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Mécanique des solides, des matériaux, des structures et des surfaces

Analyses expérimentales et modèles de simulation du comportement mécanique de structures tricotées pour applications de Bas Médicaux de Compression

par

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Contents

Acknowledgements	iii
1 Introduction	1
1.1 Problem statement	2
1.2 Research outline and scientific challenges	2
1.3 Context of the Symphonies project	3
1.4 Objectives of the thesis	5
1.5 Summary of chapters	6
2 Background and Literature Review	9
2.1 Compression therapy for venous and lymphatic diseases	10
2.1.1 Venous and lymphatic diseases	10
2.1.2 Compression principle	12
2.1.2.1 Laplace's law	13
2.1.3 Factors affecting the interface pressure	15
2.1.3.1 Knitted fabric properties	15
2.1.3.2 Patient morphology	19
2.1.3.3 Application technique	21
2.2 Medical compression stockings	22
2.2.1 French standard for manufacturing (NF S97-114)	23
2.2.2 Mechanical characterisation	26
2.2.2.1 Standard cylindrical testing (NF G30-102-B)	26
2.2.2.2 Cut-strip tensile tests	28
2.2.2.3 Biaxial tensile tests	29
2.2.2.4 Characterisation of fabric constituents	30
2.2.3 Mechanical modelling of medical compression stockings	31
2.2.3.1 Homogenisation models	31
2.2.3.2 Mesoscopic and unit cell models	35
2.2.3.3 Geometric models	37

2.3	Methods for determining and predicting interface pressure	39
2.3.1	Experimental measurements	40
2.3.1.1	Pressure sensors	40
2.3.1.2	Leg mannequins	42
2.3.2	Indirect methods based on Laplace's law	45
2.3.3	Numerical methods based on FE models	48
2.3.3.1	Models of the limb	48
2.3.3.2	Models of the medical compression stocking	52
2.3.3.3	Simulation results and validation	58
2.4	Summary and conclusions	63
2.4.1	Key findings	63
2.4.2	Future directions and research opportunities	64
3	Inlaid weft-knitted fabrics for medical compression stockings	65
3.1	Introduction	65
3.2	Medical compression stockings	67
3.2.1	Medical product and compression zones	68
3.2.2	Knitted structure and yarn materials	70
3.2.3	Geometrical parameters	71
3.3	Experimental characterisation	74
3.3.1	Structural characterisation : density of inlay yarn	75
3.3.2	Mechanical characterisation of yarns	77
3.3.2.1	Tensile testing of inlay yarns	77
3.3.2.2	Tensile response of inlay yarns	78
3.3.3	Mechanical characterisation of knitted fabrics	80
3.3.3.1	Knitted fabric samples	81
3.3.3.2	In-use circumferential strain range	83
3.3.3.3	Tensile testing of knitted fabrics	84
3.3.3.4	Tensile response of knitted fabrics	88
3.4	Interface pressure prediction	96
3.4.1	Prediction of interface pressure based on Laplace's law	97
3.4.1.1	Evolution of pressure with circumferential strain	98
3.4.1.2	Predicted pressures for reference leg perimeters	99
3.4.2	Measurements with pressure sensors	102
3.4.2.1	Measured pressures for maximal leg perimeters	105

3.4.2.2	Comparison with Laplace's predictions	105
3.5	Conclusions and research perspectives	106
4	Mechanical hybrid model of inlaid weft-knitted fabrics	109
4.1	Introduction	109
4.2	Methodology	111
4.2.1	Unit cell configuration	112
4.2.2	Constitutive laws of the model	116
4.2.3	Tensile response prediction	119
4.2.4	Parameter identification: E and ν	124
4.2.5	Validation of the hybrid model	125
4.3	Results and discussion	126
4.3.1	Identified mechanical parameters	126
4.3.2	Predicted tensile response under circumferential loading	128
4.3.2.1	Fabric without inlay yarn	129
4.3.2.2	Fabric with inlay yarn	130
4.3.3	Comparison with experimental data	132
4.3.3.1	Fabric without inlay yarn	132
4.3.3.2	Fabric with inlay yarn	135
4.3.4	Extension of the model for interface pressure prediction	138
4.4	Conclusions and research perspectives	140
5	Finite element simulations for interface pressure prediction	143
5.1	Introduction	143
5.2	Methodology	145
5.2.1	Hybrid finite element model of the medical compression stocking	146
5.2.1.1	Geometric parameters and mesh configuration	147
5.2.1.2	Material parameters	149
5.2.2	Finite element simulations for interface pressure calculation	151
5.2.2.1	Numerical leg model	151
5.2.2.2	Boundary conditions and contact formulation	153
5.2.2.3	Evaluation of interface pressure	155
5.2.2.4	Evaluation of yarn tensile response	156
5.2.2.5	Mesh-sensitivity analysis	157
5.2.3	Validation of the hybrid finite element model	158
5.3	Results	158

5.3.1	Interface pressure predictions	159
5.3.1.1	Effect of mesh ratio on predicted interface pressure	159
5.3.1.2	Predicted interface pressure for different compression zones	161
5.3.1.3	Comparison with reference pressure data	162
5.3.2	Yarn behaviour predictions	165
5.3.2.1	Effect of mesh ratio on yarn response	165
5.3.2.2	Predicted tensile yarn response	166
5.4	Conclusions and research perspectives	168
6	Conclusions, future work and dissemination	171
6.1	General conclusions of the thesis	171
6.2	Scientific contributions of the thesis	172
6.2.1	Contributions to the Symphonies project	173
6.3	Research perspectives	174
6.4	Scientific dissemination	175
	Glossary	177
	References	183
	Abstract	197

List of Figures

1.1	Logo of the Symphonies project.	4
1.2	Logos of institutions collaborating in Work Package 1 of Symphonies project.	4
2.1	Effect of valve failure on blood flow in the venous system [4].	11
2.2	Physiological mechanisms of compression therapy: (a) acceleration of venous flow, (b) and (c) reduction of venous reflux [13].	13
2.3	Circumferential equilibrium of a tubular fabric wrapped around a cylindrical limb segment, illustrating the derivation of Laplace's law [26].	14
2.4	Principle of graduated compression delivery [16].	14
2.5	Double covered yarns: (a) Schematic representation [30]; (b) Yarn covering process for elastane-core yarns [26].	15
2.6	Knitted structure of medical compression fabrics: Schematic representation of the knitted structure [16]; (b) Microscopic image of the knitted structure [26].	17
2.7	Tensile response of fabric specimens along the course direction at different height levels of a medical compression stocking: (a) tested elastic compression stocking tube with inlaid knitted structure, (b) representative tension–stretch curves measured at different zones [23].	19
2.8	Fabrication of MCS: (a) Circular knitting machine [30]; Knitting of medical compression stocking on circular seamless knitting units; (c) Seamless circular MCS [16]; (d) Graduated compression stocking with different compression levels [68].	24
2.9	Leg measurement points of the human lower limb: (I) height and (C) circumference, (a) according to the French standard [29]; (b) simplified as given by Lozo et al. [70].	25
2.10	Tensile test on cylindrical samples of medical compression stockings [30].	27
2.11	Force–Elongation curves of different knitted compression fabrics in wale and course directions [16].	28
2.12	Cut-strip test on samples of medical compression stockings [23].	29

2.13	Biaxial test on cut-strip samples of medical compression stockings [44].	30
2.14	Hybrid unit cell models of woven fabrics, combining shell elements and connectors: (a) presented by Tephany et al. [82]; (b) presented by Nasri et al. [83].	36
2.15	Unit cell model of medical compression stockings, developed by Ghorbani et al.: (a) Simulated knitted fabric; (b) The boundary conditions applied on the unit cell; (c) The force-extension curve of the elastic weft knitted fabric and its components calculated by FEM [87].	37
2.16	Geometrical model of inlaid weft-knitted fabrics, developpd by Fong et al. (a) Technical front of the knitted structure; (b) Model of a laid-in weft-knitted loop unit [41].	39
2.17	Pressure sensors used in the literature (a) PicoPress® by [30], [91]; (b)Flexiforce ®[67]; (c) MST [68]	42
2.18	View of the Textilpress measuring stand during measuring, developpd by Maklewska et al.: (1) Textile compression band; (2) Measuring matrix placed under the compression band; (3) Cylinder; (4) Cylinder holder; (5) Strain gauges amplifier; (6) Computer [28].	44
2.19	Comparison of the contact pressure measured by different lower limbs at different pressure levels: (a) Hohenstein wooden leg; (b) Fabricated leg mannequin developpd by Li et al.; (c) contact pressure distribution on the different regions measured by different leg mannequins (the fabricated leg mannequin and the wooden leg mannequin) and human legs [92].	44
2.20	Model of the lower limb presented by Zhang et al.: (a) separated parts of the lower limb; (b) cylinder model; (c) conical model [44].	49
2.21	Expanded solid model of numerical simulation, presented by Hedigalla et al. [59].	50
2.22	Geometry reconstruction from MRI scans of a lower limb unaffected by lymphedema. The two bones, tibia and fibula, were excluded from the geometry and their boundaries were treated as rigid, presented by Reda and Avril [99].	51
2.23	Simulation of clothing pressure distribution on calf (a) from the vertical direction to the surface of the calf, (b) the equivalent stress graph in the FEA model, presented by Yanmei et al. [102].	54

2.24	Geometric models of the medical compression stockings used for finite element analysis (a) MCS represented as a compilation of conical segments, proposed by Lu et al. [69]; (b) MCS represented as a single conical segment and (c) Aggregated model of MCS with leg, proposed by Ghorbani et al [80]	55
2.25	The developed geometric models of (a) FE MCS (Group A ECS, Group B ECS, Group C ECS) and (b) FE legs (WL - I, II, III), and (c) the meshing and boundary conditions applied in the FEM, presented by Ye et al. [74].	58
2.26	The simulated interface pressure profiles applied by the different groups of the studied ECSs when their 2D and 3D material mechanical properties were applied, respectively, presented by Ye et al. [74].	60
2.27	Validation of the finite element (FE) predictions by comparing the pressure values at eight sites of the skin–stocking interface: (a) locations of the eight sites; (b) comparison of the interface pressure between the FE and experimental data. [69].	62
2.28	Validation of the finite element (FE) predictions by comparing the changes in the circumference of the calf at six cross-sections of the calf: (a) locations of the six cross-sections; (b) comparison of the changes in the calf circumference between the FE and experimental data. [69].	62
3.1	Selected women’s medical compression stocking, developed and commercialised by Sigvaris France [107].	68
3.2	Microscopy of the 1x1 inlaid weft – knitted structure of the selected medical compression stocking.	70
3.3	Microscopy of the two yarns conforming the selected medical compression stocking: ground yarns and inlay yarns	71
3.4	Medical compression stocking with different compression zones (ankle, calf and thigh) in the flat configuration	72
3.5	Dimensional data from the manufacturer’s size chart for the Semi Transparent model (size M, normal length) [107].	72
3.6	Circumferential strain imposed on the fabric when placed over the leg, given by the relative difference between the initial fabric perimeter (P_0), and the leg perimeter (P_{leg})	74
3.7	Longitudinal deformation imposed on the fabric samples under in-use conditions, defining the in-use density of inlay yarn given by the number of courses (NC) for a given interline distance (d).	76

3.8	Tensile testing device used for inlay yarn characterisation [113].	78
3.9	Force–Strain response of the inlay yarn during tensile testing during three consecutive cycles.	79
3.10	Zoom of the inlay yarn Force–Strain response in the initial elastic region during tensile testing.	80
3.11	Tenacity–Strain response of the inlay yarn during tensile testing during third loading cycle before rupture.	80
3.12	Dimensional parameters of weft-knitted fabric samples from both configurations, without inlay yarn (GF) and with inlay yarn (IF): initial flat width (W_0) and interline distance (d).	82
3.13	(a) Placement of fixation pins along the longitudinal direction (E_1) at the coloured reference lines, maintaining a constant interline distance ($d = 30$ mm). (b) Attachment of the cylindrical sample to the machine along the longitudinal direction, including the stabilisers for ankle samples.	84
3.14	Loading cycle along the circumferential direction (E_2) during tensile test of cylindrical knitted fabrics, where C_0 is the initial clamp-to-clamp distance, C the clamp-to-clamp distance during test, and L the fabric length.	85
3.15	Dimensional parameters during tensile testing with controlled displacement along circumferential direction (u_2) and blocked displacements in the longitudinal and thickness directions ($u_1 = u_3 = 0$): initial fabric width (W_0) and perimeter (P_0), and maximal fabric width ($W_{\max}$) and perimeter ($W_{\max}$). . . .	86
3.16	Distribution of the total force ($F_{2\text{-total}}$) in the two fabric layers during uniaxial loading.	87
3.17	Schematic lateral representation of the different steps of the tensile test on circumferential fabric samples.	88
3.18	Evolution of registered force (F_2) as a function of fabric perimeter (P) during tensile test for ankle zone of knitted sample without inlay yarn	90
3.19	Evolution of registered force (F_2) as a function of fabric perimeter (P) during tensile test for calf zone of knitted sample without inlay yarn	90
3.20	Evolution of registered force (F_2) as a function of fabric perimeter (P) during tensile test for thigh zone of knitted sample without inlay yarn	90
3.21	Evolution of registered force (F_2) as a function of fabric perimeter (P) during tensile test for ankle zone of knitted sample with inlay yarn.	92
3.22	Evolution of registered force (F_2) as a function of fabric perimeter (P) during tensile test for calf zone of knitted sample with inlay yarn.	92

3.23	Evolution of registered force (F_2) as a function of fabric perimeter (P) during tensile test for thigh zone of knitted sample with inlay yarn.	92
3.24	Evolution of fabric tension (T) as a function of circumferential strain (ϵ_{circ}) during tensile test for the three compression zones of inlaid weft-knitted samples (IF).	94
3.25	Evolution of fabric tension (T) as a function of circumferential strain (ϵ_{circ}) during tensile test for the three compression zones of weft-knitted samples without inlay yarn (GF).	94
3.26	Evolution of Laplace's interface pressure (IP_{Laplace}) as a function of circumferential strain (ϵ_{circ}) for different compression zones of knitted fabrics with inlay yarn (IF configuration).	99
3.27	Evolution of Laplace's interface pressure (IP_{Laplace}) as a function of circumferential strain (ϵ_{circ}) for different compression zones of knitted fabrics without inlay yarn (GF configuration).	99
3.28	Calculated Laplace interface pressure (IP_{Laplace}) for reference leg perimeters of fabric with inlay yarn (IF configuration).	101
3.29	Calculated Laplace interface pressure (IP_{Laplace}) for reference leg perimeters of fabric without inlay yarn (GF configuration).	101
3.30	Three dimensional computer aided model of the leg geometry used for interface pressure measurements: cylinder height ($H1$) and diameter ($D1$), and cone height ($H2$) and diameter ($D2$)	103
3.31	Three dimensional printed rigid leg models corresponding to the maximal leg perimeters ($P_{\text{leg,max}}$) of the three compression zones	104
3.32	Placement of fabric samples with inlay yarn (IF configuration) on the rigid models with maximal leg perimeters ($P_{\text{leg,max}}$) for measurements with the PicoPress [®] sensor	104
4.1	Representation of the hybrid unit cell modelling the local knitted structure of the medical compression stocking	112
4.2	Dimensional parameters of the hybrid model in the global and local reference systems.	115
4.3	Force–Strain response of the inlay yarn during the third loading cycle in tensile test, used for defining the tensile response of the spring in the hybrid model. 119	

4.4	Displacement imposed along the circumferential direction (E_2) during tensile testing, and resulting deformation of the unit cell in the global and local reference systems.	120
4.5	Flowchart of the hybrid model: from fabric and unit cell dimensions to circumferential deformation and tensile response prediction, including elastic and spring contributions.	123
4.6	Model sensitivity for constant Poisson's ratio ($\nu = 0.10$) and variable Young's modulus (E).	127
4.7	Model sensitivity for constant Young's modulus ($E = 0.02$ MPa) and variable Poisson's ratio (ν).	127
4.8	Predicted tension (T) under circumferential deformation (ϵ_{circ}) for ankle, calf, and thigh in the GF configuration.	130
4.9	Predicted tension (T) under circumferential deformation (ϵ_{circ}) for ankle, calf, and thigh in the IF configuration.	131
4.10	Comparison between predicted and experimental tensile response for ankle zone of fabric without inlay yarn (GF).	133
4.11	Comparison between predicted and experimental tensile response for calf zone of fabric without inlay yarn (GF).	134
4.12	Comparison between predicted and experimental tensile response for thigh zone of fabric without inlay yarn (GF).	134
4.13	Comparison between predicted and experimental tensile response for ankle zone of fabric with inlay yarn (IF).	136
4.14	Comparison between predicted and experimental tensile response for calf zone of fabric with inlay yarn (IF).	137
4.15	Comparison between predicted and experimental tensile response for thigh zone of fabric with inlay yarn (IF).	137
5.1	Hybrid finite element model of the medical compression stocking: (a) shell elements modelling the continuous loop structure, (b) connector elements modelling the inlay-yarn reinforcement, (c) assembly of shells and connectors. . .	147
5.2	Parameters governing the mesh of the hybrid finite element model of the fabric.	148
5.3	Generated finite element models of the three compression zones: ankle, calf, and thigh.	149
5.4	Rigid leg models of three reference leg perimeters for the different compression zones.	153

5.5	Simulation of the placement of the fabric over the rigid leg model.	154
5.6	Element set defined on the leg surface for the evaluation of the interface pressure (<i>IP</i>) at the end of the simulation.	155
5.7	Connector total force (<i>CTF</i>), expressed in N, obtained for the average leg perimeter of ankle zone.	156
5.8	Evolution of contact pressure (<i>CPRESS</i>) with increasing element size ratio (R_{mesh}) for average leg perimeter in ankle zone.	160
5.9	Evolution of contact pressure (<i>CPRESS</i>) for two element size ratios ($R_{\text{mesh}} = 8, 9$) for the three reference leg perimeters, minimal, average and maximal, in thigh zone.	161
5.10	Numerical pressure maps resulting from the simulations for the average leg perimeters of the three compression zones, ankle, calf, and thigh. <i>CPRESS</i> is given in MPa.	162
5.11	Comparison of interface pressure (<i>IP</i>) at minimal leg perimeter ($P_{\text{leg},\text{min}}$) obtained from finite element simulations (computed) and Laplace's calculations, for the three compression zones zones.	163
5.12	Comparison of interface pressure (<i>IP</i>) at average leg perimeter ($P_{\text{leg},\text{avg}}$) obtained from finite element simulations (computed) and Laplace's calculations, for the three compression zones zones.	164
5.13	Comparison of interface pressure (<i>IP</i>) at minimal leg perimeter ($P_{\text{leg},\text{max}}$) obtained from finite element simulations (computed), Laplace's calculations, and PicoPress® measurements, for the three compression zones zones.	164
5.14	Comparison between experimental and computed tensile yarn response, for the different compression zones.	166
5.15	Computed inlay yarn tensile response for the three compression zones and reference leg perimeters.	167
5.16	Implicit simulations under development for contact pressure prediction. . . .	170

List of Tables

2.1	Compression classes defined by the French standard NF S97-114, according to the ankle interface pressure.	26
3.1	Reference leg perimeters for different compression zones: minimal ($P_{\text{leg, min}}$), average ($P_{\text{leg, avg}}$) and maximal ($P_{\text{leg, max}}$)	73
3.2	Number of courses (NC) and density of inlay yarn (D_{IY}) per compression zone under in-use conditions	77
3.3	Initial fabric dimensions for both fabric configurations, without inlay yarn (GF) and with inlay yarn (IF), measured after precondition on wooden leg: width (W_0) and perimeter (P_0)	83
3.4	Range of circumferential strain (ϵ_{circ}) for the different compression zones under in-use conditions	83
3.5	Fabric tension (T) at reference leg perimeters for fabrics without inlay yarn (GF).	95
3.6	Fabric tension (T) at reference leg perimeters for fabrics with inlay yarn (IF).	95
3.7	Dimensions of the three dimensional printed leg models for different compression zones: cylinder height ($H1$) and diameter ($D1$), and cone height ($H2$) and diameter ($D2$)	103
3.8	Measured interface pressure with pressure sensor (IP_{sensor}) for maximal leg perimeters ($P_{\text{leg, max}}$) of different compression zones	105
4.1	Geometrical parameters of the hybrid model for the different compression zones.	116
4.2	Mean absolute percentage error (MAPE) and weighted error (WEr) values for different combinations of E and ν	128
4.3	Mean absolute error (MAE) and Mean absolute percentage error (MAPE) between experimental and predicted fabric tension for the three compression zones of fabric without inlay yarn (GF configuration).	135

4.4	Mean absolute error (MAE) and Mean absolute percentage error (MAPE) between experimental and predicted fabric tension for the three compression zones of fabric with inlay yarn (IF).	138
4.5	Dimensional and mechanical parameters used in the analytical interface–pressure model.	139
4.6	Example application of the analytical model for predicting interface pressure at maximal leg perimeter for ankle and calf zones, with detailed circumferential strain ($\varepsilon_{\text{circ}}$), tension in the continuous element (T_{cont}), spring tension (T_{spring}), fabric tension (T_{model}) and experimental fabric tension (T_{exp}).	139
4.7	Comparison between analytical predictions, Laplace-based experimental estimates, and PicoPress [®] measurements.	140
5.1	Geometrical parameters defining the mesh of the three compression zones: total fabric length (L_{total}), initial fabric perimeter (P_0), density of inlay yarn (D_{IY}), element height and initial connector length ($h = l_{\text{conn}, 0}$), and number of connectors per total fabric (N_{length}).	149
5.2	Absolute error (AE) and mean absolute percentage error (MAPE) between finite element predictions and Laplace analytical estimations for the three compression zones and reference leg perimeters.	164
5.3	Connector force (CTF1) for different mesh ratios ($R_{\text{mesh}} = 7, 8, 9$) for ankle zone and three reference leg perimeters, minimal ($P_{\text{leg, min}}$), average ($P_{\text{leg, avg}}$), and maximal (P_{leg}), in the ankle zone.	165

Chapter 1

Introduction

Medical compression stockings (MCS) are widely used in the treatment of venous and lymphatic disorders. These garments apply a controlled external pressure on the lower limb to improve venous return, reduce oedema and stabilise chronic conditions. Their clinical performance depends on the magnitude and distribution of the interface pressure applied on the leg. This pressure is generated by the tension developed in the fabric when the stocking is stretched under circumferential deformation.

MCS are produced using inlaid weft-knitted fabrics. The structure combines elastic covered yarns forming the loop structure with inlay yarns acting as reinforcement. The mechanical behaviour of the fabric results from the combined action of these two yarn systems. Variations in the density of inlay yarns along the stocking produce zone-dependent stiffness, which defines the graduated compression required for therapeutic performance. Understanding how the local fabric structure affects the mechanical response and developed tension in each zone is therefore essential for predicting the pressure delivered on the leg.

The pressure exerted by an MCS can be predicted using experimental, analytical or numerical approaches. Experimental methods rely on pressure sensors or mannequins replicating leg geometry. Analytical approaches combine tensile tests with simplified formulations such as Laplace's law. Numerical simulations integrate models of the leg and of the stocking to compute the interface pressure resulting from the circumferential deformation imposed on the fabric when placed over the leg. These simulations often rely on homogenised material descriptions and simplified structural assumptions for the stocking, that do not represent the anisotropy and heterogeneity of inlaid weft-knitted fabrics.

Improving pressure prediction requires models that integrate detailed structural information and mechanical data of the two constitutive yarn systems, in conditions representative of wear. Such models would enable a better estimation of the local compression applied in each zone and support the design of stockings adapted to specific therapeutic needs and morphologies.

1.1 Problem statement

Existing approaches present several limitations for describing the mechanical behaviour of inlaid weft-knitted fabrics.

Experimental protocols found in the literature often do not reproduce the deformation imposed by the leg when the stocking is worn. Zone-specific structural parameters, such as density of inlay yarn, or in-use circumferential deformation are not always measured. This restricts the interpretation of the mechanical differences between zones and limits the accuracy of pressure prediction.

Some studies have presented the influence of the knitted structure in the mechanical response of the fabric. However, this has been rarely considered in models of inlaid weft knitted fabrics for medical compression stockings.

Many mechanical and numerical models represent the textile as a homogeneous membrane, neglecting the hybrid behaviour generated by the loop structure and the inlay yarn. The mechanical contribution of the two yarn constituents is not explicitly modelled, and the effect of variations in density of inlay yarn in fabric stiffness is often not included. These simplifications prevent a direct link between the structural parameters and the mechanical response of the MCS.

Moreover, variations in raw materials or production conditions affect the mechanical response of the fabric. Current modelling tools do not allow manufacturers to evaluate the impact of these changes or to predict the delivered pressure using structural and mechanical inputs. There is a need for models capable of representing the hybrid architecture of the knitted structure and of predicting the tension generated in the fabric as a function of structural parameters defined in each compression zone and the individual mechanical response of the constitutive yarn systems. The limitations of existing experimental, analytical and numerical approaches are discussed in detail in Chapter 2.

1.2 Research outline and scientific challenges

This thesis aims to address the current scientific challenges related to the mechanical modelling of inlaid weft-knitted fabrics used in medical compression stockings.

A **first challenge** is to characterise the combined behaviour of the loop structure and the inlay reinforcement. The deformation mechanisms of these two yarn systems must be identified in conditions representative of use, and their respective contributions to the circumferential stiffness must be described.

A **second challenge** concerns the relation between the structural parameters and the tensile response. Zone-dependent variations of density of inlay yarn and of circumferential deformation imposed by the leg modify the circumferential stiffness and explain the mechanical differences between compression zones.

A **third challenge** concerns the development of a mechanical model capable of predicting the behaviour of the fabric under circumferential deformation. The combined action of the loop structure and the inlay reinforcement must be described in a form compatible with numerical computation. The model must account for local variations in inlay density and for their effect on the circumferential stiffness, while remaining sufficiently simple to allow parameter identification from experimental data. This requires a local–mesoscopic modelling approach able to represent the hybrid knitted architecture through simplified mechanical elements, without explicitly modelling the detailed geometry of the yarns.

A **fourth challenge** is the integration of this mechanical description into a finite element framework describing the placement of the stocking on a leg. The model must be transferable to a numerical environment in which the circumferential deformation imposed by the leg geometry can be reproduced, allowing the interface pressure to be obtained from the tension developed in the fabric. This integration enables the prediction of the pressure generated once the stocking is placed on the limb and allows the analysis of pressure variations for different compression zones.

1.3 Context of the Symphonies project

This thesis was carried out within the SYMPHONIES project, funded by BPI France at the GEMTEX laboratory (ENSAIT), under the supervision of Professor Damien Soulat (thesis director), Professor Xavier Legrand (co-director), and Dr. Ahmad Rashed Labanieh (supervisor).

The SYMPHONIES project (Figure 1.1) aims to develop digital tools to support the design, evaluation and personalisation of medical compression stockings. The general objective is to link textile parameters, leg morphology and physiological behaviour through numerical simulation. The project combines industrial, clinical and academic expertise, with contributions from SIGVARIS, GEMTEX (ENSAIT), EMSE–Sainbiose and other partners.



FIGURE 1.1: Logo of the Symphonies project.

SIGVARIS, the industrial leader of the project, provides medical compression products, manufacturing specifications and experimental expertise, including knowledge of textile production processes, material selection and performance requirements. **GEMTEX**, the textile research laboratory of ENSAIT, is responsible for the mechanical modelling of the stocking fabric, integrating structural and mechanical characterisation into predictive models. GEMTEX is also involved in the evaluation of patient morphology and in the adaptation of medical compression stocking design to different morphotypes. **EMSE–Sainbiose** develops biomechanical models of the lower limb and investigates the mechanical response of soft tissues under local compression. The collaboration between these partners in Work Package 1 (Figure 1.2) aims to develop a simulation tool capable of predicting the mechanical interaction between the stocking and the leg for improving product design.



FIGURE 1.2: Logos of institutions collaborating in Work Package 1 of Symphonies project.

Work Package 1 is structured in Tasks 1.1, 1.2, and 1.3. Task 1.1, led by GEMTEX and Sigvaris, focuses on the digital design of compression products and includes the analysis of user morphology, the construction of a geometric design model and the description of the mechanical behaviour of the stocking. Task 1.1.1 addresses the representation of leg geometries and the development of dimensioning rules for stocking design. Task 1.1.2, to which this doctoral work contributes, focuses on the development of a mechanical simulation model for medical compression stockings. Work Package 1 also includes Task 1.2 and Task 1.3, led respectively by EMSE–Sainbiose and SIGVARIS, which focus on the development of biomechanical models of the lower limb and on the implementation of an interface to integrate the different modelling components in order to support and improve product design. These tasks are complementary to the work presented in this thesis.

The inlaid weft-knitted fabric used in medical compression stockings presents a hybrid architecture composed of a deformable loop structure and an inlay reinforcement. For the

simulation model to be reliable, the mechanical role of each yarn system must be represented. The model must also incorporate the variation in density of inlay yarn and structural parameters between the ankle, calf and thigh regions, since these variations define the graduated compression profile.

Within this context, Task 1.1.2 includes the following objectives:

- the analysis of the structural parameters defining the fabric configuration under in-use circumferential deformation,
- the description of the mechanical behaviour of the fabric under controlled tensile conditions,
- the development of a constitutive model considering the local knitted structure, adapted to finite element analysis,
- the numerical simulation of the stocking–leg mechanical interaction to predict interface pressure.

The present thesis contributes to Task 1.1.2 by providing structural measurements obtained in in-use conditions, including zone-specific densities of inlay yarn, and by characterising the tensile response of the yarns and of the fabric under circumferential loading. A local hybrid model is developed to describe the combined action of the loop structure and the inlay reinforcement, enabling the influence of structural parameters on the tensile behaviour and on the delivered compression to be identified. This mechanical description is then transferred to a finite element environment to simulate the placement of the stocking on a leg and to predict the interface pressure generated along the different compression zones.

These developments support the construction of a modelling chain that couples textile mechanics with a digital representation of the lower limb. The objective is to enable the analysis of how variations in textile parameters modify the mechanical behaviour of the stocking and the resulting physiological effects, such as pressure distribution or soft tissue compression.

1.4 Objectives of the thesis

In line with the problem statement, the identified scientific challenges and the context of the Symphonies project, the objectives of this thesis are defined as follows:

The **first objective** of this thesis is to characterise the mechanical behaviour of the in-laid weft-knitted fabric used in medical compression stockings and to determine how the

structural parameters influence the tensile response and the compression generated in the different zones.

A **second objective** is to develop a hybrid local–mesoscopic model capable of representing the combined action of the loop structure and the inlay yarn, and of capturing the local variations in inlay density and their effect on the circumferential stiffness of the fabric.

A **third objective** is to transfer this hybrid model into a finite element environment in order to simulate the placement of the stocking on a leg and to predict the interface pressure generated in each compression zone.

Accordingly, the work is divided in the following sub-objectives:

1. Quantify the structural parameters of the fabric, with particular focus on the inlay yarn distribution in each compression zone.
2. Measure the mechanical response of the yarns and the fabric under conditions representative of use.
3. Describe how local variations in inlay density modify the circumferential stiffness of the fabric.
4. Develop a local–mesoscopic hybrid model integrating the mechanical contribution of both the loop structure and the inlay reinforcement, and the local variations in density of inlay yarn.
5. Implement the hybrid model in a finite element simulation of the stocking–leg configuration.
6. Predict the interface pressure generated in each compression zone and compare it with analytical and experimental measurements.

1.5 Summary of chapters

The thesis is organised into six chapters, combining state of the art review, structural analysis, mechanical characterisation, local–mesoscopic modelling and finite element simulation of medical compression stockings.

Chapter 1 introduces the clinical and industrial context of medical compression. The mechanical principles governing the pressure delivered by compression stockings are presented, together with the main approaches used for pressure prediction. The limitations of experimental, analytical and numerical methods are discussed, motivating the need for

models incorporating structural information and yarn-level mechanics. The scientific objectives of the thesis are introduced, and the work is positioned within Task 1.1.2 of the Symphonies project, which focuses on the development of a mechanical simulation model for compression stockings.

Chapter 2 reviews the state of the art on medical compression products, focusing on textile structures, manufacturing processes and the mechanical behaviour of inlaid weft-knitted fabrics. The chapter analyses the principles of pressure generation, the effect of fabric tension and the influence of textile architecture. Existing models for pressure prediction are examined, including analytical formulations based on Laplace's law and finite element simulations of stocking-leg interactions. The review highlights the lack of structural measurements under in-use deformation and the limitations of homogenised material descriptions that do not represent the reinforcement effect of inlay yarns.

Chapter 3 describes the experimental characterisation of the compression stockings. The structural analysis quantifies the density of inlay yarn in each compression zone and identifies the dimensional parameters governing the circumferential deformation applied on the fabric when placed on the leg. Tensile tests are performed on inlay yarns and tubular samples with and without inlay yarn to evaluate their mechanical response under controlled uniaxial loading. The behaviour of the loop structure and the inlay reinforcement is analysed, and the combined response of the fabric is measured for the ankle, calf and thigh regions. This chapter also presents experimental measurements and mechanical estimations of interface pressure for reference leg perimeters. The chapter establishes the structural and mechanical inputs required for the development of the local-mesoscopic model. It also provides experimental data for validating the developed hybrid model and numerical simulations for pressure prediction.

Chapter 4 presents the development of the hybrid local-mesoscopic model describing the mechanical behaviour of the inlaid weft-knitted fabric. The model represents the loop structure using continuous elements and the inlay yarn reinforcement using discrete elements. Structural parameters measured under in-use conditions define the geometry of the unit cell for each compression zone. The model predicts the tensile response by combining the contribution of both yarn systems. Mechanical parameters are identified from the experimental data, and the model is validated by comparing predicted curves with the tensile response of tubular samples. The chapter demonstrates how variations in inlay density generate differences in circumferential stiffness.

Chapter 5 describes the integration of the local-mesoscopic model into a finite element simulation of the stocking placed over rigid leg geometries. The structural parameters of

each compression zone are transferred to the numerical mesh, and the circumferential deformation imposed by the leg is reproduced. The contact interaction between the fabric and the leg is modelled, and the interface pressure is computed. Mesh convergence and sensitivity analyses are performed to ensure numerical stability. The predicted pressure distributions are compared with analytical values obtained from Laplace's law and with experimental measurements using pressure sensors. The chapter evaluates the ability of the model to reproduce the pressure gradient defined by the structural variations of the stocking.

Chapter 6 summarises the main contributions of the thesis and discusses the relevance of the developments for the mechanical modelling of compression stockings. The chapter presents the contributions to the Symphonies project, emphasising the role of the structural characterisation, the local–mesoscopic model and the finite element simulations in the construction of a digital design tool. Limitations of the current work are identified, and perspectives are proposed for extending the methodology to more complex textile architectures, nonlinear material behaviour and patient-specific geometries.



Chapter 2

Background and Literature Review

Compression therapy is a well-established medical treatment for venous and lymphatic diseases of the lower limb. By applying controlled external pressure through elastic garments, the haemodynamic balance is restored, supporting venous return, and reducing swelling. Among the different therapeutic devices, medical compression stockings are the most widely used because they combine controlled pressure levels with comfort for daily life. Their effectiveness depends on precise control of the interface pressure delivered to the limb, which is governed by the interaction between textile architecture, material properties and limb geometry.

Since inadequate compression can compromise therapeutic efficacy or cause discomfort, understanding the interaction between factors affecting the delivered pressure is crucial for both clinical practice and product development. From a design perspective, predicting and optimising delivered pressure requires connecting yarn-level mechanics to fabric behaviour, and fabric mechanics to the garment–limb interaction. Over the past decades, experimental protocols, analytical formulations, and numerical models have been developed to address these challenges, each providing complementary insights but also presenting specific limitations.

This chapter reviews the current knowledge on compression therapy and the modelling of medical compression stockings. Section 2.1 introduces the clinical context, presenting the principles of compression therapy and the main factors affecting interface pressure. Section 2.2 focuses on medical compression stockings, including the French manufacturing standard, their mechanical characterisation and existing mechanical modelling approaches. Section 2.3 reviews the main methods used to determine and predict interface pressure, covering experimental, analytical and numerical approaches. Finally, Section 2.4 summarises the key findings of the literature review and identifies the main research gaps addressed in this thesis.

2.1 Compression therapy for venous and lymphatic diseases

Chronic venous insufficiency and lymphoedema are long-term disorders affecting the venous and lymphatic systems of the lower limbs. Both conditions impair the physiological mechanisms responsible for fluid return, leading to abnormal accumulation of interstitial fluid, progressive tissue remodelling and deterioration of limb function [1]–[3]. Compression therapy constitutes a cornerstone of conservative treatment for these pathologies, as it directly targets the mechanical causes of oedema and venous hypertension.

2.1.1 Venous and lymphatic diseases

In healthy individuals, venous return from the lower limbs is ensured by the combined action of venous valves and the calf muscle pump. During muscle contraction, blood is propelled proximally towards the heart, while venous valves prevent retrograde flow. During muscle relaxation, valve closure maintains unidirectional circulation and limits venous pressure. This coordinated mechanism ensures efficient drainage from both superficial and deep venous systems.

In pathological conditions, valve dysfunction alters this mechanism. Valve incompetence in the superficial, deep or perforator veins allows blood to flow back distally, reducing venous return and preventing the normal decrease in venous pressure during physical activity. As a consequence, venous pressure remains elevated, leading to venous hypertension and fluid leakage into the interstitial space [4]. Figure 2.1 illustrates the effect of valve failure on venous blood flow.

Chronic venous insufficiency Chronic venous insufficiency (CVI) is defined as a morphological or functional abnormality of the venous system of the lower limbs resulting in venous reflux, obstruction, or both [5], [6]. The primary pathophysiological mechanism is valvular incompetence, which causes persistent retrograde blood flow and elevated venous hydrostatic pressure during standing and walking.

Sustained venous hypertension increases capillary filtration, leading to plasma leakage into the interstitial space that exceeds lymphatic drainage capacity. This imbalance initiates inflammatory and structural changes in the vessel wall and surrounding tissues, including endothelial dysfunction, fibrin cuff formation and extracellular matrix remodelling [1], [2]. Progressive venous dilation further impairs the efficiency of the calf muscle pump, reducing ejection volume and prolonging venous refilling time. The resulting venous stasis induces tissue hypoxia and chronic inflammation [7], [8].

Over time, hypoxia-driven fibroblast activation and collagen deposition lead to skin thickening, fibrosis and hyperpigmentation. In advanced stages, these processes culminate in venous ulceration [9], [10]. Clinically, patients experience heaviness, swelling, pain and paraesthesias, which typically worsen throughout the day due to prolonged upright posture and fluid accumulation [5]. Post-thrombotic syndrome represents a severe secondary form of CVI following deep vein thrombosis, where venous obstruction and scarring cause chronic hypertension. Clinical studies have shown that elastic compression stockings significantly reduce the incidence and severity of post-thrombotic syndrome when used after deep vein thrombosis [11], [12].

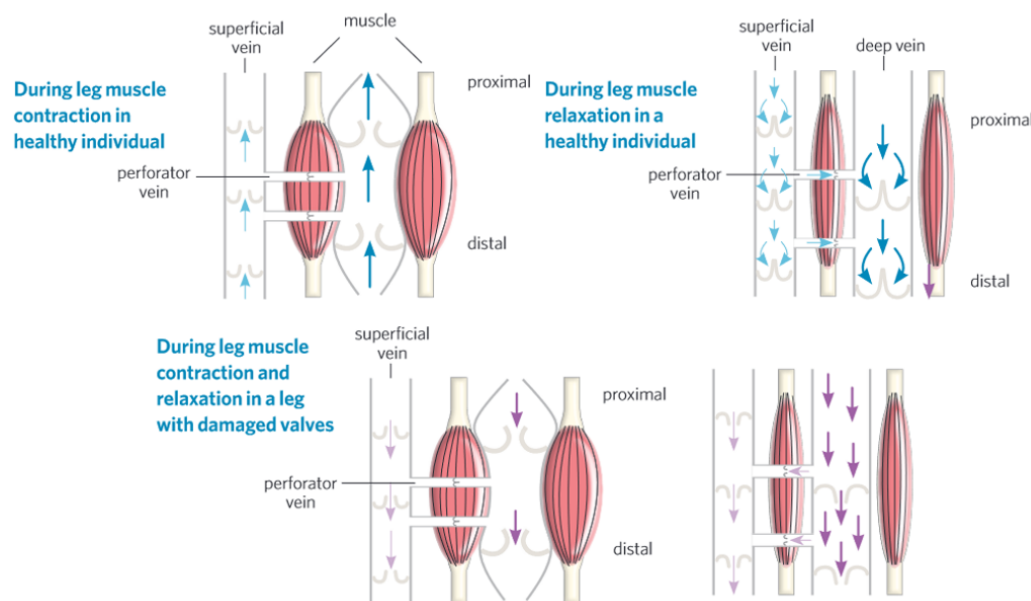


FIGURE 2.1: Effect of valve failure on blood flow in the venous system [4].

Lymphoedema Lymphoedema is a chronic condition characterised by the accumulation of protein-rich lymphatic fluid in the interstitial space due to impaired lymph transport or obstruction of lymphatic vessels. It can be primary, resulting from congenital abnormalities of the lymphatic system, or secondary, caused by surgery, infection, radiation therapy or trauma [13], [14].

Persistent lymphatic stasis leads to progressive swelling, increased tissue stiffness and inflammatory fibrosis of the subcutaneous layers. As the disease advances, the mechanical properties of the affected tissues are altered: tissue compliance decreases and viscoelastic behaviour becomes dominated by fibrotic components [14], [15]. These changes exacerbate limb enlargement, heaviness and functional impairment.

At the microscopic level, chronic oedema modifies the mechanical environment of cells and the extracellular matrix. Sustained mechanical loading stimulates fibroblast proliferation and collagen accumulation, resulting in irreversible dermal thickening and loss of elasticity. Without treatment, these mechanical alterations perpetuate lymphatic dysfunction and inflammation. Compression therapy plays a key role in long-term management by reducing capillary filtration, promoting lymphatic drainage and maintaining interstitial pressure within physiological limits, thereby helping to restore tissue homeostasis [13], [15].

2.1.2 Compression principle

Compression therapy is the primary conservative treatment for venous and lymphatic disorders. It is commonly defined as the application of a garment around a limb to exert a controlled external pressure on the tissues [16]. Attaran and Ochoa Chaar [1] emphasised that compression remains the most widely used non-invasive treatment for chronic venous disease and continues to play a central role in both therapeutic and preventive care.

The pressure delivered by a compression garment results from the interaction between textile properties, garment construction, finishing processes and application conditions [16]. When properly designed and applied, compression therapy acts through several physiological mechanisms that improve venous and lymphatic function. These mechanisms are schematically illustrated in Figure 2.2.

Clinical and experimental studies have shown that external compression reduces venous lumen diameter, thereby increasing blood flow velocity and improving venous return. Rohan et al. [17] demonstrated a reduction in vein diameter under compression, while Wu and Liu [18] quantified reductions of 13–16% in superficial veins and 3–9% in deep veins for class II compression stockings. Although moderate, these reductions are sufficient to restore venous competence. Avril et al. [19] confirmed that vein diameter reduction leads to increased blood flow velocity, while Partsch [2] showed that graduated compression stockings further enhance deep venous flow and limit superficial reflux. Wang et al. [20] reported flow velocity increases of 6.5% in healthy subjects and up to 19.6% in patients with chronic venous insufficiency.

Compression also interacts with the calf muscle pump. By limiting lateral muscle expansion during contraction, compression redirects blood flow proximally and improves pump efficiency [2]. Barufi et al. [15] clinically demonstrated this effect in lymphoedema patients, showing that walking with compression led to significantly greater volume reduction than walking without compression.

Additional clinical benefits of compression therapy have been widely reported. Improvements in symptoms such as heaviness, tension, evening oedema and discomfort have been observed in both healthy individuals and patients [7], [16]. Compression therapy has also been shown to accelerate venous ulcer healing and reduce recurrence rates [9], [10]. Moreover, the use of compression stockings following deep vein thrombosis significantly reduces the incidence and severity of post-thrombotic syndrome [11], [12].

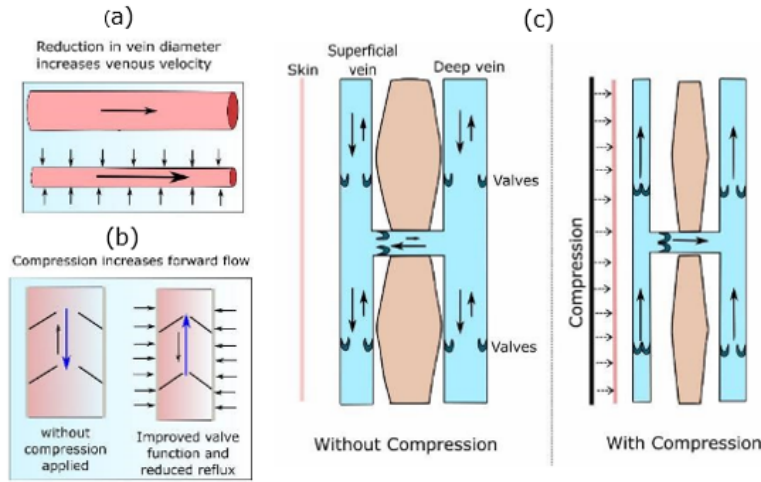


FIGURE 2.2: Physiological mechanisms of compression therapy: (a) acceleration of venous flow, (b) and (c) reduction of venous reflux [13].

The therapeutic effect of compression therapy is governed by the interface pressure (IP), defined as the pressure exerted at the contact between the garment and the skin. Interface pressure represents the mechanical interaction between the textile and the limb and is the key parameter controlling venous and lymphatic responses. Its magnitude depends on fabric tension, limb geometry and contact conditions, making it a central variable in both clinical practice and mechanical modelling [2], [21].

2.1.2.1 Laplace's law

Laplace's law provides a classical relationship between fabric tension and interface pressure in compression therapy. In a simplified biomechanical context, it relates the circumferential tension developed in a stretched fabric (T) to the radius of curvature of the underlying surface (R) and the resulting interface pressure (IP) [22]. Under the assumption of a cylindrical geometry and uniform tension distribution, Laplace's law is expressed as given in Equation 2.1.

$$IP = \frac{T}{R} \quad (2.1)$$

Figure 2.3 illustrates the derivation of this relation from the circumferential equilibrium of a tubular fabric wrapped around a rigid cylindrical limb segment. According to this formulation, interface pressure increases with fabric tension and decreases with increasing limb radius. As a result, for a given fabric tension, higher pressures are expected in regions with smaller radii of curvature, such as the ankle or the tibial crest [23]–[25].

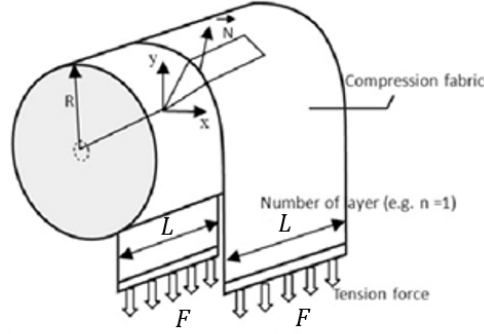


FIGURE 2.3: Circumferential equilibrium of a tubular fabric wrapped around a cylindrical limb segment, illustrating the derivation of Laplace's law [26].

Because the human leg perimeter decreases from the thigh to the ankle, a garment generating uniform fabric tension naturally produces higher pressure distally and lower pressure proximally [21], [27]. In graduated compression stockings, this pressure gradient is further controlled by introducing variations in fabric stiffness across anatomical zones. Higher inlay yarn density and elastane feed tension are typically applied in the ankle region, leading to greater circumferential tension and higher interface pressure, as illustrated in Figure 2.4.



FIGURE 2.4: Principle of graduated compression delivery [16].

Laplace's law is widely used to interpret tensile tests and mannequin-based measurements [27], [28]. However, its applicability to real compression stockings is limited. The formulation assumes a uniform cylindrical geometry and a homogeneous elastic membrane,

neglecting the complex morphology of the human leg and the heterogeneous, anisotropic architecture of knitted compression fabrics. These simplifications explain the discrepancies often observed between Laplace-based predictions and experimental pressure measurements, particularly in regions with pronounced curvature.

Consequently, a comprehensive description of interface pressure requires consideration of multiple factors, including textile structure, garment construction, limb morphology and application conditions. These factors are discussed in the following section.

2.1.3 Factors affecting the interface pressure

The magnitude and distribution of interface pressure depend on a combination of garment properties, application conditions, and patient morphology. Understanding these factors is essential for both product design and clinical use.

2.1.3.1 Knitted fabric properties

Yarn properties Medical compression stockings are manufactured using covered elastic yarns composed of an elastane core helically wrapped with one or two polyamide or polyester filaments [29]. The elastane core acts as the primary load-bearing component, while the covering filaments contribute to surface friction, abrasion resistance and energy dissipation during cyclic loading. The interaction between these constituents, shown in Figure 2.5, produces the nonlinear tensile response and hysteresis characteristic of compression fabrics [26].

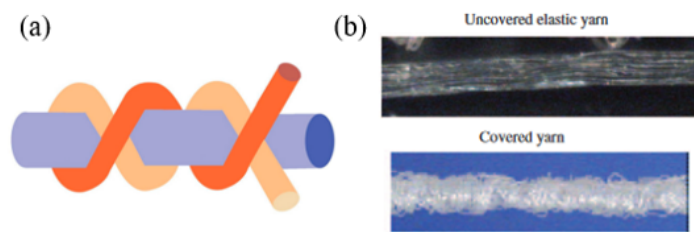


FIGURE 2.5: Double covered yarns: (a) Schematic representation [30]; (b) Yarn covering process for elastane-core yarns [26].

The mechanical behaviour of covered yarns is influenced by both elastane properties and covering configuration. Increasing elastane feed tension or yarn linear density results in higher circumferential stiffness and increased compression levels, as reported in several experimental studies. These parameters directly affect the pre-strain of the elastane core and the force generated during extension [31]–[34].

The viscoelastic behaviour of elastane governs pressure stability during wear. A rapid stress relaxation occurs shortly after loading, followed by a slower decay over time, which

explains the progressive loss of compression observed in clinical use [35], [36]. Environmental conditions further modify yarn behaviour. Increased temperature and humidity soften the elastic response and alter skin–fabric friction, influencing both comfort and pressure distribution [37], [38].

Knitted structure Medical compression stockings are typically produced as inlaid weft-knitted fabrics. This structure combines two distinct yarn systems: a ground yarn forming intermeshed loops that provide extensibility and comfort, and an inlay yarn inserted along the course direction without loop formation, acting as circumferential reinforcement. This dual configuration results in a strongly anisotropic mechanical behaviour. The inlay yarns carry most of the load during circumferential extension, while the looped ground structure accommodates deformation and ensures recovery [32], [34], [39].

The general architecture of inlaid weft-knitted compression fabrics and the two yarn system are illustrated in Figure 2.6. The figure shows the covered elastic yarn, the looped ground structure and the insertion of the inlay yarn along the course direction.

Several structural parameters govern the mechanical response of the fabric, including loop length, stitch density, inlay yarn density and inlay insertion angle. Shorter loop lengths and higher stitch or inlay densities reduce yarn crimp and increase inter-yarn friction, leading to higher circumferential stiffness and interface pressure [31], [40]. Elastane feed tension further contributes to this effect by increasing the initial pre-strain of the elastic core [33].

The inlay insertion angle plays an important role in balancing stiffness and comfort. Low insertion angles, close to the course direction, maximise circumferential stiffness, whereas higher angles increase fabric extensibility [41]. Numerical and three-dimensional modelling studies show that the inlay yarn follows a sinusoidal path within the structure. Under extension, this path progressively straightens before the yarn fully carries load, producing the characteristic nonlinear stress–strain response of compression fabrics [42].

The stitch pattern of the ground structure also affects mechanical behaviour. Plain, rib and interlock constructions differ in loop connectivity, thickness and damping capacity. Rib and interlock fabrics generally exhibit higher lateral stability and hysteresis than plain structures [39], [43]. Finishing processes, such as heat setting and mechanical stabilisation, further modify the final mechanical response by relieving residual stresses in the elastane filaments and fixing the loop geometry.

At the microscopic scale, friction within covered elastane yarns contributes to delayed recovery and time-dependent behaviour. Denser coverings and higher elastane feed tensions

increase recovery forces and reduce stress relaxation, helping to limit pressure decay during wear [26], [33], [35].

Overall, the literature consistently shows that inlay yarns dominate the circumferential stiffness of compression fabrics, while the looped ground structure governs extensibility, comfort and recovery. Structural parameters such as loop length, stitch density, inlay density and insertion angle therefore play a central role in defining interface pressure and pressure stability [32], [39], [41], [44].

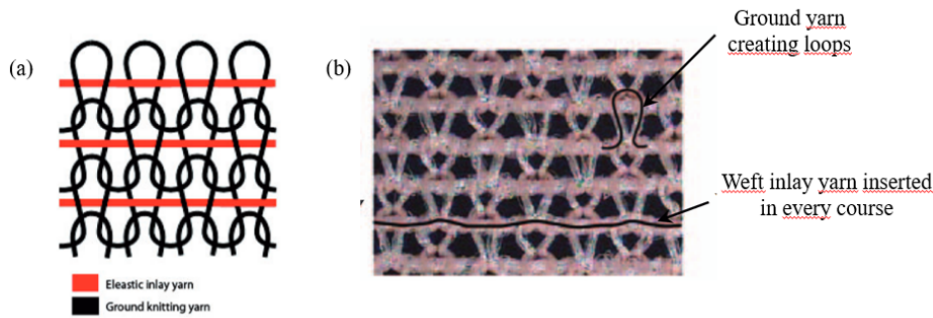


FIGURE 2.6: Knitted structure of medical compression fabrics: Schematic representation of the knitted structure [16]; (b) Microscopic image of the knitted structure [26].

Graduated pressure profile Medical compression stockings are designed to generate a graduated pressure profile along the leg, with the highest compression at the ankle and progressively lower levels toward the calf and thigh. This distal-to-proximal pressure gradient counteracts hydrostatic pressure differences, limits venous reflux and promotes venous return. To achieve this behaviour, the ankle region is knitted to be mechanically stiffer, while the upper regions are progressively more extensible. Under circumferential deformation imposed by the leg, this structural arrangement results in higher fabric tension and interface pressure at the ankle and lower pressure proximally.

Clinical studies have shown that an effective therapeutic gradient typically corresponds to a pressure reduction of approximately 20–30% between successive anatomical segments [2], [16], [20]. This gradient arises from a combination of structural heterogeneity within the fabric and variations in limb geometry. Increased inlay yarn density, thicker yarns or higher elastane feed tension in the ankle region lead to greater circumferential stiffness and higher local pressures [31], [32], [39], [44]. Experimental studies have confirmed that stockings incorporating locally reinforced inlay yarns generate stronger compression at the ankle compared with the calf and thigh regions [45].

Anatomical geometry further amplifies this effect. Smaller local radii of curvature at the ankle increase interface pressure according to Laplace's law [24]. By combining variations in fabric stiffness and limb geometry, manufacturers are able to control the pressure distribution and achieve the desired graduated compression profile required for therapeutic effectiveness.

Tensile response Circumferential tension is the direct mechanical origin of interface pressure in medical compression stockings. It develops when the garment, whose resting circumference is smaller than that of the limb, is stretched during donning. The magnitude of this tension depends on the level of circumferential deformation imposed by the leg, which explains the strong influence of garment sizing on the delivered compression. Larger extensions induce higher strains and stresses in the fabric, resulting in higher interface pressure.

The level of tension generated under extension is governed by the tensile stiffness of the knitted fabric. Standardised tensile tests, such as those defined in NF G 30-102-B [46], are commonly used to characterise the force–elongation behaviour of compression fabrics under controlled circumferential deformation. In these tests, deformation is applied along the course direction, reproducing the main loading mode experienced by the stocking during wear. Fabrics with higher tensile modulus generate higher interface pressures for a given elongation [32], [33]. Structural parameters such as course density and elastane feed tension play a key role in controlling this response [31].

The tensile behaviour of compression fabrics is nonlinear and exhibits pronounced hysteresis, reflecting the viscoelastic nature of elastane-based knitted structures [47], [48]. Shorter loop lengths or higher elastane pre-tension increase the slope of the stress–strain curve, enhancing circumferential stiffness and compression capability [40]. Zone-specific tensile analyses further show that tighter structures and higher inlay yarn densities in the ankle region lead to greater stiffness and higher local pressures compared with the calf and thigh regions [23], [39].

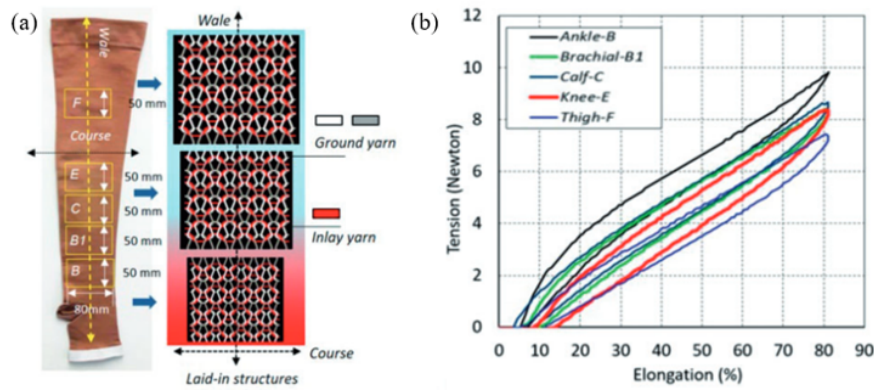


FIGURE 2.7: Tensile response of fabric specimens along the course direction at different height levels of a medical compression stocking: (a) tested elastic compression stocking tube with inlaid knitted structure, (b) representative tension–stretch curves measured at different zones [23].

Time-dependent effects further influence compression stability. A significant fraction of stress relaxation occurs within the first minutes after loading, followed by a slower decay during prolonged wear, leading to progressive pressure loss [35], [36], [49]. In addition, repeated washing and wear induce mechanical degradation of the fabric, reducing tensile stiffness and hysteresis. In some cases, this degradation is sufficient to lower the effective compression class of the product over time [50].

2.1.3.2 Patient morphology

The geometry and biomechanical properties of the limb play a decisive role in determining how the tension of the fabric translates into the interface pressure and how this pressure propagates through the tissues. Even when stockings are manufactured with identical mechanical properties, differences in limb shape, curvature, and tissue composition result in variations in the delivered pressure and therapeutic performance [23]–[25], [51].

Limb geometry Limb geometry is a major factor governing the spatial distribution of interface pressure generated by medical compression stockings. According to Laplace’s law, interface pressure is inversely proportional to the local radius of curvature, so smaller radii lead to higher pressures for a given fabric tension. This explains the recurrent pressure peaks observed at anatomical regions such as the ankle, tibial crest and malleoli [23], [24]. Experimental measurements consistently show that the ankle region experiences the highest compression, due both to its smaller radius and to the increased stiffness of the knitted structure in this zone [39]. Numerical simulations further demonstrate that local curvature variations

can significantly amplify or attenuate pressure, even when circumferential tension remains unchanged [25].

Limb geometry can be described using different modelling approaches. Classical analytical descriptions approximate the leg as a truncated cone or as a succession of circular sections [52]. More recent studies rely on three-dimensional body scanning or medical imaging to capture detailed surface curvature and anatomical transitions [53], [54]. These approaches reveal that small geometric asymmetries between medial and lateral sides can induce local pressure variations exceeding 20%, highlighting the limitations of simplified axisymmetric assumptions.

Internal morphology further modulates pressure distribution. Bone prominences and muscle bulges locally modify contact conditions and generate stress concentrations that cannot be predicted from external geometry alone [19]. Finite element simulations confirm that the combined effects of surface curvature and internal tissue organisation govern the local amplification of interface pressure under patient-specific conditions [24].

Friction at the fabric-skin interface The frictional interaction between the compression fabric and the skin plays a key role in the conversion of circumferential tension into interface pressure. When a stocking is stretched over the limb, shear stresses at the skin–fabric interface determine how tension is transmitted to the tissues. A higher coefficient of friction stabilises the garment and limits slippage, but it may also increase local shear stresses and cause discomfort or irritation [37], [38], [55]. Conversely, lower friction facilitates fabric sliding over the leg, and promotes more uniform placement, but may allow slipping and reduce effectivity of the treatment [51], [56].

Tribological studies on knitted compression fabrics report friction coefficients typically ranging from 0.3 to 0.8, depending on yarn composition, surface roughness and environmental conditions [55]. Fibre orientation, knitting pattern and moisture level strongly influence this behaviour, particularly under humid conditions [37].

In vivo measurements show that apparent friction decreases over time as moisture redistributes and micro-sliding occurs at the interface [38], [57]. Finite element studies incorporating frictional effects demonstrate that accurate friction characterisation is essential for predicting local pressure peaks, especially near highly curved regions such as the malleoli and tibial crest [56]. Clinically, excessive friction is associated with local discomfort and skin damage, particularly in patients with fragile skin or when high compression levels are applied.

Response of internal tissues The physiological effect of compression therapy depends not only on the surface pressure applied by the garment, but also on how this pressure is transmitted through the internal tissues of the limb. The leg is a multilayered structure composed of skin, subcutaneous adipose tissue, muscle, fascia and bone, each characterised by distinct stiffness and viscoelastic behaviour. The mechanical interaction between these layers governs stress redistribution and determines the effectiveness of compression in enhancing venous and lymphatic function [19], [58], [59].

Experimental and numerical studies have shown that external compression reduces superficial vein diameters and redistributes stresses toward deeper regions, but only a fraction of the applied surface pressure reaches the venous system [19]. Finite element models incorporating realistic hyperelastic properties indicate that the thickness and stiffness of subcutaneous adipose tissue are dominant factors controlling pressure transmission. Softer or thicker adipose layers attenuate internal stresses, explaining part of the inter-individual variability observed in clinical response to compression [58]. Multilayer simulations further show that pressure decays rapidly with depth, with only 40–60% of the surface pressure transmitted to venous structures [59], [60].

Reported elastic moduli range approximately from 200 to 800 kPa for skin, 20 to 100 kPa for adipose tissue, and 10 to 150 kPa for skeletal muscle, with nearly incompressible behaviour characterised by Poisson's ratios close to 0.49 [61]–[65]. These strong mechanical contrasts generate non-uniform strain fields and local stress concentrations at tissue interfaces. Numerical studies emphasise that neglecting tissue heterogeneity leads to inaccurate predictions of internal stress distribution and interface pressure, particularly when homogeneous limb models are used [19], [24].

2.1.3.3 Application technique

The method used to apply a medical compression stocking has a decisive influence on the magnitude and uniformity of the interface pressure delivered to the limb. Because compression garments rely on elastic deformation rather than rigid fixation, the tension introduced during donning directly determines the circumferential strain and, consequently, the effective compression applied.

Variations in application technique, such as stretching angle, alignment or local repositioning during donning, can lead to significant differences in interface pressure, even for identical stockings [51]. Operator skill therefore represents a major source of variability. Improper application may generate local pressure peaks, particularly around the ankle or tibial crest, increasing the risk of discomfort or skin irritation. Since the applied tension

depends on the difference between garment and limb circumferences at the moment of donning, small alignment errors can result in pressure variations of several millimetres of mercury [2]. Appropriate training of healthcare professionals and patients has been shown to improve reproducibility and therapeutic effectiveness.

In experimental studies, strict control of the application procedure is essential to obtain reliable and repeatable measurements. Consistent pressure profiles are achieved only when stockings are positioned using predefined reference marks and stretched by a controlled and reproducible amount [16], [39]. The tension applied during fitting also affects the calibration and validation of analytical and numerical models, underlining the need for standardised application protocols [66]. To address this issue, automated or guided donning systems have recently been introduced in laboratory settings to control pre-stretch and alignment. These systems improve experimental reproducibility, although their use in routine clinical practice remains limited.

Because interface pressure results from the combined effects of garment properties, limb morphology and application technique, accurate characterisation of medical compression stockings under controlled application conditions is essential for reliable mechanical modelling and pressure prediction. Such characterisation supports improved product design and contributes to the optimisation of therapeutic protocols [67].

2.2 Medical compression stockings

Medical compression stockings (MCS) are the most widely used medical devices for long-term compression therapy. Unlike bandages or adjustable wraps, which require professional application and frequent readjustment, stockings provide a stable and reproducible compression adapted to daily use [2]. Their seamless construction, comfort and adaptability to individual anatomy contribute to improved patient adherence and sustained therapeutic effectiveness [1].

Clinical studies have consistently demonstrated the superiority of stockings over other compression modalities in terms of long-term outcomes. Higher healing rates and reduced recurrence of venous leg ulcers have been reported when stockings are used instead of bandages [9], [10]. In addition, stockings have been shown to effectively reduce symptoms such as heaviness, oedema and discomfort during daily activities, even in healthy individuals [7].

The therapeutic efficacy of medical compression stockings relies on the generation of a graduated compression profile, characterised by the highest pressure at the ankle and

progressively lower pressures toward the calf and thigh. Both experimental and numerical studies have shown that this distal-to-proximal gradient improves venous return more effectively than uniform compression [20], [21]. According to Partsch [2], graduated compression also enhances comfort and adherence while maintaining clinical effectiveness [68]. More recent studies further highlighted the influence of individual morphology on pressure distribution, showing that personalised stockings can improve comfort and therapeutic outcomes [53], [54], [69].

2.2.1 French standard for manufacturing (NF S97-114)

The French standard NF S97-114 (2022) specifies the material, structural and performance requirements for medical compression stockings. According to this standard, MCS are knitted tubular garments designed to exert a controlled external pressure on the lower limb, with the highest pressure applied at the ankle and a progressive decrease toward the calf and thigh, in order to promote venous and lymphatic return [29]. The main definitions used in this work are summarised in the Glossary.

Knitted structure of medical compression stockings Medical compression stockings are manufactured using inlaid weft-knitted structures that combine ground loop-forming yarns and an inlay yarn within the therapeutic zone. The ground yarns form the loop network that ensures extensibility and comfort, while the inlay yarn is inserted along the course direction and acts as circumferential reinforcement. This configuration results in a mechanically anisotropic fabric, with high extensibility in the longitudinal direction and higher stiffness in the circumferential direction.

Knitting is performed on circular machines, producing seamless tubular fabrics, as illustrated in Figure 2.8. The inlay yarn must be double-covered and continuous throughout the therapeutic zone. All sizes within a given product range are manufactured on machines of identical needle gauge to ensure reproducible compression behaviour.

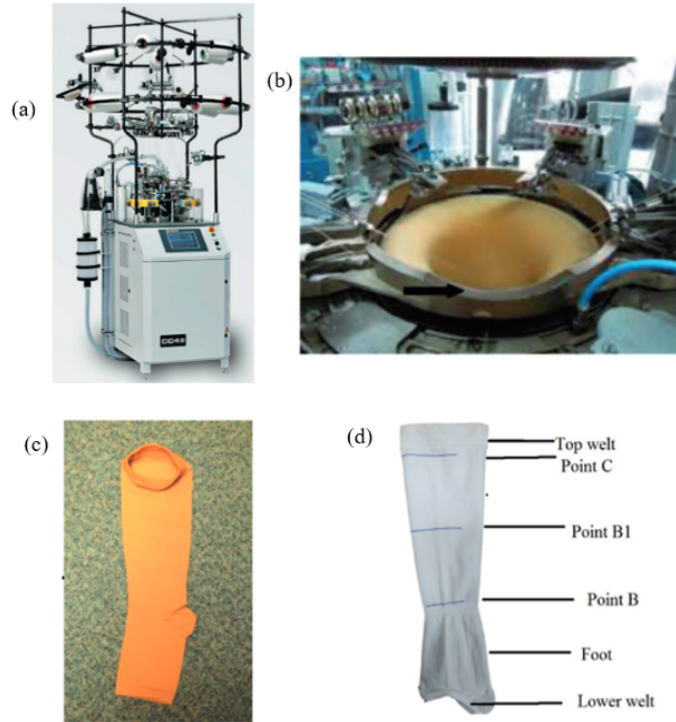


FIGURE 2.8: Fabrication of MCS: (a) Circular knitting machine [30]; Knitting of medical compression stocking on circular seamless knitting units; (c) Seamless circular MCS [16]; (d) Graduated compression stocking with different compression levels [68].

The standard further specifies the linear densities of the constituent yarns. The ground loop yarn must have a minimum linear density of 66 dtex, while the inlay yarn must reach 310 dtex. The inlay yarn consists of a 310 dtex elastane core covered by two 22 dtex filaments, ensuring sufficient extensibility, fatigue resistance and dimensional stability during repeated stretching. When two loop yarns are used, the second must also be double-covered and maintain the same 310 dtex linear density as the inlay yarn.

Two structural configurations are authorised: Structure type 1, in which the inlay yarn is inserted into every course of loops, and Structure type 2, in which the inlay yarn is inserted every two courses. Examples of Structure type 1 are shown in Figure 2.6.

Fabric stability and ageing After knitting, stockings undergo dyeing and heat-setting processes to stabilise the fabric structure. These finishing treatments must not modify the compression level by more than $\pm 10\%$ of the nominal value. Dimensional changes after washing and drying are limited to $\pm 5\%$ of the initial dimensions. Washing tests follow the 4N cycle defined in NF EN ISO 6330, with flat drying (mode C).

The mechanical and compression properties of the stockings must remain stable under repeated use. After cyclic elongation up to 120%, simulating normal wear, the variation in

ankle pressure must not exceed $\pm 10\%$ of the initial value. No visible damage to the fabric or seams is permitted, and the structure must recover its shape without permanent deformation.

Limb morphology and measurement points Compression is generated by the tension developed in the fabric when the stocking is stretched to fit the limb, as described in Section 2.1.2. The therapeutic effect therefore depends on the sizing system defined by the manufacturer and on accurate limb measurements performed by the prescriber or fitter.

For most patients, standard size ranges are sufficient to ensure appropriate compression. When limb morphology does not correspond to available sizes, custom-made stockings may be manufactured based on individual measurements. Control of compression levels thus requires the definition of application dimensions along the lower limb to ensure homogeneous pressure distribution.

According to the standard, limb dimensions must be measured at predefined locations, as illustrated in Figure 2.9. These include circumferences at the ankle (cB), calf (cC) and thigh (cF or cG), as well as corresponding heights (iF or iG) measured from the ground. These measurements define the application geometry used to determine stocking size and pressure graduation.

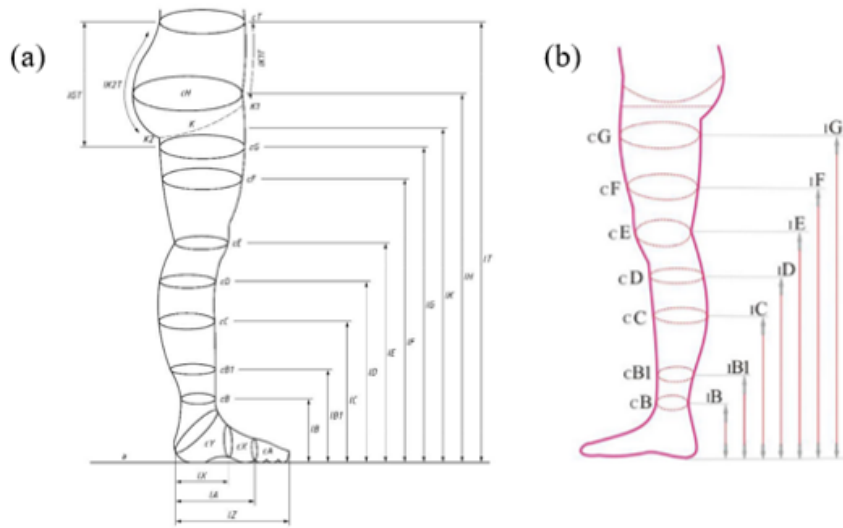


FIGURE 2.9: Leg measurement points of the human lower limb: (I) height and (C) circumference, (a) according to the French standard [29]; (b) simplified as given by Lozo et al. [70].

Compression and pressure graduation Interface pressure is determined according to NF G30-102, Annex B [46], as described in Section 2.2.2.1. Pressure measurements are performed

at the ankle, calf and thigh. The ankle pressure defines the compression class of the product, with Class I corresponding to preventive compression and Classes II to IV covering therapeutic levels for venous and lymphatic disorders [29]. Table 2.1 summarises the pressure ranges associated with each class.

TABLE 2.1: Compression classes defined by the French standard NF S97-114, according to the ankle interface pressure.

Class	Ankle pressure (mmHg)
Class I	10–15
Class II	15–20
Class III	20–36
Class IV	>36

2.2.2 Mechanical characterisation

The mechanical behaviour of medical compression stockings results from the combined effects of knitted architecture, yarn properties and manufacturing parameters. To characterise this behaviour, experimental studies have investigated the response of compression textiles at different scales, focusing either on the complete inlaid weft-knitted fabric, on simplified loop structures without inlay yarns, or on the covered elastic yarns constituting the ground and reinforcement systems. Among these approaches, tensile-based methods applied to inlaid weft-knitted fabrics remain the most representative of in-service loading conditions.

2.2.2.1 Standard cylindrical testing (NF G30-102-B)

The reference method for characterising medical compression stockings is the circumferential tensile test defined in NF G 30-102-B [46], performed under controlled environmental conditions according to EN ISO 139 [71]. In this test, tubular samples extracted from the ankle, calf or thigh regions are mounted on rigid frames with fixed axial length and stretched circumferentially at a constant rate, as illustrated in Figure 2.10. Deformation is restricted to the course direction, reproducing the primary loading mode experienced by the stocking when worn. The measured force–elongation response is used to determine circumferential stiffness, and may be converted into interface pressure using equilibrium relations based on Laplace’s law.



FIGURE 2.10: Tensile test on cylindrical samples of medical compression stockings [30].

Experimental studies consistently show that the tensile response of compression fabrics is nonlinear and exhibits pronounced hysteresis. Gaied et al. [47] reported that this behaviour reflects the progressive activation of the inlay yarns at increasing elongation levels. Piroi et al. [48] demonstrated that loop length and elastane feed tension strongly influence stiffness and hysteresis, while Kowalski et al. [22] confirmed the reproducibility of cylindrical testing for evaluating global fabric stiffness. Matsuo and Yamada [40] further showed that higher elastane pre-tension and shorter loop lengths systematically increase circumferential stiffness. Lozo et al. [39] related these trends to the structural design of graduated stockings, demonstrating that increased inlay density at the ankle explains the higher stiffness observed in this zone.

During testing, compression fabrics are typically extended over strain ranges between 20 and 120%, depending on the compression class and test objective. At low strains, deformation is dominated by loop rotation and bending, together with partial uncrimping of the inlay yarns, resulting in a compliant response governed mainly by geometric rearrangement [40]–[42]. At intermediate strains, the inlay yarns progressively straighten and carry an increasing fraction of the load, producing the characteristic nonlinear stiffening of the force–elongation curve [32], [39]. At high elongations, the response becomes dominated by elastic stretching of the yarns, particularly the elastane core.

According to NF S97-114:2022, medical compression stockings must withstand repeated extensions up to 120% without mechanical degradation or significant loss of pressure. Experimental studies confirm that fabrics remain structurally stable and elastically recoverable within this range [31], [33], [39], [40]. Beyond this threshold, irreversible mechanisms such

as yarn slippage or loss of elastane pre-strain may occur, leading to permanent softening and pressure decay. Under practical wearing conditions, however, local circumferential strains generally remain between 10 and 60%, particularly in the ankle region, ensuring operation within the elastic domain [2], [23], [39], [45].

Because compression fabrics are viscoelastic, time-dependent effects are a key aspect of their mechanical characterisation. Stress relaxation tests at fixed elongation show that most force decay occurs within the first minutes after loading, followed by a slower long-term decrease [35], [49]. Ilska et al. [36] demonstrated that relaxation may continue over several hours, explaining the gradual pressure loss observed during wear. Cyclic loading–unloading tests further reproduce in-use conditions and reveal hysteresis loops whose area reflects energy dissipation and internal friction. Progressive reduction of hysteresis over successive cycles indicates mechanical stabilisation of the fabric. Experimental studies recommend determining steady-state stiffness after several cycles, typically between five and ten, to represent realistic in-use behaviour [32], [40].

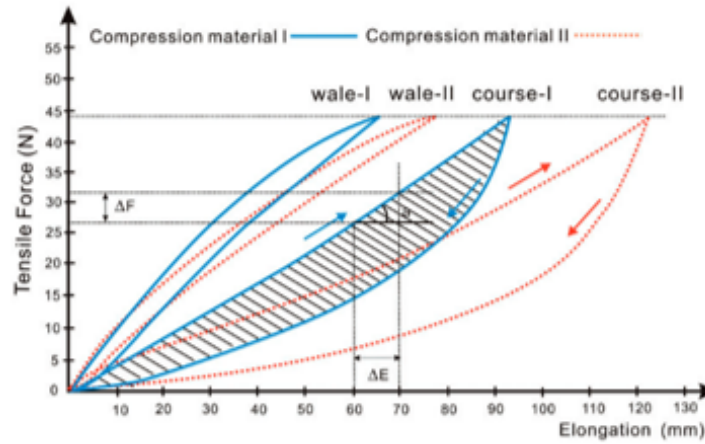


FIGURE 2.11: Force–Elongation curves of different knitted compression fabrics in wale and course directions [16].

Overall, cylindrical tensile testing provides a consistent description of the nonlinear, viscoelastic and zone-dependent behaviour of inlaid weft-knitted compression fabrics, making it the reference method for evaluating circumferential stiffness and pressure-related performance.

2.2.2.2 Cut-strip tensile tests

Cut-strip tensile tests are commonly used as a complementary method to characterise the directional mechanical properties of compression fabrics. In this approach, narrow rectangular specimens are tested the wale or course directions, as shown in Figure 2.12, allowing

independent assessment of the principal orthotropic responses. Chen et al. [33] showed that increasing inlay yarn density markedly enhances circumferential stiffness, while the ground-loop structure primarily governs extensibility. Zhang et al. [44] further demonstrated that elastane feed tension directly controls fabric stiffness and the resulting compression capability.

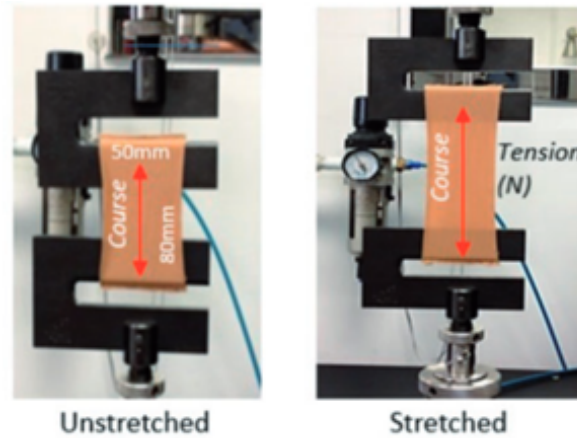


FIGURE 2.12: Cut-strip test on samples of medical compression stockings [23].

Despite their usefulness for comparative analysis and model parameter identification, cut-strip tests present significant limitations. Cutting the tubular structure alters loop continuity and boundary conditions, introducing edge effects that may underestimate circumferential stiffness. In addition, the uniaxial loading path does not reproduce the true circumferential deformation experienced by the stocking on the limb, and the measured forces cannot be directly converted into interface pressure. Siddique et al. [50] reported higher force–elongation values for cut-strip specimens compared with tubular samples tested under similar strain levels. Consequently, cut-strip tests are mainly employed for material comparison and constitutive modelling, while cylindrical testing remains the most representative method for in-service behaviour.

2.2.2.3 Biaxial tensile tests

Biaxial tensile testing has been introduced to characterise the coupled deformation occurring simultaneously in the wale and course directions of knitted compression fabrics [44]. By imposing controlled strain ratios along both axes, as shown in Figure 2.13, biaxial tests provide insight into load redistribution and yarn interaction under multiaxial extension. Jinyun et al. [72] used biaxial testing to determine directional elastic moduli and Poisson’s ratios, while Drapier and Gaied [73] combined biaxial experiments with inverse finite element analysis to

identify nonlinear anisotropic behaviour. More recently, Ye et al. [74] showed that incorporating biaxial data into orthotropic constitutive models improves the accuracy of interface pressure predictions compared with uniaxial approaches.



FIGURE 2.13: Biaxial test on cut-strip samples of medical compression stockings [44].

Although biaxial testing provides a more complete mechanical description, it remains experimentally demanding. Achieving uniform strain, avoiding slippage and controlling deformation require specialised equipment and complex protocols. As a result, biaxial tests are primarily used for research and model calibration, while cylindrical tensile tests remain the practical standard for routine characterisation of medical compression stockings.

2.2.2.4 Characterisation of fabric constituents

Several studies have investigated the mechanical response of the individual constituents of medical compression fabrics in order to isolate their respective roles in the global behaviour. These investigations focus on the looped ground structure and on the covered elastic yarns forming the reinforcement system.

Experimental studies on plain weft-knitted fabrics show that the loop network exhibits a nonlinear and anisotropic tensile response, governed at low strains by loop rotation and bending, followed by progressive stiffening as loops align under increasing elongation [43], [72]. Biaxial tensile tests further demonstrate that loop deformation mechanisms induce orthotropic behaviour and coupled responses between the wale and course directions [72]. In the context of medical compression stockings, these results indicate that the ground knitted structure mainly contributes to extensibility, conformability and comfort, while its contribution to circumferential stiffness and pressure generation remains limited compared with that of the inlay yarns [32], [39].

Covered elastane yarns, which constitute both the ground and inlay systems, have also been characterised independently through uniaxial and cyclic tensile tests. These studies

show that the elastane core governs the elastic response, while frictional interactions between the core and the covering filaments control hysteresis, stress relaxation and long-term pressure stability [26], [33], [35]. Yarn-level parameters, such as elastane feed tension during knitting, directly influence tensile stiffness and the compression delivered by the fabric under circumferential deformation [44].

In summary, mechanical characterisation studies consistently show that inlaid weft-knitted compression fabrics exhibit nonlinear, viscoelastic and strongly anisotropic behaviour. The inlay yarns dominate circumferential stiffness and pressure generation, while the looped ground structure controls extensibility and comfort. Variations in knitting parameters, inlay density and viscoelastic response lead to zone-dependent stiffness, which constitutes the mechanical basis of graduated compression profiles in medical compression stockings.

However, experimental characterisation does not allow direct prediction of fabric performance or pressure generated on the limb. To overcome this limitation, mechanical modelling approaches have been developed, relating structural parameters and material properties to fabric tension and compression delivery. This is reviewed in the following section.

2.2.3 Mechanical modelling of medical compression stockings

Experimental tests provide valuable information on the mechanical behaviour of compression fabrics, but their results remain limited to specific samples and test conditions. To generalise this knowledge and predict fabric performance in a wider range of scenarios, different mechanical models have been proposed. Their goal is to relate yarn-level mechanics and fabric architecture to macroscopic response, enabling prediction of stiffness and interface pressure from material and structural parameters.

2.2.3.1 Homogenisation models

Homogenisation models describe medical compression fabrics as continuous membranes whose global mechanical behaviour is defined by averaged parameters. These models bridge experimental characterisation and analytical or finite element (FE) simulations, enabling the prediction of interface pressure without explicitly modelling the yarn architecture. Depending on the degree of simplification, three principal approaches are distinguished: isotropic, orthotropic, and hyperelastic formulations.

Isotropic constitutive models The earliest homogenised representations of medical compression fabrics assumed isotropic elastic behaviour. In this case, the fabric was modelled as a membrane with uniform stiffness in all directions, defined by a single modulus of elasticity

identified from uniaxial tensile tests. The material is typically considered under plane stress, which is appropriate for thin knitted membranes such as medical compression stockings. The constitutive relation for a thin isotropic sheet under plane stress can be expressed as given in Equation 2.2, where E and ν are the Young's modulus and Poisson's ratio, respectively. This linear formulation defines the overall stiffness of the fabric but neglects its directional anisotropy.

$$\begin{bmatrix} \sigma_x \\ \sigma_y \end{bmatrix} = \frac{E}{1 - \nu^2} \begin{bmatrix} 1 & \nu \\ \nu & 1 \end{bmatrix} \begin{bmatrix} \varepsilon_x \\ \varepsilon_y \end{bmatrix} \quad (2.2)$$

Mirjalili et al. [75] modelled compression garments as isotropic elastic membranes fitted on simplified limb geometries. Tensile strip tests provided the modulus of elasticity, which was implemented in finite element analyses. Although predictions provided global average pressures, local variations around bony prominences were underestimated.

Wu et al. [76] adopted a similar strategy for seamless tight garments, deriving isotropic moduli from uniaxial tensile tests. Their FE analysis showed reasonable agreement with measured pressures on cylindrical mannequins but limited accuracy in regions with strong curvature.

More recently, Hedigalla et al. [59] used isotropic membrane models to investigate pressure propagation from garments into soft tissues. Validation against pneumatic sensors confirmed correct average values, but errors were observed in highly curved anatomical zones.

Overall, isotropic homogenisation provides computationally efficient models but fails to account for the directional properties of knitted fabrics.

Orthotropic constitutive models To better represent the mechanical behaviour of knitted compression fabrics, later studies introduced orthotropic constitutive models distinguishing between wale and course directions. This approach provides a macroscopic description of the fabric response by averaging the effect of its knitted architecture while maintaining directional dependence. The in-plane stress–strain relationship for an orthotropic membrane under plane stress can be expressed as given in Equation 2.3, where σ_1 and σ_2 are the normal stresses along the wale and course directions, respectively, and τ_{12} is the in-plane shear stress.

The coefficients Q_{ij} denote the stiffness terms of the fabric and are related to the orthotropic elastic constants as given in Equation 2.4, where E_1 and E_2 are the Young's moduli in the wale (longitudinal) and course (circumferential) directions, ν_{12} and ν_{21} are the Poisson's ratios, and G_{12} is the in-plane shear modulus.

$$\begin{bmatrix} \sigma_1 \\ \sigma_2 \\ \tau_{12} \end{bmatrix} = \begin{bmatrix} Q_{11} & Q_{12} & 0 \\ Q_{12} & Q_{22} & 0 \\ 0 & 0 & Q_{33} \end{bmatrix} \begin{bmatrix} \varepsilon_1 \\ \varepsilon_2 \\ \gamma_{12} \end{bmatrix} \quad (2.3)$$

$$Q_{11} = \frac{E_1}{1 - \nu_{12}\nu_{21}}, \quad Q_{22} = \frac{E_2}{1 - \nu_{12}\nu_{21}}, \quad Q_{12} = \frac{\nu_{12}E_2}{1 - \nu_{12}\nu_{21}}, \quad Q_{33} = G_{12} \quad (2.4)$$

The equivalent compliance form is given in Equation 2.5.

$$\begin{bmatrix} \varepsilon_1 \\ \varepsilon_2 \\ \gamma_{12} \end{bmatrix} = \begin{bmatrix} \frac{1}{E_1} & -\frac{\nu_{12}}{E_1} & 0 \\ -\frac{\nu_{21}}{E_2} & \frac{1}{E_2} & 0 \\ 0 & 0 & \frac{1}{G_{12}} \end{bmatrix} \begin{bmatrix} \sigma_1 \\ \sigma_2 \\ \tau_{12} \end{bmatrix} \quad (2.5)$$

Early contributions by Drapier and Gaied [73], [77] used biaxial tensile tests on elastomeric knits to determine Young's moduli and Poisson's ratios in both directions. These data were used in orthotropic membrane models, which predicted stocking pressures under circumferential extension with greater accuracy than Laplace's law. Validation against measurements on wooden legs showed that orthotropic formulations better reproduced fabric anisotropy.

Liu et al. [78] extended this concept by proposing a multiscale homogenisation approach, linking loop geometry to effective continuum parameters. Their approach enabled prediction of orthotropic responses across different knit architectures, providing a structural interpretation of the constitutive laws.

Lu et al. [79] developed 3D FE simulations of stocking-limb interaction based on orthotropic membrane laws, emphasising the importance of including anisotropy for accurate transmission of pressure into soft tissues.

Recent studies have further refined orthotropic modelling for medical compression stockings. Ye et al. [74] developed a three-dimensional orthotropic constitutive framework calibrated from biaxial tensile tests. The stockings were applied on rigid leg surrogates, and predicted pressures showed closer agreement with mannequin measurements than Laplace-based estimates.

Ghorbani et al. [80] also modelled compression stockings as orthotropic shells, identifying parameters from uniaxial and biaxial mechanical tests. Validation against Kikuhime sensor measurements confirmed good predictive accuracy for global pressure levels, though local heterogeneities remained difficult to reproduce.

Dai et al. [81] applied orthotropic formulations to graduated stockings, demonstrating that the experimentally observed pressure gradient could be predicted by varying moduli between zones, consistent with structural differences in inlay density.

Orthotropic constitutive laws thus provide an efficient and physically grounded way to represent the mechanical behaviour of knitted compression fabrics. They capture the directional stiffness introduced by inlay yarns and enable parameter identification through standard biaxial or uniaxial tests. However, they still approximate the fabric as a homogeneous layer, averaging out local structural features such as loop geometry or yarn friction.

Hyperelastic and non-linear laws More recent works have incorporated nonlinear and anisotropic hyperelastic laws to represent the large-strain behaviour of compression fabrics more accurately. These models extend orthotropic elasticity by defining the strain-energy function W , given in Equation 2.6, as a function of both isotropic and anisotropic invariants, often following formulations derived for fibre-reinforced composites. In this case, I_1 is the first invariant of the right Cauchy–Green tensor representing the matrix deformation, I_4 corresponds to the stretch along the principal reinforcement direction (inlay yarns), and C_1 , k_1 , k_2 and are material constants identified from experimental data.

$$W = \frac{C_1}{2}(I_1 - 3) + \frac{k_1}{2k_2} \left[\exp \left(k_2(I_4 - 1)^2 \right) - 1 \right] \quad (2.6)$$

Ye et al. [24] employed this type of hyperelastic orthotropic formulation to simulate the mechanical response of medical compression stockings under large deformations. The model parameters were fitted from biaxial test data and validated against PicoPress® measurements on patient-specific leg geometries, achieving deviations below 10% between predicted and measured pressures. The study demonstrated that including nonlinear anisotropic elasticity improves local accuracy, particularly in regions with high curvature such as the ankle and tibial crest. Similar conclusions were drawn by Lu et al. [69], who used hyperelastic orthotropic shells in FE analyses to evaluate the interaction between compression fabrics and multilayer leg models, confirming that nonlinear laws capture both the amplitude and gradient of interface pressure more faithfully than linear models.

These hyperelastic anisotropic formulations currently represent the most advanced level of homogenised material modelling for compression fabrics. They accurately reproduce the stress–strain behaviour under large deformations and provide reliable inputs for FE simulations of realistic garment–limb interactions. Nevertheless, they still rely on homogenisation of the looped structure and do not explicitly separate the mechanical roles of the ground

and inlay yarn systems. This motivates the development of more detailed models, such as geometric or hybrid formulations, where ground and inlay yarns are explicitly represented.

2.2.3.2 Mesoscopic and unit cell models

Mesoscopic or unit-cell models aim to represent the mechanical behaviour of knitted fabrics by describing a representative volume element (RVE) that includes the main structural features of the loops and yarns. Instead of reconstructing the entire fabric at the yarn scale, these models define a reduced periodic geometry that preserves the essential load-transfer mechanisms, allowing efficient numerical analysis while maintaining structural fidelity. They form an intermediate scale between yarn-resolved and homogenised models, linking local geometry to macroscopic response.

The hybrid finite element approach introduced by Tephany et al. [82] was originally developed for woven composites, but it established key methodological principles that were later transferred to knitted architectures. In their model, the woven matrix was represented by shell elements, while the reinforcing yarns were modelled as connector or beam elements oriented along the warp and weft directions. Mechanical parameters were derived from tensile and biaxial tests, and validation against macroscopic stress–strain curves confirmed that the model reproduced the anisotropic stiffness of the fabric in both directions. Representing reinforcement yarns separately from the matrix significantly improved prediction of the fabric response under multiaxial loading while maintaining a substantially lower computational cost than fully yarn-resolved models.

Nasri et al. [83], [84] further refined this hybrid formulation to describe the nonlinear anisotropic behaviour of woven composites at the mesoscopic scale. In their model, the resin matrix was described by deformable shell elements, while warp and weft yarns were represented by connector or beam elements transmitting load along their actual orientations. Material parameters were identified from uniaxial and biaxial tests performed on representative unit-cell specimens. Validation against experimental stress–strain curves demonstrated that the hybrid model reproduced the nonlinear stiffness and coupling effects observed in woven fabrics, while achieving an excellent balance between accuracy and computational efficiency. Although originally developed for woven reinforcements, this modelling strategy is directly transferable to inlaid weft-knitted compression fabrics, where the ground loops and inlay yarns act as structural analogues of the matrix and reinforcement phases, respectively.

Najjar et al. [85] applied a similar multiscale hybrid approach to architected textiles. In their formulation, the matrix phase was modelled using shell elements, while the reinforcing yarns were represented by truss or connector elements with unidirectional stiffness.

Yarn-level properties were measured experimentally and implemented within a unit-cell framework. Comparison with biaxial test results confirmed that the model accurately reproduced anisotropic stiffness and shear coupling, effectively distinguishing the mechanical contributions of the matrix and the reinforcement yarns.

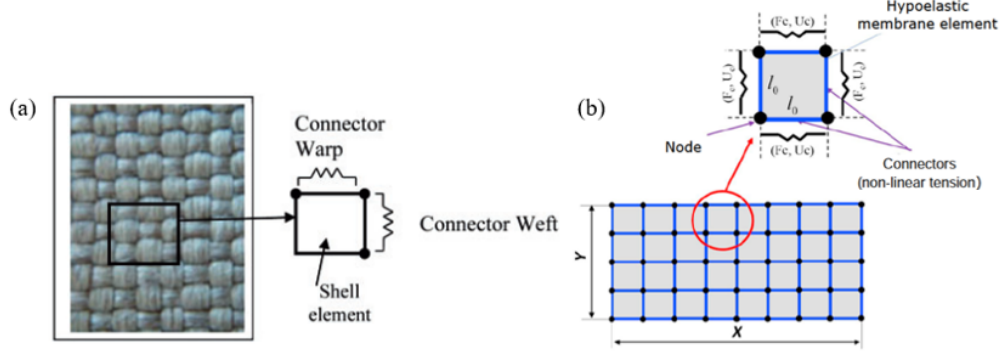


FIGURE 2.14: Hybrid unit cell models of woven fabrics, combining shell elements and connectors: (a) presented by Tephany et al. [82]; (b) presented by Nasri et al. [83].

Wadekar et al. [86] introduced an optimised geometric model of weft-knitted fabrics for finite element analysis. The model defined a representative unit cell in which yarn paths were described by continuous spline curves fitted to micro-computed tomography (μ CT) data. Contact and friction between adjacent yarns were simplified using equivalent constraint conditions, which reduced computational cost while maintaining realistic deformation modes. Tensile simulations performed on the unit cell reproduced the nonlinear load–extension response of knitted fabrics with good agreement to experimental data, confirming that mesoscopic representation can model the anisotropy and nonlinearity characteristic of looped structures.

Ghorbani et al. [87] developed a finite element unit-cell model to analyse the mechanical behaviour of inlaid weft-knitted fabrics used in medical compression stockings, presented in Figure 2.15. The periodic unit cell reproduced both the rib ground structure and the transverse inlay yarn, whose geometry was reconstructed from microscopic observations. Each yarn was discretised with 3D beam elements, with material laws identified from tensile tests on extracted covered elastic yarns.

The model simulated circumferential and longitudinal deformations under periodic boundary conditions, computing the mesoscopic stress–strain response and load sharing between yarn systems. Results matched experimental tensile curves, confirming that inlay yarns dominate circumferential stiffness, while ground loops control extensibility and recovery.

This unit-cell approach linked yarn-scale mechanics with fabric-level anisotropy, although the explicit 3D representation limited its scalability to large fabric sections.

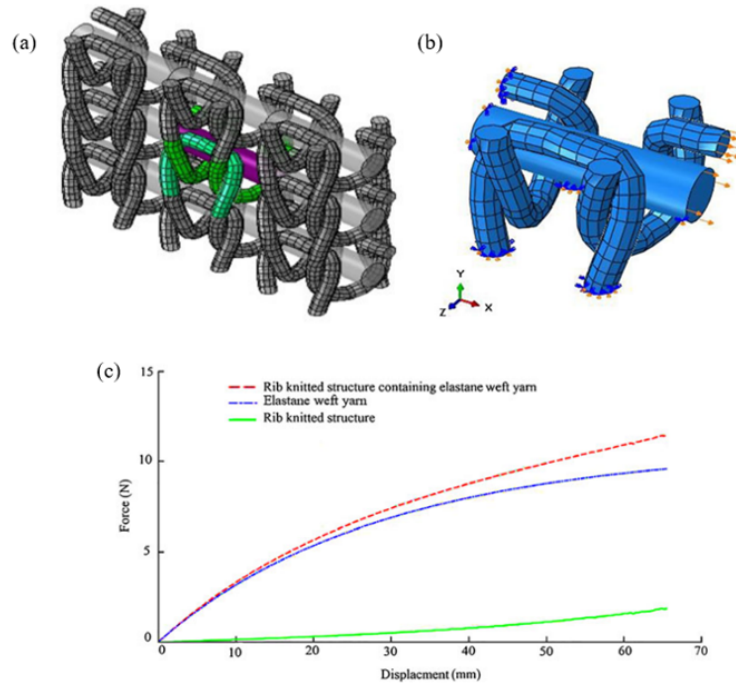


FIGURE 2.15: Unit cell model of medical compression stockings, developed by Ghorbani et al.: (a) Simulated knitted fabric; (b) The boundary conditions applied on the unit cell; (c) The force-extension curve of the elastic weft knitted fabric and its components calculated by FEM [87].

Together, these mesoscopic and hybrid models demonstrate the ability of unit-cell formulations to reproduce the anisotropic and nonlinear behaviour of complex textile architectures while remaining computationally efficient. By explicitly differentiating between ground and reinforcement yarns, they establish a direct relationship between structural parameters, such as loop length, inlay density, and yarn stiffness, and the global mechanical response.

2.2.3.3 Geometric models

Geometric models explicitly describe the looped architecture of knitted fabrics, linking yarn trajectories and loop morphology to macroscopic mechanical behaviour. In the case of medical compression stockings, these approaches make it possible to analyse how yarn geometry, inlay density, and loop dimensions influence circumferential stiffness and the pressure exerted on the limb. Unlike homogenisation models, which approximate the fabric as a continuous orthotropic material, geometric models provide a direct description of the structural origin of the mechanical response.

One of the earliest contributions was made by De Araujo et al. [43], [88], who analysed plain weft-knitted fabrics without inlay yarns. Their model defined loop geometry in three dimensions and applied classical beam elasticity theory to calculate yarn bending and extension. Load–extension curves were obtained along the wale and course directions and validated against uniaxial tensile tests, showing good agreement. These results demonstrated that loop morphology alone is sufficient to reproduce the nonlinear and anisotropic stress–strain behaviour typical of knitted fabrics. Although these studies were not directly focused on compression stockings, they introduced a methodology that was later applied to inlaid weft-knits.

Htoo et al. [42] analysed the morphology of ground loops in weft-knitted structures using 3D geometric descriptions. From this representation they calculated loop length, yarn consumption, and contact points between yarns. Circumferential stiffness was obtained as the sum of yarn tensions along their trajectories.

Subsequent work integrated the role of inlay yarns, which are essential in compression stockings. Fong et al. [41] developed a three-dimensional model of inlaid weft-knitted fabrics, describing the spatial trajectories of both ground and inlay yarns. Their study quantified loop elongation and the distribution of inlay reinforcement under strain. Results showed that inlay density and feeding tension directly control circumferential stiffness and therefore the pressure delivered by compression stockings. Validation against mannequin pressure measurements confirmed that geometric parameters explain the differences observed between ankle, calf, and thigh regions.

Geometric models thus provide a structural link between yarn-level design and the global mechanical behaviour of stockings. They also emphasise the dominant role of inlay yarns in controlling circumferential stiffness and demonstrate how variations in loop density or inlay configuration modify the delivered pressure. Their main limitation is the computational complexity involved in reconstructing large fabric sections in three dimensions, which generally restricts their application to representative samples.

In summary, geometric approaches complement homogenisation laws by explicitly relating yarn geometry to fabric behaviour. They demonstrate how structural parameters such as loop length, course density, and inlay configuration determine circumferential stiffness, offering valuable insights for the design and optimisation of medical compression stockings.

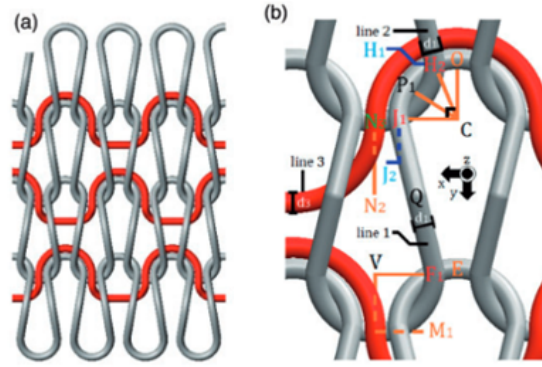


FIGURE 2.16: Geometrical model of inlaid weft-knitted fabrics, developed by Fong et al. (a) Technical front of the knitted structure; (b) Model of a laid-in weft-knitted loop unit [41].

Overall, mechanical models of medical compression stockings have evolved from simplified homogenised formulations to mesoscopic and yarn-resolved representations that account for the specific role of ground and inlay yarns. These developments have progressively improved the description of anisotropy and nonlinear behaviour, enabling more realistic simulations of the fabric's mechanical response. Overall, the circumferential stiffness, and therefore the delivered compression, is mainly governed by the density and properties of the inlay yarns, modulated by the deformability of the ground loops. These models form the mechanical basis for subsequent studies, where the predicted fabric behaviour is coupled with limb geometry to estimate the resulting interface pressure.

2.3 Methods for determining and predicting interface pressure

Accurate prediction of interface pressure is essential to ensure the therapeutic performance of medical compression stockings. Over the years, different experimental and numerical strategies have been developed to measure, estimate, and simulate the pressure applied by compression fabrics on the limb. These methods can be grouped into three main categories: experimental measurements, which provide direct pressure data; indirect analytical methods, which derive pressure from tensile behaviour and geometric relations such as Laplace's law; and numerical methods, based on finite element (FE) modelling of the stocking-limb interaction.

Each approach contributes complementary insights: experimental techniques validate models and provide reference data under controlled conditions, analytical conversions relate

tensile behaviour to pressure through simplified assumptions, and FE simulations enable detailed analysis of garment mechanics on realistic leg geometries. Together, they form a coherent framework for linking textile mechanics with delivered compression.

2.3.1 Experimental measurements

Experimental methods provide direct or indirect measurements of interface pressure, allowing the design and validation of medical compression stockings. Pressure sensors and leg mannequins are used to obtain local values of compression in controlled conditions, while standardised tensile tests provide circumferential stiffness data for indirect estimation of pressure. These measurements are not only essential for clinical evaluation but also serve as benchmarks against which analytical and numerical models are calibrated and validated. For instance, Chi et al. [89] compared Laplace-based predictions with PicoPress[®] readings on cylindrical legs, showing that direct measurements are essential for assessing the accuracy of analytical models.

Also, Kowalski et al. [22] validated analytical estimations of stocking pressure against experimental data obtained with PicoPress[®] sensors on rigid mannequins and subjects.

In addition, Shi et al. [53] combined 3D body scanning and tensile testing to predict interface pressure. They further validated their numerical simulations against PicoPress[®] measurements on mannequins and volunteers, confirming the relevance of experimental data in the development of numerical models.

2.3.1.1 Pressure sensors

Direct pressure sensors are the most established tools for quantifying the interface pressure delivered by medical compression stockings, both in clinical and laboratory environments [90]. Their use provides essential experimental data for the validation of analytical and numerical models of compression, serving as the reference method for evaluating garment performance. However, it is important to note that these sensors provide local point measurements, which are highly sensitive to placement, contact conditions, and curvature of the underlying surface.

Different sensor technologies have been developed, each with specific characteristics of sensitivity, response time, and applicability to static or dynamic measurements:

- PicoPress[®] is the most widely used pneumatic transducer for in-vitro and in-vivo assessments. It consists of a small air-filled chamber connected to a manometer. The device offers good sensitivity in the range of 10–100 mmHg, with a resolution of about

1 mmHg and a fast dynamic response suitable for monitoring limb movement. However, because of its small probe diameter (about 2–3 cm) and air layer thickness ($\approx$ 2 mm), the measurement reflects a local average over a small surface and can be influenced by positioning, garment slippage, and limb curvature. PicoPress[®] has been used extensively as a validation method for Laplace-based and FE models [16], [70], [89], showing excellent repeatability, especially at medium and high compression levels.

- Piezoresistive film sensors (e.g. Tekscan[®], FlexiForce[®]) are based on thin polymer films whose electrical resistance decreases with applied pressure. Their main advantage is their flexibility and minimal thickness (<0.2 mm), which reduces local disturbance under the garment. They allow mapping of pressure at multiple points simultaneously, making them useful for dynamic studies. However, they present lower accuracy and higher signal drift than pneumatic devices, particularly at low pressures (<20 mmHg). Chi et al. [89] compared piezoresistive films with PicoPress[®] and observed higher variability, especially in curved regions, confirming that calibration and uniform contact are critical for reliable use.
- Pneumatic sensors (AMI systems) operate on the same physical principle as PicoPress[®], using air-filled bladders that deform under external compression. Their response is stable and reproducible, and they can be miniaturised to reduce interference. Nandasiri et al. [13], [67] evaluated mini-bladder pneumatic systems as an alternative to PicoPress[®], and showed that they maintain linear response and sensitivity at low pressures where other sensors underperform. These systems are particularly suited for validating finite element simulations of heterogeneous stockings or for monitoring low-pressure garments such as lymphoedema wraps.
- Kikuhime[®] sensors consist of small air pads connected to a portable digital reader. They are widely used in clinical applications because of their simplicity, robustness, and portability. Their resolution (≈ 2 mmHg) is adequate for medical use, but the probe's air layer thickness introduces an offset similar to that of PicoPress[®]. Ghorbani et al. [80] used Kikuhime to monitor temporal pressure decay during wear and found a 15–25% reduction within the first hour, which they attributed to fabric stress relaxation. This demonstrates the usefulness of Kikuhime for time-dependent validation of viscoelastic models.
- Medical Stocking Tester (MST, Salzmänn AG) is a bench-top instrument designed for standardised laboratory evaluation. It measures circumferential pressure on stockings

mounted over rigid cylinders equipped with multiple integrated transducers. The MST is mainly used for quality control and durability testing, rather than clinical monitoring. Jamshaid et al. [68] and Siddique et al. [50] employed it to measure pressure distribution and assess the effect of repeated washing, reporting that washing reduces fabric stiffness and downgrades compression class. Its rigid setup provides highly reproducible but non-physiological results.

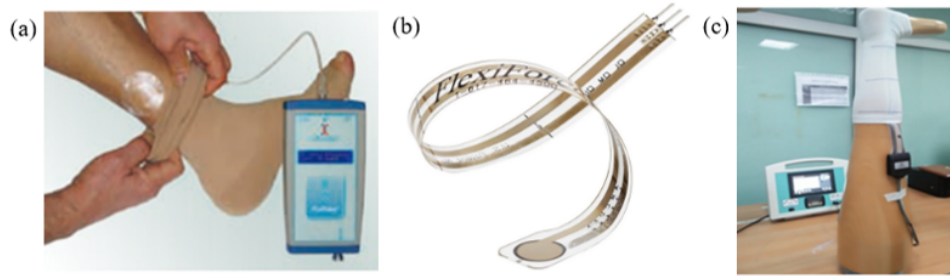


FIGURE 2.17: Pressure sensors used in the literature (a) PicoPress® by [30], [91]; (b) Flexiforce ®[67]; (c) MST [68]

Across devices, sensitivity typically ranges between 0.5 and 2 mmHg, and accuracy depends strongly on calibration, curvature, and local skin-sensor friction. Jariyapunya and Blažena [91] demonstrated through experiments and modelling that sensors with larger thickness overestimate interface pressure, particularly at low garment tensions, emphasising the need for calibration and appropriate probe selection.

Despite their limitations, pressure sensors remain indispensable for evaluating the mechanical action of compression garments. They provide quantitative and reproducible local measurements that serve as the experimental basis for validating analytical and FE models, as well as for standardising compression classes in clinical and industrial practice.

2.3.1.2 Leg mannequins

Leg mannequins are widely employed to reproduce the geometry and loading conditions of the lower limb during compression. They provide controlled and repeatable environments for evaluating the mechanical performance of medical compression stockings and serve as essential tools for validating analytical and numerical models [66], [92], [93].

Depending on their design, mannequins can be divided into two main categories:

- Conventional mannequins, rigid or deformable, used in combination with external pressure sensors that provide local measurements

- Smart mannequins, which integrate embedded sensors to record distributed pressure fields over the entire surface.

Conventional mannequins combined with pressure sensors Rigid mannequins, typically made of wood, plastic, or composite materials, represent the simplest experimental configuration. They reproduce standardised leg circumferences and are used to ensure repeatability of measurements under well-controlled extension conditions. In Europe, the Hohenstein wooden leg remains the reference form for product classification and standard testing of compression classes [29]. Stockings are fitted onto the leg, and interface pressure is measured at predefined anatomical landmarks, such as the ankle, calf, and thigh, using devices like PicoPress® or Kikuhime®. This configuration provides highly reproducible values and is essential for comparing products across laboratories. However, the rigid geometry does not replicate the compliance of human tissues, leading to overestimation of local pressures at regions of high curvature, such as the malleoli or tibial crest.

To improve physiological realism, deformable mannequins have been developed. Maklewska et al. [28] designed the Textilpress device, consisting of cylindrical legs of different diameters covered with a thin neoprene layer to mimic tissue elasticity. The stockings were mounted under controlled extension, and pressure was recorded with pneumatic probes. Results showed a linear relationship between fabric elongation, generated tension, and measured pressure, consistent with Laplace's law, and the setup provided one of the first reproducible procedures for laboratory evaluation of compression. Hirai and Partsch [94] introduced a mechanical mannequin with adjustable circumference, allowing progressive radial expansion to simulate limb swelling or reduction. By combining pressure measurements with tensile data, they demonstrated a direct correlation between fabric stiffness, limb geometry and pressure evolution.

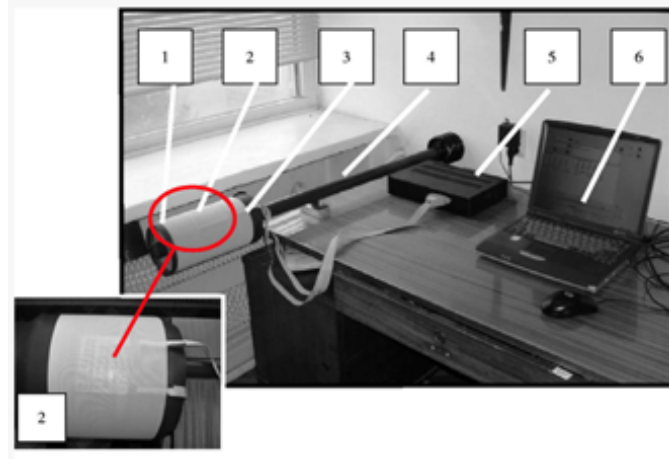


FIGURE 2.18: View of the Textilpress measuring stand during measuring, developed by Maklewska et al.: (1) Textile compression band; (2) Measuring matrix placed under the compression band; (3) Cylinder; (4) Cylinder holder; (5) Strain gauges amplifier; (6) Computer [28].

Wu and Li [66] compared rigid and compliant mannequins, showing that rigid forms systematically produced higher local pressures, particularly near curved anatomical regions. Their findings emphasised the need for compliant surrogates to obtain pressure distributions closer to in-vivo conditions. More recently, Li et al. [92] developed a soft-tissue phantom composed of a silicone gel layer over a rigid bone-like core. When fitted with medical compression stockings, this surrogate produced pressure maps closely matching those observed in clinical tests, confirming its suitability as an intermediate step between bench-top and in-vivo measurements.

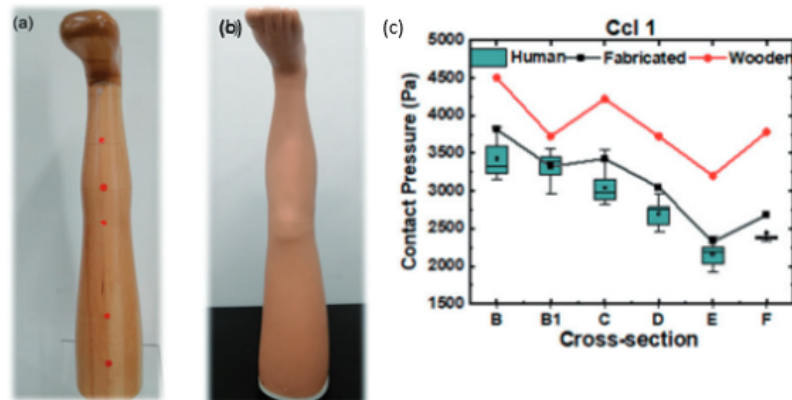


FIGURE 2.19: Comparison of the contact pressure measured by different lower limbs at different pressure levels: (a) Hohenstein wooden leg; (b) Fabricated leg mannequin developed by Li et al.; (c) contact pressure distribution on the different regions measured by different leg mannequins (the fabricated leg mannequin and the wooden leg mannequin) and human legs [92].

In these conventional systems, pressure measurement remains local, since external probes

are positioned at discrete sites. As a result, data quality strongly depends on sensor placement, curvature, and fabric slippage. Nevertheless, their simplicity and reproducibility make these mannequins indispensable for standardisation and model validation.

Overall, experimental methods remain fundamental for quantifying and validating the pressures generated by medical compression stockings. Pressure sensors provide accurate local measurements, while mannequin-based tests offer controlled and repeatable conditions for evaluating garment performance. To relate fabric mechanics and limb geometry to measured pressure, analytical methods based on Laplace's law have been developed, converting tensile-derived fabric tension into theoretical estimates of interface pressure.

2.3.2 Indirect methods based on Laplace's law

Indirect methods estimate the interface pressure of medical compression stockings from the mechanical response of the fabric and the deformation it undergoes when fitted onto the limb. These methods rely on analytical relations that transform circumferential tension measured experimentally, into interface pressure, assuming equilibrium of circumferential stresses in a thin cylindrical layer. Because they are simple, reproducible, and standardised, they are widely used for product classification, quality control, and model validation.

Experimental input In these methods, fabric tension is first measured in controlled tensile tests. The reference European protocol, NF G30-102-B [46], prescribes uniaxial circumferential tensile testing of tubular knitted specimens cut from the ankle, calf, or thigh zones of compression stockings. Each cylinder is mounted on a rigid frame equipped with axial pins to maintain constant length, ensuring deformation occurs only in the circumferential direction. Tests are performed on a conventional tensile machine under constant-rate extension, and the applied force is continuously recorded as the circumference increases. Before testing, specimens are conditioned according to EN ISO 139 [71] to standardise humidity and temperature.

The circumferential strain imposed during testing is selected to reproduce the typical deformation of the garment in use, corresponding to the mismatch between the stocking's unstretched circumference and the leg's perimeter. The measured force per unit specimen length provides the circumferential tension (T), which is the input for analytical conversion into interface pressure. This relation provides a direct link between mechanical behaviour obtained in laboratory tests and the pressure expected when the fabric is fitted onto a limb of known geometry.

Analytical conversion Once fabric tension (T) is known for a given leg perimeter, interface pressure IP can be calculated using Laplace's law, which assumes uniform circumferential tension in a thin cylindrical shell wrapped around the leg. Laplace's law is given in Equation 2.7, where ε_{circ} is the circumferential strain of the fabric, E is the Young modulus of the fabric, P_0 the initial fabric perimeter, and P is the deformed fabric perimeter.

$$IP = \frac{T}{R} = \frac{\varepsilon_{circ} \cdot E}{R} = \frac{(P - P_0) \cdot E}{P_0 \cdot R} \quad (2.7)$$

In the NF G30-102-B procedure, the equivalent interface pressure is computed directly from the measured tensile force F_2 , the effective fabric length (L), and the deformed fabric perimeter P , as given in Equation 2.8, where the factor 10 corresponds to the unit conversion to hectopascals. The result can be expressed in mmHg by dividing by 1.333.

$$IP = \frac{F_2}{L} \frac{2\pi}{P} \cdot 10 \quad (2.8)$$

Several studies have applied this method to characterise medical compression stockings and to validate analytical or numerical predictions. Gaied et al. [47] performed circumferential tensile tests on commercial stockings and compared the derived pressures with experimental measurements and simulations. They reported nonlinear force–elongation curves with pronounced hysteresis, consistent with the viscoelastic nature of knitted fabrics, and found good agreement with analytical estimates at moderate extensions.

Piroi et al. [48] evaluated axial tensile tests in order to assess the presence of a gradual compression along the stocking. They observed that circumferential stiffness and hysteresis depend strongly on loop geometry and elastane feed tension, and confirmed that the tensile response provides a reliable indicator of stocking behaviour.

Kowalski et al. [22] used the tensile test procedure to compute interface pressures from tensile data and compared them with mannequin and in-vivo measurements. They observed good agreement in global compression levels but significant deviations near curved anatomical regions, emphasising that the cylindrical assumption of Laplace's law does not account for local curvature effects.

Lun et al. [95] applied Laplace's law to stockings tested on cylindrical and anatomically inspired models to analyse the relationship between circumferential tension and limb curvature. Their results confirmed that Laplace-based predictions match experimental pressures at the ankle, where the geometry is close to cylindrical, but diverge in areas with high curvature or irregular cross-sections. This study reinforced that the classical Laplace relation

provides reliable first-order estimates but must be refined to represent anatomical complexity.

Further improvements were introduced through modified Laplace formulations. Teyeme et al. [96] incorporated knit geometry and fabric extensibility into the Laplace relation, achieving better agreement with PicoPress® measurements in experimental validation. Similarly, Zhang et al. [44] included structural parameters such as inlay yarn density and elastane feed tension, demonstrating improved correlation between predicted and measured pressures for graduated stockings.

Other studies have explored time-dependent effects. Mikucionienė and Muralienė [35], [49] and Ilska et al. [36] measured stress relaxation during constant-extension tests, showing that forces stabilise only after 120–300 s and that long-term relaxation reduces delivered pressure during wear. These findings underline the importance of viscoelastic effects when interpreting tensile data.

More recently, Shi et al. [53] combined 3D body scanning with tensile testing to estimate patient-specific interface pressure. Geometric parameters such as limb curvature and circumference were extracted from scans and combined with tensile-derived tensions through Laplace's law, improving the accuracy of predictions for individual morphologies.

Indirect methods based on Laplace's law provide a reproducible, standardised, and cost-effective approach for estimating interface pressure. They enable the mechanical response of compression fabrics, governed by structure and yarn composition, to be linked directly to the delivered pressure under controlled strain. This approach is particularly valuable for product classification and validation, where cylindrical geometry assumptions remain acceptable.

However, the simplicity of Laplace-based formulations also imposes limitations. They assume uniform circumferential tension, constant curvature, and negligible friction, overlooking the heterogeneous deformation of the fabric and the complex three-dimensional anatomy of the limb. These simplifications prevent the accurate prediction of localised pressure peaks and the redistribution of stresses across soft tissues. Furthermore, viscoelastic effects and heterogeneous stiffness between garment zones cannot be represented through analytical relations alone.

To overcome these limitations, numerical models based on finite element (FE) simulations have been developed. These models explicitly represent the mechanical interaction between the stocking and the limb, integrating material properties, geometry, and contact conditions within a unified computational approach. By coupling experimental characterisation of the fabric with realistic limb geometries, FE simulations allow the analysis of how

pressure is distributed, how tension evolves during wear, and how design parameters influence the mechanical and therapeutic performance of compression garments.

2.3.3 Numerical methods based on FE models

Finite element (FE) simulations have become essential tools for analysing and predicting the mechanical behaviour of medical compression stockings. In contrast to analytical formulations derived from Laplace's law, which assume simplified cylindrical geometry and uniform boundary conditions, FE models describe the physical interaction between the stocking and the limb in a more realistic manner. They take into account the actual leg morphology, the mechanical behaviour of the fabric and soft tissues, and the contact conditions at the skin–fabric interface [16], [19], [24], [25], [53], [58], [97].

The FE approach allows the integration of experimental data such as the tensile or biaxial response of the fabric, 3D leg geometries obtained from scanning or imaging, and boundary conditions representing garment placement. By solving mechanical equilibrium between the textile layer and the underlying tissues, these simulations provide a spatially resolved view of the interface pressure, local deformation, and internal stress distribution [19], [53], [59].

The following sections review the main strategies used in finite element analysis of compression stockings. The first part discusses the modelling of the limb, from rigid simplified shapes to deformable or patient-specific geometries. The second examines the modelling of the stocking, distinguishing between simulations where compression is imposed as an external pressure field and those where the fabric is represented as a deformable thin structure in mechanical contact with the leg.

2.3.3.1 Models of the limb

The representation of the limb is a key aspect in finite element (FE) simulations of medical compression stockings, as the geometry and mechanical properties of biological tissues strongly influence how pressure is transmitted from the garment to deeper structures. Over the years, three main approaches have been used: rigid geometries, deformable legs with simplified elasticity, and anatomically realistic models reconstructed from medical imaging or 3D scanning.

Rigid simplified models Early studies represented the leg as a rigid cylindrical or conical surface. This approach allowed the effect of fabric tension and curvature on interface pressure to be analysed independently from tissue deformation. Maklewska et al. [28] and

Yamada et al. [27] validated Laplace-based predictions using rigid cylindrical forms, demonstrating that interface pressure increases with local curvature, and that the ankle region consistently experiences the highest compression.

In most numerical studies, the leg is represented either by an analytical rigid surface or by a coarse volumetric mesh, and the simulations are conducted under quasi-static implicit conditions, since no tissue deformation is computed. Karakashian et al. [25] implemented numerical models in which uniform pressure boundary conditions were applied to rigid surfaces to explore the curvature–pressure relationship. The leg was represented as a rigid cylindrical or conical surface meshed with triangular surface elements (S3) in an implicit quasi-static simulation. This configuration enabled a direct evaluation of how local curvature influences the pressure field without including tissue deformation. Their results confirmed that rigid surface models reproduce average pressure magnitudes but cannot capture the redistribution of stresses within soft tissues, since no volumetric response is represented.

Rigid representations remain useful for benchmarking and validation of material laws, particularly when the goal is to assess the mechanical response of the stocking independently of tissue deformation. However, they overestimate local pressure peaks in highly curved regions and neglect the deformation and heterogeneity of real limbs.

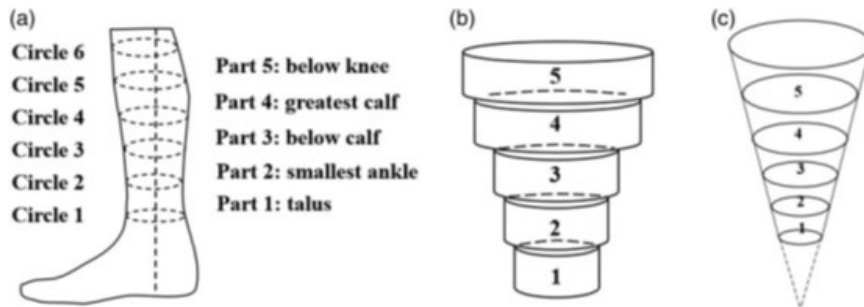


FIGURE 2.20: Model of the lower limb presented by Zhang et al.: (a) separated parts of the lower limb; (b) cylinder model; (c) conical model [44].

Deformable simplified models To overcome these limitations, deformable approaches have been introduced to reproduce the elastic response of biological tissues.

Hedigalla et al. [59] proposed a simplified model, shown in Figure 2.21 to investigate the pressure propagation patterns by a compression sleeve. They used multilayer deformable models of the calf composed of solid tetrahedral elements (C3D10) and defined contact with the stocking using a surface-to-surface frictional formulation. They reported a marked attenuation of stress with depth, confirming that only 30–50% of the surface pressure reaches the venous level.

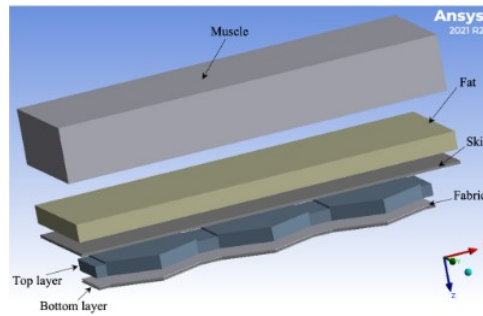


FIGURE 2.21: Expanded solid model of numerical simulation, presented by Hedigalla et al. [59].

Deformable limb geometries thus provide a physiologically relevant representation of the mechanical response of the leg, allowing the analysis of tissue deformation and pressure propagation. However, these models require accurate mechanical characterisation of biological tissues and entail higher computational costs than rigid models.

Patient-specific limb models Recent developments in imaging (MRI) and 3D scanning have enabled the creation of anatomically realistic, patient-specific limb models. In most studies, the mechanical properties of tissues are taken from experimental databases or derived by inverse identification from elastography and indentation data [65]. These databases report consistent values of elastic modulus and Poisson's ratio for skin, adipose tissue, and muscle, allowing reliable calibration of FE materials across different anatomical sites.

Avril et al. [19] developed one of the first coupled experimental–numerical representations of the leg, using MRI scans, and dividing it into layers representing skin, adipose tissue, muscle, and bone. The tissues were modelled as compressible hyperelastic solids discretised with hexahedral elements (C3D8R), with material parameters derived from experimental studies on biological tissues. Reported values vary considerably depending on the experimental technique (indentation, uniaxial or biaxial traction) and measurement conditions (in vivo or ex vivo).

Later on, Dubuis et al., Reda and Avril and Armerico de Almeida et al. [97]–[99] continued to generate three-dimensional geometries of the lower limb from MRI and CT scans, as presented in Figure 2.22. The reconstructed legs were meshed with quadratic tetrahedral elements (C3D10) in Abaqus/Standard to predict the hydrostatic pressure of the inner tissues under uniform external pressure.

Shi et al. [53] used high-resolution 3D body scanning over the leg, and the limb was represented as a deformable solid meshed with approximately 50,000 hexahedral elements (C3D8R).

Ye et al. [24] developed a fully coupled patient-specific model incorporating layered tissue representation and orthotropic stocking mechanics. Soft tissues were discretised with solid hexahedral elements (C3D8R) and described using Mooney–Rivlin hyperelastic laws.

Dan et al. [58] refined this approach by constructing multilayer leg models with tetrahedral elements (C3D4) and Ogden-type hyperelastic laws for the adipose tissue and muscle layers, calibrated from uniaxial compression tests on cadaveric specimens. The simulations were performed under static implicit conditions. Their analysis demonstrated that the internal stress field and vein compression depend strongly on the proportion of adipose tissue and muscle, explaining patient-specific differences in therapeutic efficiency.

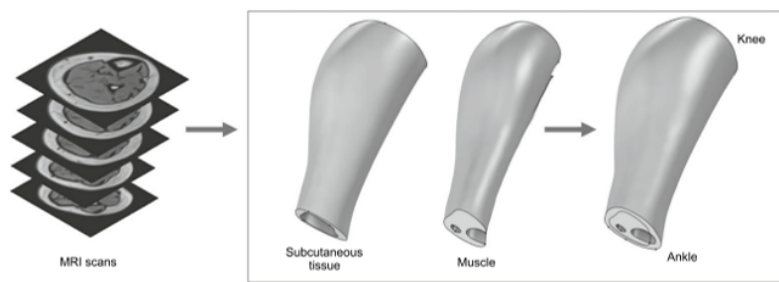


FIGURE 2.22: Geometry reconstruction from MRI scans of a lower limb unaffected by lymphedema. The two bones, tibia and fibula, were excluded from the geometry and their boundaries were treated as rigid, presented by Reda and Avril [99].

Patient-specific FE models now represent the most advanced approach, capable of integrating experimental fabric data, realistic tissue behaviour, and anatomical geometry. They have been successfully validated against in-vivo and mannequin measurements and constitute the basis for future digital twin applications in compression therapy. Nevertheless, their construction requires high-quality imaging and reliable material data for biological tissues and fabrics.

In summary, the modelling of the limb in finite element simulations has progressively evolved from rigid simplified shapes to anatomically detailed, patient-specific geometries. This progression has substantially improved the understanding of how compression is transmitted through soft tissues and veins. The next section addresses the numerical representation of the medical compression stocking itself, describing how its mechanical properties, meshing strategy, and contact behaviour with the limb are implemented to predict the resulting interface pressure.

2.3.3.2 Models of the medical compression stocking

The numerical representation of the compression stocking is a central aspect in finite element (FE) analysis of compression therapy. The way the garment is modelled determines how its mechanical behaviour, contact with the limb, and resulting interface pressure are computed. Two main approaches are commonly used: models that apply the stocking effect as external boundary pressures, and models that explicitly represent the fabric as a deformable thin structure surrounding the leg.

The following sections review the principal modelling strategies for representing medical compression stockings in FE analysis. The first subsection presents models where the stocking effect is replaced by an applied boundary pressure, primarily used to analyse tissue deformation and pressure propagation. The second subsection focuses on thin-structure models in which the stocking is explicitly meshed, allowing detailed study of fabric mechanics, contact interaction, and the influence of structural or material parameters on interface pressure.

Pressure applied as boundary condition A common strategy in finite element (FE) simulations of compression therapy is to represent the effect of the stocking not as a physical layer, but as an external pressure field applied directly on the limb surface. This simplification eliminates the need to model the fabric explicitly, while allowing detailed investigation of how external compression is transmitted through soft tissues and veins. These models are particularly suited to study physiological effects, such as venous deformation or tissue displacement, under controlled boundary conditions.

In this approach, the geometry of the limb is imported as a 2D axisymmetric or 3D solid model, and pressure loads are prescribed on its external surface to mimic the effect of the garment. The applied pressure can be uniform or vary along the length of the leg according to the desired compression class. The internal tissues, typically skin, adipose tissue, muscle, and bone, are defined as elastic or hyperelastic materials, with parameters obtained from experimental indentation or compression data reported in the literature. Numerical solvers such as Abaqus, ANSYS, or COMSOL Multiphysics are generally employed to compute deformation and stress fields at equilibrium.

You et al. [100] developed one of the earliest patient-specific FE models for compression garments using this technique. The leg was reconstructed from medical imaging and modelled as a deformable elastic continuum. External pressures, corresponding to those measured under class II garments, were applied as surface loads to analyse tissue deformation.

The study demonstrated that even nominally uniform compression generates heterogeneous strain fields, depending on local curvature and tissue stiffness.

Avril et al. [19] coupled experimental measurements with FE simulations to study the effect of compression on venous flow. Using Abaqus, they applied measured interface pressures to a multilayer leg model comprising adipose tissue, muscle, and bone. Material properties were assigned from in-vitro compression tests on biological tissues, using linear elastic and Mooney–Rivlin hyperelastic laws. MRI and ultrasound validation showed that external compression reduced superficial vein diameters and redistributed stresses toward deeper tissues, confirming the haemodynamic effectiveness of compression therapy.

Dan et al. [101] extended this work by evaluating the influence of tissue composition on the mechanical response. They modelled simplified cylindrical legs with distinct adipose tissue and muscle layers, each defined by different elastic moduli derived from experimental literature. Uniform pressures of 20–40 mmHg were applied at the surface, and the resulting displacements and internal stress fields were computed. The authors concluded that patients with higher subcutaneous adipose tissue thickness experience weaker pressure transmission to deeper veins, consistent with clinical variability in therapeutic outcomes.

Dubuis et al. [60] reconstructed patient-specific limb geometries from medical imaging and used pressure boundary conditions corresponding to measured garment compression. Their FE simulations provided detailed maps of tissue strain and stress distribution. The study highlighted that individual morphology and local curvature significantly influence how compression is transmitted, even when nominal pressures are identical. Validation against experimental displacement data confirmed the realism of the predicted deformation fields.

Yanmei et al. [102] analysed the pressure distribution on a 3D calf model by applying surface loads representing garment compression. Using ANSYS, they demonstrated that anatomical asymmetry leads to non-uniform pressure distributions, even for stockings designed to apply uniform compression. Comparisons with in-vitro pressure measurements showed close agreement, validating the model's predictive capability.

Karakashian et al. [25] revisited the Laplace law using COMSOL Multiphysics simulations in which uniform pressures were applied on anatomically curved rigid surfaces. Their objective was to quantify how curvature affects the translation between circumferential tension and interface pressure. The authors found that Laplace-based estimates underestimate local peaks in regions with small radii, confirming that curvature-dependent amplification must be considered in design and analysis.

Most recently, Reda and Avril [99] developed updated patient-specific FE simulations

incorporating realistic anatomical geometries from MRI scans. Surface pressures were assigned according to experimental measurements on the same subjects, and simulations were validated against in-vivo PicoPress® data. Their results showed excellent correspondence between predicted and measured pressures and confirmed that accounting for individual geometry is essential for accurate modelling of compression therapy.

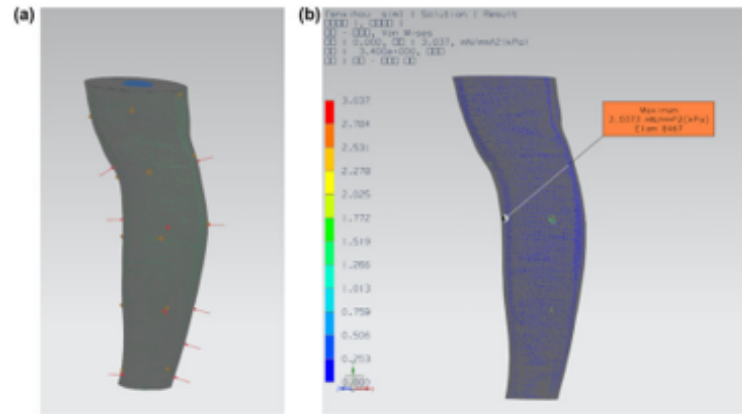


FIGURE 2.23: Simulation of clothing pressure distribution on calf (a) from the vertical direction to the surface of the calf, (b) the equivalent stress graph in the FEA model, presented by Yanmei et al. [102].

Pressure-based models offer a computationally efficient way to study the transmission of external compression into the limb. They reproduce tissue deformation, venous compression, and stress redistribution with reasonable accuracy while avoiding the complexity of meshing the stocking. However, these models neglect the mechanical properties of the fabric, its anisotropy, and the contact friction between garment and skin. Consequently, they are best suited for physiological and clinical investigations rather than for product design or textile optimisation.

Thin-structure representation A more advanced strategy in finite element (FE) simulations explicitly represents the compression stocking as a deformable thin structure in mechanical contact with the limb. In this approach, the fabric is meshed initially in a cylindrical configuration using surface or shell elements that reproduce its membrane-like behaviour. Then, the fabric mesh is positioned around the leg, that is modelled as a rigid or deformable solid, and progressively deforms to conform to the anatomical surface during the simulation. This reproduces the placement and stretching of the garment under realistic conditions. The interface pressure is then obtained directly from the contact reaction between both surfaces, allowing the simulation of realistic fabric–skin interaction, and the influence of textile parameters on the delivered pressure.

Unlike pressure-boundary models, thin-structure formulations describe the mechanical behaviour of the fabric explicitly, including anisotropy, friction, and nonlinear elasticity. The main objective of these models is to simulate garment placement and fitting, to determine the resulting interface pressure field, and to analyse how fabric properties, limb morphology, and contact conditions affect the pressure distribution and tissue deformation. They also enable virtual prototyping of compression garments by adjusting structural or material parameters before production.

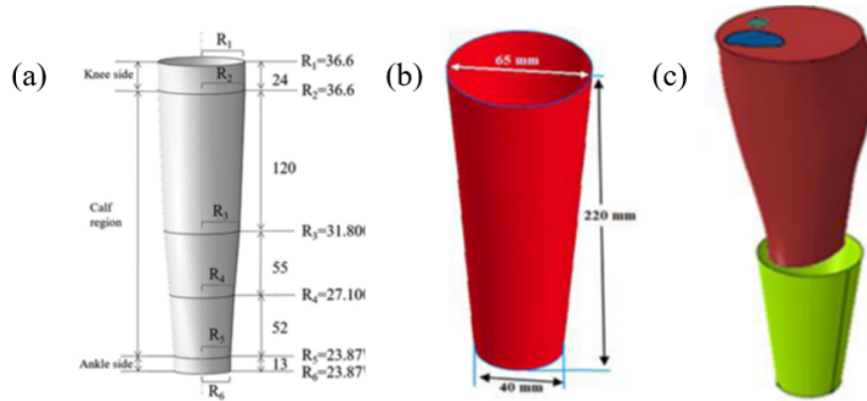


FIGURE 2.24: Geometric models of the medical compression stockings used for finite element analysis (a) MCS represented as a compilation of conical segments, proposed by Lu et al. [69]; (b) MCS represented as a single conical segment and (c) Aggregated model of MCS with leg, proposed by Ghorbani et al [80]

Element types Given the small thickness of knitted fabrics compared to the limb diameter, most studies describe the stocking as a two-dimensional layer using membrane or shell elements, while solid volumetric elements are typically reserved for the leg or soft tissues.

- Membrane elements reproduce in-plane tension without bending stiffness. They are suitable for fabrics that primarily carry tensile loads, providing computational efficiency and numerical stability. Mirjalili et al. [75] modelled compression garments using two-dimensional membrane elements (M3D4 in Abaqus) applied over rigid cylindrical legs. Wu et al. [76] adopted the same approach to represent seamless compression garments, extracting the elastic modulus from tensile strip tests. These models accurately reproduced average interface pressures and allowed efficient simulation of global garment behaviour, but were not sensitive to local curvature or anisotropy.
- Shell elements are the most commonly used for simulating medical compression stockings. They combine in-plane tensile stiffness with out-of-plane bending resistance, making them suitable for fabrics that conform to complex anatomical geometries. Dai

et al. [81] employed quadrilateral shell elements (S4R) to model graduated compression stockings, identifying moduli and Poisson ratios from uniaxial tensile data. Ghorbani et al. [80] also implemented S4R shells to represent orthotropic knitted fabrics in contact with rigid legs, achieving strong correlation with experimental Kikuhime pressure measurements. Ye et al. [24] used shell elements with a nominal thickness of 0.2–0.5 mm to model hyperelastic orthotropic behaviour in patient-specific geometries, reproducing experimental pressure gradients at the ankle and calf.

Mechanical behaviour The mechanical response of the stocking is defined through the constitutive law assigned to the surface elements. Different formulations have been used, depending on the degree of anisotropy and nonlinearity. The following three main approaches are reported in the literature:

- Isotropic formulations describe the stocking as a homogeneous elastic membrane with the same stiffness in all directions. Early studies adopted isotropic linear-elastic laws, sufficient for reproducing average compression levels. Mirjalili et al. [75] and Wu et al. [76] modelled the stocking as an isotropic membrane in Abaqus/Standard using stiffness derived from uniaxial tensile tests. Simulated interface pressures agreed with experimental means but did not represent local anisotropy.
- Orthotropic formulations distinguish between stiffness in the wale and course directions, which better reflects the mechanical anisotropy of inlaid weft-knitted textiles. Dai et al. [81] employed orthotropic shell models with elastic moduli identified from uniaxial tests in wale and course directions, successfully reproducing the graduated pressure profile. Ghorbani et al. [80] used similar orthotropic shells in Abaqus/Standard, calibrated from tensile data, and validated the predicted pressure against Kikuhime sensor measurements, reporting deviations below $\pm 10\%$. Lu et al. [69] modelled an orthotropic stocking on a multilayer leg and showed that orthotropy significantly improved the agreement between simulated and measured internal stresses.
- Hyperelastic formulations have been introduced to model the nonlinear stress–strain behaviour of compression fabrics under large deformations. Ye et al. [24] implemented a hyperelastic orthotropic model based on biaxial tensile data, with material constants identified via inverse fitting to experimental stress–strain curves. The stocking was meshed with S4R shell elements and simulated in Abaqus/Explicit to manage large sliding during placement. The model reproduced in-vivo pressure peaks within $\pm 8\%$ of sensor measurements.

Mechanical parameters for FE simulations are typically obtained from uniaxial or biaxial tensile tests. Uniaxial tests provide directional stiffness moduli and Poisson ratios, while biaxial tests capture coupling between wale and course directions. Liu et al. [103] and Zhang et al. [44] reported that course and wale moduli differ by over one order of magnitude, highlighting the need for orthotropic representation.

For nonlinear or hyperelastic formulations, constants of the Mooney–Rivlin or Ogden laws are determined by fitting simulated and experimental stress–strain data. Ye et al. [24] applied this method to achieve consistent force–elongation predictions under large deformation.

Mesh definition The stocking is meshed separately from the leg, initially as a regular cylindrical surface with uniform element distribution. This independent mesh is then positioned around the limb and progressively deformed as the garment slides upward during the simulation. The cylindrical configuration reproduces the unstressed geometry of the stocking before placement, while its deformation during sliding defines the real contact area and the circumferential stretch imposed by the leg shape.

For isotropic or simplified models, triangular shell elements (S3R) are sometimes used to improve mesh conformity over complex geometries. Wu et al. [76] adopted triangular meshes for seamless garments, which provided smoother contact transitions over sharp curvatures and facilitated automatic meshing on irregular surfaces.

More typically, the stocking is meshed using quadrilateral shell or membrane elements (S4R, M3D4 in Abaqus; SHELL181 in ANSYS). Mesh density varies across the stocking to adapt to the deformed curvature. Around the ankle and tibial crest, element sizes of 1 mm to 2 mm are typically required to model local pressure peaks, whereas 4–5 mm elements are sufficient in flatter regions such as the thigh.

Ye et al. [24] reported that reducing the element size below 1 mm did not significantly alter pressure predictions but increased computation time. Typical full-leg stocking meshes contain 30,000–80,000 elements, depending on anatomical complexity and whether the stocking covers the entire limb or only a segment.

Sensitivity analyses are frequently performed to verify mesh independence. Lu et al. [69] showed that a ratio of one element per 3–4 mm of local curvature radius provides convergence of pressure values within $\pm 5\%$. For patient-specific simulations, the same mesh density is maintained between successive anatomical scans to ensure reproducible results during parametric or time-dependent analyses.

Contact modelling The contact between the stocking and the limb defines how tension in the fabric translates into interface pressure. Most studies model this interaction using surface-to-surface contact algorithms, where the inner surface of the stocking is defined as the slave surface and the outer surface of the leg as the master.

The placement of the garment is often simulated by progressively sliding the cylindrical stocking mesh from the ankle upward along the leg surface. During this motion, the fabric stretches circumferentially and longitudinally, adapting to the limb curvature until it reaches the intended position. This procedure reproduces the real donning process in a quasi-static manner and allows the deformation field of the stocking to be obtained naturally from its interaction with the leg geometry.

Contact conditions may be defined as frictionless when the goal is to analyse pure normal pressure transmission, or with finite friction coefficients when tangential effects are considered. Chassagne et al. [51] and Shi et al. [53] reported that friction strongly influences local stress distribution and garment stability. Experimentally measured friction coefficients between textile and skin surfaces typically range between 0.2 and 0.4 under dry conditions, decreasing with humidity or lubrication.

Frictional formulations are implemented through penalty or augmented Lagrangian algorithms, ensuring stable convergence during large sliding.

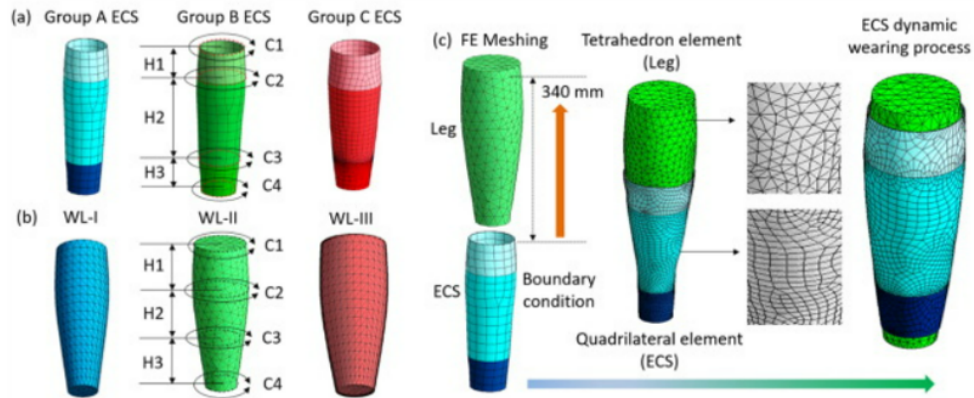


FIGURE 2.25: The developed geometric models of (a) FE MCS (Group A ECS, Group B ECS, Group C ECS) and (b) FE legs (WL - I, II, III), and (c) the meshing and boundary conditions applied in the FEM, presented by Ye et al. [74].

2.3.3.3 Simulation results and validation

The main output variables evaluated are the interface pressure, the circumferential tension within the fabric, the strain field, and the deformation of the underlying tissues. Together, these variables describe how fabric stiffness, structural parameters, and limb morphology determine the compression delivered to the body.

- The interface pressure corresponds to the normal stress transmitted at the skin–fabric interface and is the principal indicator of compression effectiveness. It is commonly extracted from the contact pressure variable (CPRESS in Abaqus or equivalent solvers) and averaged over regions corresponding to anatomical landmarks such as the ankle, calf, and thigh. Ghorbani et al. [80] computed interface pressure distributions using orthotropic shell models of compression fabrics applied to rigid mannequins, reporting deviations below 10% compared with Kikuhime sensor data. Ye et al. [24] and Shi et al. [53] extended this approach to patient-specific geometries, showing that predicted pressure gradients accurately reproduced in-vivo measurements, with local pressure concentrations at the ankle and tibial crest caused by reduced curvature radii and increased local fabric tension.
- The circumferential tension quantifies the tensile load developed along the course direction as the stocking is extended over the limb. It is derived from the in-plane stress field of shell or membrane elements and provides a direct link with experimental tensile data. Dai et al. [81] and Lu et al. [69] demonstrated that circumferential stress increases nonlinearly with applied strain, reflecting the anisotropic and nonlinear elastic response of knitted fabrics. These analyses also revealed that variations in local stiffness, arising from differences in inlay yarn density or loop geometry, are directly responsible for the pressure gradient along the limb.
- The strain distribution within the fabric characterises how the stocking adapts to the three-dimensional shape of the leg. It highlights regions subjected to greater extension, indicating heterogeneous deformation associated with knitting pattern, fabric anisotropy, or limb curvature [24], [53].
- FE simulations also quantify soft tissue deformation and internal stress transfer. Avril et al. [19] and Dan et al. [58] showed that external compression primarily affects subcutaneous tissues and that stress attenuates rapidly with depth, leading to limited transmission to deep venous regions. Hedigalla et al. [59] confirmed this attenuation, demonstrating that only a fraction of the surface pressure is transmitted to the deeper layers of muscle and vein.
- Some authors have also analysed energy-based variables, such as the strain energy stored in the fabric or the contact work at the interface, to assess the mechanical efficiency of compression [24]. These metrics quantify how effectively the elastic energy accumulated in the garment is transformed into interface pressure.

Overall, the mechanical outputs obtained from FE simulations provide a better insight on how compression garments interact with the limb. When compared to analytical methods based on Laplace, or experimental measurements, FE simulations better represent the local pressure distribution resulting from the garment deformation and the leg morphology.

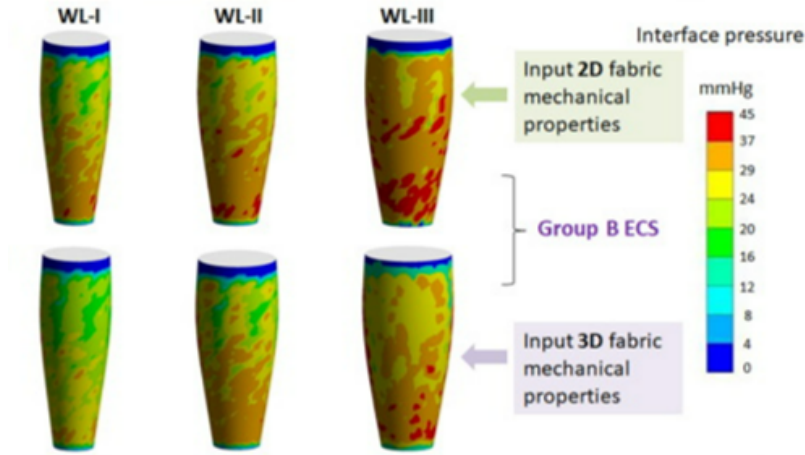


FIGURE 2.26: The simulated interface pressure profiles applied by the different groups of the studied ECSs when their 2D and 3D material mechanical properties were applied, respectively, presented by Ye et al. [74].

Validation Finite element (FE) simulations of medical compression stockings are validated through a combination of experimental characterisation, pressure measurements, and in-vivo imaging. These complementary methods ensure that numerical predictions accurately represent both surface compression and internal tissue deformation [79].

The first approach for validating FE models is based on Laplace's law, given in Equation 2.8, which describes the relationship between fabric tension, circumferential fabric deformation, and the pressure applied to the limb.

Kowalski et al. [52] used Laplace's relation to compare predicted and analytical pressures in simplified cylindrical leg models, showing that the FE simulations accurately reproduced the global pressure levels expected from geometry and fabric stiffness. Lu et al. [79] highlighted that this analytical formulation remains the first step in validating compression models, as it enables rapid verification of distal-to-proximal pressure gradients before introducing complex contact or anisotropic effects. Wang et al. [20] applied the same approach to confirm that the pressures predicted by FE and hemodynamic models were consistent with Laplace's values. More recently, Ye et al. [24], Lozo et al. [39], and Shi et al. [45] used Laplace-based calculations to validate FE-predicted pressure distributions, and confirmed that the simulated compression followed the expected gradient along the leg.

Reported deviations between Laplace-based predictions and finite element (FE) results generally remain within 10–15 % for cylindrical or mildly conical regions, increasing up to 20–25 % in areas of strong curvature or heterogeneous fabric stiffness. These differences confirm that Laplace's law provides a reliable validating method for pressure estimation, although its simplifications limit its quantitative accuracy in complex limb geometries [20], [24], [39], [45], [52], [79].

The second validation method is based on pressure measurements obtained with pressure sensors such as PicoPress[®], Kikuhime[®], Flexiforce or the Medical Stocking Tester. These devices record the local normal pressure exerted by the fabric on the limb, providing direct experimental data to compare with numerical predictions.

Lu et al. [69] validated simulations against Kikuhime[®] measurements on mannequin legs, reporting deviations below 10–12%, and confirming that calibrated fabric stiffness and contact friction parameters reproduce experimental pressure profiles with good accuracy. Shi et al. [53] compared FE-predicted pressures with PicoPress[®] readings at the ankle, calf, and thigh, observing average differences below 10%. Similarly, Siddique et al. [50] compared FE predictions with MST measurements, showing deviations of 10–15% at moderate extensions, which they attributed to viscoelastic relaxation of the fabric not included in the numerical model. Ye et al. [24] also reported that well-calibrated FE simulations reproduced in-vivo Kikuhime[®] measurements within 10%, even around curved anatomical regions such as the ankle and tibial crest.

Overall, reported differences between measured and simulated interface pressures typically remain within 10–15%, but can increase to 20% locally in zones of high curvature or frictional concentration. These results confirm that pressure sensors provide local, quantitative validation of fabric–skin contact pressures. However, the presence of the sensor itself slightly perturbs the local stress field, which can lead to a minor underestimation of the actual interface pressure [51], [69]. Its main advantages are direct measurement, ease of use, and high reproducibility. However, the presence of the sensor slightly perturbs the local stress field, producing small underestimations of true pressure, and the method cannot capture shear stresses or transient effects during movement [51], [79].

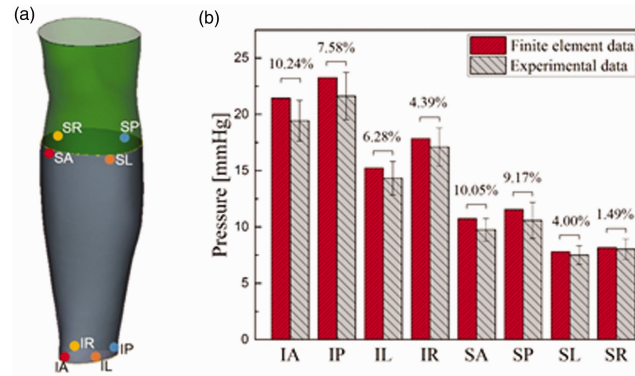


FIGURE 2.27: Validation of the finite element (FE) predictions by comparing the pressure values at eight sites of the skin–stocking interface: (a) locations of the eight sites; (b) comparison of the interface pressure between the FE and experimental data. [69].

A third validation method evaluates soft-tissue deformation within the limb using magnetic resonance imaging (MRI) or ultrasound imaging. Lu et al. [79] and Avril et al. [19] used MRI to obtain the displacement of skin, fat, and muscle before and after compression, providing a direct comparison between predicted and measured internal strain fields. This method is non-invasive and avoids the disturbances caused by surface sensors, but it requires advanced image registration to accurately align the deformed and undeformed geometries.

Overall, validation studies demonstrate that finite element simulations can reliably reproduce realistic pressure distributions when supported by accurate experimental input. Average deviations between predicted and measured pressures typically remain within 10–15%, with larger local differences mainly due to geometric irregularities, friction, or fabric stiffness variations.

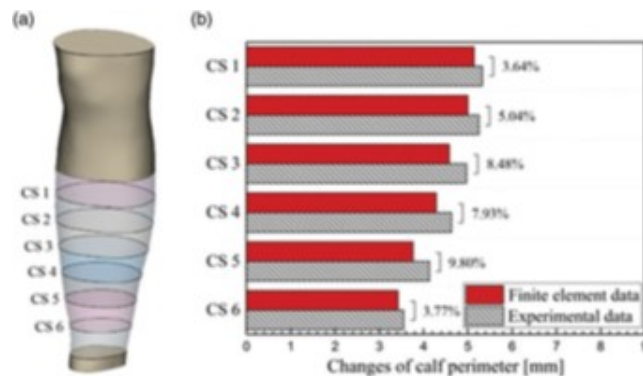


FIGURE 2.28: Validation of the finite element (FE) predictions by comparing the changes in the circumference of the calf at six cross-sections of the calf: (a) locations of the six cross-sections; (b) comparison of the changes in the calf circumference between the FE and experimental data. [69].

In summary, thin-structure formulations represent the current state of the art in FE modelling of medical compression stockings. They allow for the simulation of realistic garment placement, contact interactions, and spatial variations in pressure, providing valuable insights for both biomedical analysis and product development. However, these models remain computationally demanding and depend on accurate experimental data for material calibration and friction coefficients. In addition, they still homogenise the contribution of the ground loops and inlay yarns, meaning that structural parameters such as course density or yarn tension are not explicitly represented. The development of hybrid or mesoscopic approaches, where the fabric architecture is modelled at the loop scale, offers a promising direction for future work.

2.4 Summary and conclusions

2.4.1 Key findings

Compression therapy has proven to be clinically effective in the management of venous and lymphatic disorders. Its therapeutic performance, however, relies on precise control of the interface pressure delivered by the garment.

The fabrics used in medical compression stockings are cylindrical inlaid weft-knitted structures, that combine ground yarns creating the loop structure, and inlay yarns inserted into the courses between loops. This two-system structure has a dual behaviour, where the ground yarn is elastic and provides comfort, whereas the inlay yarn acts as reinforcement in the circumferential direction. In graduate compression stockings, the stiffness of the fabric is modulated with the knitted structure and inlay yarn distribution, to achieve a gradual pressure enhancing venous and lymphatic return.

Experimental methods are essential for characterising mechanical response of the fabric, and for validation of compression predictions. However, they are restricted to simplified conditions and cannot reproduce the complexity of in-use situations. Analytical formulations, including Laplace's law, are commonly used as tools for interface pressure prediction. Their accuracy, however, is limited to simplified geometries and uniform tension assumptions. Numerical simulations based on finite element methods, allow more realistic analyses of stocking-limb interaction, allowing the integration of fabric mechanical properties and realistic leg models. However, current fabric models rely on homogenisation and do not represent the local knitted structure nor the individual contributions of the inlay yarns or loop structure.

2.4.2 Future directions and research opportunities

Future models should integrate the local knitting structure and the combined mechanical effect of loops and inlay yarns that define the behaviour of inlaid weft-knitted compression fabrics. Hybrid mesoscopic models have been proposed to represent the dual architecture of knitted and woven composites, combining continuum representations for the matrix with discrete connectors for reinforcement yarns. The local area of the textile is modelled with a unit cell, that represents the structural configuration. This enables the explicit differentiation of matrix and reinforcement contributions, and the reproduction of anisotropy in different directions.

This methodology is transferable to medical compression stockings, where the loop structure acts as a matrix, and the inlay yarns as reinforcement, creating a highly anisotropic fabric. Moreover, the knitted structure varies across zones, by modifying the inlay yarn distribution, to achieve the graduate pressure profile. It would allow to incorporate the heterogeneous structure of inlaid weft-knitted fabrics, and to represent its variation across zones.

This literature review establishes the basis for the methodological developments presented in the following chapters. Chapter 3 will describe the structural and mechanical analysis of medical compression stockings and their components. Chapter 4 will introduce a hybrid mechanical model that combines experimental characterisation with mesoscopic representation of ground and inlay yarns. Chapter 5 will extend this model into a finite element formulation for predicting interface pressure over the limb.



Chapter 3

Inlaid weft-knitted fabrics for medical compression stockings

3.1 Introduction

Medical compression stockings are widely used as a primary therapeutic solution for venous and lymphatic disorders. The pressure exerted on the limb arises from the circumferential deformation imposed on the knitted fabric as it adapts from its relaxed perimeter to the larger perimeter of the leg. This imposed elongation induces circumferential tension within the fabric, which is transferred to the limb surface as interface pressure, generating the therapeutic compression effect.

The mechanical response of the fabric results from its specific inlaid weft-knitted architecture, which incorporates two independent yarn systems. The ground yarn forms the intermeshed loop structure that governs the axial extensibility and comfort of the fabric, while the inlay yarn is inserted into each course and acts as a circumferential reinforcement. The mechanical combination of these two systems determines the global behaviour of the fabric. The loop structure provides large extensibility under low loads, whereas the circumferentially oriented inlay yarns restrict deformation and increase stiffness. Consequently, the fabric is markedly anisotropic, exhibiting distinct mechanical responses in the longitudinal and circumferential directions.

The distribution of the inlay yarn controls the circumferential stiffness of the fabric and therefore the pressure transmitted to the leg. Local variations in the density of inlay yarn, expressed as the number of inlaid courses per unit length, modify the frequency of reinforcement insertion and govern the mechanical response in the circumferential direction. This structural modulation generates the gradual variation of circumferential tension required

for therapeutic graduated compression. As detailed in Chapter 2, this design strategy divides the product into three compression zones, ankle, calf, and thigh, each characterised by a specific value of inlay-yarn density. The ankle, with the highest reinforcement density, exhibits the greatest circumferential stiffness and thus the highest interface pressure, whereas the calf and thigh present progressively lower reinforcement, producing a controlled reduction in compression and improved comfort.

The objective of this work is to develop a mechanical model for medical compression stockings that can be implemented in finite element simulations for interface pressure prediction. The approach establishes a direct relation between the knitted architecture and the macroscopic tensile response of the fabric. The originality of the method lies in a hybrid formulation capable of separating and predicting the individual contributions of the loop structure and the inlay yarns, while also incorporating the structural variations, specifically the density of inlay yarn, that govern the circumferential stiffness of each compression zone. To enable this modelling strategy, a detailed experimental characterisation of the materials and the knitted structure is required to identify all input parameters under in-use deformation conditions.

This chapter is therefore dedicated to the mechanical characterisation of the weft-knitted fabrics used in medical compression stockings. The analysis investigates how the structural design and material composition of the fabric determine its circumferential stiffness and the resulting interface pressure. The study combines experimental characterisation with direct and indirect assessment of pressure, providing the data required for the hybrid model developed in Chapter 4.

The chapter is organised as follows. First, the physical principle linking circumferential deformation of the fabric to interface pressure is introduced. Section 3.2.1 presents the main structural parameters that govern this response. Section 3.2.2 describes the compression stocking under study, detailing its knitted architecture, yarn materials, inlay-yarn density, dimensional parameters, and the manufacturer's expected compression levels. Section 3.3.1 defines the inlay-yarn density under in-use conditions. Sections 3.3.2 and 3.3.3 present the experimental protocols used to characterise the mechanical behaviour of individual yarns and knitted fabrics. Finally, Section 3.4 describes the evaluation of interface pressure through Laplace-based calculations and direct pressure measurements.

The experimental results of this chapter were used to define the dimensional parameters of the hybrid model presented in Chapter 4, as well as the constitutive behaviour and mechanical parameters of the elements modelling the loop structure and inlay yarn. The results served also to validate the hybrid model, and the numerical simulations for interface

pressure prediction, presented in Chapter 5.

All notations and symbols used throughout this chapter are listed and defined in the glossary at the end of the manuscript.

3.2 Medical compression stockings

The principle of compression generated by medical compression stockings has been widely discussed in the literature and summarised in Chapter 2. The pressure exerted on the limb results from the circumferential deformation of the knitted fabric when the stocking is placed over the leg. This deformation corresponds to the difference between the relaxed perimeter of the fabric and the larger perimeter of the limb. The imposed elongation produces circumferential tension within the structure, which is transmitted to the skin surface as interface pressure. This mechanism was first formalised by Gaied et al. [47] and has since been adopted in numerous analytical and numerical formulations [24], [69], [96], [104].

According to Laplace's law, the interface pressure is proportional to the circumferential tension and inversely proportional to the local radius of curvature of the leg. Although this simplified description has been extensively applied to medical compression analysis [23], [105], its accuracy is limited for non-cylindrical geometries and non-homogeneous fabrics. For this reason, several authors have refined the analytical expression or developed finite element models to account for anatomical complexity and structural behaviour [69], [74], [80].

The magnitude of the pressure generated by the stocking depends simultaneously on material and geometric factors. The mechanical response of the fabric is governed by the characteristics of the constituent yarns, such as material composition, linear density and elastic behaviour, by the knitted architecture, which in this case is an inlaid weft-knitted structure, and by the dimensional parameters of both the fabric and the leg, which define the imposed circumferential strain (ε_{circ}). Among these parameters, the density of inlay yarn, that is, the number of inserted inlay yarns per unit length, plays a key role since it determines the frequency of reinforcement insertion and directly controls the circumferential stiffness of the fabric.

As described by Chen et al. [105] and Ye et al. [24], the inlaid weft-knitted structure combines two yarn systems: the ground yarn, forming the loop structure and providing extensibility, and the inlay yarn, inserted into each course and acting as circumferential reinforcement. The mechanical combination of these systems produces an anisotropic response,

with higher stiffness in the circumferential direction and greater extensibility in the longitudinal direction.

In graduated-compression stockings, the pressure gradient along the limb is achieved by varying the density of inlay yarn between anatomical zones. Modifying the number of courses per unit length changes the distribution of inlay yarns and therefore the circumferential stiffness of the structure. The ankle, which presents the highest density of inlay yarn, provides the greatest circumferential reinforcement and consequently the highest interface pressure, whereas the calf and thigh, with lower densities, deliver reduced compression and improved comfort [39], [106].

The following section presents the medical compression stocking selected for this study, describing its architecture, yarn materials, structural parameters, and the compression levels defined by the manufacturer.

3.2.1 Medical product and compression zones

The experimental study was carried out on a commercial medical compression stocking (MCS) developed and manufactured by Sigvaris France. The selected product, shown in Figure 3.1, is a women's Class II stocking, size M, model Semi transparent, designed to provide graduated compression [107]. Class II stockings correspond to medium-pressure therapeutic garments, typically prescribed for moderate venous insufficiency, oedema, or post-treatment recovery. According to the French standard NF S97-114 [29], these products are designed to deliver 15–25 mmHg (2.0–2.7 kPa) at the ankle, with progressively lower pressures along the calf and thigh.



FIGURE 3.1: Selected women's medical compression stocking, developed and commercialised by Sigvaris France [107].

This model and size were chosen because they are representative of commercially available graduated compression stockings intended for daily therapeutic use. Size M corresponds to an average leg morphology, enabling reliable experimental measurements and direct comparison with reference values reported in the literature. Although this specific product was used as a case study, the characterisation procedure and the modelling methodology developed in this work are general. By adjusting the structural parameters, such as density of inlay yarn, fabric dimensions, or yarn properties, the same approach can be directly applied to different compression classes, sizes, or textile architectures. The present study therefore establishes a general framework for analysing and modelling the mechanical behaviour of medical compression stockings independently of brand or size.

The selected product incorporates three main compression zones, ankle, calf, and thigh, each associated with a distinct level of circumferential stiffness and, consequently, a different interface pressure. The ankle zone provides the highest compression, ensuring efficient venous return from the distal part of the limb [47]. The calf maintains an intermediate pressure level to support upward blood flow, while the thigh delivers the lowest compression, improving comfort while preserving the therapeutic gradient [96], [108]. This controlled decrease in pressure along the limb is essential for physiological efficiency in graduated compression therapy [39], [109].

Structurally, all zones share the same inlaid weft-knitted architecture, composed of a looped ground yarn that provides extensibility and a circumferential inlay yarn inserted in every course as reinforcement. The yarn materials (composition, linear density, and elastic behaviour) are identical throughout the product, ensuring consistent local material properties. The density of inlay yarn, however, varies deliberately between zones and governs the local circumferential stiffness. A higher density of inlay yarn means a larger number of reinforcing yarns per unit length, increasing circumferential stiffness and resulting in higher interface pressure; conversely, a lower density increases extensibility and comfort.

The manufacturer controls the pressure distribution along the stocking by adjusting both the density of inlay yarn and the relaxed circumference of the fabric within each zone [39], [106]. Increasing the density of inlay yarn enhances the local stiffness, while modifying the relaxed circumference regulates the circumferential strain imposed when the garment is worn [47], [96]. In practice, the ankle zone combines the highest stiffness with the smallest deformation, producing the highest interface pressure, whereas the calf and thigh zones exhibit progressively lower stiffness and larger deformation, generating the desired reduction in pressure [24].

The structural values associated with these zones, together with their experimental characterisation, are presented in Section 3.3.1.

3.2.2 Knitted structure and yarn materials

The medical compression stocking analysed in this work is made of an inlaid weft-knitted fabric with a 1×1 configuration, a structure commonly used in graduated compression products. The fabric is produced on a circular knitting machine combining two yarn systems: the ground yarn, which forms the loop structure along the longitudinal direction, and the inlay yarn, inserted in each course and oriented circumferentially.

Figure 3.2 shows a microscopy image of the fabric at the local scale, in its flat configuration. In local scale, a local reference system (e_1, e_2, e_3) is used. The direction e_1 corresponds to the wale direction, visible as vertical columns of loops, e_2 corresponds to the course direction, visible as horizontal rows of loops, and e_3 represents the fabric thickness.

This local system is inclined with respect to the global reference system (E_1, E_2, E_3) of the fabric in the flat configuration, shown in Figure 3.4. In this global system, E_1 denotes the longitudinal direction of the stocking, E_2 the circumferential direction, and E_3 the thickness direction. The inclination of the local axes results from the orientation of the inlay yarn, which follows a helical trajectory along the fabric with an inclination angle (α) relative to the global system. Both reference systems are considered in the development of the hybrid mechanical model presented in Chapter 4.

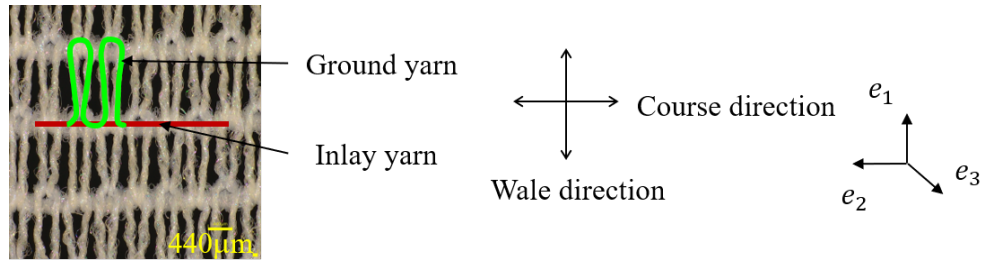


FIGURE 3.2: Microscopy of the 1x1 inlaid weft – knitted structure of the selected medical compression stocking.

The two yarn systems, shown in Figure 3.3, play different but complementary roles within the knitted architecture. The ground yarn forms the loop structure that defines the geometry of the fabric and provides extensibility and dimensional stability, while the inlay yarn is inserted in each course along the circumferential direction, ensuring structural reinforcement during stretching [24].

The ground yarn is a covered elastane yarn composed of an elastane core of 22 dtex wrapped with two 22 dtex polyamide filaments. The inlay yarn is a 210 dtex elastane core covered with two 22 dtex polyamide filaments, resulting in a total linear density of 494 dtex. The elastane core provides high recoverable stretch and governs the elastic response of the yarn under tensile loading, allowing the knitted structure to accommodate the imposed deformation while maintaining continuous tension transmission. The polyamide covering improves yarn stability during knitting, protects the elastane from mechanical damage, and provides a smooth contact surface that enhances wearer comfort [39], [96].

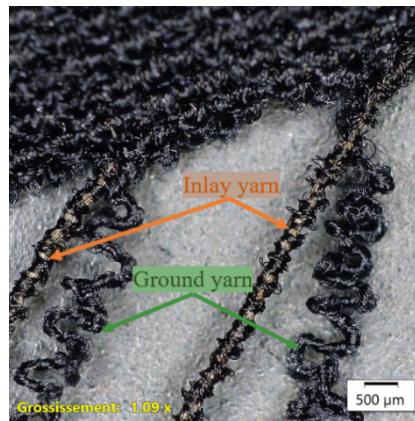


FIGURE 3.3: Microscopy of the two yarns conforming the selected medical compression stocking: ground yarns and inlay yarns

Both yarn systems remain identical across all compression zones, meaning that the material behaviour of the loops and the inlay yarns is constant along the fabric. Consequently, the variations in circumferential stiffness required to generate the graduated pressure profile arise exclusively from changes in the density of inlay yarn, which modifies the structural reinforcement per unit length without altering the material properties.

In the following sections, the deformation imposed by in-use conditions is defined, the density of inlay yarn is determined for the deformed configuration of each zone, and the mechanical behaviour of the two yarn systems, constant throughout the product, is analysed together with the tensile response of the inlaid weft-knitted fabric. This enables quantifying the influence of the structural variations between zones on the overall mechanical response of the medical compression stocking.

3.2.3 Geometrical parameters

The geometry of medical compression stockings is designed to ensure a gradual and controlled pressure distribution along the leg. The product analysed in this work, shown in Figure 3.4 in its flat configuration, presents a cylindrical shape in its three compression zones,

ankle, calf, and thigh, connected by conical transition areas. This geometry allows the stocking to progressively adapt to the anatomical profile of the leg and provides a smooth change in local stiffness between zones, which is essential for achieving the graduated pressure profile [39].

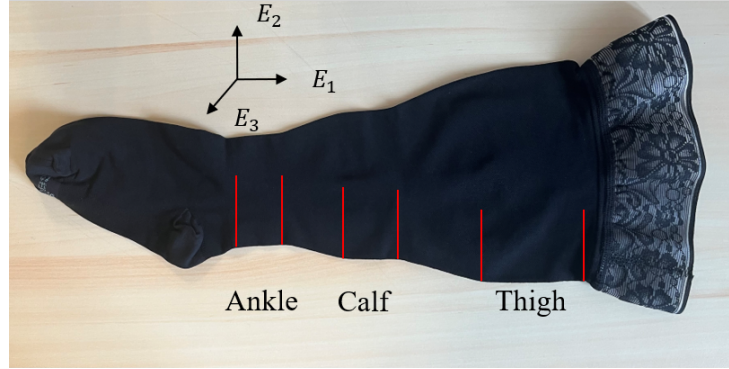


FIGURE 3.4: Medical compression stocking with different compression zones (ankle, calf and thigh) in the flat configuration

The size chart of the product defines the range of patient leg dimensions, measured according to Figure 2.9 in Chapter 2, for which each product size is designed. For the model analysed in this study, Semi Transparent, women size M, Sigvaris France, the values specified by the manufacturer are reported in Figure 3.5. The table lists the perimeters of the ankle (cheville), calf (mollet), and thigh (cuisse), as well as the half leg length (A–D), measured between the floor and two fingers below the knee, and the full leg length (A–G), measured between the floor and the crotch. These measurements define the target geometry of the stocking under in-use conditions.

Taille		Cheville B	Mollet C	Taille		Cheville B	Mollet C	Cuisse M	Hanches
Chaussettes	S	19 – 22	30 – 38	Bas auto-fixants Collant	S	19 – 22	30 – 38	45 – 58	≤ 100
	M	22 – 24	33 – 41		M	22 – 24	33 – 41	49 – 62	≤ 107
	L	24 – 26	37 – 45		L	24 – 26	37 – 45	54 – 67	≤ 114
	XL	26 – 29	40 – 48		XL	26 – 29	40 – 48	56 – 69	≤ 121

Hauteur (cm)	1/2 jambe A - D	jambe A - G	Pointure
Normal	≤ 40	≤ 75	34/40
Long	≥ 40	≥ 75	36/42

FIGURE 3.5: Dimensional data from the manufacturer's size chart for the Semi Transparent model (size M, normal length) [107].

Table 3.1 reports the reference leg perimeters for each compression zone of the selected size and product, defined as minimal ($P_{\text{leg, min}}$), average ($P_{\text{leg, avg}}$), and maximal ($P_{\text{leg, max}}$)

values. These values, derived from the manufacturer's dimensional chart in Figure 3.5, determine the three deformation states that the fabric may experience in the in-use configuration, corresponding to minimal, average, and maximal circumferential extension. They also determine the maximal deformation state of the fabric, which is reproduced during the tensile test described in Section 3.3.3, and used for the formulation of the hybrid model presented in Chapter 4. These perimeters also provide the input geometry required to model the three reference legs for to each compression zone in the finite element simulations presented in Chapter 5.

TABLE 3.1: Reference leg perimeters for different compression zones: minimal ($P_{\text{leg, min}}$), average ($P_{\text{leg, avg}}$) and maximal ($P_{\text{leg, max}}$)

Zone	$P_{\text{leg, min}}$ [mm]	$P_{\text{leg, avg}}$ [mm]	$P_{\text{leg, max}}$ [mm]
Ankle	220	230	240
Calf	330	370	410
Thigh	490	555	620

In the flat configuration shown in Figure 3.4, considered as the initial state of the stocking, the fabric is measured in terms of its width and length. The initial width (W_0) corresponds to half the relaxed perimeter (P_0), as given in Equation 3.1. Once the stocking is worn, this planar geometry is transformed into a cylindrical configuration, and the fabric stretches from its relaxed state to the deformed configuration that matches the leg shape, as illustrated in Figure 3.6. The imposed deformation is therefore governed by the relative change between the relaxed perimeter (P_0) and the deformed perimeter (P), which corresponds to the leg perimeter ($P = P_{\text{leg}}$).

To describe the deformation in this configuration, a cylindrical reference system (e_r, e_θ, e_z) is introduced, where e_r is the radial direction, e_θ the circumferential direction, and e_z the longitudinal direction, as shown in Figure 3.6. In this system, the course direction of the flat configuration (E_2) becomes the circumferential direction (e_θ) once the stocking is wrapped around the limb.

$$P_0 [\text{mm}] = 2 \cdot W_0 \quad (3.1)$$

The circumferential deformation imposed on the fabric is then expressed by Equation 3.2, where P_0 is the initial fabric perimeter and P is the deformed perimeter corresponding to the leg perimeter given in Table 3.1.

$$\varepsilon_{\text{circ}} [\text{mm/mm}] = \frac{P - P_0}{P_0} \quad (3.2)$$

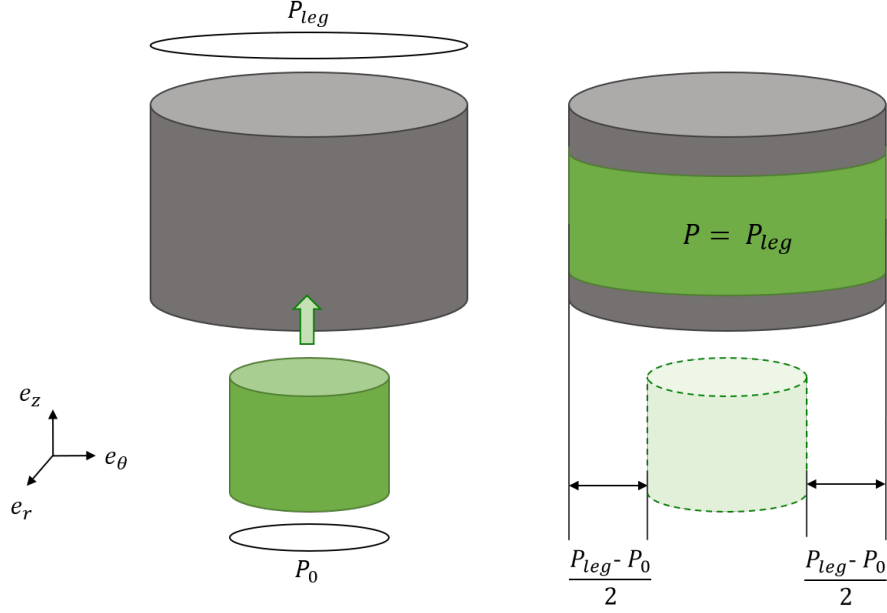


FIGURE 3.6: Circumferential strain imposed on the fabric when placed over the leg, given by the relative difference between the initial fabric perimeter (P_0), and the leg perimeter (P_{leg})

These dimensional parameters define the in-use deformation state of each compression zone and constitute the geometric input required for the mechanical characterisation presented in Section 3.3.3.

3.3 Experimental characterisation

The experimental characterisation presented in this section establishes the mechanical and structural foundation required to analyse the circumferential behaviour of the medical compression stocking under in-use conditions. After defining the dimensional configuration and knitted architecture of the product, in Sections 3.2.1–3.2.3, the next step is to quantitatively determine the material and structural parameters governing the mechanical response of the fabric. These parameters are essential for understanding the contribution of each yarn system to the global stiffness, for interpreting the tensile behaviour of knitted fabrics, and for providing the input data used in the hybrid mechanical model developed in Chapter 4.

The characterisation is carried out at two complementary scales:

- Yarn scale, where the axial behaviour of the inlay yarn is measured through uniaxial tensile tests, providing the stiffness function governing the circumferential reinforcement;
- Fabric scale, where cylindrical samples representing the ankle, calf, and thigh zones are tested under controlled circumferential deformation, reproducing the in-use strain range determined in Section 3.2.3.

In addition, the structural parameter controlling the stiffness gradient, the density of inlay yarn (D_{IY}), is experimentally determined under longitudinally deformed conditions to ensure consistency with the mechanical tests and with the in-use configuration of the stocking.

Together, these results provide the necessary experimental basis to quantify the role of the loop structure and the inlay yarn, to assess the effect of structural variations between compression zones, and to supply the input parameters required for the hybrid mechanical formulation and finite element simulations presented in Chapters 4 and 5.

3.3.1 Structural characterisation : density of inlay yarn

The architecture of inlaid weft-knitted compression fabrics, defined by the geometry of the loop structure and the frequency of inlay-yarn insertion, governs the global mechanical behaviour of the product [39], [110]. Among the structural parameters, the density of inlay yarn (D_{IY}), defined as the number of courses per unit length measured along the longitudinal direction of the stocking, plays a central role. A higher value of D_{IY} implies a greater frequency of inlay-yarn insertion, increasing the circumferential reinforcement and the corresponding local stiffness [21], [111]. In medical compression stockings, this parameter is intentionally varied between compression zones to modulate stiffness and achieve the required graduated pressure profile [68], [70]. Several studies have highlighted that such structural heterogeneity is fundamental for producing controlled pressure gradients along the limb [31], [44], [112].

In the selected product, the materials of both yarn systems and the knitted architecture remain constant across zones. Therefore, the density of inlay yarn is the key structural variable governing the evolution of circumferential stiffness between the ankle, calf, and thigh.

The density of inlay yarn was quantified to analyse its variation between zones and to assess its influence on the tensile response of the fabric. It was evaluated in the longitudinally deformed configuration, corresponding to the in-use state of the stocking, as illustrated in

Figure 3.7. This is the configuration used during tensile testing of cylindrical fabric samples, described in Section 3.3.3, ensuring full consistency between structural and mechanical characterisation.

The density of inlay yarn (D_{IY}) is calculated using Equation 3.3, where NC is the number of courses contained in a fabric segment of length (d), defined by the interline reference distance in Figure 3.7.

$$D_{IY} [\text{mm}^{-1}] = \frac{NC}{d} \quad (3.3)$$

The determination was performed on inlaid weft-knitted samples from the ankle, calf, and thigh regions. These samples reproduced the final product and served for the mechanical characterisation described in Section 3.3.3. The number of courses between two knitted reference lines (NC) was provided by the manufacturer for each zone. The reference distance was set to $d = 30 \text{ mm}$ across all zones, following the tensile testing protocol that reproduces the in-use longitudinal deformation of the fabric [29].

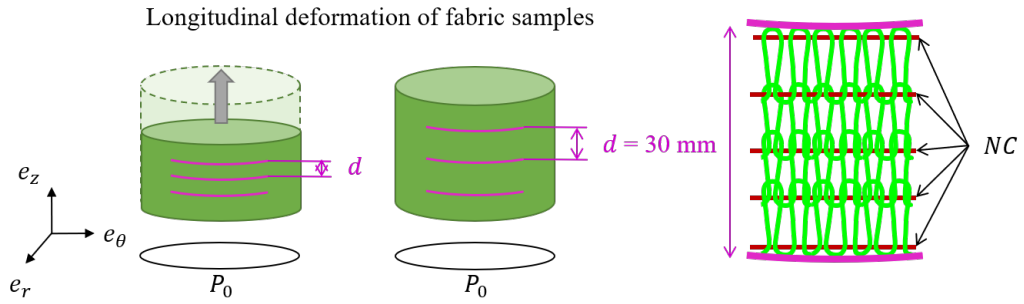


FIGURE 3.7: Longitudinal deformation imposed on the fabric samples under in-use conditions, defining the in-use density of inlay yarn given by the number of courses (NC) for a given interline distance (d).

Table 3.2 summarises the NC values provided by the manufacturer, and the corresponding densities of inlay yarn (D_{IY}) calculated for the in-use longitudinal extension. The highest value of D_{IY} is found in the ankle, reflecting the increased frequency of inlay-yarn insertion and the enhanced circumferential reinforcement in the distal region. The calf and thigh show progressively lower densities, consistent with the reduction in stiffness required to achieve graduated compression [39], [44], [112].

TABLE 3.2: Number of courses (NC) and density of inlay yarn (D_{IY}) per compression zone under in-use conditions

Zone	NC	$D_{IY} [\text{mm}^{-1}]$
Ankle	34	1.13
Calf	26	0.87
Thigh	24	0.80

These values represent the effective structural configuration of the fabric in the deformed state and are used as input parameters in the hybrid mechanical model developed in Chapter 4. This ensures that the predicted stiffness distribution accurately reflects the experimentally measured gradient between compression zones [21], [68].

3.3.2 Mechanical characterisation of yarns

The tensile mechanical behaviour of the inlay yarn was characterised through uniaxial tensile testing, in order to quantify its axial response and evaluate its contribution to the circumferential stiffness of the medical compression stocking. The results obtained here were later used to identify the parameters of the constitutive model associated with the inlay yarn, as described in Sections 4.2.2 and 4.2.4 of Chapter 4.

3.3.2.1 Tensile testing of inlay yarns

The inlay yarn consists of an elastane filament (210 dtex) covered by two polyamide filaments (22 dtex each), resulting in a total linear density of 494 dtex. Yarn samples were extracted (de-knitted) from finished medical compression stockings that had undergone the complete industrial process, including knitting, dyeing, heat-setting, and mechanical stabilisation. This ensured that the measured behaviour corresponds to the actual mechanical state of the yarn within the final product.

The tests were performed using a Statimat ME+ tensile tester (serial no. 36723) equipped with a 32 N load cell (serial no. 17309147), shown in Figure 3.8. The protocol followed the manufacturer's internal procedure for covered elastic yarns. Each specimen was mounted with an initial gauge length $YL_0 = 100$ mm, subjected to a pretension of 0.01 cN/tex, and tested at a cross-head speed of 600 mm/min. Tests were conducted under standard atmospheric conditions (21 ± 1 °C, $65 \pm 2\%$ relative humidity), in accordance with EN ISO 139 [71].



FIGURE 3.8: Tensile testing device used for inlay yarn characterisation [113].

Each specimen underwent three consecutive loading–unloading cycles. The first two cycles consisted of loading up to 80% strain and unloading, while the third cycle was extended until rupture. During testing, the load F_{yarn} was recorded as a function of the imposed elongation. The corresponding yarn strain $\varepsilon_{\text{yarn}}$ was computed according to Equation 3.4, where YL_0 is the initial gauge length and YL the yarn length at a given time:

$$\varepsilon_{\text{yarn}} [\text{mm/mm}] = \frac{YL - YL_0}{YL_0}. \quad (3.4)$$

Yarn tenacity was expressed in cN/dtex using the recorded force and the nominal linear density of 494 dtex. Five inlay yarn specimens were tested, and because sample-to-sample variability was minimal, only the average response is presented.

3.3.2.2 Tensile response of inlay yarns

The tensile response of the inlay yarn is presented in Figures 3.9 to 3.11. Figure 3.9 shows the complete Force–Strain behaviour obtained from the three consecutive loading–unloading cycles. The yarn exhibits the characteristic nonlinear elastic response of double-covered elastane yarns.

During the initial part of the curve, up to approximately $\varepsilon_{\text{yarn}} \approx 2 \text{ mm/mm}$, the behaviour is quasi-linear. This region corresponds mainly to the extension of the polyamide covering and the onset of elastane core deformation. Beyond this strain, the yarn progressively stiffens until reaching a maximum deformation of $\varepsilon_{\text{yarn}} = 4.416 \text{ mm/mm}$ and a peak force of $F_{\text{yarn}} = 182 \text{ cN}$.

Small drops in force appear near $\varepsilon_{\text{yarn}} \approx 3.2 \text{ mm/mm}$, typically associated with local rearrangements and slippage of the covering filaments. From about $\varepsilon_{\text{yarn}} = 3.75 \text{ mm/mm}$, initial signs of fibre degradation are observed, eventually leading to rupture at maximum load.

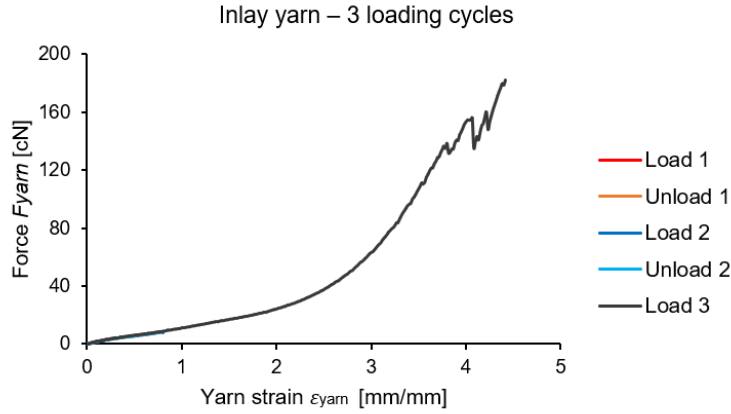


FIGURE 3.9: Force–Strain response of the inlay yarn during tensile testing during three consecutive cycles.

A zoom of the initial region of the Force–Strain curve is shown in Figure 3.10, limited to $\varepsilon_{\text{yarn}} \leq 1.3 \text{ mm/mm}$. This view highlights the reproducibility of the response over the three cycles and the very limited hysteresis. The yarn therefore exhibits a stable and reversible elastic behaviour within this moderate strain interval.

The in-use deformation of the fabric, reported in Table 3.4, corresponds to a circumferential strain range of approximately 0.49 to 1.15 mm/mm. Since the inlay yarn is inserted straight and does not form loops, its axial deformation follows the circumferential extension of the fabric with good approximation. Within this deformation range, the inlay yarn operates entirely inside the initial linear–elastic region of its tensile response.

Considering this, the third loading curve extended to failure, which provides the complete Force–Strain relation required for the modelling was used to define the stiffness function of the connector elements modelling the inlay yarn in the hybrid mechanical model in Chapter 4.

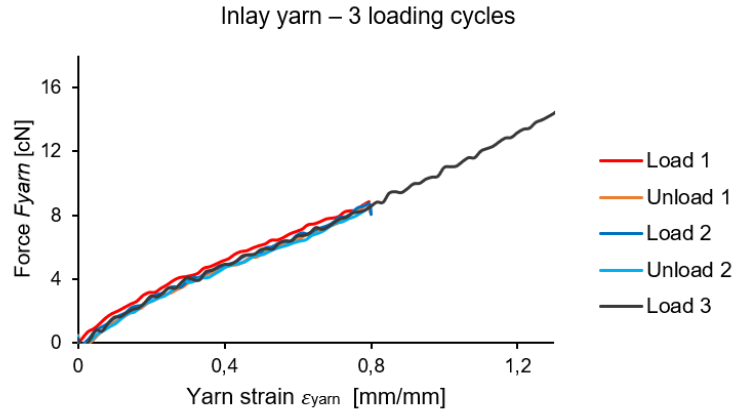


FIGURE 3.10: Zoom of the inlay yarn Force–Strain response in the initial elastic region during tensile testing.

The Tenacity–Strain curve obtained from the third loading cycle, extended to rupture, is shown in Figure 3.11. The average tenacity is 0.36 cN/dtex, consistent with reported values for polyamide-covered elastane yarns used in medical compression fabrics [24], [39], [96]. This confirms the high elastic recovery and stable tensile behaviour required for reliable circumferential reinforcement within the stocking.

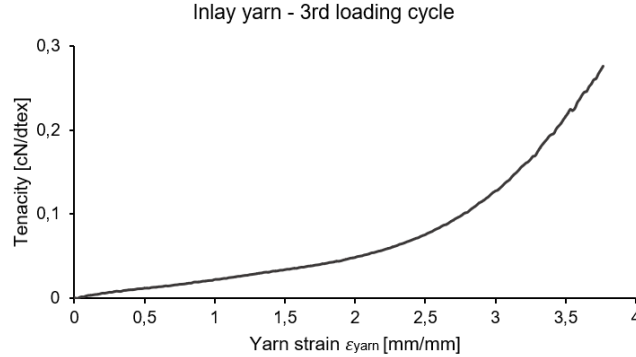


FIGURE 3.11: Tenacity–Strain response of the inlay yarn during tensile testing during third loading cycle before rupture.

3.3.3 Mechanical characterisation of knitted fabrics

The mechanical behaviour of the knitted fabrics was characterised to evaluate their circumferential response under tensile loading, reproducing the deformation experienced by the stocking during wear. This analysis complements the mechanical characterisation of the yarns by extending the study to the fabric scale, where the combined contribution of the

loop structure and the inlay yarns governs the overall stiffness and extensibility of the product. The results obtained here were later used to identify the parameters assigned to the loop structure in the hybrid model presented in Sections 4.2.2 and 4.2.4 of Chapter 4.

Three complementary aspects are analysed:

- the mechanical behaviour of the loop structure using samples knitted only with the ground yarn (GF),
- the reinforcing contribution of the inlay yarns by comparing knitted samples with and without inlay yarn (GF and IF samples),
- and the evolution of the tensile response between ankle, calf, and thigh resulting from the designed variation in D_{IY} .

3.3.3.1 Knitted fabric samples

The fabrics were tested in their cylindrical configuration, preserving the same geometry and structural conditions as in the final product. This approach offers a significant advantage over traditional cut-strip tests, which require opening and flattening the fabric, thereby removing the natural circumferential constraint and modifying the loop geometry. As highlighted by Siddique et al. [50], cut-strip testing systematically results in higher force–elongation values than cylindrical specimens because the boundary conditions no longer reflect in-use deformation. Testing the fabrics in their tubular form maintains the real knitting orientation, strain distribution, and load path, providing mechanical parameters directly applicable for pressure prediction and model validation.

All samples were knitted using the same materials and industrial parameters as the commercial stocking to ensure full representativeness of the product.

Two fabric configurations were produced:

- GF (without inlay yarn): ground-yarn-only fabrics reproducing the loop structure but without circumferential reinforcement,
- IF (with inlay yarn): fabrics containing both yarn systems, reproducing the full architecture of the commercial product.

For both configurations, samples of the three main compression zones of the product, ankle, calf and thigh, were knitted. In the IF configuration, the variation in density of inlay yarn (D_{IY}) identified in Section 3.3.1 was reproduced, following the designed gradient $D_{IY,ankle} > D_{IY,calf} > D_{IY,thigh}$.

Coloured reference lines were knitted into the fabric samples during production. The distance between two lines, denoted as d , was used both to ensure the accurate placement and alignment of the specimens during the tensile tests, and to control the axial deformation of the fabric simulating the in-use conditions. These knitted reference marks remained visible on the cylindrical specimens during testing, allowing precise alignment of the samples and ensuring consistent placement between tests. The number of courses between two reference lines, NC , was provided by the manufacturer and is listed in Table 3.2. Together, the parameters d and NC defined the density of inlay yarn (D_{IY}) for each compression zone, as introduced in Section 3.3.1.

Figure 3.12 presents the knitted samples of the three compression zones for both fabric configurations, without inlay yarn (GF) and with inlay yarn (IF), where W_0 is the flat fabric width, and d the interline reference distance.

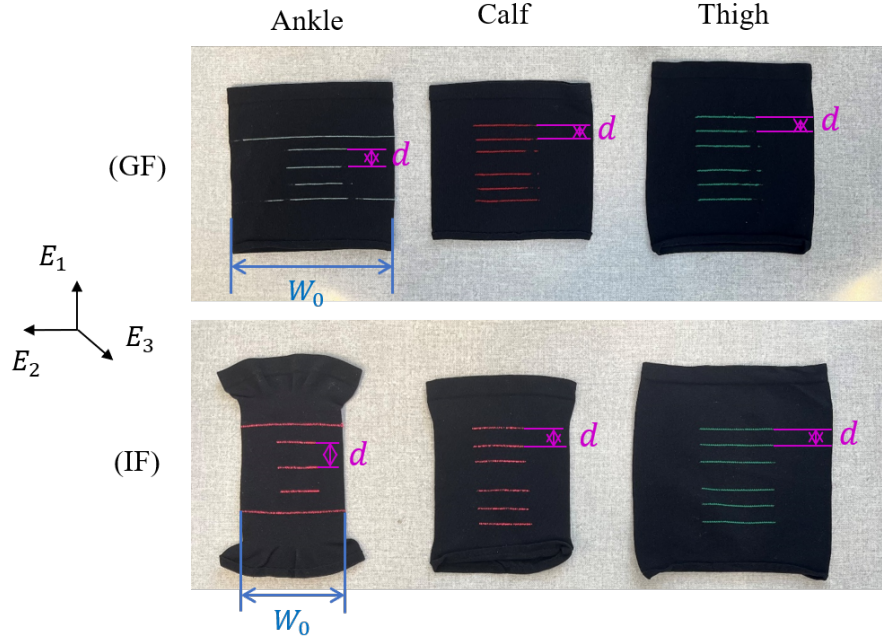


FIGURE 3.12: Dimensional parameters of weft-knitted fabric samples from both configurations, without inlay yarn (GF) and with inlay yarn (IF): initial flat width (W_0) and interline distance (d).

According to the french standard [46], prior to testing, each sample was mounted on the corresponding region of a Hohenstein wooden leg and manually massaged for 30 s. The samples were then removed, placed flat, and allowed to relax for two hours. This step preconditions the fabric, simulating the loading during use.

The flat width (W_0) was then measured and used to compute the initial fabric perimeter (P_0) according to Equation 3.1. Table 3.3 reports the measured widths and calculated initial perimeters (P_0) for all zones and both configurations. It is observed that the insertion of the

inlay yarn, constraining the circumferential deformation, results in smaller fabric width (W_0) and perimeter (P_0) in samples with inlay yarn (IF).

TABLE 3.3: Initial fabric dimensions for both fabric configurations, without inlay yarn (GF) and with inlay yarn (IF), measured after precondition on wooden leg: width (W_0) and perimeter (P_0)

Compression zone	W_0 [mm]		P_0 [mm]	
	(GF)	(IF)	(GF)	(IF)
Ankle	114	74	228	148
Calf	118.5	98	237	196
Thigh	127.5	144	255	288

3.3.3.2 In-use circumferential strain range

When the stocking is worn, the fabric stretches from its relaxed state to the deformed configuration imposed by the leg, as shown in Figure 3.6. The generated circumferential strain (ϵ_{circ}) is given by Equation 3.2, where P_0 is the initial fabric perimeter and $P = P_{\text{leg}}$ is the perimeter of the corresponding leg zone.

Using the measured initial perimeter data (P_0), from Table 3.3, and the reference leg perimeters (P_{leg}) from Table 3.1, the resulting circumferential strain ranges for the three zones are reported in Table 3.4. The strain is lowest at the ankle and increases towards the calf and thigh. This trend reflects the functional design of graduated compression: high stiffness combined with moderate deformation produces greater pressure at the ankle, whereas lower stiffness and larger deformation in the upper zones produce a controlled reduction in pressure [24].

TABLE 3.4: Range of circumferential strain (ϵ_{circ}) for the different compression zones under in-use conditions

Zone	ϵ_{circ} [mm/mm]
Ankle	0.49 - 0.62
Calf	0.74 - 1.09
Thigh	0.70 - 1.15

These deformation ranges define the in-use configuration of the medical compression stocking and were used as boundary conditions for the fabric tensile tests. They also provide the reference deformation levels used in the hybrid mechanical model in Chapter 4 and the finite element simulations developed in Chapter 5.

3.3.3.3 Tensile testing of knitted fabrics

The tensile characterisation was performed according to an in-house protocol adapted from the French standard NF G30-102, Annex B [46], specifically developed for elastic knitted fabrics in tubular configuration. This method has been widely used to assess the tensile response of medical compression stockings [32], [33]. The resulting data are frequently employed to predict interface pressure using Laplace's law [22], [47], [48], [96] and to identify the mechanical parameters of behaviour models of compression fabric mechanics [69], [76], [81].

Tensile tests were carried out on cylindrical fabric samples using a Zwick/Roell Z005 universal testing machine equipped with a 100 N XP-Force load cell. The samples were clamped along the longitudinal direction (E_1) using metallic pins inserted in the coloured reference lines, as shown in Figure 3.13.a. The pins imposed and maintained a constant interline distance (d), simulating the in-use longitudinal deformation of the fabric. The value $d = 30$ mm was used for all compression zones. This ensured consistency with the structural characterisation performed in Section 3.3.1, where the in-use density of inlay yarn (D_{IY}) was determined. As a result, the mechanical response measured during tensile testing reflected the actual knitted geometry of the fabrics during wear.

In the ankle samples, where the knitted structure is more compact, stabilisers were used to prevent secondary displacements in the longitudinal direction (E_1) and ensure that deformation occurred only along the circumferential direction (E_2). These stabilisers, shown in Figure 3.13.b, maintained the sample length (L) across the entire width (W), and ensured uniform strain distribution. For the calf and thigh samples, the use of stabilisers was unnecessary, as edge effects were negligible.

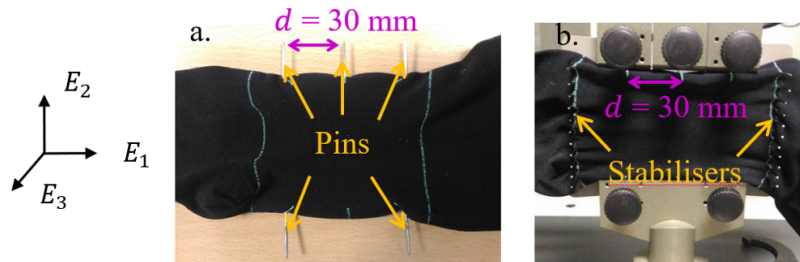


FIGURE 3.13: (a) Placement of fixation pins along the longitudinal direction (E_1) at the coloured reference lines, maintaining a constant interline distance ($d = 30$ mm). (b) Attachment of the cylindrical sample to the machine along the longitudinal direction, including the stabilisers for ankle samples.

The imposed upper-clamp displacement (u_2) was applied along the circumferential direction (E_2), as shown in Figures 3.14, 3.15, which corresponds to the circumferential direction once the stocking is worn. In the flat test configuration, this displacement increased the clamp-to-clamp distance (C) and, consequently, the instantaneous fabric width (W) through Equation 3.5. Since the circumference of a cylindrical stocking is twice the flat width ($P = 2W$), the imposed displacement (u_2) directly produced a controlled variation in fabric perimeter. The test therefore reproduced the circumferential deformation experienced in use, where the perimeter increases from its relaxed value (P_0) to the leg perimeter ($P = P_{\text{leg}}$).

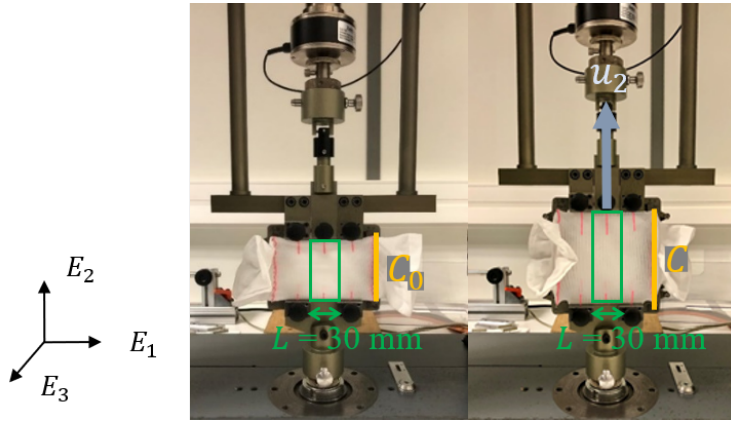


FIGURE 3.14: Loading cycle along the circumferential direction (E_2) during tensile test of cylindrical knitted fabrics, where C_0 is the initial clamp-to-clamp distance, C the clamp-to-clamp distance during test, and L the fabric length.

The relation between the deformed fabric width (W) and the clamp-to-clamp distance (C) was given by Equation 3.5, where $\phi_{\text{pin}} = 6.05 \text{ mm}$ is the pin diameter. The initial clamp-to-clamp distance (C_0) was set to 60 mm for ankle samples and 50 mm for calf and thigh samples.

$$W [\text{mm}] = C + \frac{\pi \cdot \phi_{\text{pin}}}{2} \quad (3.5)$$

For data processing, a preload of 0.01 N was considered for GF samples and 0.1 N for IF samples to account for residual slack. The instantaneous fabric perimeter (P) was then calculated using Equation 3.6, and the circumferential strain (ϵ_{circ}) as the relative change in width, given in Equation 3.7.

$$P [\text{mm}] = 2W \quad (3.6)$$

$$\varepsilon_{\text{circ}} [\text{mm/mm}] = \frac{P - P_0}{P_0} = \frac{W - W_0}{W_0} \quad (3.7)$$

During the test, the fabric was stretched until its deformed perimeter reached the maximal leg perimeter ($P = P_{\text{leg,max}}$) for the corresponding compression zone, as defined in Table 3.1 and represented in Figure 3.15. The upper clamp of the testing device, connected to the load cell, was displaced at a constant rate of 500 mm/min for the IF samples and 200 mm/min for the GF samples. In the GF ankle sample, where P_0 and $P_{\text{leg,max}}$ were very close, the target deformation was extended to 320 mm to enable a more complete analysis.

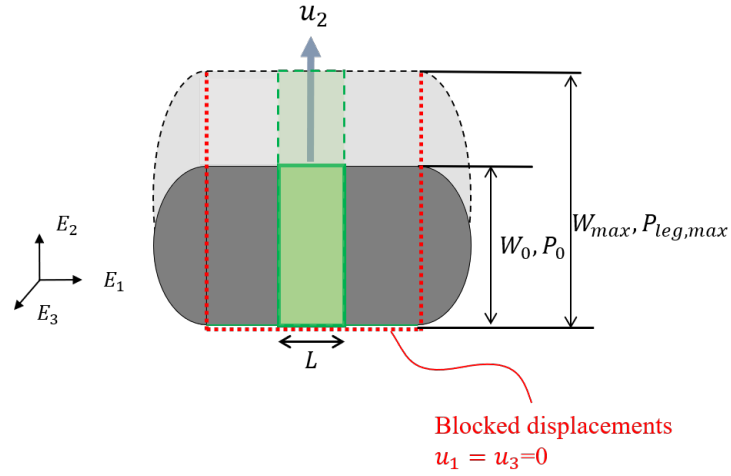


FIGURE 3.15: Dimensional parameters during tensile testing with controlled displacement along circumferential direction (u_2) and blocked displacements in the longitudinal and thickness directions ($u_1 = u_3 = 0$): initial fabric width (W_0) and perimeter (P_0), and maximal fabric width (W_{max}) and perimeter (W_{max}).

During the loading, the total force ($F_{2-\text{total}}$) was recorded by the load cell. Because the cylindrical sample consists of two fabric layers aligned in the loading direction, the measured force represents the combined contribution of both layers, as represented in Figure 3.16. For consistency with analytical modelling and interface-pressure calculations, where the fabric is treated as a single-layer membrane [50], [80], [105], the internal force in one layer (F_2) was computed with Equation 3.8.

$$F_2 [\text{N}] = \frac{F_{2-\text{total}}}{2} \quad (3.8)$$

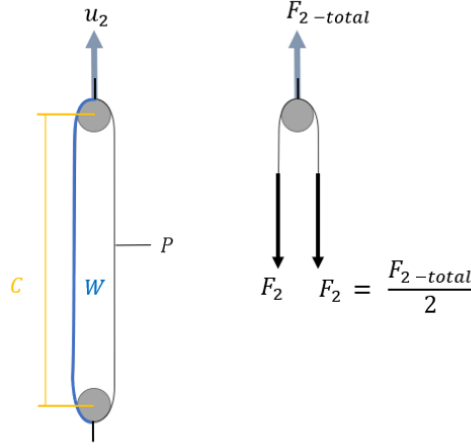


FIGURE 3.16: Distribution of the total force ($F_{2-total}$) in the two fabric layers during uniaxial loading.

The effective fabric length ($L = 30$ mm) was defined by the central pin width, as represented in Figure 3.14. The fabric tension (T) was then computed as a function of the force (F_2) per unit length (L).

Figure 3.17 shows the different steps of the tensile test. The initial instant t_0 represents the clamping of the fabric sample to the machine, with the initial clamp-to-clamp distance C_0 set according to the compression zone, and the fabric in a relaxed configuration (not fully extended). At instant t_1 , the axial load along the circumferential direction (F_2) is applied and the fabric is fully extended but not yet deformed, with the perimeter still equal to its initial value $P_0 = 2 \cdot W_0$. The instant t_2 represents an intermediate stage of deformation, where the fabric is partially stretched with a deformed perimeter $P = 2 \cdot W$. The final instant t_f corresponds to the maximum imposed elongation, at which the fabric perimeter matches the maximum leg perimeter $P = P_{leg, max} = 2 \cdot W_{max}$, defined in the manufacturer's size chart for each compression zone.

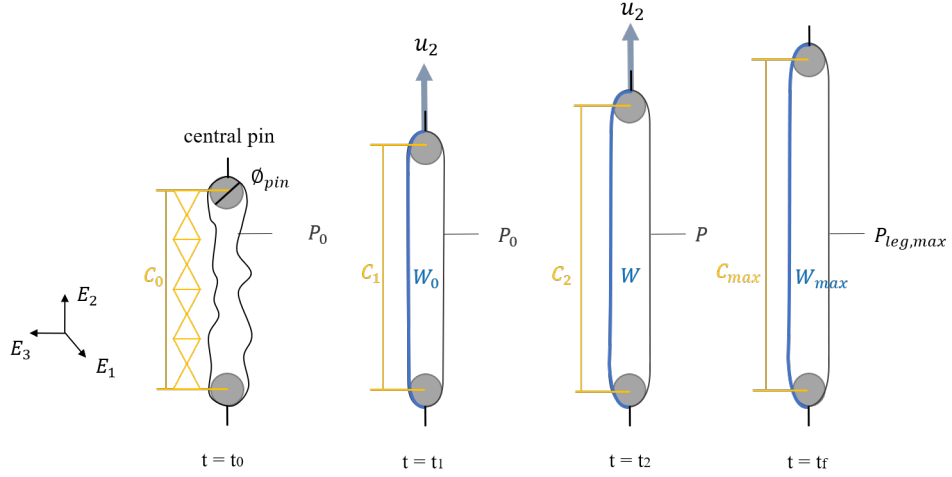


FIGURE 3.17: Schematic lateral representation of the different steps of the tensile test on circumferential fabric samples.

Two loading–unloading cycles were applied. The first cycle served to stabilise the fabric by releasing residual stresses and internal rearrangements. The second cycle was used to evaluate the tensile response representative of in-use conditions.

The results were obtained as Force–Perimeter curves and subsequently converted into Tension–Circumferential strain curves using Equation 3.7. Although only one specimen per zone was available for each configuration (GF and IF), the results were validated by the manufacturer and found consistent with expected mechanical performance.

GF samples provided the behaviour of the loop structure alone and were later used to identify the shell mechanical parameters in Chapter 4. IF samples characterised the combined action of loops and inlay yarns and enabled assessment of the influence of density of inlay yarn (D_{IY}). These results were also used for interface-pressure prediction (Section 3.4.1) and for validating the hybrid model in Chapter 4.

3.3.3.4 Tensile response of knitted fabrics

The circumferential tensile response of the knitted fabrics was analysed separately for the configurations without inlay yarn (GF) and with inlay yarn (IF), in order to identify the contributions of the loop structure and the inlay reinforcement to the overall fabric behaviour. For each configuration, the evolution of the applied force with leg perimeter was first analysed for the three compression zones (ankle, calf, and thigh), followed by a comparison of the corresponding Tension–Circumferential strain curves obtained from the second loading cycle.

Knitted fabrics without inlay yarn - GF Figures 3.18–3.20 present the Force–Perimeter curves obtained for the weft-knitted fabric samples without inlay yarn (GF) in the three compression zones. The mechanical response of the GF configuration is characterised by very low force levels, never exceeding 0.5 N even at maximum deformation, and by a quasi-linear evolution with limited circumferential stiffness.

During the first loading cycle, pronounced hysteresis and a marked preload effect appear in all zones, indicating substantial internal rearrangements of the loop structure. These rearrangements include yarn slippage, loop opening, and progressive reorientation of the knitted geometry. Such mechanisms are typical of extensible weft-knitted structures with low lateral constraint, and have been widely reported for elastomeric loop-dominated fabrics [40], [114].

The unloading branches exhibit significant residual strain, confirming the viscoelastic behaviour of the knitted ground structure. This response arises from frictional losses inside the loops and from irreversible micro-reorganisation of the elastane–polyamide system under tension.

A key result is that the ankle, calf, and thigh samples exhibit nearly identical responses. Since both the material composition and the knitted architecture of the ground fabric are the same across the stocking, the circumferential stiffness is governed exclusively by the loop structure, which shows a homogeneous behaviour along the product. In the absence of inlay yarns, all circumferential load is carried by the loop structure, which therefore determines the stiffness measured in the GF configuration. This confirms that the loop system alone cannot generate the stiffness gradient required for graduated compression, highlighting the essential role of the inlay reinforcement in the mechanical differentiation between zones.

Moreover, the maximal forces measured on GF samples remain an order of magnitude lower than those obtained in the inlaid configuration, presented below in Section 3.3.3.4. This reinforces the observation that the ground-loop structure contributes only marginally to circumferential reinforcement.

These GF results provide a direct experimental reference for isolating the mechanical behaviour of the loop subsystem. They were subsequently used to identify the mechanical parameters of the shell elements representing the loop structure in the hybrid mechanical model developed in Chapter 4.

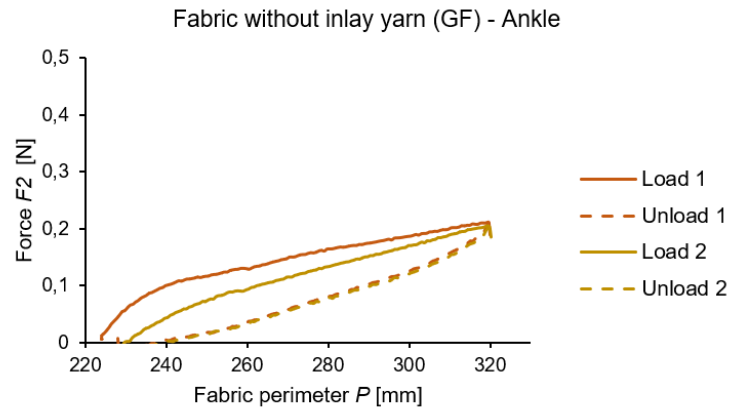


FIGURE 3.18: Evolution of registered force (F_2) as a function of fabric perimeter (P) during tensile test for ankle zone of knitted sample without inlay yarn

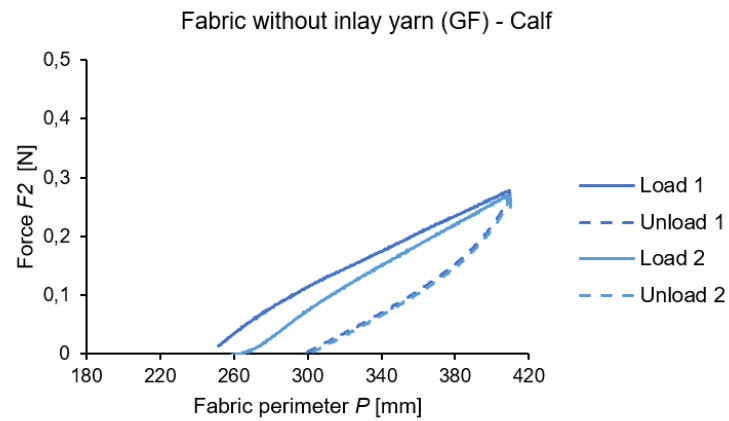


FIGURE 3.19: Evolution of registered force (F_2) as a function of fabric perimeter (P) during tensile test for calf zone of knitted sample without inlay yarn

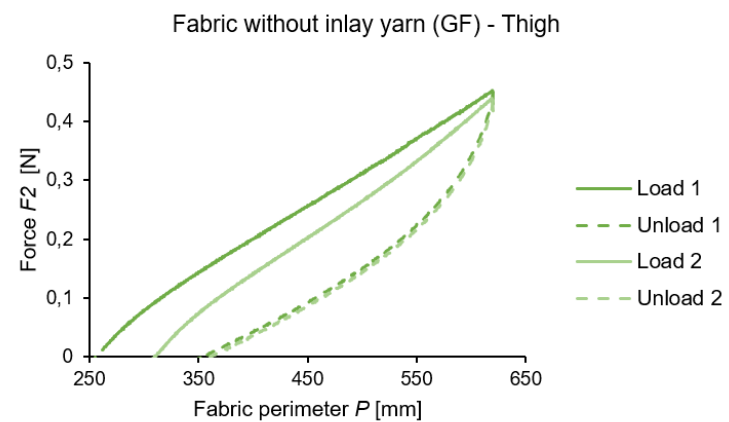


FIGURE 3.20: Evolution of registered force (F_2) as a function of fabric perimeter (P) during tensile test for thigh zone of knitted sample without inlay yarn

Knitted fabrics with inlay yarn - IF Figures 3.21–3.23 present the Force–Perimeter curves obtained for the inlaid weft-knitted fabrics (IF) in the three compression zones during the two loading–unloading cycles. In contrast with the GF configuration, the IF samples exhibit a clearly non-linear response, characterised by a progressive increase in stiffness with deformation. This behaviour reflects the hybrid nature of the structure: at low strains, the knitted loop matrix governs the response, whereas at higher strains the inlay yarns progressively engage and become the dominant load-bearing component.

At the ankle (Figure 3.21), the loading and unloading branches of both cycles almost coincide, showing minimal hysteresis and excellent repeatability. This indicates a mechanically stabilised structure with limited internal rearrangement, consistent with the high density of inlay yarn (D_{IY}) in this zone, which restricts loop mobility and constrains transverse deformation. Because the circumferential strain range experienced at the ankle is relatively small, the curves remain mostly within the low-to-moderate deformation regime and do not fully capture the stiffening phase that becomes prominent at higher strain levels.

In the calf and thigh zones (Figures 3.22 and 3.23), the evolution of stiffness with deformation is much more pronounced. After an initial low-stiffness section, associated with the alignment of the loops and the straightening of the inlay yarns, the curves exhibit a progressive increase in slope as the inlay yarns become fully tensioned. This transition illustrates the gradual load transfer from the knitted ground structure to the inlay reinforcement, which governs the high-strain behaviour. A moderate preload effect is observed in the first cycle, reflecting internal reorganisation during the initial extension. The second cycle shows reduced hysteresis and negligible residual strain, indicating that the structure reaches a stable mechanical state once the internal geometry is aligned. Such non-linear stiffening and viscoelastic features have been consistently reported for inlaid weft-knitted structures designed for compression applications [41], [68], [115].

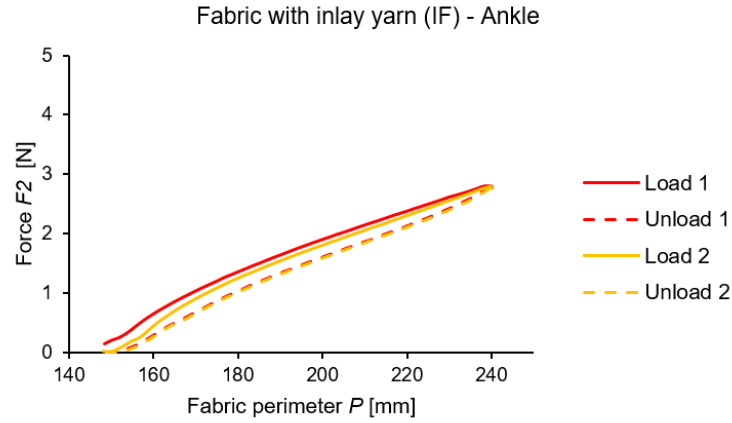


FIGURE 3.21: Evolution of registered force (F_2) as a function of fabric perimeter (P) during tensile test for ankle zone of knitted sample with inlay yarn.

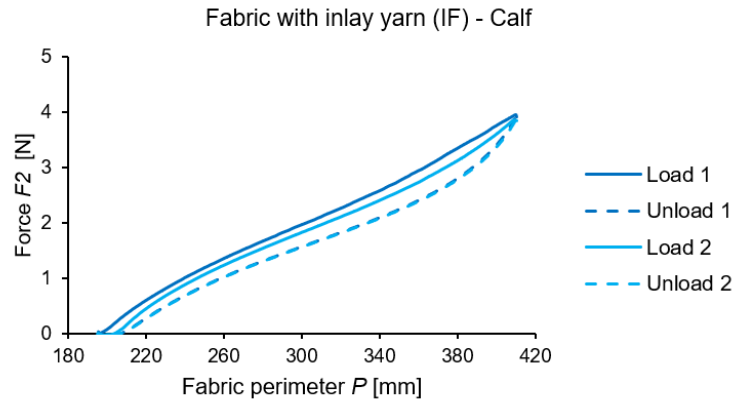


FIGURE 3.22: Evolution of registered force (F_2) as a function of fabric perimeter (P) during tensile test for calf zone of knitted sample with inlay yarn.

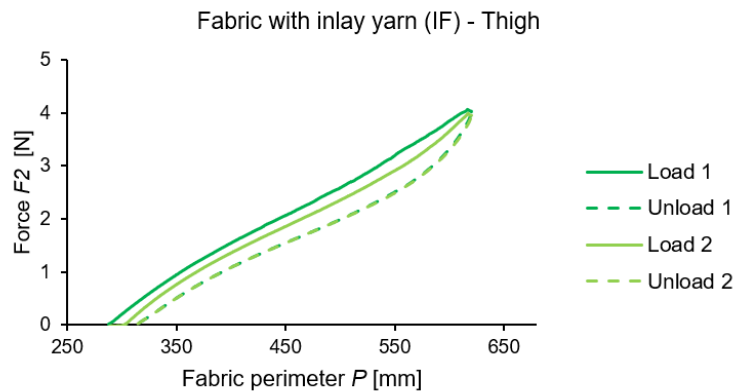


FIGURE 3.23: Evolution of registered force (F_2) as a function of fabric perimeter (P) during tensile test for thigh zone of knitted sample with inlay yarn.

A comparison between GF and IF configurations reveals that the overall energy dissipation is significantly lower in the IF samples. This reduced hysteresis confirms the stabilising effect of the inlay reinforcement, which limits loop reorganisation and provides a more repeatable and mechanically controlled circumferential response.

The comparison between the GF and IF configurations highlights the fundamental role of the inlay reinforcement in governing the circumferential behaviour of the fabric. In the absence of inlay yarn (GF), the response is soft, highly hysteretic, and dominated by loop rearrangement, with negligible differences between the anatomical zones. When the inlay yarn is inserted (IF), the stiffness increases by more than one order of magnitude, the response becomes non-linear, and the deformation mechanisms shift from loop-driven extensibility to yarn-controlled stiffening. These contrasting behaviours demonstrate that the circumferential mechanics of medical compression fabrics cannot be attributed to the loop structure alone and must be understood as the combined contribution of the two yarn systems. The following sections quantify the specific influence of the inlay reinforcement and analyse how the controlled variation of the density of inlay yarn (D_{IY}) between zones generates the stiffness gradient required for graduated compression.

Effect of inlay yarn reinforcement To quantitatively compare the knitted fabrics and assess the contribution of the inlay reinforcement, the Load–Perimeter data were converted into Tension–Circumferential strain curves according to the relationships defined in Section 3.3.3.3. The registered force (F_2) was normalised by the effective fabric length ($L = 30$ mm) to obtain the fabric tension (T), while the circumferential strain (ϵ_{circ}) was computed from the relative change in perimeter using Equation 3.2.

Figures 3.24 and 3.25 show the resulting curves for the inlaid (IF) and ground (GF) configurations, respectively, obtained from the second loading cycle.

The IF curves, given in Figure 3.24, exhibit a distinctly nonlinear tensile response, characterised by a progressive increase in stiffness as strain grows. The ankle zone reaches the highest tension levels at relatively low ϵ_{circ} , reflecting its increased stiffness and the higher density of inlay yarn (D_{IY}) in this region. In contrast, the calf and thigh zones display more gradual tension increases and extend over wider deformation ranges, consistent with their lower reinforcement content and reduced circumferential stiffness. At larger strains, both zones show a clear stiffening regime, confirming that load is gradually transferred from the compliant loop structure to the inlay system as the inlay yarns align and elongate. This evolution in slope reflects the hybrid mechanical behaviour of the knitted architecture and the mesostructural origin of its nonlinearity.

The GF curves, given in Figure 3.25, display a quasi-linear response over the entire deformation range, with substantially lower tension values and minimal differences between compression zones. This confirms that, in the absence of inlay reinforcement, the circumferential stiffness of the fabric is dominated by the extensibility of the loop structure, which is nearly identical across zones. The comparison between GF and IF responses thereby demonstrates the essential role of the inlay yarns in increasing the circumferential stiffness and enabling the generation of significant tensile forces required for therapeutic compression. The loop structure primarily contributes to extensibility and comfort, whereas the inlay system controls the mechanical efficiency of the fabric in producing pressure, in agreement with previous studies on the functional design of compression garments [21], [39], [112].

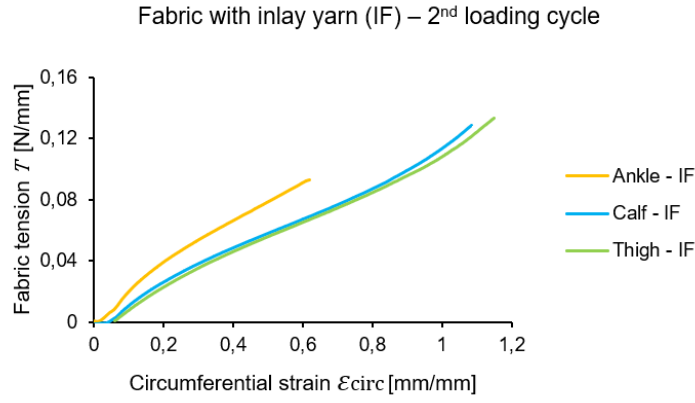


FIGURE 3.24: Evolution of fabric tension (T) as a function of circumferential strain (ϵ_{circ}) during tensile test for the three compression zones of inlaid weft-knitted samples (IF).

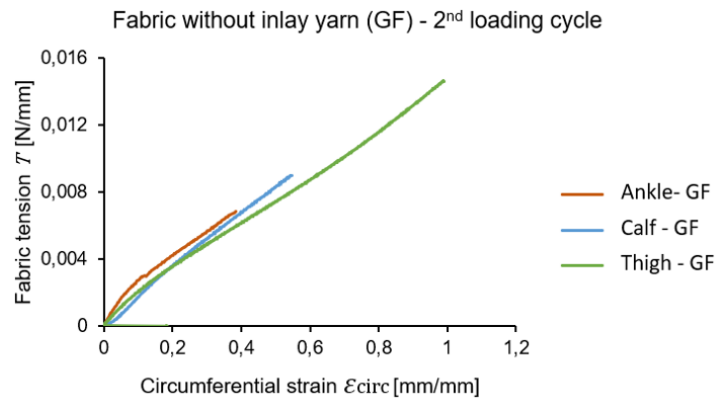


FIGURE 3.25: Evolution of fabric tension (T) as a function of circumferential strain (ϵ_{circ}) during tensile test for the three compression zones of weft-knitted samples without inlay yarn (GF).

It should be noted that the circumferential strain ranges achieved in the tests differ between the two configurations because the same leg perimeters (P_{leg}) were imposed on GF

and IF fabrics, while their initial relaxed perimeters (P_0) were not identical. In particular, for the GF ankle zone, the relaxed perimeter exceeded the minimal leg perimeter ($P_{\text{leg,min}}$), preventing the measurement of forces in this lower deformation range. These differences do not affect the overall interpretation: even at their maximum extension, the GF samples produce tensions approximately one order of magnitude lower than the IF samples. This observation further confirms the dominant role of the inlay reinforcement in defining the circumferential stiffness and its central contribution to the graduated compression effect.

Tables 3.5 and 3.6 report the tension values (T) at the three reference leg perimeters, minimal ($P_{\text{leg,min}}$), average ($P_{\text{leg,avg}}$), and maximal ($P_{\text{leg,max}}$), for the GF and IF configurations.

Across both configurations, the absolute tension values increase from ankle to thigh, because the imposed perimeters (P_{leg}) are larger for proximal zones, resulting in greater overall deformation. However, the effective stiffness, reflected by the slope of the T - $\varepsilon_{\text{circ}}$ curves, remains highest at the ankle and decreases steadily toward the thigh, mirroring the reduction in D_{IY} .

TABLE 3.5: Fabric tension (T) at reference leg perimeters for fabrics without inlay yarn (GF).

Compression zone	$T(P_{\text{leg,min}})$ [N/mm]	$T(P_{\text{leg,avg}})$ [N/mm]	$T(P_{\text{leg,max}})$ [N/mm]
Ankle	–	0.0000383	0.00145
Calf	0.00436	0.00678	0.00830
Thigh	0.00836	0.01140	0.01390

TABLE 3.6: Fabric tension (T) at reference leg perimeters for fabrics with inlay yarn (IF).

Compression zone	$T(P_{\text{leg,min}})$ [N/mm]	$T(P_{\text{leg,avg}})$ [N/mm]	$T(P_{\text{leg,max}})$ [N/mm]
Ankle	0.077	0.085	0.092
Calf	0.079	0.104	0.128
Thigh	0.085	0.114	0.134

The following section examines in detail how the variation in the density of inlay yarn (D_{IY}) between compression zones governs the progressive modulation of circumferential stiffness and, consequently, the graduated compression profile.

Effect of density-of-inlay-yarn variation The influence of the density of inlay yarn (D_{IY}) on the circumferential stiffness of the knitted fabrics was examined by comparing the mechanical responses of the three compression zones. As reported in Table 3.2, D_{IY} decreases

progressively from ankle to thigh, following the structural design of the product. This variation is directly reflected in the $T - \varepsilon_{\text{circ}}$ curves of the IF configuration, given in Figure 3.24, where the slope of the response decreases toward the proximal zones.

The ankle, which presents the highest D_{IY} , shows the steepest increase in tension with strain, indicating greater circumferential stiffness and a more constrained deformation of the knitted matrix. In contrast, the calf and thigh, where the inlay yarns are more spaced, develop tension more gradually and exhibit a less stiff initial response. This change in slope demonstrates that D_{IY} is the primary structural parameter controlling the rate of tension development with circumferential deformation. The absolute tension values, however, also depend on the imposed leg perimeter (P_{leg}). Thus, even though the thigh reaches higher tensions at $P_{\text{leg,max}}$, its stiffness, defined by the tangent modulus of the $T - \varepsilon_{\text{circ}}$ curve, remains lower than that of the ankle, fully consistent with its reduced reinforcement content.

In the GF configuration, the $T - \varepsilon_{\text{circ}}$ curves, given in Figure 3.25, are nearly identical for all three zones. This confirms that the loop structure alone is unable to generate the stiffness gradient required for graduated compression. Therefore, the mechanical differentiation of the stocking arises exclusively from the modulation of D_{IY} along the leg, since the yarn materials and knitting parameters remain constant across zones.

The progressive reduction in stiffness from ankle to thigh reflects the core design principle of medical compression stockings: high circumferential stiffness in the distal region to generate the prescribed compression at the ankle, followed by a controlled decrease toward the thigh to preserve comfort while maintaining an effective pressure gradient. These experimental findings establish a direct relationship between inlay-yarn distribution, circumferential stiffness, and the graduated compression effect. They also provide the structural foundation for the hybrid mechanical model developed in Chapter 4, where the variation of D_{IY} was explicitly implemented to reproduce the stiffness gradient and the resulting pressure distribution along the stocking.

3.4 Interface pressure prediction

The interface pressure (*IP*) exerted by a medical compression stocking on the leg is the primary therapeutic parameter governing its clinical effectiveness. It results from the internal circumferential tension of the fabric (T), generated by the imposed circumferential strain $\varepsilon_{\text{circ}}$ when the stocking is stretched from its relaxed perimeter to the larger leg perimeter, and from the local geometry of the limb. In this work, the interface pressure was evaluated using two complementary approaches:

- an analytical prediction based on the tensile response of the knitted fabrics combined with Laplace's law, and
- direct measurements using pressure sensors.

3.4.1 Prediction of interface pressure based on Laplace's law

Laplace's law provides a classical analytical method for estimating the average pressure exerted by a thin membrane wrapped around a cylindrical surface [44], [45], [95]. Under the assumptions of uniform tension distribution, circular cross-section, and full contact between fabric and limb, the interface pressure (IP_{Laplace}) can be expressed as a function of the circumferential tension (T) and the deformed fabric perimeter (P). Following the membrane formulation, the pressure is calculated with Equation 3.9.

$$IP_{\text{Laplace}} = T \cdot \frac{2\pi}{P} \quad (3.9)$$

The experimental inputs required for this calculation were obtained from the tensile tests performed on cylindrical samples of the knitted fabric. As described in Section 3.3.3.3, the samples were elongated along the circumferential direction (E_2), reproducing the circumferential deformation experienced during wear. The deformation was controlled through the clamp displacement applied along E_2 , and the samples were stretched until the deformed perimeter matched the maximal leg perimeter ($P = P_{\text{leg,max}}$) defined by the manufacturer for each compression zone, given in Table 3.1. For each intermediate value of perimeter (P), the axial force (F_2) was registered, providing the Force–Perimeter curves reported in Figures 3.21–3.23 for the inlaid configuration, and Figures 3.18–3.20 for the ground-fabric configuration.

The conversion of these data for the calculation of Laplace's interface pressure followed the definitions established in Section 3.3.3.3. The circumferential strain was obtained from the relative perimeter variation using Equation 3.2, where (P_0) is the initial fabric perimeter and (P) the deformed perimeter. The tension (T) was computed from the axial force according to the membrane definition, by normalising F_2 with the effective fabric length (L) equal to 30 mm, as performed in the mechanical characterisation of the fabrics. The deformed perimeter (P) was calculated as twice the instantaneous fabric width, using the geometric relation of Equation 3.6, where W was obtained from the clamp-to-clamp distance through Equation 3.5. This procedure provided, for each measurement point, the pair (T , P) required for the evaluation of IP_{Laplace} through Equation 3.9.

The resulting Interface pressure–Circumferential strain curves describe the evolution of the pressure generated under controlled deformation for the three compression zones and for both fabric configurations. The comparison between the inlaid (IF) and ground (GF) fabrics reveals the reinforcing effect of the inlay yarn on the delivered pressure, while the comparison between ankle, calf, and thigh zones within the IF configuration highlights the influence of the density of inlay yarn (D_{IY}) on the pressure gradient produced along the leg.

Although Laplace’s law relies on simplified assumptions regarding geometry and stress distribution, it remains an effective analytical tool for estimating the average interface pressure delivered by the knitted fabrics. In the present work, this analytical approach provides a direct link between the mechanical characterisation of the knitted structures and the pressure generated under in-use deformation, and it serves as a reference for the validation of the finite element simulations developed in Chapter 5.

3.4.1.1 Evolution of pressure with circumferential strain

Figures 3.26 and 3.27 present the predicted evolution of the interface pressure ($IP_{Laplace}$) as a function of the circumferential strain (ϵ_{circ}) for both fabric configurations: with inlay yarn (IF) and without inlay yarn (GF). These Interface pressure–Strain curves directly reflect the mechanical behaviour of the knitted fabrics under circumferential extension and reproduce the non-linear stiffening observed in the tensile tests.

For the IF configuration (Figure 3.26), the ankle zone, characterised by the highest density of inlay yarn (D_{IY}) and the smallest leg perimeter, exhibits the steepest increase in pressure with strain and reaches the highest pressure levels. The calf zone displays an intermediate response, while the thigh zone, with the lowest D_{IY} and bigger leg perimeter, shows the most gradual pressure increase. This progression confirms the direct mechanical role of the inlay-yarn reinforcement in controlling the circumferential stiffness of the fabric and, consequently, the local compression delivered along the leg.

The differences between zones appear not only in the pressure magnitude but also in the curvature of the $IP_{Laplace}-\epsilon_{circ}$ curves: as D_{IY} decreases towards the top of the leg, the response becomes progressively softer and less non-linear. This behaviour mirrors the structural design of the product and the experimentally observed tension–strain responses.

In contrast, the GF configuration (Figure 3.27), with no effect of inlay yarn reinforcement, results in interface pressures below 0.16 kPa across all strain levels and zones. These extremely low values are consistent with the limited circumferential stiffness of the loop structure alone. When inlay yarns are present (IF configuration), the predicted pressures increase by more than one order of magnitude, reaching up to 2.4 kPa.

This comparison highlights the dominant contribution of the inlay yarn to the circumferential reinforcement of the knitted fabric and confirms its essential role in generating the therapeutic pressure required for graduated compression.

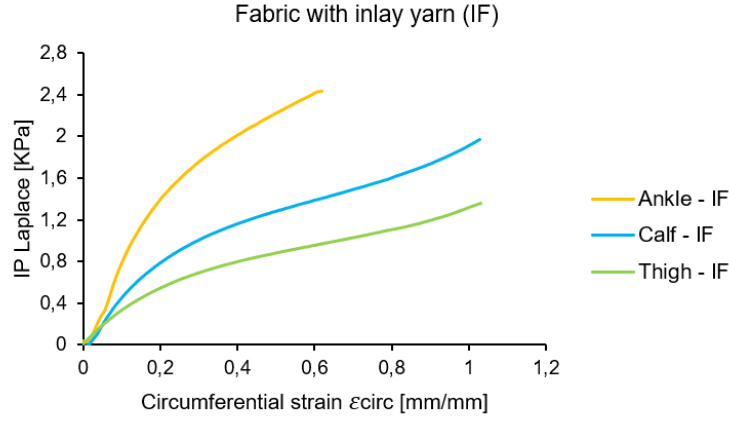


FIGURE 3.26: Evolution of Laplace's interface pressure (IP_{Laplace}) as a function of circumferential strain (ϵ_{circ}) for different compression zones of knitted fabrics with inlay yarn (IF configuration).

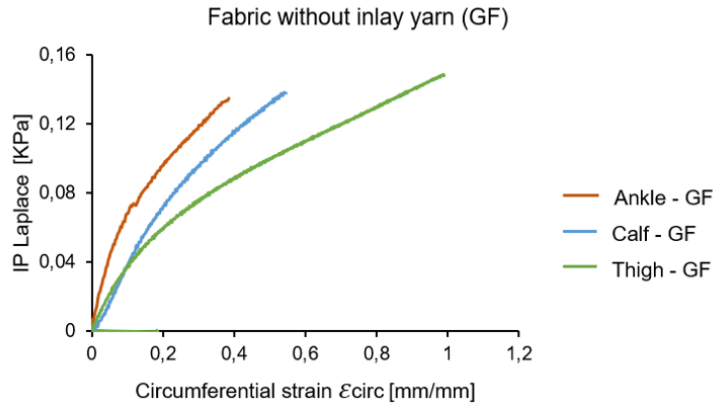


FIGURE 3.27: Evolution of Laplace's interface pressure (IP_{Laplace}) as a function of circumferential strain (ϵ_{circ}) for different compression zones of knitted fabrics without inlay yarn (GF configuration).

3.4.1.2 Predicted pressures for reference leg perimeters

To illustrate the interface pressure values at representative deformation levels, Figures 3.28 and 3.29 present the estimated Laplace pressure (IP_{Laplace}) for the three reference leg perimeters, minimal ($P_{\text{leg,min}}$), average ($P_{\text{leg,avg}}$), and maximal ($P_{\text{leg,max}}$), defined from the manufacturer's size table (Table 3.1). These perimeters determine the range of circumferential deformation experienced by the fabric in use (Table 3.4) and provide direct insight into the combined influence of leg geometry and structural stiffness on the delivered compression.

For the fabric with inlay yarn (IF), the predicted interface pressure (Figure 3.28) increases systematically with P_{leg} for all compression zones. As the leg perimeter grows from its minimal to maximal reference value, the circumferential strain imposed on the fabric increases, leading to higher tension and, consequently, higher pressure according to Laplace's law. The predicted pressures range from 2.19–2.41 kPa at the ankle, 1.43–1.96 kPa at the calf, and 0.95–1.34 kPa at the thigh. The ankle consistently exhibits the highest pressure for each perimeter, reflecting its higher density of inlay yarn (D_{IY}) and smaller perimeter. This confirms that the graduated compression profile results from the combined effects of structural reinforcement and leg geometry.

In contrast, the ground-fabric configuration (GF), which does not include inlay-yarn reinforcement, does not display a consistent decreasing gradient from ankle to thigh (Figure 3.29). The predicted pressures remain below 0.15 kPa in all zones and vary irregularly with P_{leg} , primarily due to differences in the relaxed perimeters of the GF samples and their very low circumferential stiffness. Because the same P_{leg} values were imposed for IF and GF samples, the resulting circumferential strains were not identical, producing small apparent variations in the pressure curves. Nevertheless, the resulting pressures remain negligible, confirming that the loop structure alone cannot generate therapeutic compression.

Overall, the analysis across leg perimeters highlights that the circumferential strain, defined by the difference between P_{leg} and P_0 , is the primary driver of pressure generation, while the density of inlay yarn (D_{IY}) controls the stiffness of the fabric and the sensitivity of pressure to deformation. Together, these parameters ensure that the delivered compression adapts to the leg geometry while maintaining the intended gradient between zones—a behaviour consistent with previous studies on compression garment mechanics and structural design [21], [27], [31], [33], [39], [44], [45], [103], [111], [116].

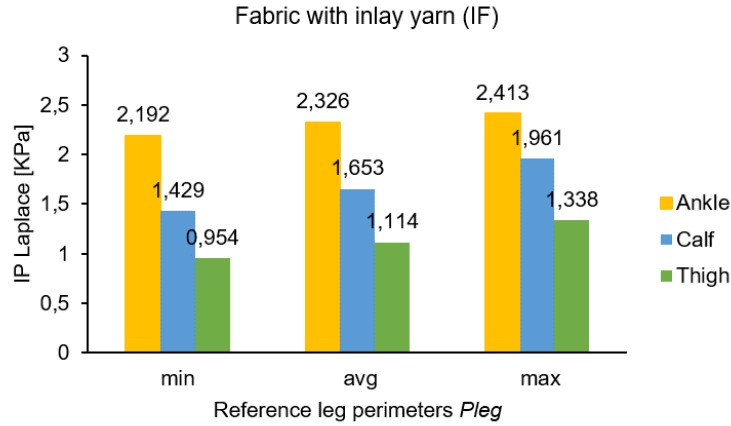


FIGURE 3.28: Calculated Laplace interface pressure ($IP_{Laplace}$) for reference leg perimeters of fabric with inlay yarn (IF configuration).

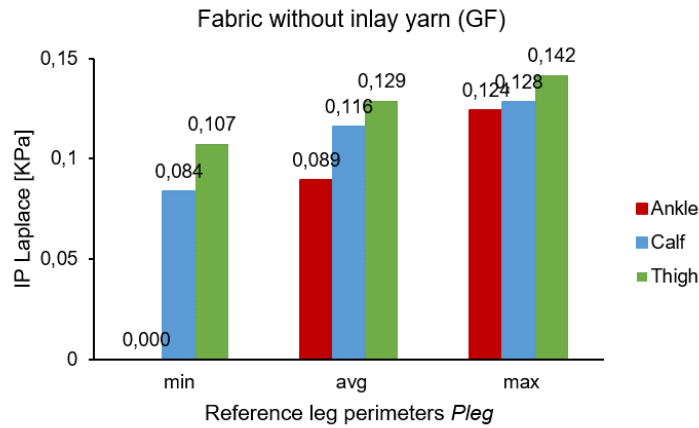


FIGURE 3.29: Calculated Laplace interface pressure ($IP_{Laplace}$) for reference leg perimeters of fabric without inlay yarn (GF configuration).

The predicted ankle pressure for the IF configuration, ranging from approximately 2.2 to 2.4 kPa (17–18 mmHg), is within the pressure interval of Class II compression products (15–20 mmHg) defined by the AFNOR and NF G30-102 standards [46]. This agreement confirms that the mechanical parameters identified through tensile testing provide a reliable prediction of the interface pressures generated by the knitted fabrics. It also demonstrates that the structural modulation of D_{IY} , combined with the geometric variation of leg perimeters (P_{leg}), governs the graduated compression profile characteristic of medical compression stockings.

The Laplace-based predictions were next compared with direct pressure measurements obtained using the PicoPress[®] sensor, as described in Section 3.4.2, and subsequently used for validating the finite element simulations presented in Section 5.2.3 of Chapter 5.

3.4.2 Measurements with pressure sensors

The interface pressure (IP) delivered by the medical compression stocking was experimentally measured using the PicoPress[®] pressure sensor (Microlab Elettronica, Italy) [117]. The device consists of a pneumatic membrane of 50 mm diameter and 0.2 mm thickness, connected to a microprocessor that converts the applied force into pressure with a resolution of ± 3 mmHg. This sensor is widely used for the assessment of medical compression systems and provides reliable measurements of fabric to skin pressure under static and quasi static conditions [70], [108], [109], [118]. Measurements were performed on the fabric samples with inlay yarn, denoted as IF, which represent the actual mechanical configuration of the final product.

To reproduce realistic in use conditions, three rigid leg models were designed, each corresponding to the maximal leg perimeter ($P_{\text{leg,max}}$) of the ankle, calf, and thigh zones. The models were idealised as cylindrical surfaces, an approach consistent with previous studies on compression devices [22], and a conical end was added to facilitate the placement and upward sliding of the stocking.

The geometrical dimensions of the cylindrical and conical sections are given in Table 3.7, and the three dimensional model is shown in Figure 3.30. The cylinder diameters ($D1$) were assigned from the corresponding maximal leg perimeters using Equation 3.10.

$$D1 \text{ [mm]} = \frac{P_{\text{leg,max}}}{\pi} \quad (3.10)$$

The height of the cylindrical section ($H1$) was defined to allow the fabric to be fully extended while maintaining the interline distance $d = 30$ mm, reproducing the longitudinal deformation and the density of inlay yarn, measured under in use conditions and reported in Section 3.3.1. For the conical part, the lower diameter ($D2$) was defined smaller than the relaxed fabric perimeter (P_0), to facilitate insertion and to avoid excessive stretching during mounting. The cone height ($H2$) was selected to ensure a gradual transition of the fabric from the relaxed region to the cylindrical measurement area, preventing local stress concentrations during placement. This height was set approximately 50 to 60 mm greater than the sample lengths, ensuring sufficient space for progressive extension before reaching the test region.

These geometrical definitions were subsequently used to define the leg models in the finite element simulations presented in Chapter 5.

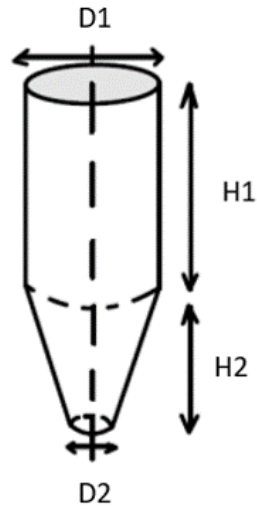


FIGURE 3.30: Three dimensional computer aided model of the leg geometry used for interface pressure measurements: cylinder height ($H1$) and diameter ($D1$), and cone height ($H2$) and diameter ($D2$)

TABLE 3.7: Dimensions of the three dimensional printed leg models for different compression zones: cylinder height ($H1$) and diameter ($D1$), and cone height ($H2$) and diameter ($D2$)

Compression zone	$H1$ [mm]	$D1$ [mm]	$H2$ [mm]	$D2$ [mm]
Ankle	300	76.4	150	40
Calf	350	130.5	180	50
Thigh	380	197.4	200	80

The leg models were produced by three dimensional printing in rigid polylactic acid, with dimensional tolerances below 0.2 mm, ensuring accurate reproduction of the target perimeters, and providing a smooth and homogeneous surface suitable for repeatable pressure measurements. The printed models are shown in Figure 3.31.



FIGURE 3.31: Three dimensional printed rigid leg models corresponding to the maximal leg perimeters ($P_{\text{leg,max}}$) of the three compression zones

During testing, the PicoPress[®] sensor was positioned at the central height of each cylindrical section. Each IF fabric sample was then mounted on its corresponding leg model, ankle, calf, or thigh, as shown in Figure 3.32. The knitted reference lines were aligned so that the interline distance remained equal to $d = 30$ mm, which reproduced the longitudinal deformation and the associated density of inlay yarn determined for the in use configuration and given in Table 3.2. This ensured full consistency between the pressure measurements, the structural characterisation, and the analytical predictions obtained from Laplace's law.

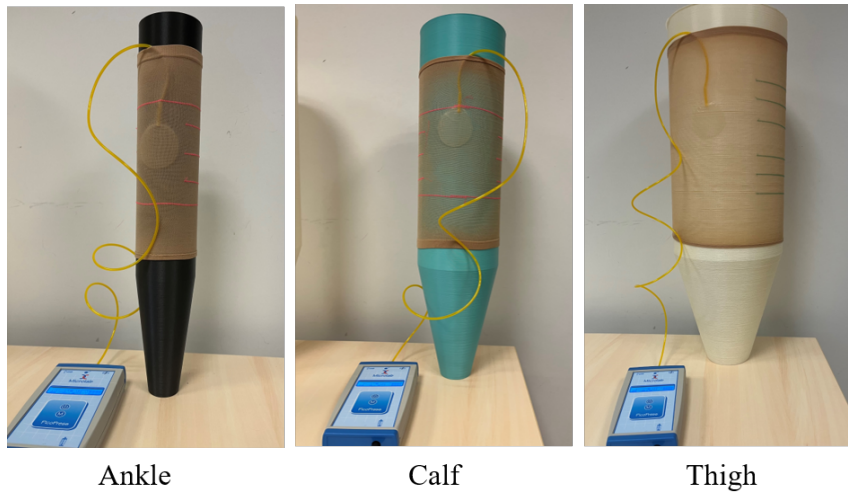


FIGURE 3.32: Placement of fabric samples with inlay yarn (IF configuration) on the rigid models with maximal leg perimeters ($P_{\text{leg,max}}$) for measurements with the PicoPress[®] sensor

For each compression zone, five measurements were taken after a stabilisation period of

thirty seconds. The average of these readings was taken as the experimental interface pressure, denoted as IP_{sensor} . All pressures were converted to kilopascals to allow direct comparison with the analytical predictions based on Laplace's law. These sensor based measurements were later used to validate the finite element simulations presented in Section 5.2.3 of Chapter 5.

3.4.2.1 Measured pressures for maximal leg perimeters

The interface pressures measured for the three compression zones are reported in Table 3.8. A clear decreasing trend is observed from ankle to thigh, consistent with the structural design of the product and with the variation in circumferential stiffness resulting from the distribution of inlay yarns, as described in Sections 3.3.1 and 3.3.3.4. The highest pressure was measured at the ankle, where the density of inlay yarn (D_{IY}) and the circumferential stiffness are greatest. These measurements confirm that, under the in-use configuration corresponding to the maximal leg perimeters, the stocking delivers the intended graduated compression profile.

TABLE 3.8: Measured interface pressure with pressure sensor (IP_{sensor}) for maximal leg perimeters ($P_{\text{leg,max}}$) of different compression zones

Compression zone	IP_{sensor} [kPa]
Ankle	2.45 ± 0.41
Calf	1.88 ± 0.32
Thigh	1.42 ± 0.28

The variability reflected in the standard deviations mainly arises from local differences in fabric–surface contact conditions during measurement. The obtained values fall within the ranges typically reported for PicoPress[®] assessments of medical compression garments [70], [108], [109], [118].

3.4.2.2 Comparison with Laplace's predictions

To quantify the agreement between analytical predictions and experimental measurements, the relative deviation between the sensor-based pressure (IP_{sensor}) and the pressure predicted by Laplace's law (IP_{Laplace}) was calculated according to Equation 3.11. The comparison was performed for the maximal leg perimeter configuration ($P_{\text{leg,max}}$), which represents the highest deformation state applied in both the tensile tests and the sensor measurements.

$$\text{Deviation [\%]} = \frac{|IP_{\text{sensor}} - IP_{\text{Laplace}}|}{IP_{\text{Laplace}}} \times 100 \quad (3.11)$$

The analytical and experimental values showed excellent agreement across all compression zones. The relative deviations were 6.5% for the ankle, 5.6% for the calf, and 6.0% for the thigh. The pressures measured with PicoPress[®] were slightly higher than the Laplace predictions at the ankle and thigh, and slightly lower at the calf, but all deviations remained well within the sensor's measurement uncertainty. These results confirm that Laplace's formulation provides an accurate prediction of the interface pressure for inlaid weft-knitted fabrics when cylindrical geometry and controlled deformation are ensured.

The present findings are consistent with previous work demonstrating strong agreement between analytical and experimental methods when material parameters are experimentally characterised and when testing is performed on rigid cylindrical supports. Partsch [109] and Mosti et al. [108] reported high reproducibility of PicoPress[®] measurements for medical compression applications. Teyeme et al. [96] observed deviations below 10% between analytical predictions and sensor measurements on cylindrical leg models, while Zhang et al. [44] and Lozo et al. [39] confirmed that Laplace-based calculations accurately capture pressure trends when the circumferential mechanical behaviour of the fabric is well characterised. More recently, Shi et al. [45] emphasised the importance of combining experimental and analytical approaches to fully assess the biomechanical performance of compression garments.

In this study, deviations remained below 6% for all zones, demonstrating that the combination of tensile characterisation and Laplace-based modelling provides a reliable and consistent estimation of the interface pressure. The measured pressures were subsequently used to validate the local pressure distributions obtained from the finite element simulations presented in Chapter 5, ensuring continuity between analytical, experimental, and numerical descriptions of the mechanical response of medical compression stockings.

3.5 Conclusions and research perspectives

Conclusions This chapter presented a comprehensive experimental characterisation of a commercial medical compression stocking, covering its knitted architecture, constituent yarns, dimensional parameters, and mechanical response under in-use deformation.

The analysed product, a women's Class II graduated compression stocking (Size M) manufactured by Sigvaris France, was studied in its three reference compression zones (ankle, calf and thigh). The knitted structure features an inlaid weft-knitted architecture composed of two polyamide-covered elastane yarns: the ground yarn, forming the intermeshed loop

structure and providing extensibility and comfort, and the inlay yarn, inserted in each course as circumferential reinforcement.

A designed gradient of density of inlay yarn (D_{IY}) was identified as the primary structural parameter governing the circumferential stiffness of the fabric. Structural characterisation confirmed its decrease from 1.13 mm^{-1} at the ankle to 0.80 mm^{-1} at the thigh, establishing the mechanical basis for the graduated compression profile.

The deformation imposed during wear was determined from the manufacturer's size chart and French standards, defining the in-use circumferential strain ranges: approximately 0.49–0.63 for the ankle, 0.67–0.95 for the calf, and 0.96–1.15 for the thigh. These values were used to define the experimental tensile testing range and to ensure that the mechanical characterisation reproduced realistic deformation conditions.

Uniaxial tensile tests on single inlay yarns extracted from finished stockings characterised their intrinsic mechanical response. The yarns exhibited a nonlinear force–strain behaviour with strong stiffening at high deformation, typical of covered elastane filaments, confirming their central role in providing resistance to stretch when embedded in the knitted architecture. These results provided the constitutive inputs required to model the inlay system in subsequent chapters.

The mechanical behaviour of the knitted fabrics was then evaluated through tensile tests on cylindrical samples with (IF) and without (GF) inlay yarn. The inclusion of the inlay reinforcement substantially increased circumferential stiffness and introduced a nonlinear stiffening response. The IF samples displayed a clear differentiation between compression zones: the ankle, with the highest D_{IY} , exhibited the steepest slope in the tension–strain curves, while the calf and thigh, with lower inlay densities, showed more compliant behaviour. Conversely, the GF samples exhibited a nearly linear and nearly identical response across zones, with very low stiffness values.

This absence of variation among GF samples confirms that the loop structure alone cannot generate the stiffness gradient required for graduated compression. The comparison between GF and IF configurations demonstrated that the inlay yarn not only increases the overall stiffness of the fabric but also dictates the local mechanical differentiation between zones.

Interface pressure was then estimated using the tensile response of IF fabrics combined with Laplace's law. The predicted pressures ranged from approximately 2.4 kPa at the ankle to 1.4 kPa at the thigh for the maximal leg perimeters, consistent with the therapeutic range of Class II compression. These predictions were compared with direct measurements obtained using PicoPress® sensors placed on rigid leg models. The excellent agreement

between analytical and experimental values, with deviations below 6%, confirmed the reliability of the combined tensile–Laplace methodology for predicting interface pressure under controlled cylindrical conditions.

Overall, this chapter established the first mechanical dataset for the studied inlaid weft-knitted fabric, quantifying the influence of D_{IY} on circumferential stiffness and pressure generation. The structural parameters, material properties, and mechanical results obtained here constitute the experimental foundation for the hybrid mechanical model developed in Chapter 4 and for the finite element simulations of interface pressure prediction presented in Chapter 5.

Research perspectives Future work should extend the mechanical database established in this study by incorporating additional compression products and configurations, including other compression classes (Class I and Class III), products designed for different morphologies, and fabrics manufactured with alternative yarn systems or knitting parameters. Such an extension will enable a more comprehensive assessment of how density of inlay yarn (D_{IY}), yarn composition, loop geometry and structural design influence stiffness, deformability, and pressure generation.

Further investigations should also analyse the effect of yarn material properties and covering techniques, as these parameters strongly affect the stiffness of both the loop and inlay systems. Additionally, the influence of user morphology on circumferential deformation and pressure distribution should be examined, since local leg geometry directly affects the mechanical response of the garment.

Integrating these parameters into the hybrid modelling framework developed in the next chapters will allow the formulation of generalised relationships linking fabric structure, mechanical response, and pressure generation. Such methodological development can support the classification and comparison of compression products and guide the design of new knitted architectures with optimised performance.

Ultimately, this progress will support the design of next-generation compression garments featuring improved comfort, controlled pressure distribution, and enhanced therapeutic efficacy, enabling patient-specific adaptability and more efficient clinical performance.



Chapter 4

Mechanical hybrid model of inlaid weft-knitted fabrics

4.1 Introduction

Medical compression stockings are knitted fabrics designed to apply controlled pressure on the lower limb. As demonstrated experimentally in Chapter 3, their mechanical response results from the combined effect of two distinct yarn systems organised within an inlaid weft-knitted structure. The ground yarn forms the intermeshed loop structure, which governs extensibility in the longitudinal direction, while the inlay yarn is inserted in every course and provides reinforcement along the circumferential direction. This specific arrangement gives the fabric its characteristic anisotropy, with high compliance axially and increased stiffness circumferentially due to the presence of the inlay yarns.

Accurate prediction of the mechanical response of medical compression stockings is crucial for ensuring therapeutic efficiency and guiding product design. Several modelling strategies have been proposed in the literature. Homogenised membrane models, generally isotropic, orthotropic or hyperelastic, have been widely used to simulate the global behaviour of knitted fabrics [73]–[75], [79], [80], [119]. Their material parameters are typically identified from tensile tests performed on the complete fabric, a methodology also used in Chapter 3 to characterise the stocking under in-use deformation. Although efficient for finite element simulations of interface pressure, these homogenised approaches cannot distinguish between the mechanical contributions of the two fabric constituents and cannot represent the effect of structural variations such as the density of inlay yarn (D_{IY}), identified in Chapter 3 as the key parameter controlling circumferential stiffness between compression zones.

More detailed approaches, such as unit-cell or geometrical models [42], [44], [87], explicitly describe the yarn paths and their interaction. While these models capture the local architecture with greater fidelity, they remain computationally demanding, which limits their implementation in full-scale finite element frameworks for pressure prediction.

Consequently, current finite element models used to simulate compression stockings still rely on homogenised membrane descriptions. These simplified formulations do not capture the local mechanical effects of the ground and inlay yarns, nor the structural differentiation introduced by the varying density of inlay yarn that was experimentally demonstrated in Chapter 3.

The objective of this chapter is therefore to develop a hybrid model that predicts the circumferential tensile response of inlaid weft-knitted fabrics by explicitly accounting for the contributions of the loop structure, the inlay yarn, and the local knitted architecture associated with each compression zone. The model is based on a representative unit cell in which the ground yarn is modelled using continuous elements, and the inlay yarn is represented by discrete elements oriented along its insertion trajectory. The constitutive laws governing each constituent are defined and calibrated using the experimental characterisation of yarns and fabrics presented in Chapter 3. The density of inlay yarn (D_{IY}) is incorporated as a structural input controlling the circumferential stiffness, ensuring that the mechanical distinctions observed experimentally between zones are reproduced by the model.

The hybrid model is then used to simulate the tensile response of the knitted fabric under the same loading conditions as the experimental tensile tests described in Chapter 3, where deformation was applied uniaxially along the course direction (E_2). The predictions are compared with the experimental responses for both configurations, with and without inlay yarn, to assess the accuracy of the model and its ability to reproduce the structural effects associated with the variation in D_{IY} .

The chapter first introduces the geometrical configuration of the knitted unit cell and its relation to the global reference system of the fabric. It then presents the equations allowing the prediction of the tensile response in Section 4.2.3, including the projection of deformations and forces. The constitutive behaviour of the continuous and discrete elements is described next, in Section 4.2.2, followed by the identification of their mechanical parameters in Section 4.2.4 based on the experimental results of Chapter 3. The predicted tensile responses for the three compression zones are then detailed and discussed in Sections 4.3.2.1 and 4.3.2.2. Lastly, the hybrid model is validated through comparison with the experimental tensile curves of both fabric configurations in Section 4.3.3.

The results obtained in this chapter define the structural and mechanical parameters of

the hybrid model for predicting the tensile response of the medical compression stocking. This is latter transferred into a finite element model, as described in Chapter 5, to simulate the tensile response of the fabric when placed over a leg. This enables the prediction of the interface pressure generated by the stocking, for each compression zone, using the local parameters of the fabric.

All notations and symbols used throughout this chapter are listed and defined in the glossary at the end of the manuscript.

4.2 Methodology

The hybrid mechanical model was developed to reproduce the tensile test described in Chapter 3 and to predict the circumferential tensile response of the inlaid weft-knitted fabrics used in medical compression stockings. The modelling strategy combines the geometric description of the knitted architecture with the experimentally identified mechanical behaviour of its two constituent systems. The loop structure and the inlay yarns are represented through separate mechanical elements, enabling the prediction of the macroscopic tensile response from their local behaviour. A unit-cell formulation links the structural scale of the knitted architecture to the global mechanical response, through a combination of continuous elastic elements modelling the loop structure and discrete elements modelling the inlay yarns.

This approach follows the principle of hybrid finite element formulations previously proposed for woven and composite textiles, in which the matrix is represented by continuous elements and the reinforcement yarns by discrete or beam elements **tephanyInfluencenon-linearityFabric2016**, [83]–[85]. This concept, successfully validated for woven composites, was later extended to knitted fabrics by Ghorbani et al. [87], demonstrating that hybrid configurations can reproduce the anisotropic and non-linear response of inlaid weft-knitted structures.

The structural parameters characterised in Chapter 3, including the density of inlay yarn (D_{IY}), the initial and maximal perimeters (P_0 and $P_{\text{leg,max}}$), and the sample length (L), are directly introduced into the hybrid formulation, ensuring that the model reflects the real configuration of each compression zone.

The following sections describe the development of the hybrid model, the definition of the hybrid unit cell, the constitutive behaviour assigned to the continuous and discrete elements, the analytical equations used to predict the circumferential tensile response, the identification of the model parameters from the experimental results of Chapter 3, and the

validation of the hybrid formulation against the tensile tests performed on the knitted fabrics.

4.2.1 Unit cell configuration

At the local structural scale, the inlaid weft-knitted fabric consists of two constituents: the loop structure, formed by the ground yarns, and the inlay yarns, inserted in every course of the loop structure. The hybrid model represents this configuration through a unit cell, corresponding to a small repeating segment of the fabric. The ground yarns are not modelled individually. Instead, the looped network is represented as a continuous elastic element. The inlay yarns, inserted in every course, are modelled with discrete elements, aligned with the course direction, linking two opposite edges of the elastic element.

The unit cell, shown in Figure 4.1, combines a continuous and discrete element, to model the local mechanical response of the fabric, as a superposition of the response of its constituents.

The unit cell dimensions are defined according to the local knitted structure. The unit cell height (h) is considered as the spacing between courses of the inlay yarn, along e_1 . The spacing is given by the inverse of density of inlay yarns (D_{IY}), reported in Table 3.2 for the three compression zones. The unit cell height is therefore calculated as given in Equation 4.1. Since the discrete element links the two bottom edges of the continuous elastic element, the unit cell width is equal to the spring length (l_{spring}).

$$h [\text{mm}] = \frac{1}{D_{IY}} \quad (4.1)$$

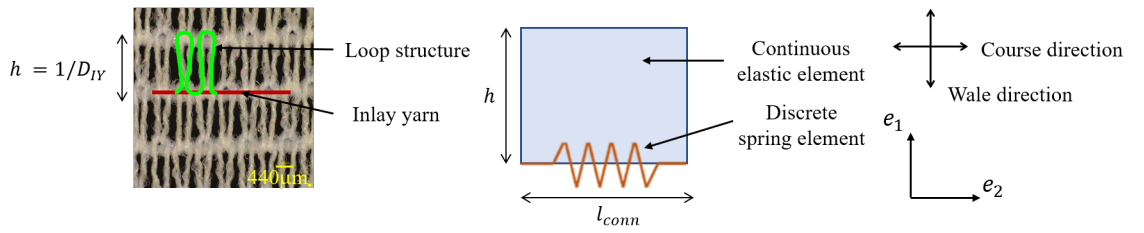


FIGURE 4.1: Representation of the hybrid unit cell modelling the local knitted structure of the medical compression stocking

This hybrid configuration enables the anisotropic mechanical behaviour of the fabric to be described through the combined action of continuous and discrete elements. Variations in local knitted architecture are introduced by adjusting the unit cell dimensions. Regions with higher inlay yarn density, such as the ankle zone, are therefore represented by smaller

unit cells, resulting in a higher number of springs per unit length and an increased circumferential stiffness.

The geometrical parameters of the model, specific for modelling the different compression zones; are described in Section 4.2.1. The constitutive laws and mechanical parameters of the continuous and discrete elements, identified from the experimental characterisation presented in Chapter 3, are described below in Sections 4.2.2 and 4.2.4.

Coordinate reference systems

During the tensile tests described in Chapter 3, the cylindrical fabric samples were elongated along the circumferential direction to reproduce the deformation imposed during wear. Although the samples are tubular, the imposed deformation develops tangentially to the cylindrical surface and can therefore be described under plane stress conditions. For this reason, the fabric is represented as a flat surface in the model, and Cartesian coordinate systems are adopted instead of cylindrical ones. This assumption allows the tensile deformation to be described using planar boundary conditions, as illustrated in Figures 3.15 and 3.17.

In the cylindrical geometry of the stocking, the inlay yarns do not follow a purely circumferential path. Due to the knitting process, they form a helical trajectory along the fabric length, resulting in a slight inclination with respect to the circumferential direction. This inclination varies between compression zones and defines the orientation of the course direction within the fabric.

To account for this orientation, two Cartesian coordinate systems are introduced in the model, as shown in Figures 4.2 and 4.4: a global reference system and a local reference system.

The global reference system (E_1, E_2, E_3) is defined at the macroscopic fabric scale. Direction E_1 corresponds to the longitudinal direction of the stocking, E_2 to the circumferential direction, and E_3 to the through-thickness direction.

The local reference system (e_1, e_2, e_3) is attached to the unit cell and follows the local knitted architecture. Direction e_1 corresponds to the wale direction, e_2 to the course direction along which the inlay yarn is inserted, and e_3 is normal to the fabric surface.

The orientation of the local reference system with respect to the global one is defined by an inclination angle (α), corresponding to the angle between the course direction and the circumferential direction. This angle describes the rotation between the two reference systems and allows the transformation of geometrical and mechanical quantities from the local unit cell description to the global fabric description.

The inclination angle (α) is determined from the knitting geometry and is expressed as a function of the density of inlay yarns (D_{IY}) and the fabric perimeter (P). Considering that two inlay yarns are inserted during knitting ($n = 2$), with each individual yarn appearing every two courses, the angle (α) is given by Equation 4.2.

$$\tan \alpha \text{ [rad]} = \frac{n}{D_{IY} \cdot P} = \frac{2}{D_{IY} \cdot P} \quad (4.2)$$

The initial inclination angle (α_0) is obtained by replacing the fabric perimeter (P) with the initial perimeter (P_0) in Equation 4.2, leading to Equation 4.3.

$$\tan \alpha_0 \text{ [rad]} = \frac{2}{D_{IY} \cdot P_0} \quad (4.3)$$

Geometrical parameters of the model

The geometrical parameters of the model are given in the two Cartesian reference systems.

At the fabric scale, the geometry of the compression stocking is described in the global reference system (E_1, E_2, E_3). The fabric length (L) is defined along the longitudinal direction (E_1), while the fabric width (W) is defined along the circumferential direction (E_2), as illustrated in Figure 3.15 and Figure 4.2. The fabric perimeter (P) is given by its width (W), with Equation 3.5.

In the initial flat configuration, the fabric perimeter (P_0) is given by the initial width (W_0), by Equation 3.6. In the deformed configuration, the fabric is stretched circumferentially to match the maximal leg perimeter ($P = P_{\text{leg,max}}$). The fabric length (L) is kept constant, considering the in-use longitudinal elongation. The density of inlay yarn (D_{IY}) is defined from the in-use configuration and remains constant within each compression zone.

The global geometrical parameters (L , P_0 , $P_{\text{leg,max}}$ and D_{IY}) were identified experimentally for each compression zone and are reported in Tables 3.2, 3.1 and 3.3 of Chapter 3.

At the unit cell scale, the geometry is defined from the local knitted architecture in the local reference system (e_1, e_2, e_3), as presented in Section 4.2.1. The unit cell height (h) corresponds to the spacing between two successive inlay yarns along the wale direction (e_1) and is directly defined from the inlay yarn density with Equation 4.1. Since D_{IY} varies between compression zones, the unit cell height varies accordingly.

Considering this, the total number of unit cells distributed along the fabric length (N_{length}) varies between compression zones. The number of unit cells along the fabric length is given by Equation 4.4.

$$N_{\text{length}} = \frac{L}{h \cdot \cos \alpha} \quad (4.4)$$

This parameter reflects the density of unit cell repetitions along the fabric length and governs the number of discrete elements contributing to the circumferential stiffness in each compression zone.

The unit cell width is equal to the spring length (l_{spring}), aligned with the course direction (e_2).

Because the local knitted architecture is inclined with respect to the global fabric directions, the unit cell defined by (h, l_{spring}) in the local reference system appears with inclined dimensions in the global reference system. These projected dimensions are obtained through the rotation defined by the inclination angle (α) given in Equation 4.3. This transformation provides the geometric link between the local unit cell description and the global fabric representation used in the finite element model. The correspondence between local and global unit cell dimensions is illustrated in Figure 4.2.

In the initial configuration, a square reference cell is considered in the global, such that the unit cell height is equal to the width. The corresponding local dimensions are obtained using the initial inclination angle (α_0) given in Equation 4.3.

The fabric thickness t was measured experimentally in the relaxed state and averaged 0.75 mm for the different compression zones. In this study, stresses are expressed per unit width and normalised for $t = 1$ mm. Since thickness variations are negligible in service, t was considered constant in the model.

The geometrical parameters defined at the fabric and unit cell scales for each compression zone are summarised in Table 4.1.

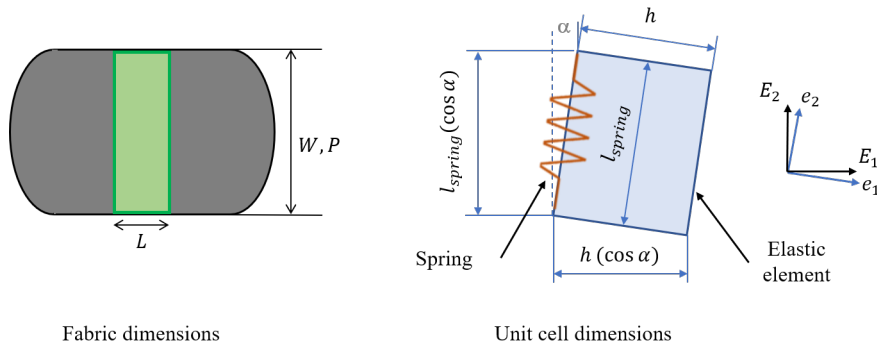


FIGURE 4.2: Dimensional parameters of the hybrid model in the global and local reference systems.

TABLE 4.1: Geometrical parameters of the hybrid model for the different compression zones.

Parameter	Symbol	Ankle	Calf	Thigh
<i>Fabric scale</i>				
Fabric length along E_1 [mm]	L	30	30	30
Initial fabric perimeter along E_2 [mm]	P_0	148	196	288
Maximal leg perimeter along E_2 [mm]	$P_{\text{leg,max}}$	240	410	620
Thickness along E_3 [mm]	t	1	1	1
Density of inlay yarn [mm^{-1}]	D_{IY}	1.13	0.87	0.80
<i>Unit cell scale</i>				
Unit cell height along e_1 [mm]	h	0.885	1.149	1.250
Initial spring length along e_2 [mm]	$l_{\text{spring},0} = h$	0.885	1.149	1.250
Initial inclination angle [rad]	α_0	0.012	0.0114	0.0083
Number of elements per fabric length	N_{length}	34	26	24

4.2.2 Constitutive laws of the model

This section presents the mechanical behaviour assigned to the two constituents of the hybrid model. The loop structure created by the ground yarns and the inlay yarn reinforcement are represented separately using continuous elements and discrete elements. For each component, the constitutive behaviour is defined from the experimental characterisation presented in Chapter 3 and formulated predict the fabric response under circumferential deformation imposed during the tensile testing.

Loop structure: continuous elements The loop structure formed by the ground yarns is not modelled at the yarn scale but represented as a continuous medium. The fabric is described using continuous elements with isotropic membrane behaviour. Since the fabric thickness is small compared with its in-plane dimensions, a plane-stress membrane assumption is adopted.

Only circumferential loading along the global direction (E_2) is considered. This choice reduces model complexity while remaining consistent with the uniaxial tensile tests performed on tubular knitted samples in Chapter 3.

The tensile response of weft-knitted fabrics without inlay yarns (GF configuration), presented in Figures 3.18–3.20 of Chapter 3, showed pronounced hysteresis and non-elastic effects associated with loop rearrangement during the first loading cycle. To simplify the

model and focus on the contribution of the loop structure to circumferential stiffness, the continuous behaviour is approximated by an equivalent linear elastic response, identified from the second loading cycle.

The continuous elastic element constitutive behaviour is defined by Hooke's law under plane-stress conditions and expressed in terms of logarithmic strains (LE_{ij}), as given in Equation 4.5. For the range of deformations considered, logarithmic strains are close to the corresponding engineering strains.

$$\begin{bmatrix} \sigma_{11} \\ \sigma_{22} \\ \tau_{12} \end{bmatrix} = \frac{E}{(1+\nu)(1-2\nu)} \begin{bmatrix} 1-\nu & \nu & 0 \\ \nu & 1-\nu & 0 \\ 0 & 0 & \frac{1-2\nu}{2} \end{bmatrix} \begin{bmatrix} LE_{11} \\ LE_{22} \\ LE_{12} \end{bmatrix} \quad (4.5)$$

Under the applied boundary conditions, the deformation is uniaxial along the circumferential direction (E_2), the longitudinal strain is fixed ($LE_{11} = 0$), and no in-plane shear is imposed ($LE_{12} = 0$). The stress-strain relations reduce to Equations 4.6 and 4.7.

$$\sigma_{11} = \frac{\nu E}{(1+\nu)(1-2\nu)} LE_{22} \quad (4.6)$$

$$\sigma_{22} = \frac{(1-\nu) E}{(1+\nu)(1-2\nu)} LE_{22} \quad (4.7)$$

The logarithmic strain along the global circumferential direction (LE_{22}), is obtained from the relative change in unit-cell width. Considering the periodic representation of the fabric through unit cells, the normalised unit cell stretch ratio along E_2 is equal to the global stretch ratio. By accounting for the inclination of the knitted structure, the circumferential logarithmic strain is expressed with Equation 4.8.

$$LE_{22} = \ln\left(\frac{l_{\text{spring}}}{l_{\text{spring},0}}\right) = \ln\left(\frac{P \cos(\alpha)}{P_0 \cos(\alpha_0)}\right) \quad (4.8)$$

The identification of the elastic parameters of the continuous elements (E and ν) is presented below in Section 4.2.4.

Inlay yarn: discrete spring elements Unlike the ground yarns forming the loop structure, the inlay yarns are inserted into the knitted network without forming loops. As a result, their deformation under circumferential loading is not governed by geometric rearrangement but by axial stretching. It is therefore assumed that, under tensile test conditions, the circumferential deformation of the fabric is directly transmitted to the inlay yarn. In this sense, the

deformation experienced by a single inlay yarn is considered equal to the circumferential deformation imposed on the fabric.

In the hybrid model, each inlay yarn is represented by a discrete spring element aligned with the local course direction e_2 . Each spring links two opposite edges of the unit cell and carries axial load only. The initial spring length ($l_{\text{spring},0}$) corresponds to the inclined unit-cell width.

The axial deformation of a spring (Δl_{spring}) is defined by the change in its length, and the corresponding axial strain is expressed with Equation 4.9

$$\varepsilon_{\text{spring}} [\text{mm/mm}] = \frac{\Delta l_{\text{spring}}}{l_{\text{spring},0}} \quad (4.9)$$

By accounting for the unit-cell inclination and periodic deformation of the fabric, the spring strain can be expressed as a function of the global fabric deformation with Equation 4.10

$$\varepsilon_{\text{spring}} [\text{mm/mm}] = \left(\frac{P \cdot \cos \alpha}{P_0 \cdot \cos \alpha_0} - 1 \right) \quad (4.10)$$

The axial force carried by the discrete element (f^{spring}) is written in the general form of a linear spring law with Equation 4.11, where k is the stiffness.

$$f^{\text{spring}} = k \cdot \Delta l_{\text{spring}} \quad (4.11)$$

For consistency with the tensile characterisation of the inlay yarn, this relation is not defined using a constant stiffness value. Instead, the force–displacement law is normalised and expressed in terms of strain, assuming that the spring strain ($\varepsilon_{\text{spring}}$) is equal to the tensile strain of a single inlay yarn ($\varepsilon_{\text{yarn}}$). For each imposed circumferential deformation of the fabric, the corresponding spring strain is computed, and the axial force is obtained directly from the experimental Force–Strain response of the inlay yarn.

In practice, the non-linear relation between axial force and strain is defined by interpolation of the experimental tensile curve of the inlay yarn, using the stabilised loading cycle reported in Figure 4.3 and in Chapter 3. This approach ensures that the discrete elements model the tensile behaviour of a single inlay yarn under the same deformation conditions as those imposed during the fabric tensile tests. The total contribution of the inlay-yarn system to the circumferential tension of the fabric is obtained by adding the axial forces of all discrete elements along the fabric length.

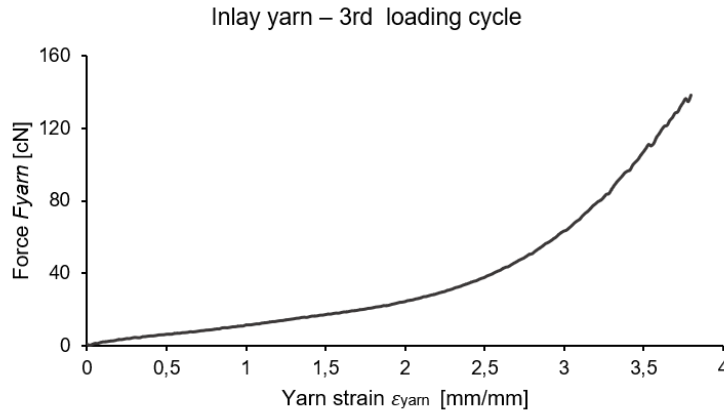


FIGURE 4.3: Force–Strain response of the inlay yarn during the third loading cycle in tensile test, used for defining the tensile response of the spring in the hybrid model.

4.2.3 Tensile response prediction

The hybrid model predicts the circumferential tensile response of the fabric by combining the mechanical contributions of the continuous and discrete elements within the unit cell.

As in the experimental tensile test described in Chapter 3, the deformation is imposed along the circumferential direction (u_2). Longitudinal displacements are constrained ($u_1 = 0$ and $L = \text{const}$), and the thickness remains constant ($u_3 = 0$). When this displacement is applied globally along E_2 , the inclined unit cell undergoes coupled displacements along both local axes (e_1, e_2), because the inlay yarn, and therefore the cell, are rotated by the angle α . This projection, illustrated in Figure 4.4, is necessary to compute the local strains acting on the continuous and discrete elements.

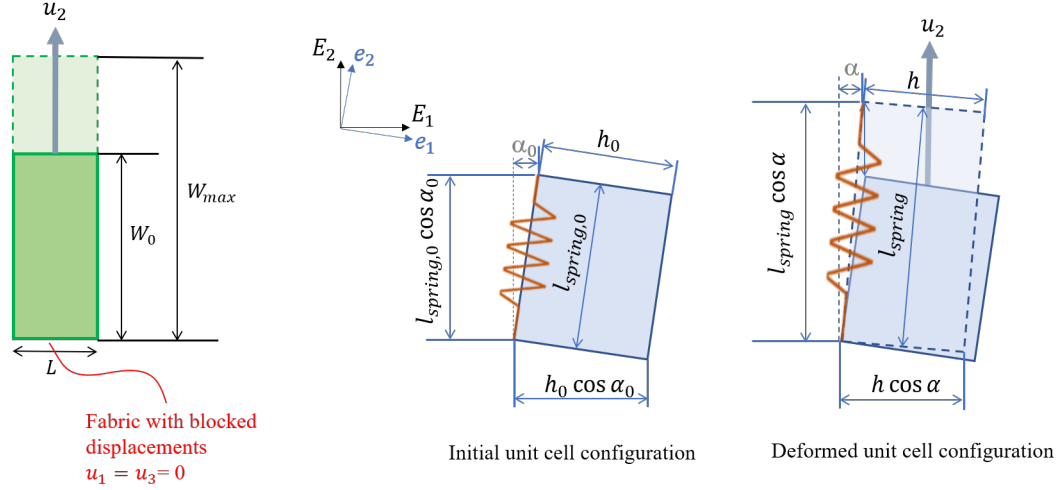


FIGURE 4.4: Displacement imposed along the circumferential direction (E_2) during tensile testing, and resulting deformation of the unit cell in the global and local reference systems.

Total circumferential force and fabric tension The total circumferential force (F_2) generated during the imposed deformation is obtained by adding the forces carried by all unit cells arranged along the longitudinal direction (E_1). Being F_2^{cell} the contribution of a single unit cell and N_{length} the number of unit cells along the length of the fabric (E_1), the circumferential force (F_2) is calculated with Equation 4.12.

$$F_2 [\text{N}] = F_2^{cell} N_{length} \quad (4.12)$$

The fabric tension per unit length (T), consistent with the tensile test definition in Chapter 3, is obtained as the force per unit length $T = F_2/L$.

Projection of local forces into the global reference system Within the unit cell, the local forces (f_1^{cell}, f_2^{cell}) along (e_1, e_2) result from the contributions of the continuous and discrete elements modelling the response of the loop structure and the inlay yarn. These forces are projected into the global axes (E_1, E_2) using the rotation matrix defined by the inclination angle (α), as given in Equation 4.13.

$$\begin{bmatrix} F_1^{cell} \\ F_2^{cell} \end{bmatrix}_{(E_1, E_2)} = \begin{bmatrix} \cos \alpha & -\sin \alpha \\ \sin \alpha & \cos \alpha \end{bmatrix} \begin{bmatrix} f_1^{cell} \\ f_2^{cell} \end{bmatrix}_{(e_1, e_2)}. \quad (4.13)$$

The resulting force of a single unit cell along the circumferential direction (F_2), which drives the tensile response, is given in Equation 4.14, where f_1^{cell} is the cell force in the local wale direction (e_1) and f_2^{cell} is the cell force in the local course direction (e_2).

$$F_2^{cell} = \sin \alpha f_1^{cell} + \cos \alpha f_2^{cell} \quad (4.14)$$

Local contributions of continuous and discrete element The local forces originate from the two distinct mechanical contributions of the elastic and spring element. The continuous element carries forces in both local directions (e_1 and e_2). The spring, aligned with e_2 , contributes only in this direction. Therefore, the local forces along the wale direction (f_1^{cell}) correspond to the force of the continuous elastic element in this direction (f_1^{cont}), and the local forces along the course direction (f_2^{cell}) correspond to the addition of the continuous element force (f_2^{cont}) and spring force (f^{spring}), as expressed in Equation 4.15.

$$f_1^{cell} = f_1^{cont} ; \quad f_2^{cell} = f_2^{cont} + f^{spring} \quad (4.15)$$

Membrane forces of the continuous elastic element The membrane forces in the continuous elastic element are obtained from the stresses (σ_{11}, σ_{22}) and the element cross-sectional areas (A_1, A_2), using Equation 4.16.

$$f_1^{cont} = \sigma_{11} A_1 ; \quad f_2^{cont} = \sigma_{22} A_2 \quad (4.16)$$

The areas A_1 and A_2 are considered for the initial unit cell dimensions ($h, l_{spring, 0}$) and cell thickness (t), as expressed in Equation 4.17.

$$A_1 = l_{spring, 0} t ; \quad A_2 = h t \quad (4.17)$$

The elastic stresses (σ_{11} and σ_{22}) are given by Equations 4.6 4.7, following the isotropic plane-stress law defined in Section 4.2.2, with the logarithmic circumferential strain LE_{22} obtained from Equation 4.8.

Axial forces of the spring discrete element The spring discrete element represents the inlay yarn and carries only axial load. Its strain is defined by the change in inclined width, according to Equation 4.9.

The spring force follows the non-linear spring law defined in Equation 4.11. Replacing the spring strain from Equation 4.10, the spring force is expressed as a function of the fabric perimeter variation, giving Equation 4.18.

$$f^{spring} = k \left(\frac{P \cos \alpha_0}{P_0 \cos \alpha} - 1 \right) \quad (4.18)$$

Resulting circumferential force and fabric tension Substituting the local forces from Equations 4.16 and 4.18 into Equation 4.15, to obtain the projection in the global reference system with Equation 4.14, and finally into the global summation with Equation 4.12, provides the total circumferential force (F_2) and the tension (T) for any imposed deformation.

The analytical hybrid model was then used to predict the tensile response of the knitted fabrics in the three compression zones. For each zone, the dimensional and structural parameters were taken from the experimental characterisation of Chapter 3, and listed in Table 4.1. These zone-specific parameters were introduced directly into the hybrid formulation, while the mechanical parameters of the continuous and discrete elements were kept constant across all zones, as they depend solely on the material properties of the ground and inlay yarns. This ensures that the variations in tensile stiffness between ankle, calf, and thigh arise strictly from the structural and geometrical differences of each zone, in agreement with the fabric architecture.

For the fabric configuration without inlay yarn (GF), only the elastic contribution modelling the loop structure was considered, following the same assumptions used for parameter identification. For the inlaid configuration (IF), both elastic and spring contributions were included, reproducing the combined behaviour of the loop structure and the inlay reinforcement.

Using the imposed circumferential deformation between P_0 and $P_{\text{leg,max}}$, the hybrid formulation was applied to compute the local strains and stresses at the unit-cell scale according to the relations established in Section 4.2.3. The circumferential force per unit length was then derived from the sum of the continuous and discrete components projected into the global reference system, giving the analytical Tension–Circumferential strain curves over the same deformation range as the experimental tests.

Figure 4.5 presents the described method for predicting the tensile response of the weft-knitted fabrics of medical compression stockings, based on the developed hybrid model. The identification of the mechanical parameters of the continuous elastic element, required to perform these predictions, is detailed in the following Section 4.2.4.

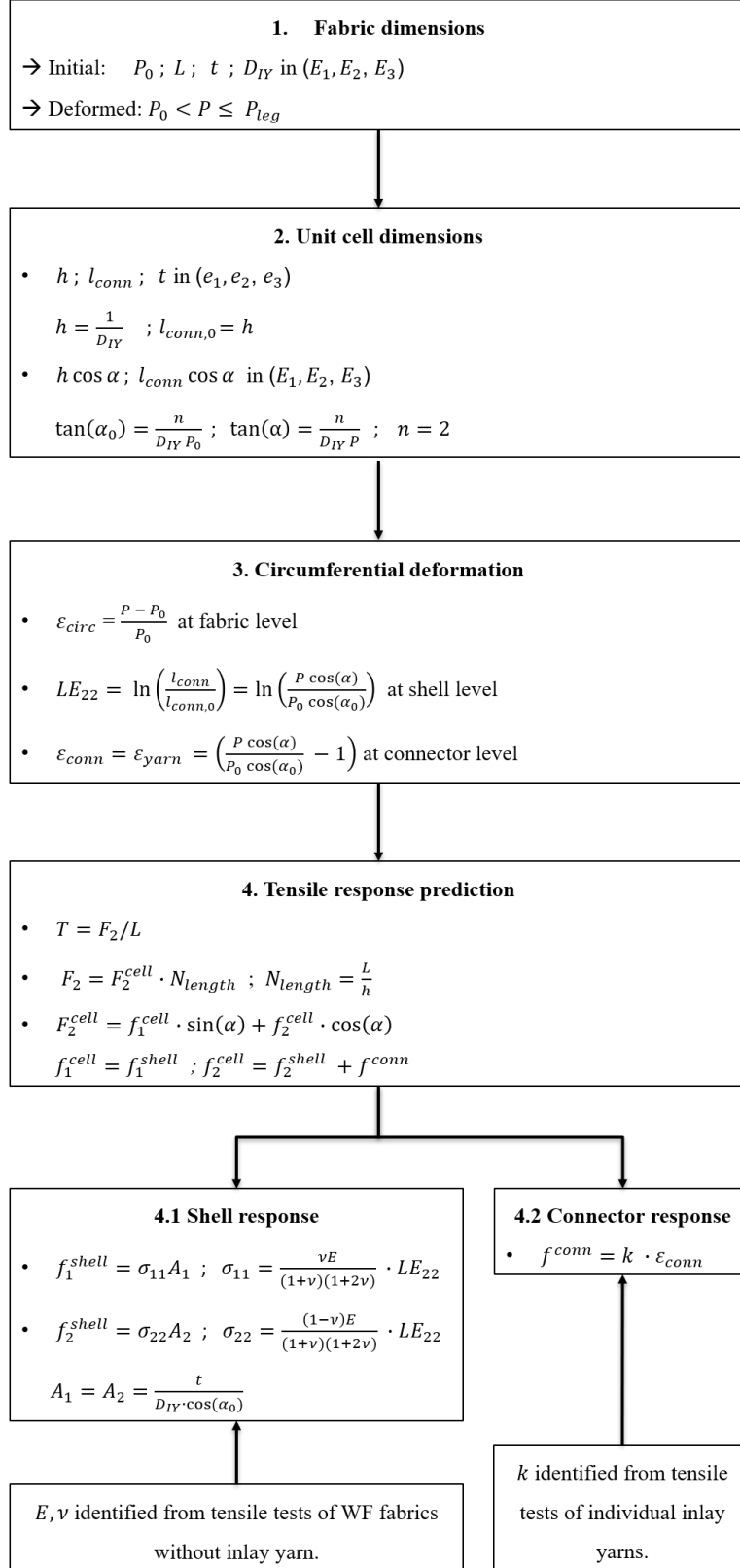


FIGURE 4.5: Flowchart of the hybrid model: from fabric and unit cell dimensions to circumferential deformation and tensile response prediction, including elastic and spring contributions.

4.2.4 Parameter identification: E and ν

The loop structure was modelled using three-dimensional continuous elements with isotropic elastic behaviour. The corresponding mechanical parameters, the Young's modulus (E) and the Poisson's ratio (ν), were identified from the tensile tests on cylindrical fabric samples without inlay yarn (GF configuration), presented in Figure 3.25 of Chapter 3. Since the experimental tensile curves obtained for the ankle, calf, and thigh were almost identical, a homogeneous behaviour of the loop structure was assumed for the entire stocking, and constant values of E and ν were assigned to all compression zones.

The identification procedure was carried out in two stages. First, an initial estimate of E and ν was obtained using the *Solver* tool in Microsoft Excel, by minimising the Mean Absolute Percentage Error (MAPE) between the predicted and experimental Tension–Strain curves for the GF configuration, as defined in Equation 4.19. In this expression, T_i^{model} is the tension predicted by the hybrid model at iteration i , T_i^{exp} is the corresponding experimental tension, and Q is the total number of experimental points. Both parameters were treated as free variables under the constraints $E > 0$ and $\nu > 0$.

$$MAPE [\%] = \frac{100}{Q} \sum_{i=1}^Q \frac{|T_i^{model} - T_i^{exp}|}{T_i^{exp}} \quad (4.19)$$

In the second step, a sensitivity analysis was performed to refine the identification and to evaluate the influence of each parameter on the predicted tensile response. Based on the preliminary estimates, a discrete set of values was defined around the initial range: $E = 0.015, 0.020, 0.025$, and 0.030 MPa, and $\nu = 0.05, 0.10, 0.15$, and 0.20 . All possible combinations were tested. For each pair (E, ν), the hybrid model was used to simulate the tensile response in the three compression zones, and the MAPE was computed for each zone separately.

To obtain a global measure of accuracy, a weighted error ($WErr$) was computed according to Equation 4.20, combining the errors of the three zones using weighting coefficients (C_{ankle} , C_{calf} , and C_{thigh}). Following the findings of Chapter 3, and given the predominant functional role of the ankle zone in the compression profile, the coefficients were set to $C_{ankle} = 0.6$, $C_{calf} = 0.3$, and $C_{thigh} = 0.1$, fulfilling the condition $C_{ankle} + C_{calf} + C_{thigh} = 1$.

$$WErr [\%] = C_{ankle} MAPE_{ankle} + C_{calf} MAPE_{calf} + C_{thigh} MAPE_{thigh}. \quad (4.20)$$

The sensitivity analysis provided the distribution of MAPE values for each zone and the global weighted error for each parameter pair. The optimal values of E and ν corresponded

to the minimum weighted error, ensuring the best simultaneous agreement with the experimental curves of the three compression zones. These identified parameters were subsequently assigned to the continuous elements in all zones, as the ground-yarn loop structure is homogeneous throughout the stocking.

4.2.5 Validation of the hybrid model

The hybrid model was validated by comparing its analytical predictions with the experimental tensile tests performed on knitted fabrics with and without inlay yarn. The objective of this comparison was to assess the ability of the model to reproduce the circumferential tensile behaviour of the three compression zones, ankle, calf, and thigh, under uniaxial loading along the circumferential direction while imposing the in-use longitudinal deformation.

For the weft-knitted fabric without inlay yarn (GF configuration), the analytical predictions were compared with the experimental Tension–Circumferential strain curves obtained from the cylindrical GF samples. This first validation stage assessed the capacity of the continuous elements to reproduce the tensile response of the loop structure alone, in the absence of any contribution from the inlay yarns.

For the inlaid weft-knitted fabric (IF configuration), the predicted tensile response resulted from the combined contribution of the continuous and discrete elements. The analytical Tension–Circumferential strain curves were compared with the corresponding experimental curves for each compression zone. This second validation stage evaluated the capability of the hybrid model to capture the effect of the inlay reinforcement on the circumferential stiffness, as well as the mechanical differentiation introduced by the variation of density of inlay yarn between zones.

In both configurations, the accuracy of the predictions was quantified using the mean absolute error (MAE) and the mean absolute percentage error (MAPE), defined in Equations 4.21 and 4.19. In these expressions, T_i^{model} is the predicted fabric tension at iteration i , T_i^{exp} is the experimental tension, and N is the number of data points used for the comparison.

$$MAE \text{ [N/mm]} = \frac{1}{N} \sum_{i=1}^N |T_i^{model} - T_i^{exp}| \quad (4.21)$$

These indicators are widely used in the assessment of mechanical models for textile and composite materials [73], [74], [83], [87].

4.3 Results and discussion

This section presents the results from the development and application of the hybrid mechanical model for predicting the circumferential tensile response of medical compression stockings. First, the identified mechanical parameters of the continuous elements are presented, including a sensitivity analysis of the elastic response with respect to E and ν . Then, the predicted tensile responses for fabrics with and without inlay yarn are presented, followed by a comparison with the experimental data for validation.

4.3.1 Identified mechanical parameters

The mechanical parameters assigned to the continuous elements, representing the loop structure, were identified from the tensile tests performed on the weft-knitted fabrics without inlay yarn (GF configuration). As shown in Chapter 3, the tensile responses of the ankle, calf, and thigh GF samples were nearly identical, confirming that the loop structure behaves homogeneously across the stocking. Consequently, a single pair of parameters, the Young's modulus (E) and the Poisson's ratio (ν), was identified and applied to all compression zones.

A preliminary estimation was carried out using the Solver tool (MS Excel), by minimizing the mean absolute percentage error (MAPE) between the experimental and predicted Tension–Circumferential strain curves of the weft knitted fabrics without inlay yarn (GF configuration), where only the elastic contribution was considered in the model. The error metric, defined in Equation 4.19, compares the predicted tension (T_i^{model}) with the experimental values (T_i^{exp}) over all evaluation points i .

This first step resulted an initial estimate of $E = 0.02$ MPa and $\nu = 0.01$, which provided a good approximation of the experimental response and served as the reference point for the subsequent sensitivity analysis.

Sensitivity analysis

To assess how E and ν influence the tensile response of the loop structure, a parametric analysis was performed around the preliminary estimates. Two series of calculations were carried out considering only the ankle zone:

- First, ν was fixed at 0.10 while E was varied in the range 0.015–0.030 MPa.
- Second, E was fixed at 0.020 MPa while ν was varied between 0.05 and 0.20.

Figures 4.6 and 4.7 illustrate the main trends for the two combinations of parameters. As observed in Figures 4.6, the model is highly sensitive to the elastic modulus E . For $E = 0.030$

and 0.025 MPa, the predicted GF response becomes considerably stiffer than the experimental one, with tension levels clearly overestimated. For $E = 0.015$ MPa the behaviour is too compliant and underestimates the forces. The best agreement is obtained for $E = 0.020$ MPa, which closely follows the experimental curve over the entire deformation range.

In contrast, as observed in Figures 4.7, the influence of ν is very limited. Variations of ν between 0.05 and 0.20 produce only small changes in the predicted curves, which is consistent with the predominantly uniaxial loading applied in the GF tests. This confirms that the circumferential tensile behaviour of the loop structure is governed primarily by E , whereas ν plays a secondary role.

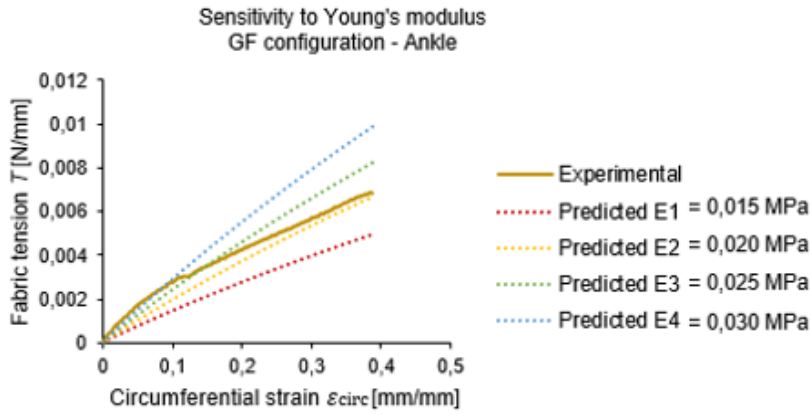


FIGURE 4.6: Model sensitivity for constant Poisson's ratio ($\nu = 0.10$) and variable Young's modulus (E).

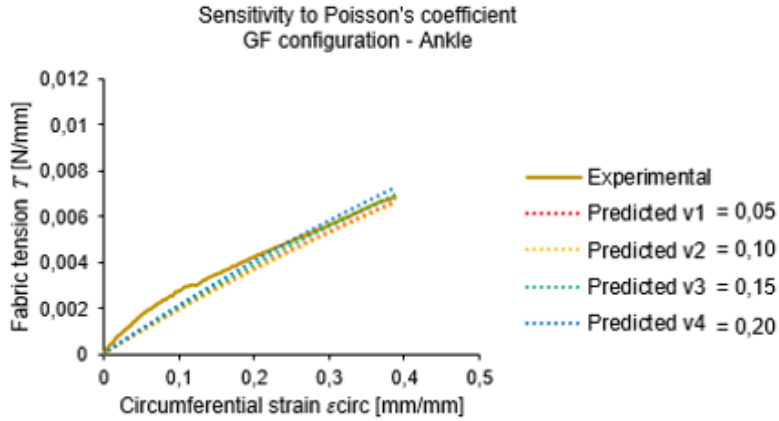


FIGURE 4.7: Model sensitivity for constant Young's modulus ($E = 0.02$ MPa) and variable Poisson's ratio (ν).

Evaluation of (E, ν) pairs

Finally, all combinations of E and ν were evaluated using the weighted error ($WErr$), defined in Equation 4.20. The weighting coefficients ($C_{ankle} = 0.6$, $C_{calf} = 0.3$, $C_{thigh} = 0.1$)

were chosen to reflect the relevance of each compression zone in therapeutic performance, as discussed in Chapter 3. The resulting MAPE and WErr values for all parameter pairs are reported in Table 5.2.

According to the results, for all ν values, $E = 0.015$ MPa systematically underestimates the tensile response, while $E = 0.025$ and 0.030 MPa produce very large deviations, particularly in the calf and thigh, where MAPE exceeds 30% and may reach 60%. Only $E = 0.020$ MPa provides low errors in all zones. Within this set, the combination $(E, \nu) = (0.020 \text{ MPa}, 0.10)$ gives the lowest weighted error (12.73%), confirming the preliminary estimation.

TABLE 4.2: Mean absolute percentage error (MAPE) and weighted error (WErr) values for different combinations of E and ν .

E [MPa]	Zone	$\nu = 0.05$		$\nu = 0.10$		$\nu = 0.15$		$\nu = 0.20$	
		MAPE [%]	WErr [%]	MAPE [%]	WErr [%]	MAPE [%]	WErr [%]	MAPE [%]	WErr [%]
0.015	Ankle	36.58		36.09		35.28		34.10	
	Calf	21.88		21.44		26.79		19.61	
	Thigh	21.86	30.70	20.51	30.14	17.92	31.00	13.64	27.71
0.020	Ankle	16.80		16.32		15.50		16.95	
	Calf	7.14		7.61		8.60		10.55	
	Thigh	5.40	12.76	6.52	12.73	9.43	12.82	15.15	14.85
0.025	Ankle	15.79		16.20		16.93		17.96	
	Calf	32.98		34.08		35.71		38.19	
	Thigh	30.23	22.39	32.49	23.19	36.79	24.55	43.94	26.63
0.030	Ankle	29.14		29.82		31.10		33.12	
	Calf	59.58		60.80		62.85		65.82	
	Thigh	56.77	41.04	59.99	42.03	64.14	43.93	72.72	46.90

The distribution of errors therefore demonstrates that the elastic modulus governs the overall stiffness of the loop structure, whereas ν has only a marginal influence. Based on these results, the elasticity parameters for the continuous element are $E = 0.02$ MPa and $\nu = 0.01$.

These values are consistent with those reported in the literature for thin knitted membranes used in compression garments [73], [75], [79].

4.3.2 Predicted tensile response under circumferential loading

The hybrid model, combining continuous and discrete elements, was used to predict the circumferential tensile response of the medical compression fabric. The mechanical parameters identified in Section 4.2.4, together with the dimensional inputs listed in Table 4.1, were introduced into the model for each compression zone.

As described in Sections 3.3.3.3 to 2.2.2 of Chapter 3, both the knitted architecture and the imposed circumferential strain differ between the ankle, calf, and thigh zones. Accordingly, the model was adapted to each zone by assigning the corresponding structural parameters, in particular the density of inlay yarn (D_{IY}). The calculations were carried out up to the maximal circumferential strain $\varepsilon_{\text{circ,max}}$ defined in Table 3.4, ensuring consistency with the experimental test range.

Two fabric configurations were analysed:

- Without inlay yarn (GF): only the loop structure was represented, using continuous elements.
- With inlay yarn (IF): the loop structure and inlay yarns were represented, combining continuous and discrete elements.

For each case, the maximal circumferential strain was computed from the ratio between the initial fabric perimeter (P_0) and the maximal leg perimeter ($P_{\text{leg,max}}$) according to Equation 3.2. This procedure reproduced the in-use deformation levels introduced in Chapter 3. The loading was applied in ten increments, each corresponding to 10% of the total imposed strain.

At each strain increment, stresses and internal forces were computed at the unit-cell scale from the constitutive laws defined in Section 4.2.2. For the continuous elastic element, membrane stresses were obtained from the isotropic plane-stress formulation using the identified values of E and ν . For the springs, the axial force f_2^{spring} was calculated from the stiffness function derived from the inlay-yarn tensile curve (Figure 3.9).

The continuous and discrete contributions were then combined to obtain the total force in the local coordinate system. The circumferential force at the fabric scale was calculated by multiplying the unit-cell force by the number of cells along the circumferential direction, following the procedure described in Section 4.2.3. The resulting analytical Force–Circumferential strain curves were used for comparison with the experimental results, as presented in the validation section below.

4.3.2.1 Fabric without inlay yarn

The hybrid model was first applied to the weft-knitted fabrics without inlay yarn (GF configuration). In this case, only the loop structure contributed to the circumferential response, since the discrete elements representing the inlay yarns were not activated. The calculations

were performed for the ankle, calf, and thigh zones using the same mechanical parameters for the continuous elements, while the dimensional inputs and unit-cell geometry were adapted to each zone, as defined in Section 4.2.1 and Table 4.1.

The predicted Tension–Circumferential strain curves are shown in Figure 4.8. All three zones exhibited very similar responses, characterised by low stiffness over the entire deformation range, consistent with the experimental observations reported in Chapter 3. This behaviour reflects the compliance of the loop structure, which deforms mainly through loop opening, yarn bending, and inter-loop rotation, rather than through significant yarn extension [40], [114].

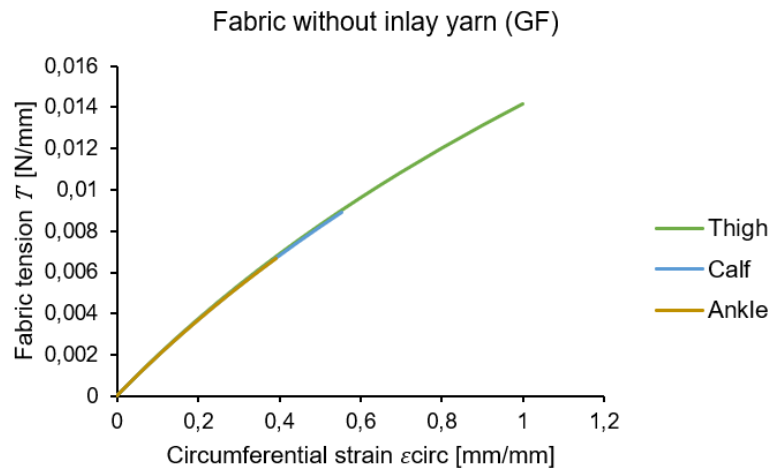


FIGURE 4.8: Predicted tension (T) under circumferential deformation (ϵ_{circ}) for ankle, calf, and thigh in the GF configuration.

Minor differences between zones appeared due to their distinct initial and maximal perimeters, which influence the number of unit cells involved in load transfer, and the global circumferential strain imposed during testing. However, these geometric effects did not substantially modify the overall stiffness, confirming that the loop structure alone behaves as a compliant, nearly homogeneous membrane under circumferential loading. This agrees with previous studies reporting that knitted fabrics without inserted reinforcements exhibit low and relatively uniform circumferential stiffness [39], [111].

4.3.2.2 Fabric with inlay yarn

The hybrid model was then applied to the inlaid weft-knitted fabrics (IF configuration). In this case, the global response resulted from the combined action of the continuous elements, representing the loop structure, and the discrete elements, representing the inlay yarns. The dimensional parameters and the unit-cell geometry were adapted to each compression zone according to the structural data reported in Table 4.1. These variations directly modified the

spring length and the number of elements per unit circumferential length, thereby influencing the stiffness of each zone.

The predicted Tension–Circumferential strain curves for the ankle, calf, and thigh are presented in Figure 4.9. Compared with the GF configuration, the IF fabrics exhibited a substantial increase in stiffness and a clear stiffness gradient between zones: the ankle showed the highest stiffness, followed by the calf and the thigh. This trend reflects the structural differences identified experimentally in Chapter 3, in particular the higher density of inlay yarn at the ankle and its reduction toward the proximal regions.

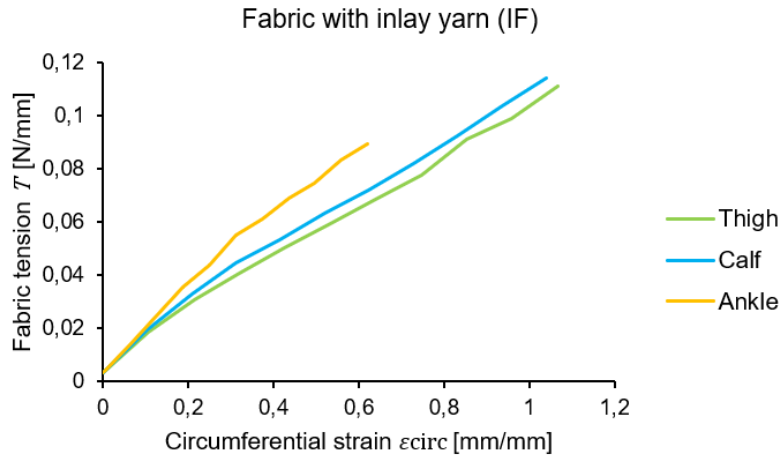


FIGURE 4.9: Predicted tension (T) under circumferential deformation (ϵ_{circ}) for ankle, calf, and thigh in the IF configuration.

The predicted curves exhibited a characteristic non-linear profile. At low strain, the response is governed mainly by the loop structure, which offers limited resistance to circumferential deformation due to the dominance of loop rotation, bending, and intra-loop sliding. As the strain increases, the inlay yarns progressively are activated, and begin to carry most of the circumferential load, producing the progressive stiffening observed in the curves. Such sequential activation of the looped and inserted yarns has been reported in previous studies on inlaid or hybrid knitted structures **tephanyInfluencenon-linearityFabric2016**, [74], [79], [87]. Similar non-linear stiffening behaviour has also been documented in compression fabrics where the circumferential reinforcement dominates the macroscopic tensile response [75], [83].

The clear stiffness gradient between zones was directly associated with the structural parameter D_{IY} : a higher density of inlay yarn, as in the ankle zone, results in shorter spring length and a larger number of load-bearing elements per unit circumferential width, producing a stiffer response. Conversely, the reduced inlay density in the thigh leads to a more compliant behaviour. These results are fully consistent with the experimental observations

in Chapter 3, where the inlay yarn system was shown to be the dominant contributor to circumferential stiffness and to the graduated mechanical behaviour of the stocking.

The combined analysis of the GF and IF predictions highlights the respective mechanical roles of the two fabric constituents. The loop structure, modelled by the continuous elements, contributes only modestly to the circumferential stiffness and exhibits a nearly uniform response across all zones. In contrast, the inlay yarns, represented by the discrete elements, introduce a strong directional reinforcement that dominates the macroscopic tensile behaviour and generates the stiffness gradient characteristic of graduated compression fabrics. The hybrid model successfully reproduced these mechanisms by coupling the elastic and spring responses within the unit cell, and by incorporating the zone-specific structural parameter D_{IY} . This demonstrates the capacity of the model to translate local structural variations into the global mechanical response of the fabric.

A detailed quantitative comparison with the experimental Tension–Circumferential strain curves is provided in Section 4.2.5.

4.3.3 Comparison with experimental data

The analytical predictions obtained with the hybrid model were validated using the experimental tensile tests presented in Chapter 3.

Two fabric configurations were examined to assess the model performance. The weft-knitted fabrics without inlay yarn were used to validate the representation of the loop structure, while the inlaid fabrics served to evaluate the combined mechanical response of the loop structure and the inlay yarns.

4.3.3.1 Fabric without inlay yarn

Model predictions for the fabric without inlay yarn (GF configuration) were compared with the experimental Tension–Circumferential strain curves for the three compression zones. The resulting curves per compression zone are presented in Figures 4.10 to 4.12. Overall, the hybrid model reproduced the low-stiffness behaviour of the loop structure with good accuracy across all zones, confirming the overall model's ability to reproduce the compliant behaviour of the loop structure under circumferential loading. This agreement is consistent with the experimental observations discussed in Chapter 3, where the absence of inlay yarn resulted in low stiffness and minimal variation between zones. However, a closer examination of the curves reveals zone-dependent differences related to the deformation range explored during testing.

In the ankle, the experimental curve, shown in Figure 4.10, exhibits a pronounced curvature at low strain, reflecting loop rearrangements in the initial loading phase. Because the hybrid model assigns a linear elastic behaviour to the continuous element, this early curvature cannot be captured, resulting in deviations before the response becomes more linear. Moreover, the ankle samples were tested over a relatively small deformation range, as defined in Table 3.4, which accentuates the discrepancy in the initial part of the curve and leads to a higher relative prediction error.

For the calf and thigh, the experimental curves presented in Figure 4.11) and (Figure 4.12, show a more gradual curvature and converge toward a quasi-linear trend as deformation increases. Since these zones reached higher circumferential strain levels, the part dominated by loop reorientation represents a smaller part of the total deformation. Consequently, the model reproduces the experimental behaviour more accurately, particularly beyond the initial strain range, where the response of the loop structure is more elastic.

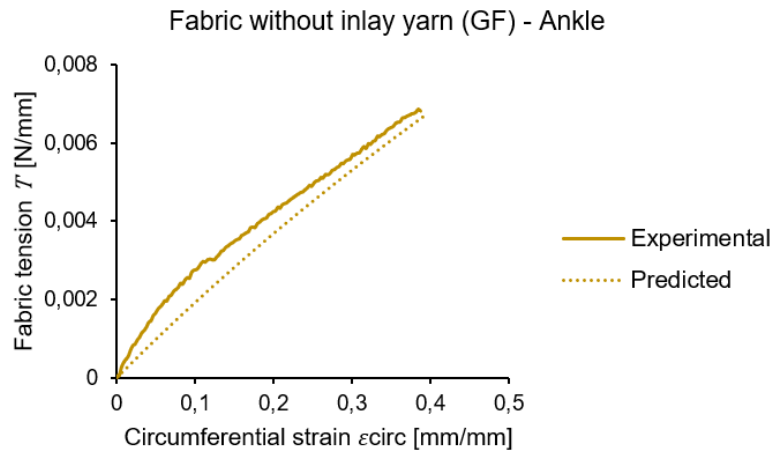


FIGURE 4.10: Comparison between predicted and experimental tensile response for ankle zone of fabric without inlay yarn (GF).

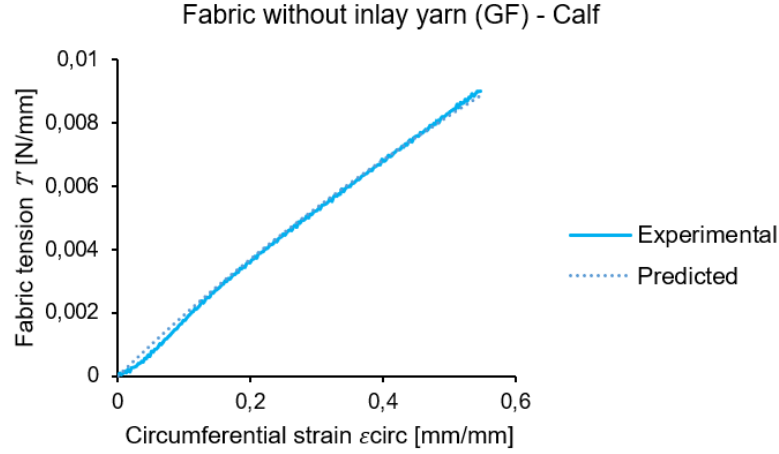


FIGURE 4.11: Comparison between predicted and experimental tensile response for calf zone of fabric without inlay yarn (GF).

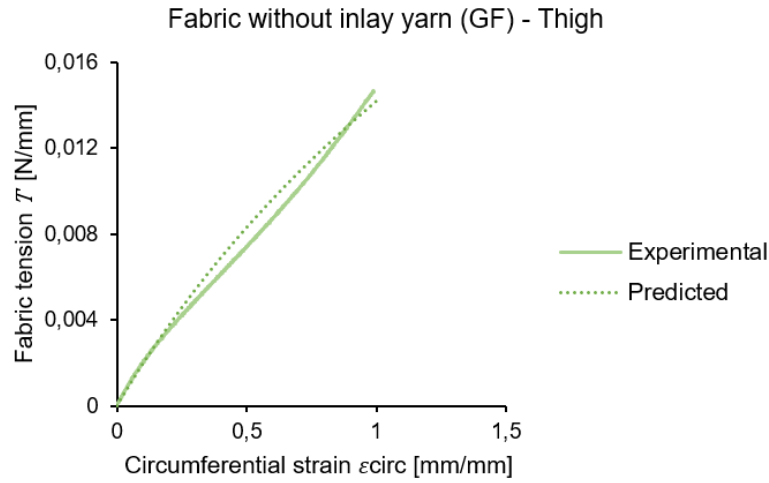


FIGURE 4.12: Comparison between predicted and experimental tensile response for thigh zone of fabric without inlay yarn (GF).

These qualitative observations are consistent with the quantitative results obtained with the accuracy indicators. The Mean Absolute Error (MAE) and Mean Absolute Percentage Error (MAPE), defined in Equations 4.21 and 4.19, measure the absolute and relative deviations between the predicted and experimental tensions. The corresponding values for the three zones are reported in Table 4.3.

The MAPE values confirm the trends observed in the curves: the highest error is found in the ankle, followed by the calf and the thigh. Despite these differences, the absolute deviations remain very small ($\text{MAE} < 0.0006 \text{ N/mm}$), showing that the prediction error is limited in magnitude even where the percentage error is higher.

At the highest deformation levels, slight discrepancies are observed, especially in the thigh. These deviations arise from the simplified definition of the loop structure as a linear isotropic continuous element, whereas real knitted fabrics present non-linear mechanisms such as loop reorientation, yarn sliding, and local buckling [42], [114], [120], [121]. Such effects become more pronounced at high strains [114], [121], which explains the slightly stiffer experimental response that is not fully captured by the simplified formulation.

TABLE 4.3: Mean absolute error (MAE) and Mean absolute percentage error (MAPE) between experimental and predicted fabric tension for the three compression zones of fabric without inlay yarn (GF configuration).

Compression zone	MAE [N/mm]	MAPE [%]
Ankle	0.00051	16.32
Calf	0.00018	7.61
Thigh	0.00047	6.52

Overall, the good agreement across zones demonstrates that the identified elastic parameters ($E = 0.02$ MPa, $\nu = 0.10$) provide an appropriate representation of the tensile response of the loop structure over the deformation range relevant for in-use conditions.

4.3.3.2 Fabric with inlay yarn

The model validation was then extended to the fabric with inlay yarn (IF configuration), where both yarn systems contribute to the circumferential response. The predicted Tension–Circumferential strain curves for the ankle, calf, and thigh are compared with the experimental results in Figures 4.13–4.15. In this configuration, the discrete elements representing the inlay yarns are activated, leading to a marked increase in circumferential stiffness compared with the GF configuration, as previously reported from the experimental analyses in Chapter 3.

Overall, the hybrid model reproduced the non-linear stiffening trends and the stiffness hierarchy between zones, although differences appeared depending on the deformation range and local knitted architecture.

In the ankle zone (Figure 4.13), the predicted curve closely follows the experimental response across the entire strain range tested. This zone is characterised by the smallest circumferential deformation, which results in a moderate initial curvature and a smoother transition towards the stiffer regime. The model reproduces these features well, indicating that the spring stiffness derived from the single-yarn tests (Figure 3.9) accurately models the activation of the inlay reinforcement. Small deviations at higher strain levels remain limited,

and are consistent with local structural mechanisms such as friction between the inlay and the loop structure, partial tightening of the ground yarns and geometric reorientation effects, which are not explicitly represented in the simplified spring formulation.

For the calf and thigh zones (Figures 4.14 and 4.15), the predicted curves reproduce the initial slope and overall non-linear evolution, and correctly reflect the decrease in stiffness from ankle to thigh induced by the reduction in density of inlay yarn (D_{IY}). However, the discrepancy between model and experiment increases with deformation. This trend is particularly visible in the thigh, where the experimental curve exhibits a more pronounced stiffening at large circumferential strains. Such behaviour is well documented in the literature and is attributed to deformation mechanisms not included in the hybrid representation, such as loop-reorientation, local buckling, frictional interaction between yarn systems and non-linear tightening of the knitted network [114], [120], [121]. Despite these effects, the model successfully reproduces the dominant contribution of the inlay yarns and the progressive load transfer from the loop structure to the inlay system, which is characteristic of inlaid and hybrid knitted fabrics **tephanyInfluencenon-linearityFabric2016**, [74], [79], [87].

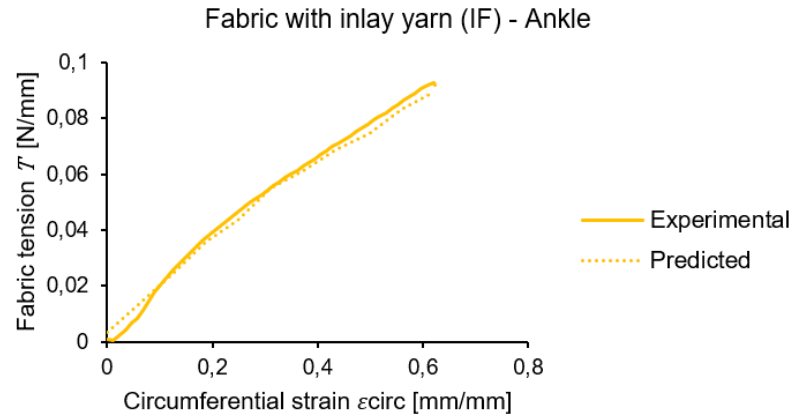


FIGURE 4.13: Comparison between predicted and experimental tensile response for ankle zone of fabric with inlay yarn (IF).

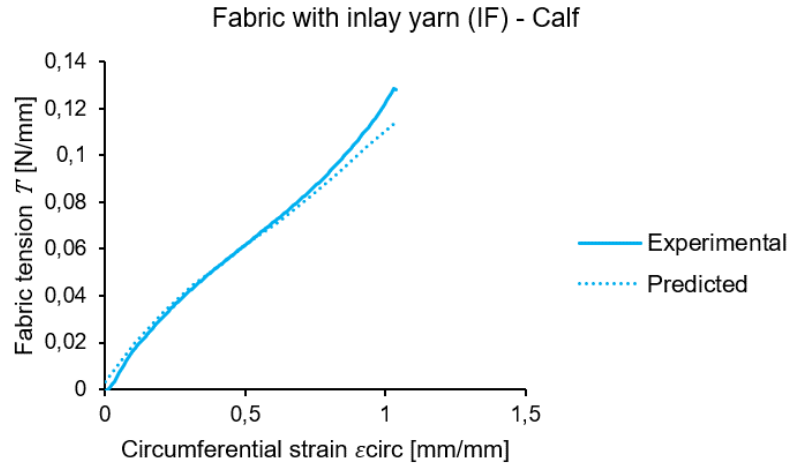


FIGURE 4.14: Comparison between predicted and experimental tensile response for calf zone of fabric with inlay yarn (IF).

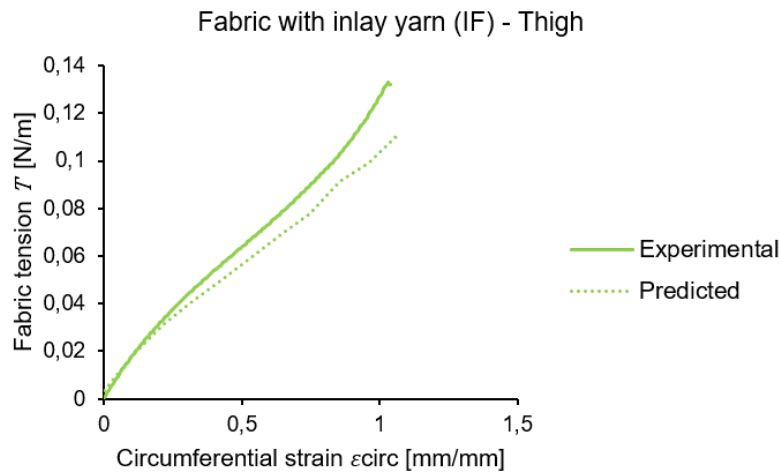


FIGURE 4.15: Comparison between predicted and experimental tensile response for thigh zone of fabric with inlay yarn (IF).

The quantitative comparison between predicted and experimental responses is summarised in Table 4.4. The MAE and MAPE values, computed according to Equations 4.21 and 4.19, remained below 0.01 N/mm and 11% respectively for all zones. The lowest deviation was observed for the ankle, where the imposed deformation remained moderate and the model was able to reproduce both the initial curvature and the progressive stiffening of the experimental response. In contrast, the deviations increased for the thigh at large strains, where the experimental curve displayed a sharper stiffening that the simplified hybrid formulation could not fully capture. This behaviour is consistent with the modelling assumptions discussed above, namely the use of linear elastic isotropic laws for the loop

structure and axial spring behaviour for the inlay yarns.

These error levels are within, or below, the typical ranges reported for analytical or hybrid models of compression fabrics (10–20%) [74], [75], [87], confirming the ability of the developed hybrid model to predict the circumferential tensile response of medical compression stocking with accuracy.

TABLE 4.4: Mean absolute error (MAE) and Mean absolute percentage error (MAPE) between experimental and predicted fabric tension for the three compression zones of fabric with inlay yarn (IF).

Compression zone	MAE [N/mm]	MAPE [%]
Ankle	0.0025	9.69
Calf	0.0038	5.76
Thigh	0.0095	10.50

4.3.4 Extension of the model for interface pressure prediction

The hybrid mechanical model developed in this chapter provides the circumferential tensile response of the knitted fabric for each compression zone, based on its local structural parameters and the constitutive behaviour of the loop structure and inlay yarn. A natural extension of this formulation is its direct application to the prediction of the interface pressure exerted by the stocking on the leg. This extension enables a continuous link between the experimentally identified fabric behaviour, the hybrid mechanical model developed in this chapter, and the pressure delivered under in-use conditions.

To establish this connection, an analytical interface–pressure model was prototyped. Its objective is double: (i) to provide a rapid estimation of the interface pressure from patient-specific or zone-specific dimensional inputs, and (ii) to complement the finite element simulations that will be introduced in Chapter 5. In its current implementation, the calculation is performed independently for each compression zone, using zone-specific structural parameters and the tensions predicted by the hybrid model.

Table 4.5 summarises the dimensional parameters defined earlier in Section 4.2.1 and the mechanical parameters identified in Section 4.3.1. These parameters constitute the complete set of inputs required for the analytical interface–pressure model.

TABLE 4.5: Dimensional and mechanical parameters used in the analytical interface–pressure model.

Parameter	Symbol
Initial perimeter	P_0
Leg perimeter (in-use)	P_{leg}
Fabric length (in-use)	L
Inlay yarn density	D_{IY}
Thickness	t
Young’s modulus	E
Poisson ratio	ν
Stiffness law	k

The hybrid model provides the local tensions for a given circumferential deformation. These tensions are combined to obtain the global circumferential fabric tension T , which is then converted into interface pressure using Laplace’s law, as given in Equation 3.9, considering the maximal leg perimeter as the deformed fabric perimeter ($P = P_{\text{leg,max}}$).

To illustrate the application of the analytical interface–pressure model, Table 4.6 reports the calculation steps for the ankle and calf zones at maximal leg perimeter ($P_{\text{leg,max}}$), given in Table 3.1. The resulting circumferential strains (ϵ_{circ}) given in Table 3.4 are introduced first, followed by the computed local tensions (T_{cont} , T_{spring}), with the resulting fabric tension (T) and experimental fabric tension (T_{exp}).

TABLE 4.6: Example application of the analytical model for predicting interface pressure at maximal leg perimeter for ankle and calf zones, with detailed circumferential strain (ϵ_{circ}), tension in the continuous element (T_{cont}), spring tension (T_{spring}), fabric tension (T_{model}) and experimental fabric tension (T_{exp}).

Quantity	Ankle	Calf
ϵ_{circ} [mm/mm]	0.62	1.09
T_{cont} [N]	0.368	0.618
T_{spring} [N]	2.21	2.645
T_{model} [N]	2.68	3.75
T_{exp} [N]	2.78	2.645

Finally, Table 4.7 compares the analytical predictions with the Laplace-based values and the direct pressure measurements obtained with the PicoPress[®] sensor in Chapter 3.

TABLE 4.7: Comparison between analytical predictions, Laplace-based experimental estimates, and PicoPress[®] measurements.

Compression zone	IP_{Laplace}	IP_{sensor}	IP_{model}
Ankle	2.41	2.53	2.35
Calf	1.96	1.95	1.91

These results show that the tension predicted by the hybrid model, once coupled to Laplace’s law, provides an accurate evaluation of the interface pressure under the in-use circumferential deformation. This analytical tool offers a fast first-step estimator for compression levels, and establishes a direct link between structural parameters, mechanical behaviour, and therapeutic pressure.

A natural continuation of this work is the extension of the model to predict the pressure distribution along the entire height of the stocking. By expressing the dimensional parameters (P , L , D_Y) as continuous functions along the longitudinal direction, the hybrid model may be integrated into a one-dimensional continuous formulation, enabling the prediction of the pressure profile delivered by the garment under realistic in-use conditions. This development will complement the full finite element simulations presented in Chapter 5, where the hybrid formulation is incorporated into a complete fabric–leg numerical model for interface pressure prediction.

4.4 Conclusions and research perspectives

Conclusions This chapter presented a hybrid mechanical model for predicting the circumferential tensile behaviour of inlaid weft-knitted fabrics used in medical compression stockings. The model computes the fabric tension (T) from measurable structural parameters of each compression zone by combining the membrane response of the loop structure, represented by continuous elements, with the axial response of the inlay yarn, modelled through discrete elements.

The formulation reproduces the tensile test protocol introduced in Chapter 3, where cylindrical fabric samples were elongated in the circumferential direction. Although the samples were tubular, the imposed deformation occurred tangentially to the cylindrical surface and could therefore be represented under plane-stress conditions, assuming a longitudinally fixed configuration corresponding to the in-use state of the stocking on the leg.

The loop structure was idealised as a linear isotropic elastic membrane, while the inlay yarns were assigned a non-linear axial behaviour consistent with the experimental stiffness

function $k(\varepsilon_{\text{circ}})$. Zone-specific dimensional parameters, such as the initial and maximal perimeters (P_0 , $P_{\text{leg,max}}$) and the inlay yarn density (D_{IY}), defined the unit-cell geometry and directly controlled the number of elemets. These structural inputs governed the local stiffness of the fabric and reproduced the gradient needed for graduated compression.

The identification of the mechanical parameters of the loop structure was carried out through a combined inversion and sensitivity analysis. Young's modulus had a dominant effect on the elastic response, whereas Poisson's ratio had negligible influence. A weighted error criterion prioritising the ankle zone ($C_{\text{ankle}} = 0.6$, $C_{\text{calf}} = 0.3$, $C_{\text{thigh}} = 0.1$) led to the selection of a single parameter set for all zones, $E = 0.02$ MPa and $\nu = 0.10$. The spring behaviour was obtained from the tensile response of the inlay yarns and governed the non-linear stiffening of the fabric.

The predicted tensile responses confirmed that the loop structure alone produces a compliant and nearly homogeneous circumferential behaviour, while the inclusion of the inlay yarn introduces non-linear stiffening and a distinct stiffness hierarchy between ankle, calf, and thigh. These trends demonstrated the necessity of modelling the fabric as a hybrid system in which each yarn family contributes differently to the global mechanical response.

Validation against the experimental results of Chapter 3 showed very good agreement. For the GF configuration, MAE remained below 0.00051 N/mm and MAPE below 17%. For the IF configuration, MAE remained below 0.0095 N/mm and MAPE values were 9.69% for ankle, 5.76% for calf, and 10.50% thigh. The deviations observed at large strains, particularly in the calf and thigh, were attributed to the simplifying assumptions of the model, such as the linear isotropic behaviour assigned to the loop structure and the absence of non-linear mechanisms such as loop reorganisation, friction, and local buckling.

Overall, the hybrid analytical model accurately predicted the circumferential tensile response of the medical compression fabric under in-use deformation. By explicitly linking the structural parameters of each zone to the resulting tensile stiffness, the model provides a mechanistic understanding of the role of the inlay yarn density in shaping the graduated compression profile. This formulation was subsequently transferred to the finite element model developed in Chapter 5, enabling the simulation of the fabric-leg interaction and the prediction of the interface pressure distribution.

Research perspectives The hybrid modelling strategy developed in this work, based on dimensional parameters and unit-cell description, is not restricted to medical compression stockings. It can be extended to other compression devices such as sleeves, bandages, or post-operative garments, provided that suitable structural parameters and yarn properties

are defined. Different yarn types, fibre compositions, or knitted architectures may require adapted constitutive laws for both continuous and discrete elements to maintain the model's predictive capability.

The present formulation assumed a linear elastic behaviour for the loop structure, which was adequate for moderate strains but less accurate in the high-strain regime. Introducing non-linear and possibly rate-dependent constitutive laws for the continuous elements would allow better reproduction of loop reorientation, frictional sliding, and other deformation mechanisms observed experimentally, thereby improving prediction accuracy at large strains.

At the analytical scale, the current interface-pressure model relies on Laplace's law and simplified cylindrical geometries. Although this provides a rapid first-order estimation of the delivered pressure, it does not account for variations in limb curvature or anatomical irregularities. A natural continuation of this work is therefore the numerical implementation of the hybrid formulation into a finite element framework, coupling continuous and discrete elements and the zone-specific structural properties. This numerical hybrid model ;developed in the following Chapter 5, will enable simulation of stocking placement over complex leg geometries and prediction of the resulting pressure distribution under realistic conditions, providing a powerful tool for the design and optimisation of compression garments.



Chapter 5

Finite element simulations for interface pressure prediction

5.1 Introduction

The interface pressure (IP) is defined as the pressure exerted by a compression fabric on the surface of the limb when the garment is stretched into its in-use configuration. It represents the mechanical transfer between the textile and the body, and is the main indicator of the therapeutic efficiency and mechanical performance of medical compression stockings (MCS). The level and gradient of IP determine the compression class of the product and directly influence venous return, lymphatic drainage, and wearing comfort. Accurate prediction of this pressure is therefore essential for evaluating, designing, and optimizing compression garments.

Different approaches have been developed to estimate the pressure applied by compression fabrics. Analytical methods based on Laplace's law [47], [96], [104], [122] relate the interface pressure to the circumferential tension and limb radius, assuming uniform cylindrical geometries and homogeneous material properties. While these models are widely used for classification and design evaluation, they fail to capture the local variations associated with the fabric's architecture or the complex anatomical shape of the leg. Experimental methods using pressure sensors, such as PicoPress[®], provide direct measurement but remain limited to discrete locations and can be affected by sensor thickness, local curvature, and fabric-sensor friction [67], [90].

Finite element (FE) simulations have become an effective tool to overcome these limitations. They allow coupling material behaviour, fabric structure, and limb geometry to predict the mechanical interaction between garment and body. Early FE studies represented compression fabrics as thin continuum layers with isotropic or orthotropic constitutive laws

identified from tensile tests [24], [69], [105]. These approaches reproduced the global deformation of the stocking but relied on homogenized assumptions, neglecting the internal fabric structure and the distinct contributions of the loop and inlay yarn systems. Later studies introduced nonlinear or viscoelastic formulations [81], [87], but the fabric was still treated as a continuous medium.

The mechanical behaviour of medical compression fabrics is governed by their inlaid weft-knitted architecture, which combines two yarn systems with distinct functions: the loop structure, formed by the ground yarn, provides elasticity and comfort, while the inlay yarns, inserted in each course, reinforce the fabric circumferentially. The density of inlay yarn D_{IY} varies along the stocking to generate the required pressure gradient, producing higher stiffness at the ankle and lower stiffness towards the thigh. The experimental analyses presented in Chapter 3 defined the structural parameters and mechanical response of each compression zone under in-use conditions. The hybrid mechanical model developed in Chapter 4 provided a mesoscopic description of the fabric by combining shell and connector elements to represent the loop structure and inlay yarn, respectively, and identified the mechanical parameters required for numerical prediction, namely the elastic modulus (E), Poisson's ratio (ν), and connector stiffness (k).

The present chapter extends this modelling strategy to a finite element framework for predicting the interface pressure generated by a medical compression stocking. The model integrates the structural and mechanical parameters identified in the previous chapters and simulates the placement of the fabric over rigid cylindrical leg geometries. Each compression zone, ankle, calf, and thigh, is represented using its specific dimensional and structural parameters, such as the initial fabric perimeter (P_0) and the density of inlay yarn (D_{IY}). This approach enables the prediction of circumferential deformation and the resulting interface pressure directly from the fabric architecture, without relying on homogenised material descriptions.

This chapter first presents the finite element implementation of the hybrid model, including the definition of the element types, the mesh configuration, and the material parameters associated with the loop structure and the inlay yarn. The simulation procedure used to reproduce the placement of the stocking on the leg is then described, together with the evaluation of the interface pressure obtained from the contact variable *CPRESS*. In addition, the response of the inlay yarn system is analysed by extracting the axial force carried by the connector elements, enabling direct assessment of the yarn tension as a function of the imposed circumferential deformation. A mesh-sensitivity analysis is also performed to verify the numerical stability of the contact solution and to ensure that both the predicted

pressure and the connector-force response are independent of the discretisation. Finally, the numerical predictions are validated through comparison with analytical estimations based on Laplace's law and with experimental measurements obtained using PicoPress[®] sensors, in order to assess the accuracy of the model and its ability to reproduce the graduated compression profile of medical compression stockings.

All notations and symbols used throughout this chapter are listed and defined in the glossary at the end of the manuscript.

5.2 Methodology

The methodology presented in this chapter aims to implement the hybrid mechanical model developed in Chapter 4 within a finite element simulation, in order to predict the interface pressure between the medical compression stocking and the leg under in-use conditions.

This approach preserves the local mesoscopic representation of the inlaid weft knitted fabric introduced previously, in which the mechanical response of the fabric results from the combined action of the loop structure and the inlay yarn reinforcement. In the finite element model, the two yarn systems are explicitly represented. The loop structure is modelled as a continuous elastic membrane, while the inlay yarns are described as axial connectors. The constitutive laws and mechanical parameters identified from the experimental characterisation in Chapter 3 and from the hybrid analytical model in Chapter 4 are directly implemented in the numerical model.

The local structural parameters of the fabric, such as the density of inlay yarn (D_{IY}), the initial fabric perimeter (P_0), and the sample length (L), were identified experimentally in Chapter 3, and were used to define the geometry and the meshing of each compression zone. The mechanical parameters of the elements, elastic modulus (E), Poisson's ratio (ν), and stiffness (k), were identified in Chapter 4 from experimental data of Chapter 3, and were kept constant for the different zones, as they describe the intrinsic material behaviour of the yarn systems. As shown with the analytical hybrid model presented in Chapter 4, this strategy allows the circumferential stiffness of the fabric to be controlled through structural parameters, without the need for homogenisation.

The numerical simulations reproduce the placement of the stocking over rigid leg geometries corresponding to reference leg perimeters (P_{leg}), identified from the manufacturer's size table. This allows the validation of the fabric model through direct comparison with experimental measurements performed on rigid leg models, presented in Section 3.4.2 of Chapter

3. The circumferential deformation imposed by the leg is applied to the fabric through contact interactions, inducing tensile forces within the hybrid model. The resulting interface pressure is computed from the contact pressure field at the fabric–leg interface, following approaches previously adopted in similar works [19], [58].

The proposed methodology therefore enables the prediction of the interface pressure generated in the different compression zones, ankle, calf and thigh, from structural and mechanical parameters defined independently for each region.

Section 5.2.1 first describes the finite element model of the fabric, including the definition of the element types, the mesh configuration derived from the unit cell geometry, and the material parameters assigned to the continuous and discrete elements. Section 5.2.2 then details the numerical simulations used to reproduce the placement of the stocking over the rigid leg geometry, describing the leg model, the boundary conditions, and the methods used to compute both the interface pressure and the tensile response of the inlay yarn. A mesh sensitivity analysis is also performed to verify the numerical stability of the simulations and to ensure that both the predicted pressure and the connector force response are independent of the meshing. Finally, Section 5.2.3 presents the validation procedures used to compare the numerical results with analytical predictions and experimental measurements.

5.2.1 Hybrid finite element model of the medical compression stocking

The hybrid finite element model of the medical compression stocking is derived from the mechanical formulation developed and validated in Chapter 4. The inlaid weft knitted fabric is represented through a mesoscopic description, accounting for its hybrid architecture by combining continuous and discrete finite elements.

The loop structure of the fabric is modelled using four node shell elements (S4), which describe the continuous membrane behaviour of the ground yarn network under plane stress conditions. Bending stiffness is neglected, as the mechanical response of the fabric is dominated by in plane tensile forces.

The reinforcement introduced by the inlay yarns is modelled using two-node connector elements (CONN3D2), which transmit only axial forces and reproduce the discrete tensile contribution of the inlay yarns laid along the course direction. As described in Section 5.2.2.4, this formulation enables the post-processing of the axial forces in the connectors and the direct evaluation of the tensile force carried by each inlay yarn.

The assembled mesh of the fabric forms a cylindrical structure representative of the tubular geometry of the medical compression stocking, as commonly adopted in numerical models of tubular textiles and compression garments [24], [45], [79], [80]. As shown in Figure 5.1,

the shell elements representing the loop structure define a continuous cylindrical surface, while the connector elements modelling the inlay yarns follow helical trajectories along the fabric, consistent with the course-wise insertion of the inlay yarns in the knitted structure. This geometric description allows the circumferential deformation imposed by the leg to be applied directly to the fabric and ensures mechanical compatibility between the continuous shell network and the discrete inlay yarn reinforcement.

The geometrical parameters defining the local and global mesh configuration, together with the material parameters governing the behaviour of the elements, are described in the sections below.

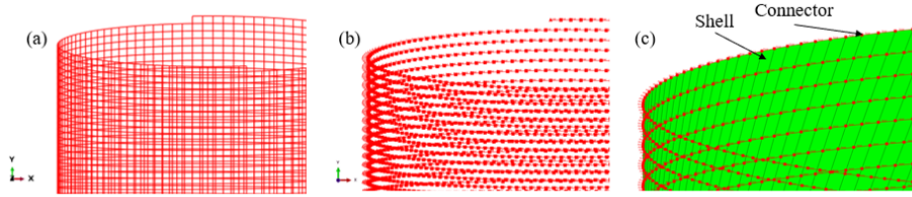


FIGURE 5.1: Hybrid finite element model of the medical compression stocking: (a) shell elements modelling the continuous loop structure, (b) connector elements modelling the inlay-yarn reinforcement, (c) assembly of shells and connectors.

5.2.1.1 Geometric parameters and mesh configuration

The dimensions of the shell and connector elements define the local configuration of the fabric mesh. As previously considered for the hybrid model in Chapter 3, the local configuration is controlled by the density of inlay yarn, which defines the longitudinal spacing between consecutive inlay yarn courses. The height of the shell element (h), is given by this spacing and calculated as the inverse of the density of inlay yarn. In the initial configuration, the shell elements are defined with a square geometry, with a height and width equal to h , consistent with the hybrid model configuration introduced in Chapter 4. Since each connector links two opposite nodes of adjacent shell elements, the initial connector length is equal to the unit cell height ($l_{\text{conn},0} = h = 1/D_{\text{IY}}$).

By considering the in-use density of inlay yarn, given in Table 3.2 of Chapter 3, the longitudinal extension of the fabric is implicitly accounted for in the mesh configuration. As a result, only the circumferential deformation needs to be imposed during the simulations, which simplifies the numerical procedure while remaining consistent with the experimental and analytical configurations of Chapters 3 and 4.

The shell elements define a continuous cylindrical surface, whose initial perimeter (P_0) is defined from the experimental data, reported in Table 3.3 in Chapter 3. The total length

of the cylindrical model is fixed as $L_{\text{total}} = 50$ mm for all compression zones. As described below in Section 5.2.2, interface pressure is evaluated over a central region of length $L = 30$ mm, corresponding to the effective fabric length considered in the tensile tests and in the hybrid modelling, as described in Chapters 3 and 4.

The connector elements form two helical paths that correspond to the two inlay yarns inserted during circular knitting. The inclination angle of the connectors (α) is defined as a function of the density of inlay yarn and the fabric perimeter, following the geometrical relations introduced in Section 4.2.1 of Chapter 4. The initial inclination angle (α_0) corresponds to the undeformed configuration at given by the initial fabric perimeter (P_0). During the simulations, the inclination angle is updated as a function of the deformed fabric perimeter, reproducing the geometrical reorientation of the inlay yarns induced by circumferential deformation.

Figure 5.2 summarises the parameters governing the local and global mesh of the finite element model of the medical compression stocking.

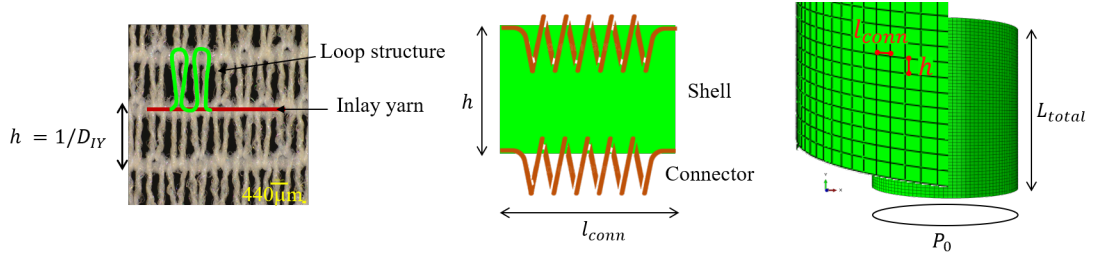


FIGURE 5.2: Parameters governing the mesh of the hybrid finite element model of the fabric.

Most studies simulated a single cylindrical region of the stocking in contact with a leg-shaped surface to predict local pressure distribution [87], [105], while few others represented the full stocking geometry [24], [69]. In the present work, independent finite element models are developed for each compression zone of the stocking. This choice allows the hybrid finite element formulation to be established and validated at the zone level, before extending the approach to a full-scale model of the medical compression stocking.

Figure 5.3 presents the finite element meshes generated for the ankle, calf and thigh compression zones. Each model is defined from the experimentally measured structural parameters, namely the total length (L_{total}), the initial fabric perimeter (P_0) and the density of inlay yarn (D_{IY}). These parameters, summarised in Table 5.1, define the local and global meshes of the three compression zones.

Variations in the density of inlay yarn between compression zones result in different element sizes and, consequently, in a higher number of connector elements per unit length

in zones with higher reinforcement, directly reflecting the local distribution of the inlay yarns in the knitted structure. In the ankle region, where the density of inlay yarn is higher, the resulting unit cell dimensions are smaller, leading to a higher number of connector elements along the longitudinal direction (N_{length}), as reflected in Table 5.1. This configuration reflects the increased inlay yarn reinforcement in this zone and results in a higher circumferential stiffness of the fabric compared with the calf and thigh regions.

TABLE 5.1: Geometrical parameters defining the mesh of the three compression zones: total fabric length (L_{total}), initial fabric perimeter (P_0), density of inlay yarn (D_{IY}), element height and initial connector length ($h = l_{\text{conn},0}$), and number of connectors per total fabric (N_{length}).

Zone	L_{total} [mm]	P_0 [mm]	D_{IY} [mm]	$h = l_{\text{conn},0}$ [mm]	N_{length}
Ankle	50	148	1.13	0.89	56
Calf	50	196	0.87	1.15	43
Thigh	50	288	0.8	1.25	40

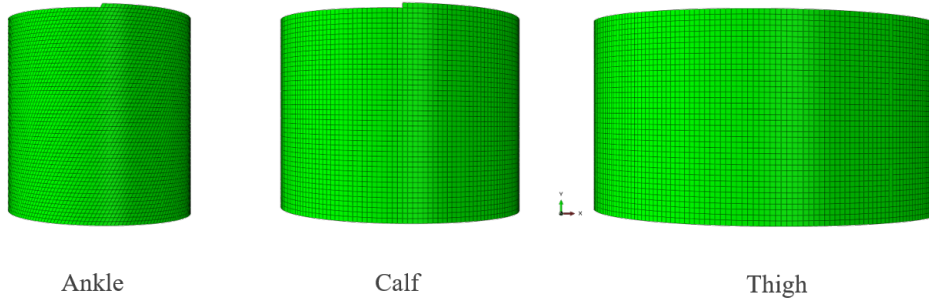


FIGURE 5.3: Generated finite element models of the three compression zones: ankle, calf, and thigh.

5.2.1.2 Material parameters

Previous finite element models of medical compression stockings have generally described the fabric as a continuous shell or membrane with homogenised mechanical properties [24], [69], [87], [105]. Most of these models assumed isotropic linear elastic behaviour for simplicity, with elastic modulus values typically ranging from $E = 0.01$ to 0.5 MPa and Poisson's ratio between $\nu = 0.3$ and $\nu = 0.49$, identified from the global tensile response of the fabric. Although some studies introduced anisotropic or orthotropic formulations to better approximate the knitted architecture [24], [69], the fabric was still treated as a single homogeneous layer, without explicitly distinguishing the mechanical contributions of the loop structure and the inlay yarns. In most cases, the mechanical parameters were identified through inverse calibration on global force–displacement curves.

In the proposed hybrid finite element model, the mechanical parameters assigned to each element type are obtained directly from the experimental characterisation and identification procedures described in Chapters 3 and 4. This approach allows the individual mechanical contributions of the two yarn systems to be represented explicitly within the numerical model.

The loop structure, modelled with shell elements, is assigned an isotropic linear elastic behaviour characterised by an elastic modulus E and a Poisson's ratio ν . These parameters are identified by inverse analysis from uniaxial tensile tests performed on weft-knitted fabrics without inlay yarns (GF configuration), as described in Sections 4.2.4 and 4.3.1 of Chapter 4. The identified values, $E = 0.02$ MPa and $\nu = 0.1$, are used for all compression zones, as they describe the intrinsic mechanical behaviour of the loop structure.

The inlay yarns were modelled using connector elements with an axial spring behaviour. The axial force carried by each connector (f^{conn}) was defined as a function of the connector stiffness (k) and the axial displacement (Δl_{conn}), according to Equation 5.1.

$$f^{\text{conn}} = k \cdot \Delta l_{\text{conn}} \quad (5.1)$$

To define the axial behaviour of the connectors, force–displacement laws are introduced in the finite element model. These laws are derived from the experimental tensile response of the inlay yarn, initially characterised in terms of Force–Strain for the hybrid analytical model. In contrast to the analytical formulation, where the connector force is expressed as a function of normalised strain, the finite element implementation requires the axial force to be defined as a function of displacement. Since the axial displacement depends on the initial length of the connector, which varies between compression zones due to differences in density of inlay yarn, three Force–Displacement input files are generated, one for each compression zone.

Because the inlay yarns are inserted into the loop structure without forming loops, their deformation is assumed to be equal to the circumferential deformation of the fabric. For a given fabric deformation, the axial displacement of each connector is computed from its initial length, and the corresponding axial force is obtained from the associated Force–Displacement law. This formulation ensures consistency between the analytical hybrid model and its finite element implementation.

Overall, the mechanical parameters assigned to each element type are defined independently from experimental characterisation and are kept constant for all compression zones,

while local structural parameters, in particular the density of inlay yarn, are used to introduce spatial variations in the mesh configuration. These variations result in different numbers of connector per unit length, governing the local circumferential stiffness of the fabric.

The coupling of the hybrid fabric models with leg geometries, described in the following section, enables the prediction of the interface pressure as a function of the mechanical parameters of the fabric constituents and the local structural parameters defining the knitted architecture.

5.2.2 Finite element simulations for interface pressure calculation

Finite element simulations were performed to predict the interface pressure generated by the medical compression stocking when placed over a rigid leg model. Since the leg perimeter (P_{leg}) is larger than the relaxed fabric perimeter (P_0), the fabric undergoes circumferential elongation during placement, inducing tensile loads within the knitted structure. These tensile loads are transmitted to the leg surface through contact interactions, resulting in an interface pressure (IP).

The magnitude and distribution of the interface pressure depend on the mechanical parameters of the fabric constituents, the local structural parameters defining each compression zone, and the contact conditions at the fabric–leg interface. The numerical simulations therefore provide a direct link between fabric structure, material behaviour, and pressure generation.

The following sections describe the finite element configuration adopted to simulate the placement of the fabric over the leg, including the leg geometry, boundary conditions, and contact formulation. The procedures used to evaluate the interface pressure from the contact variable $CPRESS$ and to analyse the tensile response of the inlay yarn through the connector axial force (CTF) are also presented. Finally, a mesh sensitivity analysis is performed to verify the numerical stability of the contact solution and the convergence of the predicted results.

5.2.2.1 Numerical leg model

To reproduce the mechanical interaction between the medical compression stocking and the leg during placement, each compression zone was associated with an independent numerical leg model. The ankle, calf, and thigh regions were therefore represented by separate cylindrical geometries, corresponding to the anatomical locations of the different compression zones of the stocking.

The leg surface was modelled as a rigid body and discretised using quadrilateral rigid shell elements (R3D4). This simplified representation was adopted as a first modelling approach to reduce computational cost and to isolate the mechanical contribution of the fabric, without introducing additional complexity associated with soft-tissue deformation. The size of the rigid leg elements was selected according to the ratio between the rigid and fabric meshes, as discussed in the mesh sensitivity analysis presented in Section 5.2.2.5.

Similar simplified leg geometries and rigid-surface assumptions have been widely used in previous numerical studies to evaluate interface pressure in compression garments [81], [87], [105]. These works have shown that rigid-leg models are sufficient to reproduce the fabric-induced pressure field when the objective is to analyse the mechanical behaviour of the stocking rather than the response of the underlying tissues. More advanced approaches have incorporated deformable leg models to simulate the coupled fabric–tissue interaction [24], [69]. While these models provide a more realistic description of tissue deformation, they require higher computational resources. In this context, the use of a rigid leg model represents a reasonable compromise for the mechanical analysis of compression fabrics.

The dimensions of the leg models were defined from reference leg perimeters obtained from the manufacturer’s size table, presented in Chapter 3 and summarised in Table 3.1. For each compression zone, three reference perimeters were considered, corresponding to the minimum, average, and maximum values of the size range. This variation within a given region allows the influence of circumferential deformation on the interface pressure to be evaluated, while the comparison between regions provides insight into the effect of the knitted structure and the inlay-yarn density on pressure generation.

To facilitate the numerical placement of the stocking and to ensure consistency with the experimental setup, the leg geometry includes a conical transition at its lower end. This feature reproduces the geometry of the 3D-printed leg forms used for the PicoPress[®] pressure measurements described in Section 3.4.2 of Chapter 3.

The dimensions of the rigid leg models were previously summarised in Table 3.7 of Chapter 3. Figure 5.4 illustrates the resulting numerical leg geometries corresponding to the three reference leg perimeters, minimal $P_{\text{leg,min}}$, average $P_{\text{leg,avg}}$, and maximal $P_{\text{leg,max}}$, for the ankle, calf, and thigh compression zones.

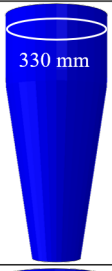
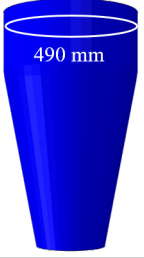
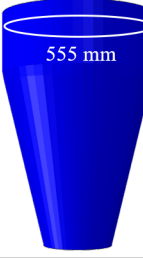
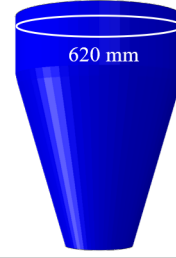
Compression zone	$P_{leg,min}$	$P_{leg,avg}$	$P_{leg,max}$
Ankle			
Calf			
Thigh			

FIGURE 5.4: Rigid leg models of three reference leg perimeters for the different compression zones.

5.2.2.2 Boundary conditions and contact formulation

The simulation reproduces the placement of the fabric over the rigid leg model through a controlled sliding motion. As illustrated in Figure 5.5, the fabric is initially positioned below the lower conical part of the leg and then displaced vertically upwards until reaching its final position. The final configuration is defined such that the fabric is centred on the cylindrical region of the leg, with its upper edge located 10 mm below the top boundary of the cylinder. This setup reproduces the experimental placement procedure adopted for the PicoPress[®] measurements described in Chapter 3.

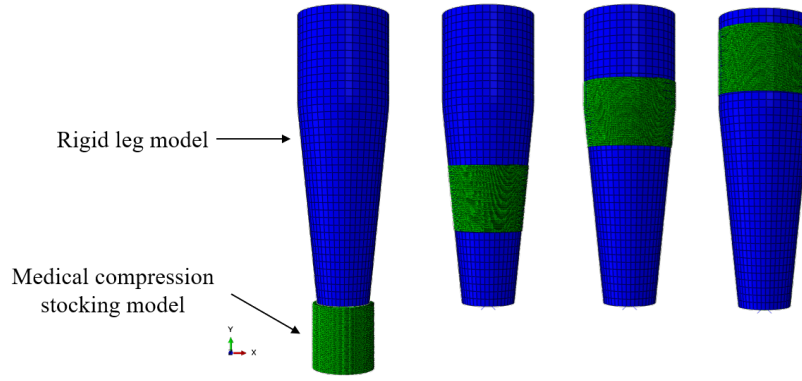


FIGURE 5.5: Simulation of the placement of the fabric over the rigid leg model.

During the simulation, the rigid leg is fully constrained, with all translational and rotational degrees of freedom blocked. The fabric mesh is displaced vertically until full contact with the leg surface is achieved. Both longitudinal ends of the fabric are left free, avoiding artificial stress concentrations and allowing uniform circumferential deformation along the length of the cylinder. The fabric is free to deform in the circumferential direction, where the imposed strain ϵ_{circ} results from the difference between the leg perimeter and the relaxed fabric perimeter. This imposed deformation defines the tensile state developed in the fabric and, consequently, the pressure transmitted to the leg surface.

A frictionless surface-to-surface contact formulation is adopted, in agreement with numerical strategies reported in the literature [24], [69]. These studies showed that neglecting friction ensures a stable and conservative transfer of normal contact stresses without introducing artificial tangential constraints at the fabric–leg interface. The rigid leg is defined as the master surface and the outer face of the fabric as the slave surface, ensuring a stable transfer of normal contact forces while preserving the in-plane membrane behaviour of the shell elements.

The simulations are solved using an explicit solver (ABAQUS/EXPLICIT) under quasi-static conditions, as commonly adopted in finite element analyses of compression stockings [87], [105]. This solver is well suited for contact problems involving large deformations, as it avoids iterative equilibrium procedures and provides consistent convergence under complex contact conditions. In the present case, it ensures numerical stability during the sliding phase and prevents instabilities associated with nonlinear contact and large circumferential deformation.

The input files for all simulations are generated using a dedicated PYTHON script developed for ABAQUS. This script enables a fully parametric definition of the models, ensuring reproducibility and consistency across the different compression zones and leg perimeters.

For each configuration, the script defines the fabric mesh and element types, generates the rigid leg geometry with the corresponding dimensions, assigns boundary conditions and contact interactions, and specifies the output variables to be stored. The recorded outputs include nodal displacements, contact pressure (*CPRESS*), and connector axial force (*CTF*), allowing both interface pressure and inlay yarn response to be analysed.

5.2.2.3 Evaluation of interface pressure

The interface pressure (*IP*) was obtained from the normal contact pressure variable *CPRESS* computed on the master surface of the rigid leg model. In ABAQUS/EXPLICIT, the contact pressure acting on the master surface is reported as the variable *CPRESS* with a negative sign convention (*CPRESS SNEG*), which corresponds to compressive normal contact stresses. This variable has been widely used in numerical studies to evaluate interface pressure in compression garments and textile–body contact problems [69], [87], [105].

The interface pressure was evaluated over a central region of length $L = 30$ mm on the leg surface, as illustrated in Figure 5.6. This avoids boundary effects near the fabric edges while remaining consistent with the fabric length considered in the tensile tests, shown in Figure 3.14 in Chapter 3.

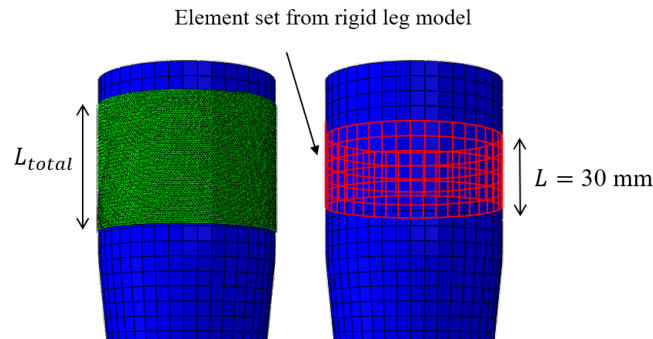


FIGURE 5.6: Element set defined on the leg surface for the evaluation of the interface pressure (*IP*) at the end of the simulation.

The contact pressure results were obtained and processed using a second PYTHON script. For each simulation, the script identified the nodes and elements belonging to the element set on the leg surface, obtained the corresponding *CPRESS – SNEG* values from the output database, and computed the average and standard deviation. This automated procedure ensured rapid, consistent and reproducible pressure evaluation across all compression zones and leg geometries.

5.2.2.4 Evaluation of yarn tensile response

The tensile response of the inlay yarn was evaluated to verify the mechanical consistency of the hybrid finite element model and to establish a link between the yarn tensile force, the local density of inlay yarn reinforcement, and the resulting interface pressure. As shown experimentally in Chapter 3 and demonstrated with the hybrid mechanical model in Chapter 4, the inlay yarn is the main load-bearing component of the fabric in the circumferential direction and governs the tensile stiffness associated with compression generation.

In the knitted structure, each inlay yarn is inserted along one course and provides the tensile reinforcement associated with that course. In the finite element model, the inlay yarns are represented by axial connector elements aligned with the course direction. For a given compression zone, the number of connectors distributed along the fabric length, denoted N_{length} and reported in Table 5.1, directly reflects the number of courses and thus the number of inlay yarns contributing to the circumferential tensile response.

To characterise the tensile contribution of an individual inlay yarn independently of the number of connectors, the connector force was analysed using the connector total force output variable ($CTF - Local$). The average value of CTF over all connector elements was computed as the tensile force carried by a single inlay yarn under the imposed circumferential deformation. Figure 5.7 illustrates the distribution of connector forces obtained for the ankle zone at the average leg perimeter.

As for the interface pressure evaluation, a dedicated PYTHON post-processing script was used to obtain the axial force from all connector elements and to compute the corresponding average value and standard deviation over the entire fabric mesh. These results were subsequently used to analyse the predicted yarn tensile response across compression zones and to validate the connector-based formulation through comparison with experimental yarn data.

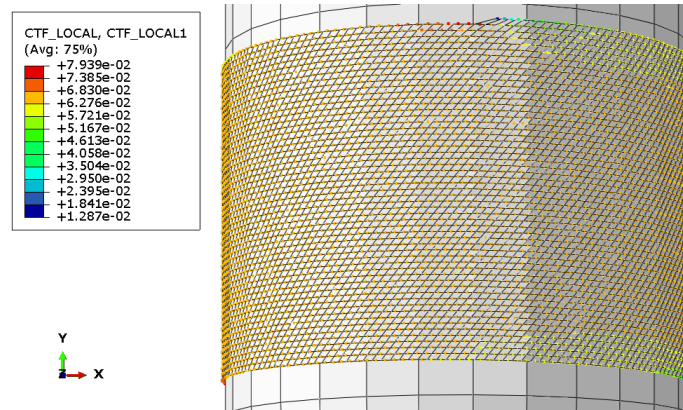


FIGURE 5.7: Connector total force (CTF), expressed in N, obtained for the average leg perimeter of ankle zone.

5.2.2.5 Mesh-sensitivity analysis

A mesh-sensitivity analysis was performed to quantify the influence of the discretisation ratio between the rigid leg and the fabric on the predicted interface pressure. The objective was to ensure that the calculated pressure field was independent of the mesh density of the rigid surface and to verify the numerical stability and convergence of the contact formulation.

The mesh-size ratio (R_{mesh}) was defined as the ratio between the element height of the rigid leg mesh (h_{rigid}) and that of the fabric mesh (h_{fabric}), as given in Equation 5.2. Since both meshes are composed of square-shaped elements, the element height and width are identical for each part.

$$R_{\text{mesh}} = \frac{h_{\text{rigid}}}{h_{\text{fabric}}} \quad (5.2)$$

For each mesh-size ratio, the average interface pressure was computed from the *CPRESS* values extracted over the central element set of length 30 mm defined on the leg surface, as described in Section 5.2.2.3.

The analysis was conducted in two successive stages. In a first stage, a sensitivity study was carried out using the ankle configuration and the average leg perimeter $P_{\text{leg,avg}}$, which corresponds to the most critical compression zone. The fabric mesh was defined from the structural parameters of this zone and kept constant, while the element size of the rigid leg surface was progressively increased. The mesh-size ratio was varied over the range $R_{\text{mesh}} = 2, 4, 6, 7, 8, 9$, and 10. In all cases, the element size of the rigid leg was kept larger than that of the fabric mesh. This choice is consistent with the master–slave contact formulation, to avoid artificial penetration during large deformations.

Based on the results of this first analysis, the two most suitable mesh-size ratios were selected. In a second stage, these ratios were further evaluated using the calf configuration for the three reference leg perimeters, minimal, average, and maximal. This second analysis allowed verification that the selected mesh-size ratios provide stable and mesh-independent predictions not only for different compression zones but also for different levels of circumferential deformation.

In addition to the interface pressure, the sensitivity of the fabric tensile response was also examined by analysing the connector total force for the selected mesh-size ratios. The connector force response was evaluated for $R_{\text{mesh}} = 7$ and 9 to verify that the predicted tensile behaviour of the inlay yarn was not affected by the discretisation of the rigid surface.

This analysis allowed the identification of a mesh-size ratio for which both the predicted interface pressure and the connector force were no longer affected by further refinement of

the rigid leg discretisation, ensuring mesh-independent numerical results.

5.2.3 Validation of the hybrid finite element model

The hybrid finite element model pressure predictions were validated by comparison with analytical estimations based on Laplace's law and with experimental measurements obtained using PicoPress[®] sensors, both presented in Chapter 3. This validation aimed to assess the predictive capability of the numerical model over the range of circumferential deformations relevant for medical compression stockings.

The analytical reference pressure was obtained from the Laplace-based estimations reported in Figure 3.26 in Chapter 3, derived from the experimental tensile response of the fabrics with inlay yarn (IF configuration). The comparison between finite element predictions and analytical estimations was performed using the average interface pressure extracted from the contact surface for the three reference leg perimeters, minimal, average, and maximal, for each compression zone. This allowed evaluation of the model accuracy over the full in-use range of circumferential deformation.

Experimental validation was carried out by comparison with PicoPress[®] measurements reported in Table 3.8 in Chapter 3 for the maximal leg perimeter only, corresponding to the geometry available in the experimental setup. The average interface pressure predicted by the finite element model was compared with the measured pressure values, ensuring consistency between numerical and experimental configurations.

The accuracy of the finite element predictions was quantified using the absolute error (AE) and the mean absolute percentage error ($MAPE$), according to Equation 5.3, where IP_{FE} is the pressure predicted by the finite element model and IP_{ref} corresponds either to Laplace's pressure ($IP_{Laplace}$) or to the sensor's measured pressure (P_{sensor}).

$$AE = |IP_{FE} - IP_{ref}|, \quad MAPE = \frac{|IP_{FE} - IP_{ref}|}{P_{ref}} \times 100 \quad (5.3)$$

5.3 Results

This section presents the numerical results obtained from the hybrid finite element simulations and is divided into two main parts.

The first part focuses on the prediction of the interface pressure generated by the medical compression stocking. It includes the results of the mesh sensitivity analysis, the predicted pressure fields for each compression zone, and the comparison with analytical estimations based on Laplace's law and with experimental measurements presented in Chapter 3.

The second part addresses the prediction of the tensile response of the fabric through the inlay yarn behaviour. It presents the mesh sensitivity analysis for the connector force, the comparison with experimental yarn force data, and the evolution of the inlay yarn force for different leg perimeters. This analysis provides further insight into the mechanical mechanisms governing the circumferential stiffness of the fabric.

5.3.1 Interface pressure predictions

The interface pressure predictions are first analysed to assess the ability of the hybrid finite element model to predict the interface pressure generated in the different compression zones. The different mesh configurations used for the ankle, calf and thigh, reflect the structural variations responsible for the zone-dependent circumferential stiffness of the fabric.

5.3.1.1 Effect of mesh ratio on predicted interface pressure

The mesh sensitivity analysis was first conducted for the ankle configuration and the average leg perimeter, selected as a representative and critical compression condition. The evolution of the average interface pressure was analysed as a function of the mesh-size ratio R_{mesh} , defined as the ratio between the element size of the rigid leg and that of the fabric mesh, as given in Equation 5.2.

Figure 5.8 presents the evolution of the average interface pressure predicted for the ankle zone as the mesh-size ratio is increased, while keeping the fabric discretisation constant. For mesh-size ratios ranging from $R_{\text{mesh}} = 2$ to 8, fluctuations are observed in the predicted pressure values. This variability reflects the sensitivity of the contact solution to the discretisation of the rigid surface for relatively fine rigid meshes. For higher ratios, between $R_{\text{mesh}} = 8$ and 10, the predicted interface pressure progressively approaches the analytical reference obtained from Laplace's law.

In parallel with the pressure predictions, the computational cost was strongly influenced by the mesh-size ratio. For low ratios ($R_{\text{mesh}} = 2$ to 6), the fine discretisation of the rigid leg resulted in a large number of contact elements and led to significantly increased computation times, which complicated both the simulations and the post-processing of the results. As the mesh-size ratio increased, the number of rigid elements decreased and the computation time was substantially reduced, facilitating the numerical analysis.

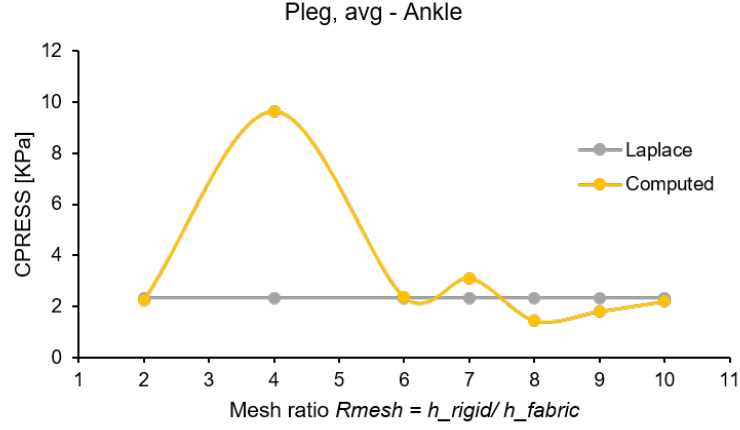


FIGURE 5.8: Evolution of contact pressure (CPRESS) with increasing element size ratio (R_{mesh}) for average leg perimeter in ankle zone.

Based on these observations, the mesh-size ratios $R_{\text{mesh}} = 8$ and 9 were identified as suitable compromises between numerical accuracy and computational efficiency. As shown in Figure 5.8, these ratios slightly underestimate the interface pressure with respect to the analytical reference, but provide sufficiently accurate predictions while ensuring reasonable computation times. The ratio $R_{\text{mesh}} = 7$ was found to remain computationally expensive, whereas $R_{\text{mesh}} = 10$ led to an overly coarse discretisation of the rigid leg, resulting in a slight loss of the circular geometry of the contact surface.

In a second stage of the analysis, the mesh-size ratios $R_{\text{mesh}} = 8$ and 9 were further evaluated for the thigh configuration and for the three reference leg perimeters. The corresponding results are shown in Figure 5.9. This analysis aimed to assess the consistency of the pressure predictions across different compression zones and deformation levels. The results indicate that the most stable evolution of the interface pressure is obtained for $R_{\text{mesh}} = 9$, which provides predictions closest to the analytical Laplace estimations.

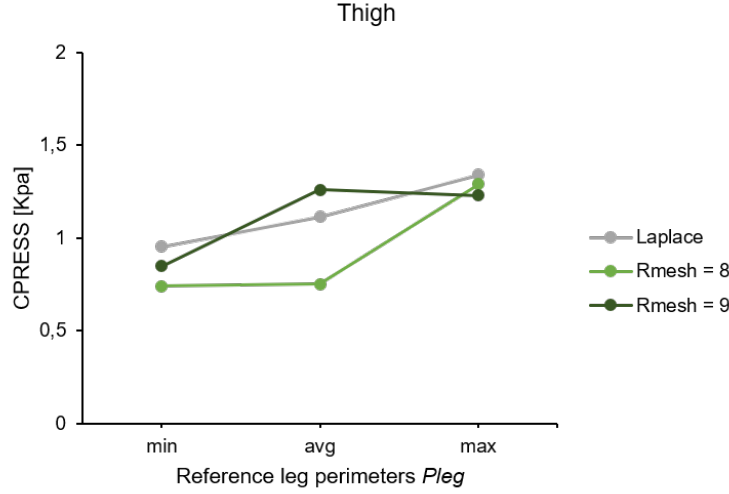


FIGURE 5.9: Evolution of contact pressure ($CPRESS$) for two element size ratios ($R_{mesh} = 8, 9$) for the three reference leg perimeters, minimal, average and maximal, in thigh zone.

Nevertheless, as illustrated in Figure 5.9, even for $R_{mesh} = 9$, the predicted interface pressure does not always exhibit an increase with increasing leg perimeter. This behaviour highlights the intrinsic difficulty of contact modelling in the presence of large deformations and complex textile architectures, and illustrates the sensitivity of the solution to the discretisation choices.

Despite these limitations, the mesh configuration corresponding to $R_{mesh} = 9$ was selected for the full set of simulations. This ratio provides a suitable compromise between numerical accuracy and computational efficiency, while providing interface pressure predictions that remain close to the analytical reference values.

5.3.1.2 Predicted interface pressure for different compression zones

The finite element results obtained for the average leg perimeters of the three anatomical regions of the leg, ankle, calf, and thigh, are presented in Figure 5.10. The figure shows the contact pressure distributions computed on the rigid leg surface at the end of the simulations. Each map corresponds to the final equilibrium configuration reached after the placement of the fabric over the leg model.

As observed in Figure 5.10, the contact pressure is localised in the region where the fabric is in contact with the leg at the end of the simulation. This pressure distribution is consistent with the simulated placement procedure illustrated in Figures 5.5 and 5.6, and reflects the correct definition of the boundary conditions, contact formulation, and evaluation region adopted in the numerical model.

The pressure maps also exhibit a non-uniform spatial distribution of the contact pressure within the contact zone. Such spatial variability of the contact pressure is commonly reported in numerical simulations of compression garments and textile–body interactions, particularly in the presence of large deformations and frictionless contact formulations, and motivates the use of averaged pressure values for quantitative comparison [69], [87], [105]. This variability reflects the sensitivity of the contact solution to local discretisation and contact interactions in complex textile architectures.

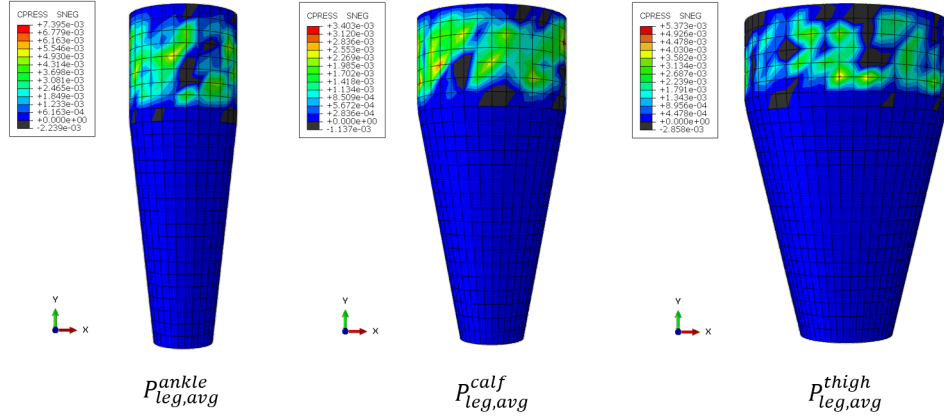


FIGURE 5.10: Numerical pressure maps resulting from the simulations for the average leg perimeters of the three compression zones, ankle, calf, and thigh. CPRESS is given in MPa.

In the following sections, the interface pressure predicted for the different leg perimeters and compression zones is presented and compared with analytical estimations and experimental reference data from Chapter 3, to assess the predictive capability of the hybrid finite element model.

5.3.1.3 Comparison with reference pressure data

The finite element predictions of average interface pressure were assessed by comparison with analytical Laplace estimations and experimental PicoPress® measurements reported in Chapter 3.

Figures 5.11 and 5.12 compare the average interface pressure predicted by the finite element model with the analytical Laplace estimations for the minimal and average reference leg perimeters, respectively. Results are shown for the three compression zones, ankle, calf, and thigh.

For both leg perimeters, the finite element model reproduces the zone-dependent pressure levels predicted by Laplace’s law, with the highest pressures obtained in the ankle zone and lower pressures in the calf and thigh zones. This trend reflects the variation in circumferential stiffness between zones induced by differences in the knitted structure and by the

local density of inlay yarns, and indicates that the model captures the relationship between circumferential deformation and pressure generation.

The quantitative agreement between finite element predictions and analytical estimations depends on both the compression zone and the leg perimeter, as summarised in Table 5.2. The ankle zone exhibits the lowest errors overall, with negligible absolute errors for the minimal leg perimeter and mean absolute percentage errors below 13% for the maximal leg perimeter. This behaviour is associated with the higher structural homogeneity of this zone and the reduced influence of geometric effects.

Larger discrepancies are observed for the calf zone, for which the mean absolute percentage error ranges from approximately 28% to 36% depending on the leg perimeter. The calf region is characterised by a larger curvature radius, that increase the sensitivity of the contact formulation and lead to a moderate underestimation of the interface pressure by the finite element model. The thigh zone exhibits an intermediate and more stable behaviour, with mean absolute percentage errors generally below 15% for the three leg perimeters.

Figure 5.13 presents the comparison between finite element predictions, analytical Laplace estimations, and experimental PicoPress® measurements for the maximal reference leg perimeter. The relative deviation between numerical and experimental average interface pressure amounts to 14.2% for the ankle zone, 26.4% for the calf zone, and 13.5% for the thigh zone. These values remain within the range typically reported for experimental validation of compression garments and reflect the combined effects of contact sensitivity and geometric variability.

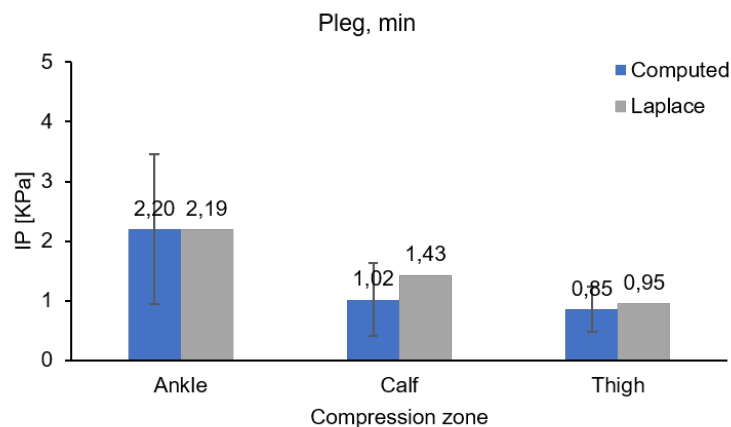


FIGURE 5.11: Comparison of interface pressure (IP) at minimal leg perimeter ($P_{leg,min}$) obtained from finite element simulations (computed) and Laplace's calculations, for the three compression zones zones.

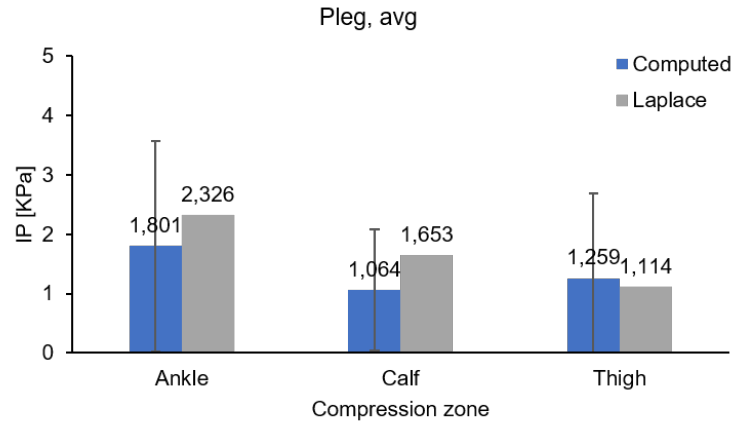


FIGURE 5.12: Comparison of interface pressure (IP) at average leg perimeter ($P_{leg,avg}$) obtained from finite element simulations (computed) and Laplace's calculations, for the three compression zones zones.

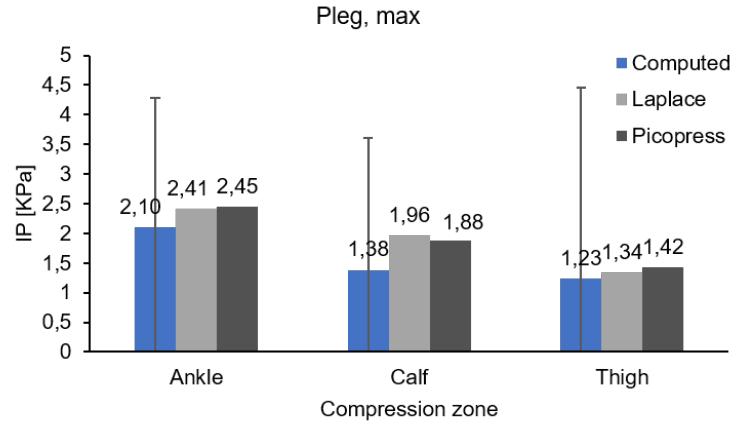


FIGURE 5.13: Comparison of interface pressure (IP) at minimal leg perimeter ($P_{leg,max}$) obtained from finite element simulations (computed), Laplace's calculations, and PicoPress® measurements, for the three compression zones zones.

TABLE 5.2: Absolute error (AE) and mean absolute percentage error (MAPE) between finite element predictions and Laplace analytical estimations for the three compression zones and reference leg perimeters.

Zone	$P_{leg, min}$		$P_{leg, avg}$		$P_{leg, max}$	
	AE [KPa]	MAPE [%]	AE [KPa]	MAPE [%]	AE [KPa]	MAPE [%]
Ankle	0.008	0.365	0.525	22.571	0.31	12.847
Calf	0.412	28.3135	0.589	35.632	0.581	29.628
Thigh	0.106	11.111	0.145	13.016	0.11	8.221

Overall, the agreement between the finite element predictions, the analytical estimations,

and the experimental measurements is within the range reported in the literature for numerical models of medical compression garments, and in several cases exhibits improved accuracy. Reported mean absolute percentage errors typically are between 20% and 40%, depending on the modelling assumptions and the contact formulation adopted [24], [69], [87], [105].

The present results demonstrate that the hybrid finite element model is able to represent the evolution of the compression levels required for therapeutic compression across the different zones of the stocking, by explicitly accounting for the mechanical contributions of the loop structure and the inlay yarn reinforcement, and by integrating the local variation in density of inlay yarn.

5.3.2 Yarn behaviour predictions

The tensile response of the inlay yarn was analysed through the connector behaviour predicted by the hybrid finite element model. The axial force was extracted from the connector total force output variable ($CTF1$) and averaged over all connector elements distributed along the fabric length. This average value represents the equivalent tensile force carried by a single inlay yarn, corresponding to one course of the knitted structure.

5.3.2.1 Effect of mesh ratio on yarn response

The influence of the mesh-size ratio on the predicted yarn response was first evaluated to verify the numerical stability of the connector-based formulation. Table 5.3 reports the average connector force (CTF) obtained for three mesh-size ratios ($R_{\text{mesh}} = 7, 8, 9$), for the ankle zone and for the three reference leg perimeters.

The results show that the predicted connector force is essentially independent of the mesh-size ratio. For all leg perimeters, the variations in $CTF1$ remain within the numerical scatter associated with the standard deviation. This confirms that the predicted tensile response of the inlay yarn is not affected by the discretisation of the rigid leg surface.

TABLE 5.3: Connector force ($CTF1$) for different mesh ratios ($R_{\text{mesh}} = 7, 8, 9$) for ankle zone and three reference leg perimeters, minimal ($P_{\text{leg, min}}$), average ($P_{\text{leg, avg}}$), and maximal (P_{leg}), in the ankle zone.

R_{mesh}	$CTF1 (P_{\text{leg, min}}) [\text{N}]$	$CTF1 (P_{\text{leg, avg}}) [\text{N}]$	$CTF1 (P_{\text{leg, max}}) [\text{N}]$
7	0.0578 ± 0.0021	0.0636 ± 0.0023	0.0692 ± 0.0025
8	0.0577 ± 0.0021	0.0635 ± 0.0023	0.0690 ± 0.0025
9	0.0576 ± 0.0021	0.0634 ± 0.0023	0.0690 ± 0.0025

Based on these results, the mesh-size ratio $R_{\text{mesh}} = 9$ was retained for the remainder of the simulations. This ensured consistency with the interface pressure predictions and an optimal compromise between numerical accuracy and computational efficiency.

5.3.2.2 Predicted tensile yarn response

The tensile response of the inlay yarn predicted by the finite element model was first validated by comparison with the experimental force–strain curve of the yarn, which was identified experimentally and used as input for the hybrid mechanical model described in Chapter 4. Figure 5.14 compares the numerical and experimental yarn responses.

In Figure 5.14, the deformation levels corresponding to the reference leg perimeters of each compression zone are indicated. A very good agreement is observed over the entire deformation range, confirming that the connector-based formulation accurately reproduces the tensile behaviour of the inlay yarn. These results validate the mechanical consistency of the connector-based representation of the inlay yarn under circumferential deformation.

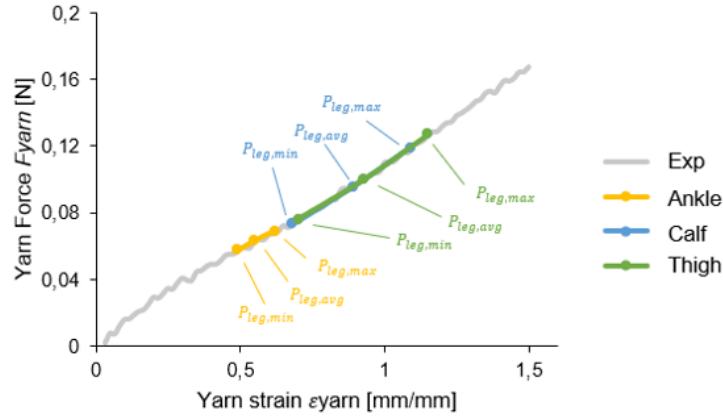


FIGURE 5.14: Comparison between experimental and computed tensile yarn response, for the different compression zones.

Figure 5.15 presents the average tensile force predicted for a single inlay yarn for each compression zone and for the three reference leg perimeters. For all zones, the yarn force increases with leg perimeter, reflecting the increase in circumferential deformation imposed by the leg.

For a given leg perimeter, the tensile force carried by a single inlay yarn is lowest in the ankle zone and increases in the calf and thigh zones. This trend is consistent with the force–strain behaviour shown in Figure 5.14 and reflects the smaller circumferential deformation experienced by the fabric in the ankle region.

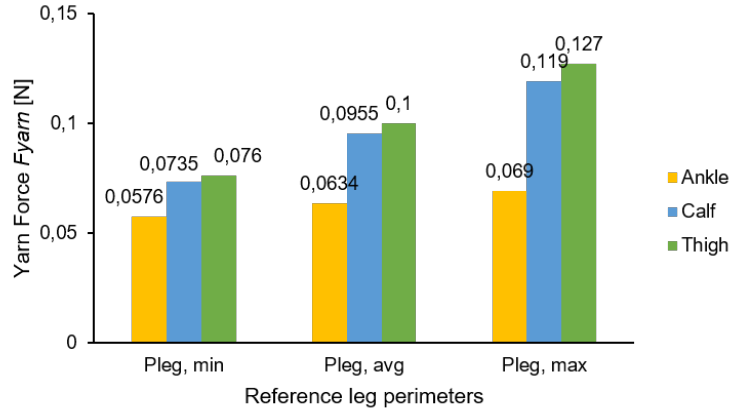


FIGURE 5.15: Computed inlay yarn tensile response for the three compression zones and reference leg perimeters.

Although a single inlay yarn in the ankle zone is subjected to a lower tensile force, the higher density of inlay yarns in this region leads to a higher overall circumferential stiffness and, consequently, to higher interface pressure levels. These results highlight the importance of accounting for the local density of inlay yarns within the fabric. Although a single inlay yarn in the ankle zone is subjected to a lower tensile force, the higher number of inlay yarns per unit length in this region, represented in the model by a larger number of connector elements, results in a higher overall circumferential stiffness and, consequently, in higher interface pressure levels. This observation is consistent with the interface pressure predictions presented in the previous section.

The combined analysis of the predicted yarn force and interface pressure provides a better understanding of compression generation in medical compression stockings. The results indicate that the compression level in each zone is governed by the combined effect of the yarn tensile behaviour and the local density of inlay yarn reinforcement.

By explicitly representing the inlay yarns as discrete tensile elements, the hybrid finite element model establishes a direct relationship between the experimentally identified mechanical properties of the yarns and the macroscopic compression generated by the fabric. This formulation reflects the influence of structural parameters, such as the density of inlay yarns, on the circumferential stiffness of the fabric and the resulting pressure distribution.

From a modelling perspective, this approach allows compression levels to be predicted from yarn mechanical properties and local structural parameters. Since the model enables independent control of yarn behaviour and local density of inlay yarn, it is particularly relevant for the design of graduated compression profiles, and supports the extension to other compression devices and full-stocking simulations.

5.4 Conclusions and research perspectives

Conclusions This chapter developed a hybrid finite element model to predict the interface pressure generated by medical compression stockings when placed on the leg under in-use conditions. The proposed model transfers the hybrid mechanical description introduced in Chapter 4 into a numerical formulation that explicitly represents the two yarn systems of inlaid weft-knitted fabrics, namely the loop structure and the inlay yarn reinforcement.

The fabric was modelled using a mesoscopic description combining shell elements for the loop structure and axial connector elements for the inlay yarns. This representation preserves the distinct mechanical roles of both constituents and allows the circumferential stiffness of the fabric to be governed directly by local structural parameters, in particular the density of inlay yarns, rather than by homogenised fabric properties. The mechanical behaviour assigned to each element type was identified independently from experimental characterisation.

Finite element simulations reproduced the placement of the stocking over rigid leg geometries defined by reference leg perimeters. These perimeters were selected from the manufacturer size table for the selected product, ensuring the representation of the in-use configuration. The interface pressure was computed from the contact interaction between the fabric and the leg and was validated against analytical estimations based on Laplace's law and experimental PicoPress[®] measurements. The predicted average interface pressures showed good agreement with reference data across compression zones and leg perimeters, with deviations within the range reported in the literature.

In addition to interface pressure prediction, the model enabled direct evaluation of the tensile response of the inlay yarn through the connector forces. This provided an independent validation of the numerical formulation by comparison with experimental yarn data. The results demonstrated that the compression generated by the stocking arises from the combined effect of the tensile behaviour of a single inlay yarn and the local density of inlay yarn reinforcement, accounting for the zone-dependent compression levels.

Overall, the hybrid finite element model developed in this chapter provides a consistent tool to analyse and predict the mechanical behaviour of medical compression stockings. By linking experimentally identified yarn properties, local fabric structure, and interface pressure, the proposed model offers accurate prediction of compression levels and is suitable for guiding the design of knitted compression products.

Perspectives The modelling strategy developed in this chapter opens several perspectives for extending the analysis of medical compression stockings and improving the predictive

capabilities of the numerical approach.

A first perspective concerns the inclusion of deformable leg models. In the present work, the leg was assumed to be rigid in order to isolate the mechanical response of the fabric and to validate the hybrid formulation under controlled conditions. An extension of the model to deformable leg representations, using simplified hyperelastic constitutive laws for soft tissues, would allow investigation of the coupled fabric–tissue interaction. This approach would enable analysis of the redistribution of interface pressure due to tissue compliance and provide a more realistic description of in-use conditions, while preserving the fabric modelling strategy developed in this thesis.

A second perspective relates to the consideration of leg morphologies. The current formulation relies on simplified cylindrical geometries defined by reference leg perimeters, which are appropriate for zone-based analysis and model validation. Extending the model to non-cylindrical and patient-specific leg geometries would allow evaluation of the influence of local curvature, anatomical transitions, and morphological variability on pressure distribution. This extension would provide a direct link between the numerical model and clinical practice, and would support the analysis of sizing strategies and personalised compression solutions.

Finally, an important perspective concerns the development of a reduced implicit formulation to decrease the computational cost of the simulations. The explicit contact-based approach adopted in this chapter provides a detailed description of fabric placement and contact mechanics but remains computationally demanding. An alternative strategy, currently under development, consists in decomposing the simulation into two successive stages: first, a radial extension of the fabric to impose the target circumferential deformation, followed by a relaxation step allowing the fabric to conform to the leg geometry, either rigid or deformable. Figure 5.16 represents the current developments, with the initial configuration of the fabric and leg models, together with the two steps considered for the simulation. Implemented within an implicit framework, this approach would significantly reduce numerical complexity and computation time, while maintaining the hybrid modelling approach proposed in this thesis. Such a formulation would facilitate large parametric studies and the simulation of full-stocking configurations.

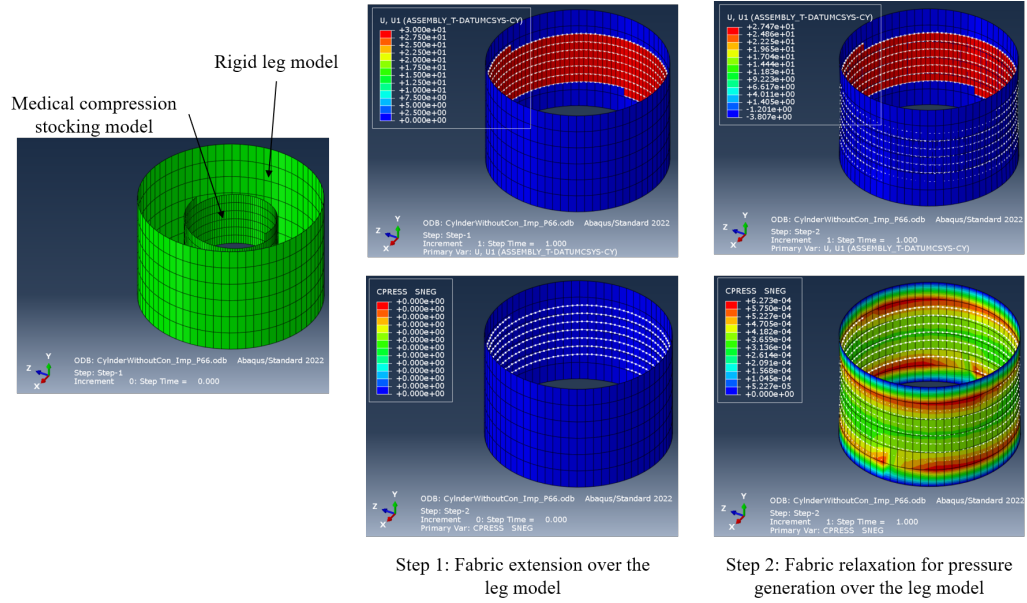


FIGURE 5.16: Implicit simulations under development for contact pressure prediction.

Together, these perspectives outline the extension of the proposed hybrid modelling approach towards more realistic geometries, improved computational efficiency, and broader applicability to the design and analysis of medical compression devices.



Chapter 6

Conclusions, future work and dissemination

6.1 General conclusions of the thesis

Medical compression stockings generate interface pressure as a result of the tension developed in the fabric when it is stretched around the leg. This tension depends on the mechanical response of the knitted structure and on the structural parameters defining the fabric configuration under circumferential deformation. A reliable prediction of compression therefore requires a mechanical description that relates textile structure, tensile behaviour and pressure generation, essential for ensuring stable and reliable pressure levels and for improving the design of compression products.

Previous studies have analysed compression behaviour using experimental measurements, analytical formulations and numerical simulations. These approaches provide useful information on global pressure levels and tensile properties. However, the structural parameters defining the in-use configuration of inlaid weft-knitted fabrics, such as the distribution of inlay yarns and the circumferential deformation imposed by the leg, are not always explicitly considered. This limits the ability to relate the textile architecture to the mechanical response and to the compression delivered in each zone.

The work presented in this thesis addresses this limitation through a combined experimental and modelling approach. Structural measurements obtained under in-use conditions were used to describe the configuration of the fabric in the different compression zones. Tensile tests performed on yarns and fabric samples provided the mechanical data required to analyse the circumferential response. These results were used to develop a local–mesoscopic hybrid model describing the combined contribution of the loop structure and the inlay reinforcement to the fabric stiffness.

The mechanical model was then implemented in a finite element framework to simulate the placement of the stocking on a leg. This numerical description allowed the interface pressure to be obtained from the tension developed in the fabric under circumferential deformation. The predicted pressure distributions reproduced the pressure gradient defined by the structural variations along the stocking and showed agreement with analytical estimations and experimental measurements.

Overall, the work presented in this thesis demonstrates that predicting the compression delivered by medical compression stockings requires a combined experimental, mechanical and numerical approach. Structural and mechanical parameters defined under in-use conditions were shown to be essential for describing the circumferential response of the fabric. The hybrid local–mesoscopic model established a direct relation between textile structure and tensile behaviour, while the finite element simulations enabled this description to be transferred to the stocking–leg configuration.

By validating the predicted interface pressure against analytical estimations and experimental measurements, the thesis demonstrates that compression levels can be predicted from local textile parameters and leg geometry. This integrated approach moves beyond descriptive analyses and provides a predictive methodology for analysing and designing medical compression stockings.

6.2 Scientific contributions of the thesis

The work presented in this thesis contributes to the mechanical modelling of medical compression stockings by addressing several limitations identified in the existing literature. Previous studies often relied on homogenised descriptions of the fabric or on analytical formulations that do not explicitly account for the hybrid architecture of inlaid weft-knitted structures. As a result, the link between textile structure, tensile behaviour and delivered compression remained partially described.

A first contribution of this thesis is the experimental characterisation of the fabric under conditions representative of wear. Structural parameters were measured in in-use configurations, with particular attention to zone-specific inlay yarn density and circumferential deformation imposed by the leg. This approach differs from many existing studies that rely on flat fabric characterisation or global material properties, and it provides a more relevant description of the mechanical state of the fabric during use.

A second contribution is the development of a hybrid local–mesoscopic mechanical model

describing the combined action of the loop structure and the inlay reinforcement. Unlike homogenised membrane models, the proposed formulation distinguishes the mechanical roles of the two yarn systems and relates variations in inlay density to changes in circumferential stiffness. This modelling strategy allows the tensile response of the fabric to be predicted from structural parameters measured experimentally, establishing a direct link between textile architecture and mechanical behaviour.

A third contribution concerns the transfer of the mechanical description to a finite element simulation of the stocking–leg configuration. The integration of the hybrid model into a numerical environment enabled the prediction of interface pressure from the tension developed in the fabric under circumferential deformation. The numerical formulation was validated by comparison with analytical estimations and experimental pressure measurements, demonstrating that the approach provides reliable pressure predictions within the investigated conditions.

6.2.1 Contributions to the Symphonies project

The work carried out in this thesis contributes directly to Task 1.1.2 of the Symphonies project, which focuses on the development of a mechanical simulation model for medical compression stockings. This task aims to provide a numerical description of the fabric behaviour that can be coupled with a digital representation of the lower limb to predict interface pressure.

The thesis provides the structural and mechanical inputs required for this task by delivering measurements obtained under in-use conditions, including zone-specific inlay yarn densities and tensile responses of yarns and fabric samples. These data form the basis for defining the mechanical behaviour of the stocking fabric within the simulation chain of the project.

The hybrid local–mesoscopic model developed in this work constitutes the mechanical component required for representing the inlaid weft-knitted fabric in the Symphonies framework. The model links structural parameters to tensile behaviour and can be transferred to a finite element environment. Its integration into numerical simulations of the stocking placed on a leg enables the prediction of interface pressure in each compression zone.

By establishing a consistent link between textile structure, mechanical response and pressure prediction, the thesis directly supports the modelling strategy defined in the Symphonies project. The developments presented provide the textile mechanical component required to couple the behaviour of the stocking with the biomechanical models of the lower limb developed in Task 1.2 and with the design and integration tools developed in Task 1.3.

The results enable the analysis of how modifications in textile parameters, such as inlay yarn density or material substitution, affect the mechanical behaviour of the stocking and the resulting interface pressure. In this context, the work contributes to the development of a simulation-based design approach for medical compression stockings, supporting product optimisation and adaptation to different morphologies and therapeutic objectives.

6.3 Research perspectives

The developments presented in this thesis open several perspectives for extending the proposed methodology and improving the modelling of medical compression products.

A first perspective is the extension of the approach to other compression products, textile architectures and material choices. The experimental characterisation and the hybrid local-mesoscopic modelling strategy can be applied to different knitted structures, to alternative inlay configurations, and to variations in the constituent yarns of medical compression stockings, including changes in inlay yarns, ground yarns and elastic coverings. Considering different knitting patterns and material combinations makes it possible to analyse the influence of textile design parameters on mechanical behaviour and delivered compression. In this perspective, the numerical tool developed in this thesis can be used not only to analyse existing products, but also to support the optimisation of the materials constituting medical compression stockings, enabling informed selection of yarn types and knitted structures to achieve targeted compression levels.

A second perspective relates to the representation of a complete compression product. In the present work, the modelling and simulations were performed zone by zone, using representative configurations for the ankle, calf and thigh regions. Extending the approach to a continuous description of the stocking along its entire length would allow the global mechanical behaviour of the product to be analysed. Such a development would require the interpolation of structural parameters between zones and the integration of spatially varying properties within the numerical model. This perspective would enable a more realistic simulation of graduated compression and facilitate the analysis of transitions between compression zones.

A third perspective concerns the coupling of the textile model with deformable leg models. In the present work, the leg was represented as a rigid geometry. Introducing deformable simplified parts would allow the interaction between the fabric and compliant tissues to be analysed. Such simulations would provide a better representation of the mechanical interaction between the stocking and the leg and offer insight into the redistribution

of pressure induced by tissue deformation. This coupling represents an intermediate step towards more advanced biomechanical simulations while maintaining a controlled level of numerical complexity.

A fourth perspective concerns the extension of the simulations to different leg morphologies. The modelling strategy developed in this thesis could be combined with leg geometries representative of various morphotypes. This would allow the analysis of how variations in leg shape and perimeter influence the circumferential deformation of the fabric and the resulting pressure distribution. Such an extension would support the study of inter-individual variability and contribute to the design of compression products adapted to specific morphologies.

In addition, future developments could address the use of implicit numerical formulations for the simulation of the stocking–leg interaction. While the present work relied on an explicit contact-based approach, an implicit formulation could significantly reduce computational cost while preserving the hybrid textile description. Such an approach could be based on a different simulation strategy, in which the circumferential deformation of the fabric is imposed prior to its relaxation onto the leg geometry, either rigid or deformable. This extension would facilitate large parametric studies, the simulation of full-stocking configurations, and the integration of the model into design-oriented numerical tools.

6.4 Scientific dissemination

The work presented in this thesis has been disseminated in the following scientific journals and conference contributions, presented in reverse chronological order:

Scientific journals

- I. Pita Miguelez, A. R. Labanieh and D. Soulat, “Hybrid finite element model for predicting the interface pressure of medical compression stockings,” *International Journal for Numerical Methods in Biomedical Engineering*, (in press) DOI: [10.1002/cnm.70151](https://doi.org/10.1002/cnm.70151)
- I. Pita Miguelez, A. R. Labanieh, C. Mesmin, D. Dupuy, and D. Soulat, “Development of a hybrid model of the tensile behaviour of weft-knitted structures for medical compression stockings,” vol. 20, p. 15 589 250 251 352 034, Jul. 2025, ISSN: 1558-9250. *Journal of Engineered Fibers and Fabrics*. DOI: [10.1177/15589250251352034](https://doi.org/10.1177/15589250251352034)

Conferences with proceedings

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Glossary

List of Definitions

Compression	Pressure exerted by the stocking on the limb.
Compression class	Range of pressure delivered by the product at the ankle level.
Durability	Ability of the stocking to maintain its pressure properties over time.
Elastic yarn	Yarn that elongates under the application of a force and returns to its initial length after the removal of that force.
Inlay yarn (weft yarn)	Additional elastic yarn inserted between the loops of a knitted fabric during knitting, without forming loops or binding stitches.
Application dimension	Circumference or height at a specific point of the limb on which the stocking is applied.
Pressure profile	Representation of the pressures exerted by the stocking along the length of the limb.
Decreasing pressure (graduation)	Residual pressure measured at a point compared with the pressure measured at the ankle reference point, expressed as a percentage.
Linear density (count)	Mass per unit length, expressed in tex or its multiples or sub-multiples (e.g., dtex). One dtex equals 1 g/km.
Covered yarn	Yarn obtained by regular helical winding of a filament (the covering) around another filament (the core).
Double-covered yarn	Covered yarn obtained using two covering filaments wound in opposite directions.
Gauge	Number of needles per unit length on a knitting machine.

Knitting machine	Machine that produces a fabric composed of loops, called a knitted fabric.
Circular knitting machine	Knitting machine with a cylindrical design allowing the production of tubular knitted fabrics.
Flat knitting machine	Knitting machine with a flat-bed design allowing the production of flat knitted fabrics.
Therapeutic zone	Part of the stocking intended to cover the lower limb area to be treated by compression.
Terminal part	Part of the stocking outside the therapeutic zone, intended either to maintain the product on the limb or to close it, for example, at the toe or foot end.
Pitch	Distance between two courses of the same inlay yarn, given in mm.

List of Abbreviations

MCS	M edical C ompression S tocking
CVI	C hronical V enous I nsufficiency
IP	I nterface P ressure
MRI	M agnetic R essonance I mage
IF	I nlaïd weft-knitted fabric/ W eft-knitted fabric with inlay yarn
GF	W eft-knitted fabric without inlay yarn
MAE	M ean A bsolute E rror
MAPE	M ean A bsolute P ercentage E rror
AE	A bsolute E rror
WErr	W eighted E rror
AFNOR	A ssociation F rançaise de N ORmalisation
ISO	I nternational O rganization for S tandardization
FE	F inite E element

Glossary of Terms

Reference systems

(E_1, E_2, E_3)	Global Cartesian reference system for fabric flat configuration: E_1 longitudinal direction, E_2 circumferential direction, E_3 thickness.
(e_1, e_2, e_3)	Local unit cell reference system for fabric flat configuration: e_1 wale direction, e_2 course/inlay yarn direction, e_3 normal.
α [rad]	Inclination angle between global and local cartesian reference systems.
α_0 [rad]	Initial inclination angle between global and local cartesian reference systems.
(e_z, e_θ, e_r)	Cylindrical reference system for tubular configuration.

Fabric dimensions

W_0 [mm]	Initial fabric width along E_2 in flat configuration.
W [mm]	Deformed fabric width along E_2 during tensile test.
P_0 [mm]	Initial fabric perimeter: $P_0 = 2W_0$.
P [mm]	Deformed fabric perimeter: $P = 2W$.
P_{leg} [mm]	Leg perimeter defining the deformed fabric perimeter ($P = P_{\text{leg}}$).
$P_{\text{leg,min}}, P_{\text{leg,avg}}, P_{\text{leg,max}}$ [mm]	Minimal, average and maximal reference leg perimeters.
$\varepsilon_{\text{circ}}$ [mm/mm]	Circumferential fabric strain: $\varepsilon_{\text{circ}} = (P - P_0)/P_0$.

Fabric structure

d [mm]	Interline reference distance, considered constant: $d = 30\text{mm}$.
NC	Number of courses between knitted reference lines.
D_{IY} [mm ⁻¹]	Density of inlay yarn.

Fabric tensile testing

L [mm]	Fabric length along E_1 , considered constant: $L = 30\text{mm}$.
C [mm]	Clamp-to-clamp distance along E_2 .
C_0 [mm]	Initial clamp distance: 60 mm ankle, 50 mm calf/thigh).

ϕ_{pin} [mm]	Pin diameter: 6.05 mm.
u_2 [mm]	Imposed displacement along circumferential direction E_2 .
$F_{2\text{-total}}$ [N]	Total measured force along circumferential direction E_2 .
F_2 [N]	Force along E_2 in one fabric layer.
T [N/mm]	Fabric circumferential tension: $T = F_2/L$.

Yarn tensile testing

YL [mm]	Deformed yarn length.
YL_0 [mm]	Initial yarn (gauge) length: $YL_0 = 100$ mm.
$\varepsilon_{\text{yarn}}$ [mm/mm]	Yarn axial strain.
F_{yarn} [N]	Measured yarn force.

Unit cell geometry

h [mm]	Unit cell height along e_1 , $h = 1/D_{\text{TY}}$.
l_{spring} [mm]	Spring length along e_2 .
$l_{\text{spring},0}$ [mm]	Initial spring length along e_2 : $l_{\text{spring},0} = h$
t [mm]	Unit cell thickness along E_3 and e_3 .
A_1 [mm ²]	Unit cell cross-section normal to e_1 : $A_1 = l_{\text{spring},0} \cdot t$.
A_2 [mm ²]	Unit cell cross-section normal to e_2 : $A_2 = h \cdot t$.
N_{length} [-]	Number of unit cells along the fabric length (E_1): $N_{\text{length}} = L/h$.

Unit cell forces: global system

F_1^{cell} [N]	Unit cell force along E_1 .
F_2^{cell} [N]	Unit cell force along E_2 .

Unit cell forces: local system

f_1^{cell} [N]	Unit cell force along local wale direction e_1 .
f_2^{cell} [N]	Unit cell force along local course direction e_2 .
f_1^{cont} [N]	Continuous membrane force along e_1 .
f_2^{cont} [N]	Continuous membrane force along e_2 .
f^{spring} [N]	Axial spring force along e_2 .

Unit cell stress and strain

σ_{11} [MPa]	Elastic stress along e_1 .
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σ_{22} [MPa]	Elastic stress along e_2 .
LE_{11} [mm/mm]	Logarithmic strain along e_1 .
LE_{22} [mm/mm]	Logarithmic strain along e_2 .
$\varepsilon_{\text{spring}}$ [mm/mm]	Spring axial strain.
Δl_{spring} [mm]	Spring axial displacement.

Mechanical parameters

E [MPa]	Young's modulus.
ν [-]	Poisson ratio.
k [N/mm]	Spring stiffness.



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Abstract

Knitted structures used in medical compression stockings exhibit complex mechanical behaviour under large deformations, due to the reorientation of yarns and the interaction between looped ground yarns and inlay yarn reinforcement. This non-linear response makes the prediction of compression levels challenging and requires modelling approaches able to account for the fibrous architecture of the material. At the macroscopic scale, existing numerical approaches for textile materials, mainly developed for woven fabrics, rely on hybrid modelling strategies that combine continuous finite elements to describe in-plane deformation and discrete elements to represent tensile behaviour. While these approaches have been successfully applied to woven structures, their extension to knitted fabrics, and in particular to inlaid weft-knitted fabrics used in medical compression stockings, remains limited. The objective of this thesis is to develop a finite element simulation tool for the numerical design of medical compression stockings, accounting for variations in interface pressure associated with patient morphologies. This work combines experimental characterisation of representative medical compression fabrics, with the development of a hybrid finite element model explicitly accounting for the respective roles of the loop structure and the inlay yarn. The proposed approach provides an alternative to current homogenised models and supports the numerical design and optimisation of medical compression stockings.

Title : Experimental analysis and numerical models of the mechanical behaviour of knitted fabrics for Medical Compression Stockings

Keywords: Mechanical behaviour, Constitutive modelling, Finite element simulation, Knitted structures, Medical compression stockings

Résumé en français

Les structures tricotées utilisées dans les bas de compression médicale présentent un comportement mécanique complexe sous grandes déformations, en raison de la réorientation des fils et de l'interaction entre la maille et le renfort par fil tramé. Cette réponse non linéaire rend la prédiction des niveaux de compression difficile et nécessite le développement d'approches de modélisation capables de prendre en compte l'architecture fibreuse du matériau. À l'échelle macroscopique, les approches numériques existantes pour les matériaux textiles, principalement développées pour les structures tissées, reposent sur des stratégies de modélisation hybrides, combinant des éléments finis continus pour décrire le comportement en cisaillement plan et des éléments discrets pour représenter le comportement en tension. Bien que ces approches aient été appliquées avec succès aux structures tissées, leur extension aux structures tricotées, et en particulier aux tricotés avec fil tramé utilisés dans les bas de compression médicale, reste limitée. L'objectif de ces travaux est de développer un outil de simulation par éléments finis permettant la conception numérique des bas de compression médicale, en tenant compte des variations de pression d'interface associées aux morphologies des patients. Ce travail combine une caractérisation expérimentale de tricotés représentatifs de bas de compression médicale avec le développement d'un modèle hybride par éléments finis prenant explicitement en compte les rôles respectifs de la maille et du fil tramé. L'approche proposée constitue une alternative aux modèles homogénéisés actuellement utilisés et vise à soutenir la conception et l'optimisation numériques des bas de compression médicale.

Mots clés: Comportement mécanique, Modélisation du comportement, Simulation par éléments finis, Structures tricotées, Bas médicaux de compression