*, vol. 695, pp. L10–10, 2001.
- [226] R. Avazmohammadi, D. S. Li, T. Leahy, *et al.*, "An integrated inverse model-experimental approach to determine soft tissue three-dimensional constitutive parameters: Application to post-infarcted myocardium," *Biomechanics and Modeling in Mechanobiology*, vol. 17, pp. 31–53, 2018.
- [227] B. Pierrat, D. MacManus, J. Murphy, and M. Gilchrist, "Indentation of heterogeneous soft tissue: Local constitutive parameter mapping using an inverse method and an automated rig," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 78, pp. 515–528, 2018.
- [228] D. Tabor, "A simple theory of static and dynamic hardness," *Proceedings of the Royal Society of London. Series A. Mathematical and Physical Sciences*, vol. 192, no. 1029, pp. 247–274, 1948.
- [229] D. Tabor, *The hardness of metals*. Oxford University press, 1951.
- [230] K. L. Johnson, *Contact mechanics*. Cambridge university press, 1987.
- [231] D. C. Lin, E. K. Dimitriadis, and F. Horkay, "Elasticity of rubber-like materials measured by afm nanoindentation," *Express Polymer Letters*, vol. 1, pp. 576–584, 9 2007.
- [232] B. Briscoe, "Isolated contact stress deformations of polymers: The basis for interpreting polymer tribology," *Tribology International*, vol. 31, no. 1-3, pp. 121–126, 1998.
- [233] M. Garcia, K. D. Schulze, C. S. O'Bryan, T. Bhattacharjee, W. G. Sawyer, and T. E. Angelini, "Eliminating the surface location from soft matter contact mechanics measurements," *Tribology-Materials, Surfaces & Interfaces*, vol. 11, no. 4, pp. 187–192, 2017.
- [234] R. Hill, "On constitutive inequalities for simple materials—I," *Journal of the Mechanics and Physics of Solids*, vol. 16, no. 4, pp. 229–242, 1968.
- [235] R. Hill, "Constitutive inequalities for isotropic elastic solids under finite strain," *Proceedings of the Royal Society of London. A. Mathematical and Physical Sciences*, vol. 314, no. 1519, pp. 457–472, 1970.
- [236] L. Li, L. Gao, K. K. Yap, A. Phinikaridou, and M. Masen, "Characterization of mouse artery tissue properties using experimental testing combined with finite element modelling," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 166, p. 106953, 2025, ISSN: 1751-6161.

-
- [237] F. Liu, C. M. Haeger, P. B. Dieffenbach, *et al.*, “Distal vessel stiffening is an early and pivotal mechanobiological regulator of vascular remodeling and pulmonary hypertension,” *JCI Insight*, vol. 1, 8 2016.
- [238] F. Savvopoulos, M. C. Keeling, D. Carassiti, *et al.*, “Assessment of the nano-mechanical properties of healthy and atherosclerotic coronary arteries by atomic force microscopy,” *Journal of the Royal Society Interface*, vol. 21, 211 Feb. 2024, ISSN: 17425662.
- [239] E. H. Yoffe, “Modified hertz theory for spherical indentation,” *Philosophical Magazine A: Physics of Condensed Matter, Structure, Defects and Mechanical Properties*, vol. 50, pp. 813–828, 6 1984.
- [240] C. E. Wu, K. H. Lin, and J. Y. Juang, “Hertzian load-displacement relation holds for spherical indentation on soft elastic solids undergoing large deformations,” *Tribology International*, vol. 97, pp. 71–76, 2016.
- [241] R. Long, M. S. Hall, M. Wu, and C.-Y. Hui, “Effects of gel thickness on microscopic indentation measurements of gel modulus,” *Biophysical Journal*, vol. 101, no. 3, pp. 643–650, 2011.
- [242] A. R. Harris and G. Charras, “Experimental validation of atomic force microscopy-based cell elasticity measurements,” *Nanotechnology*, vol. 22, no. 34, p. 345 102, 2011.
- [243] C. Johnson and A. Dunn, “Tribological characterization of gradient-density polyacrylamide hydrogel surfaces,” *Experimental Mechanics*, vol. 61, pp. 829–842, 2021.
- [244] P. L. Chandran, E. K. Dimitriadis, E. L. Mertz, and F. Horkay, “Microscale mapping of extracellular matrix elasticity of mouse joint cartilage: An approach to extracting bulk elasticity of soft matter with surface roughness,” *Soft Matter*, vol. 14, pp. 2879–2892, 15 2018.
- [245] T. Yuan, W. Zhan, and D. Dini, “Linking fluid-axons interactions to the macroscopic fluid transport properties of the brain,” *Acta Biomater.*, vol. 160, pp. 152–163, Apr. 2023, ISSN: 1742-7061.
- [246] D. He, D. Malu, and Y. Hu, “A comprehensive review of indentation of gels and soft biological materials,” *Applied Mechanics Reviews*, vol. 76, 5 2024.
- [247] H. Miyazaki and K. Hayashi, “Atomic force microscopic measurement of the mechanical properties of intact endothelial cells in fresh arteries,” *Medical & Biological Engineering & Computing*, vol. 37, pp. 530–536, 1999.

-
- [248] Q. Zhang and Q.-S. Yang, "Effects of large deformation and material nonlinearity on spherical indentation of hyperelastic soft materials," *Mechanics Research Communications*, vol. 84, pp. 55–59, 2017.
- [249] A. M. Dagro and K. Ramesh, "Nonlinear contact mechanics for the indentation of hyperelastic cylindrical bodies," *Mechanics of Soft Materials*, vol. 1, no. 1, p. 7, 2019.
- [250] Y. Du, P. Stewart, N. A. Hill, *et al.*, "Nonlinear indentation of second-order hyperelastic materials," *Journal of the Mechanics and Physics of Solids*, vol. 171, p. 105 139, 2023.
- [251] F. Meng, I. Mambetsariev, Y. Tian, *et al.*, "Attenuation of lipopolysaccharide-induced lung vascular stiffening by lipoxin reduces lung inflammation," *American Journal of Respiratory Cell and Molecular Biology*, vol. 52, pp. 152–161, 2 2015.
- [252] E. Melo, N. Cárdenes, E. Garreta, *et al.*, "Inhomogeneity of local stiffness in the extracellular matrix scaffold of fibrotic mouse lungs," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 37, pp. 186–195, 2014.
- [253] A. J. Engler, L. Richert, J. Y. Wong, C. Picart, and D. E. Discher, "Surface probe measurements of the elasticity of sectioned tissue, thin gels and polyelectrolyte multilayer films: Correlations between substrate stiffness and cell adhesion," *Surface Science*, vol. 570, no. 1-2, pp. 142–154, 2004.
- [254] M. S. Chen, J. M. John, D. P. Chew, D. S. Lee, S. G. Ellis, and D. L. Bhatt, "Bare metal stent restenosis is not a benign clinical entity," *American heart journal*, vol. 151, no. 6, pp. 1260–1264, 2006.
- [255] B. Doyle, C. S. Rihal, C. J. O'Sullivan, *et al.*, "Outcomes of stent thrombosis and restenosis during extended follow-up of patients treated with bare-metal coronary stents," *Circulation*, vol. 116, no. 21, pp. 2391–2398, 2007.
- [256] C. Rogers and E. R. Edelman, "Endovascular stent design dictates experimental restenosis and thrombosis," *Circulation*, vol. 91, no. 12, pp. 2995–3001, 1995.
- [257] H. Hara, M. Nakamura, J. C. Palmaz, and R. S. Schwartz, "Role of stent design and coatings on restenosis and thrombosis," *Advanced drug delivery reviews*, vol. 58, no. 3, pp. 377–386, 2006.
- [258] D. Martin and F. J. Boyle, "Computational structural modelling of coronary stent deployment: A review," *Computer Methods in Biomechanics and Biomedical Engineering*, vol. 14, no. 04, pp. 331–348, 2011.

-
- [259] M. S. Kim and L. S. Dean, "In-stent restenosis," *Cardiovascular therapeutics*, vol. 29, no. 3, pp. 190–198, 2011.
- [260] J. Aoki and K. Tanabe, "Mechanisms of drug-eluting stent restenosis," *Cardiovascular intervention and therapeutics*, vol. 36, pp. 23–29, 2021.
- [261] J. Clare, J. Ganly, C. A. Bursill, H. Sumer, P. Kingshott, and J. B. de Haan, "The mechanisms of restenosis and relevance to next generation stent design," *Biomolecules*, vol. 12, no. 3, p. 430, 2022.
- [262] L. Gruberg, M. K. Hong, G. S. Mintz, *et al.*, "Optimally deployed stents in the treatment of restenotic versus de novo lesions," *The American journal of cardiology*, vol. 85, no. 3, pp. 333–337, 2000.
- [263] A. Kraitzer, Y. Kloog, and M. Zilberman, "Approaches for prevention of restenosis," *Journal of Biomedical Materials Research Part B: Applied Biomaterials: An Official Journal of The Society for Biomaterials, The Japanese Society for Biomaterials, and The Australian Society for Biomaterials and the Korean Society for Biomaterials*, vol. 85, no. 2, pp. 583–603, 2008.
- [264] F. Alfonso, J. Coughlan, D. Giacoppo, A. Kastrati, and R. A. Byrne, "Management of in-stent restenosis," *EuroIntervention*, vol. 18, no. 2, e103, 2022.
- [265] G. Giustino, A. Colombo, A. Camaj, *et al.*, "Coronary in-stent restenosis: Jacc state-of-the-art review," *Journal of the American College of Cardiology*, vol. 80, no. 4, pp. 348–372, 2022.
- [266] I. Shimizu, A. Wada, and M. Sasaki, "A study on designing balloon expandable magnesium alloy stent for optimization of mechanical characteristics," vol. 2, no. 8, p. 523, 2018.
- [267] C. Conway, F. Sharif, J. P. McGarry, and P. E. McHugh, "A computational test-bed to assess coronary stent implantation mechanics using a population-specific approach," *Cardiovascular Engineering and Technology*, vol. 3, pp. 374–387, 4 2012.
- [268] M. I. Okereke, R. Khalaj, A. G. Tabriz, U. Nandi, N. Scoutaris, and D. Douroumis, "Development of 3d printable bioresorbable drug eluting coronary stents: An experimental and computational investigation," *Journal of Drug Delivery Science and Technology*, vol. 79, 2023.
- [269] R. G. Pauck and B. D. Reddy, "Computational analysis of the radial mechanical performance of plla coronary artery stents," *Medical Engineering & Physics*, vol. 37, pp. 7–12, 1 2015.

-
- [270] B. Hoddy, N. Ahmed, K. Al-Lamee, N. Bullett, N. Curzen, and N. W. Bressloff, "Investigating the material modelling of a polymeric bioresorbable scaffold via in-silico and in-vitro testing," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 120, 2021.
- [271] F. AURICCHIO*, M. Di Loreto, and E. Sacco, "Finite-element analysis of a stenotic artery revascularization through a stent insertion," *Computer Methods in Biomechanics and Biomedical Engineering*, vol. 4, no. 3, pp. 249–263, 2001.
- [272] J. L. Berry, E. Manoach, C. Mekkaoui, P. H. Rolland, J. Moore Jr, and A. Rachev, "Hemodynamics and wall mechanics of a compliance matching stent: In vitro and in vivo analysis," *Journal of Vascular and Interventional Radiology*, vol. 13, no. 1, pp. 97–105, 2002.
- [273] G. C. Cheng, H. M. Loree, R. D. Kamm, M. C. Fishbein, and R. T. Lee, "Distribution of circumferential stress in ruptured and stable atherosclerotic lesions. a structural analysis with histopathological correlation.," *Circulation*, vol. 87, no. 4, pp. 1179–1187, 1993.
- [274] D. E. Kiousis, T. C. Gasser, and G. A. Holzapfel, "A numerical model to study the interaction of vascular stents with human atherosclerotic lesions," *Annals of Biomedical Engineering*, vol. 35, pp. 1857–1869, 11 2007.
- [275] G. A. Holzapfel and R. W. Ogden, "'constitutive modelling of arteries'," *Proceedings of the Royal Society A: Mathematical, Physical and Engineering Sciences*, vol. 466, no. 2118, pp. 1551–1597, 2010.
- [276] F. Auricchio, M. Conti, M. D. Beule, G. D. Santis, and B. Verhegghe, "Carotid artery stenting simulation: From patient-specific images to finite element analysis," *Medical Engineering and Physics*, vol. 33, pp. 281–289, 3 2011.
- [277] S. Zhao, L. Gu, and S. R. Froemming, "Performance of self-expanding nitinol stent in a curved artery: Impact of stent length and deployment orientation," *Journal of Biomechanical Engineering*, vol. 134, 7 2012.
- [278] I. Pericevic, C. Lally, D. Toner, and D. J. Kelly, "The influence of plaque composition on underlying arterial wall stress during stent expansion: The case for lesion-specific stents," *Medical Engineering and Physics*, vol. 31, pp. 428–433, 4 2009.
- [279] F. J. Gijssen, F. Migliavacca, S. Schievano, *et al.*, "Simulation of stent deployment in a realistic human coronary artery," *Biomedical Engineering Online*, vol. 7, pp. 1–11, 2008.
- [280] M. Azaouzi, N. Lebaal, A. Makradi, and S. Belouettar, "Optimization based simulation of self-expanding nitinol stent," *Materials and Design*, vol. 50, pp. 917–928, 2013.

-
- [281] D. Mohan and J. W. Melvin, "Failure properties of passive human aortic tissue. i—uniaxial tension tests," *Journal of Biomechanics*, vol. 15, no. 11, pp. 887–902, 1982.
- [282] C. Lendon, M. Davies, P. Richardson, and G. Born, "Testing of small connective tissue specimens for the determination of the mechanical behaviour of atherosclerotic plaques," *Journal of Biomedical Engineering*, vol. 15, no. 1, pp. 27–33, 1993.
- [283] L. Cardoso and S. Weinbaum, "Changing views of the biomechanics of vulnerable plaque rupture: A review," *Annals of Biomedical Engineering*, vol. 42, pp. 415–431, 2014.
- [284] G. A. Holzapfel, J. J. Mulvihill, E. M. Cunnane, and M. T. Walsh, "Computational approaches for analyzing the mechanics of atherosclerotic plaques: A review," *Journal of Biomechanics*, vol. 47, no. 4, pp. 859–869, 2014.
- [285] M. Walsh, E. Cunnane, J. Mulvihill, A. Akyildiz, F. Gijssen, and G. A. Holzapfel, "Uniaxial tensile testing approaches for characterisation of atherosclerotic plaques," *Journal of Biomechanics*, vol. 47, no. 4, pp. 793–804, 2014.
- [286] F. Nematzadeh and S. K. Sadrnezhaad, "Effects of material properties on mechanical performance of nitinol stent designed for femoral artery: Finite element analysis," *Scientia Iranica*, vol. 19, pp. 1564–1571, 6 2012.
- [287] F. Nematzadeh, "Computational evaluation of the impact of friction coefficient on self-expanding stent and peripheral artery during contact," *Journal of Environmental Friendly Materials*, vol. 2, no. 2, pp. 29–35, 2018.
- [288] L. Gu, S. Zhao, and S. R. Froemming, "Arterial wall mechanics and clinical implications after coronary stenting: Comparisons of three stent designs," *International Journal of Applied Mechanics*, vol. 4, 2 2012.
- [289] B. Istrate, D. Mareci, C. Munteanu, *et al.*, "In vitro electrochemical properties of biodegradable ysz-coated mgca alloy," *Environmental Engineering & Management Journal (EEMJ)*, vol. 15, no. 5, 2016.
- [290] C. H. Tan, N. Muhamad, and M. M. Abdullah, "Surface topographical modification of coronary stent: A review," in *IOP Conference Series: Materials Science and Engineering*, vol. 209, Institute of Physics Publishing, 2017.
- [291] T. Y. Qiu, M. Song, and L. G. Zhao, "A computational study of crimping and expansion of bioresorbable polymeric stents," *Mechanics of Time-Dependent Materials*, vol. 22, pp. 273–290, 2 2018.

-
- [292] N. Filipovic, D. Nikolic, V. Isailovic, *et al.*, “In vitro and in silico testing of partially and fully bioresorbable vascular scaffold,” *Journal of Biomechanics*, vol. 115, p. 110 158, 2021.
- [293] M. Paggi and J. Reinoso, “A variational approach with embedded roughness for adhesive contact problems,” *Mechanics of Advanced Materials and Structures*, vol. 27, pp. 1731–1747, 20 2020.
- [294] D. Tang, C. Yang, S. Kobayashi, and D. N. Ku, “Effect of a lipid pool on stress/strain distributions in stenotic arteries: 3-d fluid-structure interactions (FSI) models,” *Journal of Biomechanical Engineering*, vol. 126, pp. 363–370, 3 2004.
- [295] V. A. Yastrebov, J. Durand, H. Proudhon, and G. Cailletaud, *Rough surface contact analysis by means of the finite element method and of a new reduced model*, 2011.
- [296] M. B. Amor, S. Belghith, and S. Mezlini, “Finite element modeling of rms roughness effect on the contact stiffness of rough surfaces,” *Tribology in Industry*, vol. 38, no. 3, p. 392, 2016.
- [297] A. Ferretti, M. Giacomini, L. Mastrandrea, and D. Dini, “Investigation of the influence of different asperity contact models on the elastohydrodynamic analysis of a conrod small-end/piston pin coupling,” *SAE International Journal of Engines*, vol. 11, pp. 919–934, 6 2018.
- [298] J. Bonari, M. Paggi, and D. Dini, “A new finite element paradigm to solve contact problems with roughness,” *International Journal of Solids and Structures*, vol. 253, 2022.
- [299] J. J. Wentzel, J. Kloet, I. Andhyiswara, and *et al.*, “Shear-stress and wall-stress regulation of vascular remodeling after balloon angioplasty: Effect of matrix metalloproteinase inhibition,” *Circulation*, vol. 104, no. 1, pp. 91–96, 2001.
- [300] T. Saleem and S. Raju, “An overview of in-stent restenosis in iliofemoral venous stents,” *Journal of Vascular Surgery: Venous and Lymphatic Disorders*, vol. 10, no. 2, 492–503.e2, 2022.
- [301] E. M. J. Hartman, G. D. Nisco, F. J. H. Gijssen, *et al.*, “The definition of low wall shear stress and its effect on plaque progression estimation in human coronary arteries,” *Scientific Reports*, vol. 11, 1 2021.
- [302] H. Zahouani, C. Pailler-Mattei, B. Sohm, R. Vargiolu, V. Cenizo, and R. Debret, “Characterization of the mechanical properties of a dermal equivalent compared with human skin in vivo by indentation and static friction tests,” *Skin Research and Technology*, vol. 15, no. 1, pp. 68–76, 2009.

-
- [303] C. Zhang, Y. Li, L. Yang, and H. Zhao, “Dynamic indentation and microstructural analysis of artery under pre-strain in different directions,” *Materials Today Communications*, p. 111 041, 2024.
- [304] N. F. A. Manan, M. H. M. Ramli, M. N. A. A. Patar, *et al.*, “Determining hyperelastic parameters of human skin using 2d finite element modelling and simulation,” in *2012 IEEE Symposium on Humanities, Science and Engineering Research*, IEEE, 2012, pp. 805–809.
- [305] T. Marinopoulos, L. Zani, S. Li, and V. V. Silberschmidt, “Modelling indentation of human lower-limb soft tissue: Simulation parameters and their effects,” *Continuum Mechanics and Thermodynamics*, pp. 1–17, 2020.
- [306] H. Wei, X. Liu, L. Li, *et al.*, “Visual indentation apparatus and finite element modelling as a method to characterize 3d mechanical properties of facial skin in vivo,” *Mechanics of Materials*, vol. 157, p. 103 852, 2021.
- [307] R. B. Groves, S. Coulman, J. C. Birchall, and S. L. Evans, “Quantifying the mechanical properties of human skin to optimise future microneedle device design,” *Computer Methods in Biomechanics and Biomedical Engineering*, vol. 15, no. 1, pp. 73–82, 2012.
- [308] D. Bader and P. Bowker, “Mechanical characteristics of skin and underlying tissues in vivo,” *Biomaterials*, vol. 4, no. 4, pp. 305–308, 1983.
- [309] J. T. Iivarinen, R. K. Korhonen, P. Julkunen, and J. S. Jurvelin, “Experimental and computational analysis of soft tissue stiffness in forearm using a manual indentation device,” *Medical Engineering & Physics*, vol. 33, no. 10, pp. 1245–1253, 2011.
- [310] J. T. Iivarinen, R. K. Korhonen, and J. S. Jurvelin, “Experimental and numerical analysis of soft tissue stiffness measurement using manual indentation device—significance of indentation geometry and soft tissue thickness,” *Skin Research and Technology*, vol. 20, no. 3, pp. 347–354, 2014.
- [311] T. Marinopoulos, L. Zani, S. Li, and V. V. Silberschmidt, “Modelling indentation of human lower-limb soft tissue: Simulation parameters and their effects,” *Continuum Mechanics and Thermodynamics*, vol. 35, pp. 939–955, 3 2023.
- [312] A. Lundkvist, E. Lilleodden, W. Siekhaus, J. Kinney, L. Pruitt, and M. Balooch, “Viscoelastic properties of healthy human artery measured in saline solution by afm-based indentation technique,” *MRS Online Proceedings Library*, vol. 436, pp. 353–358, 1996.
- [313] R. A. Macrae, K. Miller, and B. J. Doyle, “Methods in mechanical testing of arterial tissue: A review,” *Strain*, vol. 52, no. 5, pp. 380–399, 2016.

-
- [314] M. A. Panteleev, N. Korin, K. D. Reesink, *et al.*, “Wall shear rates in human and mouse arteries: Standardization of hemodynamics for in vitro blood flow assays: Communication from the isth ssc subcommittee on biorheology,” *Journal of Thrombosis and Haemostasis*, vol. 19, no. 2, pp. 588–595, 2021.
- [315] L. M. Riou, A. Broisat, C. Ghezzi, *et al.*, “Effects of mechanical properties and atherosclerotic artery size on biomechanical plaque disruption—mouse vs. human,” *Journal of Biomechanics*, vol. 47, no. 4, pp. 765–772, 2014.
- [316] A. C. Akyildiz, L. Speelman, and F. J. Gijsen, “Mechanical properties of human atherosclerotic intima tissue,” *Journal of Biomechanics*, vol. 47, no. 4, pp. 773–783, 2014.
- [317] D. M. Ebenstein, D. Coughlin, J. Chapman, C. Li, and L. A. Pruitt, “Nanomechanical properties of calcification, fibrous tissue, and hematoma from atherosclerotic plaques,” *Journal of Biomedical Materials Research Part A: An Official Journal of The Society for Biomaterials, The Japanese Society for Biomaterials, and The Australian Society for Biomaterials and the Korean Society for Biomaterials*, vol. 91, no. 4, pp. 1028–1037, 2009.
- [318] P. Mortier, G. A. Holzapfel, M. De Beule, *et al.*, “A novel simulation strategy for stent insertion and deployment in curved coronary bifurcations: Comparison of three drug-eluting stents,” *Annals of Biomedical Engineering*, vol. 38, pp. 88–99, 2010.
- [319] C. Conway, J. P. McGarry, and P. E. McHugh, “Modelling of atherosclerotic plaque for use in a computational test-bed for stent angioplasty,” *Annals of Biomedical Engineering*, vol. 42, pp. 2425–2439, 12 2014.
- [320] H.-H. Choi, S.-M. Hwang, Y. Kang, J. Kim, and B. Kang, “Comparison of implicit and explicit finite-element methods for the hydroforming process of an automobile lower arm,” *The International Journal of Advanced Manufacturing Technology*, vol. 20, pp. 407–413, 2002.
- [321] S. Sjöstrand, A. Widerström, I. Svensson, *et al.*, “The impact of geometry, intramural friction, and pressure on the antegrade longitudinal motion of the arterial wall: A phantom and finite element study,” *Physiological Reports*, vol. 11, 12 2023.
- [322] K. Takashima, T. Kitou, K. Mori, and K. Ikeuchi, “Simulation and experimental observation of contact conditions between stents and artery models,” *Medical Engineering & Physics*, vol. 29, no. 3, pp. 326–335, 2007.
- [323] C. Lally, F. Dolan, and P. Prendergast, “Cardiovascular stent design and vessel stresses: A finite element analysis,” *Journal of Biomechanics*, vol. 38, no. 8, pp. 1574–1581, 2005.

-
- [324] W. Wu, M. Qi, X. P. Liu, D. Z. Yang, and W. Q. Wang, "Delivery and release of nitinol stent in carotid artery and their interactions: A finite element analysis," *Journal of Biomechanics*, vol. 40, pp. 3034–3040, 13 2007.
- [325] M. Cilla, E. Peña, and M. A. Martínez, "3d computational parametric analysis of eccentric atheroma plaque: Influence of axial and circumferential residual stresses," *Biomechanics and Modeling in Mechanobiology*, vol. 11, pp. 1001–1013, 7 2012.
- [326] A. Schiavone, L. G. Zhao, and A. A. Abdel-Wahab, "Effects of material, coating, design and plaque composition on stent deployment inside a stenotic artery—finite element simulation," *Materials Science and Engineering C*, vol. 42, pp. 479–488, 2014.
- [327] M. De Beule, P. Mortier, S. G. Carlier, B. Verhegghe, R. Van Impe, and P. Verdonck, "Realistic finite element-based stent design: The impact of balloon folding," *Journal of Biomechanics*, vol. 41, no. 2, pp. 383–389, 2008.
- [328] F. Gervaso, C. Capelli, L. Petrini, S. Lattanzio, L. Di Virgilio, and F. Migliavacca, "On the effects of different strategies in modelling balloon-expandable stenting by means of finite element method," *Journal of Biomechanics*, vol. 41, no. 6, pp. 1206–1212, 2008.
- [329] C. Dumoulin and B. Cochelin, "Mechanical behaviour modelling of balloon-expandable stents," *Journal of Biomechanics*, vol. 33, no. 11, pp. 1461–1470, 2000.
- [330] F. Migliavacca, L. Petrini, M. Colombo, F. Auricchio, and R. Pietrabissa, "Mechanical behavior of coronary stents investigated through the finite element method," *Journal of Biomechanics*, vol. 35, no. 6, pp. 803–811, 2002.
- [331] F. Migliavacca, L. Petrini, V. Montanari, I. Quagliana, F. Auricchio, and G. Dubini, "A predictive study of the mechanical behaviour of coronary stents by computer modelling," *Medical Engineering & Physics*, vol. 27, no. 1, pp. 13–18, 2005.
- [332] M. De Beule, R. Van Impe, B. Verhegghe, P. Segers, and P. Verdonck, "Finite element analysis and stent design: Reduction of dogboning," *Technology and Health Care*, vol. 14, no. 4-5, pp. 233–241, 2006.
- [333] G. J. Hall and E. P. Kasper, "Comparison of element technologies for modeling stent expansion," *Journal of Biomechanical Engineering*, pp. 751–756, 2006.
- [334] M. Early, C. Lally, P. J. Prendergast, and D. J. Kelly, "Stresses in peripheral arteries following stent placement: A finite element analysis," *Computer Methods in Biomechanics and Biomedical Engineering*, vol. 12, no. 1, pp. 25–33, 2009.

-
- [335] M. A. Kendall, Y.-F. Chong, and A. Cock, "The mechanical properties of the skin epidermis in relation to targeted gene and drug delivery," *Biomaterials*, vol. 28, no. 33, pp. 4968–4977, 2007.
- [336] J. Van Kuilenburg, M. A. Masen, and E. Van Der Heide, "Contact modelling of human skin: What value to use for the modulus of elasticity?" *Proceedings of the Institution of Mechanical Engineers, Part J: Journal of Engineering Tribology*, vol. 227, no. 4, pp. 349–361, 2013.
- [337] K. L. Johnson, K. Kendall, and A. Roberts, "Surface energy and the contact of elastic solids," *Proceedings of the Royal Society of London. A. mathematical and physical sciences*, vol. 324, no. 1558, pp. 301–313, 1971.
- [338] B. V. Derjaguin, V. M. Muller, and Y. P. Toporov, "Effect of contact deformations on the adhesion of particles," *Journal of Colloid and Interface Science*, vol. 53, no. 2, pp. 314–326, 1975.
- [339] A. Dai, W. He, S. Wang, *et al.*, "In vivo adhesive behavior of human facial skin by a modified indentation test," *Journal of the Mechanical Behavior of Biomedical Materials*, vol. 92, pp. 172–178, 2019.
- [340] Y. Qiao, Y. Zeng, Y. Ding, J. Fan, K. Luo, and T. Zhu, "Numerical simulation of two-phase non-newtonian blood flow with fluid-structure interaction in aortic dissection," *Computer Methods in Biomechanics and Biomedical Engineering*, vol. 22, pp. 620–630, 6 2019.
- [341] I. Larrabide, M. Kim, L. Augsburger, M. C. Villa-Uriol, D. Rüfenacht, and A. F. Frangi, "Fast virtual deployment of self-expandable stents: Method and in vitro evaluation for intracranial aneurysmal stenting," *Medical Image Analysis*, vol. 16, pp. 721–730, 3 2012.