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Mechanical characterization and modeling of prosthetic meshes

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Mechanical characterization and modeling of prosthetic meshes

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presented by

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Abstract

The present thesis is aimed at a mechanical characterization and at a model formulation for prosthetic meshes considering different length scales. The main focus is on providing methodologies, experimental and theoretical, to evaluate aspects of the meshes’ mechanical biocompatibility.

Prosthetic meshes are synthetic network fabrics. They are implanted for soft body tissue support, such as in case of pelvic organ prolapse (POP). POP describes a descensus of the pelvic organs, herniating into the vagina. So far, the pathogenesis of POP is not fully understood as it involves a complex interplay between tissues and structures at different pelvic floor length scales. Surgical techniques for POP repair include native tissue repair and mesh repair. The major objective is to reconstruct the pelvic floor anatomy and relieve the associated symptoms in order to increase the patients’ quality of life.

Due to a simplified FDA approval for these devices in the past, originating from the remarkable success of meshes in hernia repair, there is a large variety of mesh types and precut mesh kits on the market. However, recently the FDA issued a health notification warning of severe complications associated with the transvaginal placement of surgical meshes. Professional societies, such as the IUGA pleaded for a more thoughtful clinical introduction of novel implants. The need for a standardized and comprehensive product description for meshes, also including their mechanical properties was emphasized. There are theoretical and clinical indications that optimized mechanical properties of meshes will lead to an improved mechanical biocompatibility and ultimately will affect the clinical outcome of the treatment.

This thesis is intended to contribute to the development of experimental techniques and protocols to assess the mechanical properties of pelvic floor tissues and prosthetic meshes in a physiologically relevant way. The notion “mechanical biocompatibility” should be sharpened by including different mechanical phenomena at different length scales and a model description to deduce respective criteria, which should be considered for clinical applications and mesh design optimization. These objectives were realized through different studies.

A clinical study was conducted using the aspiration technique, an in vivo technique, to characterize the mechanical behavior of the anterior vaginal wall in women with different
clinical presentations. The new aspiration device and the corresponding data analysis procedures are introduced. Mechanical parameters were evaluated describing the local tissue compliance in cyclic load. Measurements were performed on pre- and postmenopausal women with and without anterior vaginal prolapse. Additionally, for women undergoing native tissue prolapse repair, pre- and postoperative findings were compared. The results show that the evaluated mechanical parameters do not allow to assess significant differences between the mechanical properties of the anterior vaginal wall of women with and without POP. However, differences are observed between the mechanical properties in pre- and postoperative state. Major challenges associated with this transvaginal in vivo measurement technique are described and discussed in context with similar studies reported in the current literature.

An ex vivo experimental study was performed to mechanically characterize mesh explants in a physiologically relevant way. Explants are mesh-tissue complexes after the integration of the mesh in the host tissue, herein derived from a rabbit abdominal wall model. Uniaxial and biaxial experimental protocols and corresponding mechanical parameters, describing the samples’ tangent stiffness at different load levels are proposed. Tests were performed on rabbit abdominal wall native tissue (control) and on explants from two different types of mesh, SPMM (heavy-weight) and Gynemesh M (light-weight, partly absorbable). The explant stiffnesses were compared to the response of native tissue and meshes were ranked according to the corresponding explant stiffnesses in different load cases (uniaxial or biaxial) and at different load levels. The findings show that mechanical parameters describing the sample stiffness as well as a corresponding ranking of meshes depend on the load case (uniaxial or biaxial). Thus, different rankings are obtained from the uniaxial as compared to the biaxial load case. Moreover, stiffnesses depend on the load level at which they are evaluated. These findings demonstrate that an evaluation of the mechanical biocompatibility of prosthetic meshes should be based on an experimental configuration (uniaxial or biaxial tension) which reproduces the expected in vivo conditions of mechanical loading and deformation.

An accordingly designed study was conducted using meshes embedded in an elastomeric (VHB) matrix. It is shown that stiffnesses obtained for this non-biological model system are predictive for the corresponding explant stiffnesses. Moreover it is demonstrated that the uniaxial stiffness of dry mesh is generally not predictive for the corresponding explant stiffness. Mesh elastomer composites are shown to provide reproducible results helping to clarify load case specific mechanisms of respective explants. They are useful for mesh design optimization with respect to improved mechanical properties, prior to validation in animal studies.

Dry Gynemesh M (light-weight, resorbed) was mechanically characterized in uniaxial and biaxial loading conditions. Protocols are presented to evaluate global macroscale phenomena, such as anisotropy, the nonlinear force response, hysteresis and preconditioning effects. Therefore, eight different loading configurations were defined (two loading conditions, uniax-
ial and biaxial, and four material directions). Furthermore, procedures for local deformation analysis and evaluation of corresponding homogenized kinematic measures are described. This local analysis allowed to identify mesh specific phenomena due to mesoscale mechanisms, such as inhomogeneous deformation patterns, edge effects or a structure-specific anisotropy. The findings offer an understanding of the global and local mesh kinematics and suggest to see meshes more as structures than as continua. Contributions of these mesh phenomena to deformation mismatches between mesh and underlying structures, decreasing the meshes’ mechanical biocompatibility at different length scales are discussed and according recommendations for clinical applications are deduced. The presented protocols and data analysis procedures allow to acquire a comprehensive picture of the mechanical behavior of the dry mesh, which can be applied for any kind of prosthetic mesh. In this way, this study might contribute new aspects to the definition of standardized protocols for a mechanical characterization of meshes to be considered for comprehensive mesh product descriptions.

This broad experimental data base was used to develop a structural model of the prosthetic mesh. A representative unit cell was modeled based on the theory of multibody systems. The model geometry was chosen to abstract the knitting pattern in a physical way. The derivation of the respective system equations, the kinematics, the force elements and constitutive laws as well as kinematic and periodic constraints are described. Constitutive laws were defined based on physical considerations and examinations. The 20 parameters determining the force laws, were adjusted to fit the global force response per unit cell and the global and local kinematics provided by the experimental data. A comparison between the experimental data and the model response in eight different loading configurations show excellent descriptive capabilities of the model. Moreover, based on the model observations, preconditioning effects are interpreted as a change of the material mesoscale structure. The model provides a deeper understanding of mesoscale mechanisms behind the macroscopic mechanical response. Therefore, it represents a valuable tool to discuss aspects of mechanical biocompatibility at a lower scale. The level of non-affine deformations of a unit cell is an appropriate mesoscale criterion to quantify new aspects of the mesh’s mechanical biocompatibility, independent of the underlying tissue properties.

From the experiences and the outcomes of these studies, it is concluded that pelvic organ prolapse is a multiscale event. An appropriate treatment, e.g. using surgical meshes, has to consider the pelvic floor mechanics at different length scales. In vivo techniques to assess mechanical properties of pelvic floor tissues are most challenging as they are affected by aspects from all length scales. An appropriate separation of scales has to be well defined and at the same time still physiologically relevant. Nevertheless, it is shown, that a stepwise reduction of the system complexity, allows for relevant and mechanically well-founded contributions to an evaluation of quantitative criteria for an improved mechanical biocompatibility of prosthetic meshes.
Zusammenfassung

Das Ziel der vorliegenden Arbeit ist eine mechanische Charakterisierung und eine Modellformulierung für Netzimplantate unter Berücksichtigung verschiedener Längenskalen. Der Fokus liegt dabei auf der Entwicklung von experimentellen als auch theoretischen Methoden, um die mechanische Biokompatibilität der Netze zu evaluieren.


Diese Arbeit leistet durch verschiedene Studien einen Beitrag zur Entwicklung von experimentellen Techniken und Protokollen, um die mechanischen Eigenschaften von Beckenboden- und Netzimplantaten zu messen und zwar in einer Weise die physiologisch relevant ist. Das Konzept der “mechanischen Biokompatibilität” soll durch die Untersuchung ver-
Zusammenfassung

schiedener mechanischer Phänomene der Netze auf verschiedenen Längenskalen, sowie durch das Einbeziehen von numerischen Modellen konkretisiert werden. Entsprechende Kriterien werden abgeleitet, die dann in die klinischen Anwendungen und bei der Optimierung des Netzdesign einfliessen sollen.


Eine entsprechend gestaltete Studie wurde mit Netzen durchgeführt, die in eine Elastomermatrix (VHB) eingebettet wurden. Die Auswertung der lastfall- und lastniveauspezifischen Steifigkeiten dieses nicht-biologischen Modellsystems zeigen eine gute Vorhersagefähigkeit.


Zusammenfassung


Introduction

The present thesis is aimed at a mechanical characterization and at a model formulation for prosthetic meshes considering different length scales. The main focus is on providing appropriate methodologies, experimental and theoretical, to evaluate aspects of the meshes’ mechanical biocompatibility. Appropriate meaning with respect to the field of application, such as the selection of a case-specific mesh type or mesh design optimization.

1.1 Prosthetic meshes

Prosthetic meshes are network fabrics [1]. They are used for soft body tissue repair, e.g. in abdominal wall or hernia repair, or for pelvic floor reconstruction in case of a pelvic organ prolapse (POP). The present study mainly focuses on the latter application. POP is a weakening of the pelvic floor, described as fascial tear and muscle laxity, followed by a descensus of pelvic organs with downward pressure [2, 3]. Pelvic organs protrude into the vagina, bulging the vaginal wall, which may result in a complete eversion of the vagina [3, 4]. Depending on the prolapsed pelvic floor compartment and the organs involved, POP is classified as anterior vaginal prolapse (herniation of the urethra or bladder), apical vaginal prolapse (herniation of the uterus or the vaginal vault) or posterior vaginal prolapse (herniation of the rectum) [3] (see chapter 2, section 2.2 for more detailed information about POP).

The pathogenesis of POP is not fully understood [6] and apparently complex as both pelvic floor structures and tissues are involved. According to Drutz [3], intrinsic and extrinsic risk factors are differentiated. Intrinsic factors include race, anatomy, connective tissue changes or neurologic abnormalities, and extrinsic factors include pregnancy, childbirth, aging, hormone effects or increases in intra-abdominal pressure. Tissue trauma due to vaginal delivery, in particular forceps delivery, is known to significantly increase the risk for POP [7–9]. The estimated lifetime risk for women to develop a POP is 30-50%, with 2% of women becoming symptomatic [6, 10].

*Parts of these considerations are taken from [5].
Symptomatic POP, although not life-threatening, significantly decreases women’s quality of life due to discomfort or disability in everyday life situations, such as physical or sexual activities. There are conservative non-surgical therapies to “prevent” POP through pelvic floor exercise, or to treat POP by vaginal inserts (pessaries) [3, 11]. Moreover, surgical alternatives exist aiming at (i) a long term reconstruction of the anatomy, (ii) a subjective relief of symptoms and thus at an increase of women’s quality of life.

Surgical repair techniques can be categorized according to the access route, into abdominal or vaginal approaches. There are indications that abdominal procedures are more effective with respect to the clinical outcome and the recurrence rates [12], whereas vaginal techniques are less invasive, cause less morbidity and require shorter hospitalization [3]. Moreover, surgical techniques are differentiated according to the material used for repair, i.e. native tissue or surgical mesh [3, 10, 11]. Native tissue repair, also called traditional tissue repair, has its origins in the late 1950s [13]. According to the clinical indications, patients’ own tissue is used to re-suspend organs within the pelvic cavity (e.g. transvaginal sacrospinous vault suspension or abdominal sacral colpopexy) and/or the vaginal wall is plicated to restore the barrier against herniation (colporrhaphy). Mainly due to the fact, that already weakened pelvic floor structures and tissues are used, recurrence rates are high and scattered [14], reported in the range up to 63% [15, 16].

The use of synthetic meshes represents an alternative, which started with applications for hernia repair, in the late 1960s [17]. In the following 30-40 years, mesh repair was promoted as the preferred method for hernia repair due to the significant decrease in recurrences without an increase of complications when compared to traditional suture repair [18–21]. More recently, synthetic meshes were more and more also applied for prolapse repair, to suspend the organs and support the pelvic floor tissues. Mesh repair was mainly aimed at increasing the longevity of the surgical repair and at providing a substitute supporting structure in case of a severe or a recurrent prolapse [10, 11].

Driven by the remarkable success of hernia meshes and the enthusiasm of the initially promising results with respect to pelvic anatomical reconstruction [22, 23], and encouraged by a simplified FDA approval for substantially equivalent devices, the market was flushed by a large variety of different types of meshes. Large porous and monofilament meshes (for both hernia and prolapse repair) were reported to be superior to small porous, multifilament meshes with respect to infection, pain and recurrence rates [24, 25]. Meshes for prolapse repair were introduced as sheets to be self-tailored by the surgeons [26–28] or, more recently, as kits with precut meshes and tools for a more standardized implantation (Figure 1.1) [22, 23, 29, 30].

Abdominal sacrocolpopexy, more frequently using synthetic mesh strips, has become the gold standard for the treatment of vaginal apical prolapse due to low recurrence rates and the low risk of complications [31–33]. In contrast, transvaginal placement of surgical meshes has lately started to be discussed conversely based on a large number of publications, system-
Prosthetic meshes

Figure 1.1: Mesh kit for a complete transvaginal and minimally invasive pelvic floor reconstruction, a) Schematic view. The mesh system is precut from a sheet. The shape is characterized by planar sheets in the midregion to support the pelvic organs and slender strips to anchor the implant within the bony structure of the pelvis. The mesh system is cut without regarding the varying material directions of the strips. (This figure is a redrawn schematic, based on commercially available mesh kits). b) Mesh system in the implanted configuration, Gynecare Prolift+M* Pelvic Floor Repair System by Ethicon Inc. (a Johnson & Johnson company, Somerville, NJ, United States). Figure with permission of Ethicon.

1.1 Prosthetic meshes

At the time of this writing, surgical mesh is used in multiple clinical settings, including repair of pelvic organ prolapse, stress urinary incontinence, and hernia repair. Despite the widespread use of meshes, the long-term outcomes and complications associated with mesh implants have been a subject of concern. The FDA has been systematically reviewing current trials to evaluate the anatomical and functional outcome of mesh application compared to traditional surgery [34–36]. Although the anatomical outcome is improved, the risk for severe mesh-related complications, such as erosions and contractions, is critically high [26,27,30,34–39]. While data describing the long-term outcome (> 1 year) are still rare [10].

Responding to these facts and due to major concerns arising from a continuously increasing number of entries to the MAUDE (Manufacturer and User Facility Device Experience) database, reporting adverse events associated with the transvaginal placement of surgical meshes, the FDA recently issued a health notification [10]. An accompanied review of the respective literature exposed concerning erosion rates of 10%-20% in a one to three years follow-up. Even more distressing is the fact that, so far, there is no proven benefit of mesh repair with respect to women’s quality of life, when compared to native tissue repair [26,40]. In the absence of sufficient safety and effectiveness of prosthetic meshes, the FDA currently considers a re-classification from class II to class III, requiring premarket approval and scientific review for new devices [41].

Also professional societies, such as the IUGA have pleaded for a more thoughtful clinical introduction of novel implants, which starts with a standardized and comprehensive product
description, also including their mechanical properties [42]. There are theoretical and clinical indications that optimized mechanical properties of meshes will lead to an improved mechanical biocompatibility and ultimately will affect the clinical outcome of the treatment [43–45]. Animal studies are recommended taking into consideration the host response in controlled conditions, in particular also with respect to a mechanical characterization [42].

1.2 Prosthetic mesh mechanics - current status and future challenges

A review of the current literature, which is outlined in greater detail in the subsequent chapters of this thesis, shows that there is an increasing interest in:

- Reference data for mechanical properties of human pelvic floor structures and tissues [46–67],
- In vivo methods to (i) objectively quantify mechanical properties of human pelvic floor tissues in physiologically relevant conditions, (ii) evaluate the long term follow-up of a treatment, (iii) diagnose predispositions to POP based on the mechanical properties of pelvic floor tissues and structures [46–51, 61–63, 65–67],
- Structural mechanical properties of explants from animal models, to evaluate the structural mechanical biocompatibility of meshes [43, 68–77],
- Structural mechanical properties and mechanical phenomena of dry meshes in order to compare and categorize different meshes [68, 73, 78–92].

Given the broad variety of studies, there is a need for the definition of standards, in particular with respect to a description of the mechanical behavior of prosthetic meshes. Moreover, there is a demand for standardized criteria to evaluate the mechanical biocompatibility of different mesh types and to categorize them accordingly. In detail, there is a need for:

- Standardized devices and methodologies for an in vivo mechanical characterization of human pelvic floor tissues and structures [42],
- Standardized experimental procedures and structural mechanical measures and parameters to describe, compare and categorize the mechanical response of explants from animal models [42, 93],
- An agreement on physiologically relevant loading conditions,
- Standardized experimental procedures, mechanical measures and parameters for dry meshes to be included in a comprehensive product description [42, 88],
- A discussion on the relevance and impact of different mechanical mesh phenomena, such as the nonlinear force response, anisotropy, time-dependence and load history dependence,

- A discussion on the relevance and impact of mechanical phenomena at different length scales, (so far only the macroscale has been considered).

1.3 Structure of the thesis

The present thesis is intended to contribute to these fields of current research. The prosthetic mesh *Gynecare Gynemesh M* (Ethicon Inc., a Johnson & Johnson company, Somerville, NJ, United States) is characterized, which is an approved mesh for hernia repair (known as *Ultrapro*). The current classification of prosthetic meshes (class II) allows to introduce sufficiently equivalent devices into the market without premarket approval. Therefore, *Gynecare Gynemesh M* (in the following called *Gynemesh M*) is currently available for gynecological applications to be self-tailored or as mesh kit, named *Gynecare Prolift+M*\(^1\) (Figure 1.1), which has been subject to a recently finished clinical trial [37].

This thesis is structured according to different studies conducted in order to contribute to the identified fields of interest.

**Chapter 2** A clinical study was performed using the aspiration technique, an in vivo technique, to characterize the mechanical behavior of the anterior vaginal wall. As the vaginal wall mechanics is seen as important and representative for pelvic floor tissues, an in vivo characterization is aimed at providing physiologically relevant reference data and experience to discuss aspects of mechanical biocompatibility for prosthetic meshes. The new device and the corresponding data analysis procedures are presented. Measurements were performed on pre- and postmenopausal women with and without cystocele (prolapse of the bladder). Additionally, for women undergoing traditional prolapse repair, pre- and postoperative findings were compared. Based on the findings, POP is described as a multiscale problem. Hypotheses are formulated on factors at different pelvic floor length scales possibly influencing the present measurements. These considerations contribute to a better understanding of the pelvic floor mechanics.

**Chapter 3** Uniaxial and biaxial ex vivo experimental protocols and corresponding mechanical parameters are proposed to characterize and categorize mesh explants from animal models. Tests were performed on rabbit abdominal wall native tissue and on two different

\(^1\)In 2012, Ethicon stopped selling *Gynecare Prolift+M* Pelvic Floor Repair System, responding to the FDA warning and to a number of lawsuits initiated by women affected by severe complications.
types of explants, i.e. two different meshes (SPMM, Gynemesh M), ingrown into native tissue, gained from a rabbit abdominal wall model (part I). Mechanical parameters were evaluated describing the sample stiffness in uniaxial and biaxial loading conditions and at different load levels. The experimental findings are discussed in context with their physiological relevance and with respect to differences between explant and native tissue behavior.

An accordingly designed study was conducted using meshes embedded in an elastomeric matrix (part II). Uniaxial and biaxial stiffness parameters obtained from this non-biological model system are evaluated with respect to their predictive capabilities for corresponding explant stiffnesses.

Chapter 4 Uniaxial and biaxial protocols are presented to evaluate global macroscale mechanical phenomena of dry Gynemesh M, such as anisotropy, the nonlinear force response, hysteresis and preconditioning effects. Furthermore, procedures for local deformation analysis and evaluation of corresponding homogenized kinematic measures are described allowing to identify mesh specific phenomena due to mesoscale mechanisms. The findings are shown to offer an understanding of the global and local mesh kinematics and allow to see the mesh more as a structure than as a continuum. This study might contribute new aspects to the definition of standardized protocols to be considered for comprehensive mesh product descriptions.

Chapter 5 A structural model of a representative unit cell of the knitted prosthetic mesh, Gynemesh M, is proposed based on the theory of multibody systems. The derivation of the respective system equations, the kinematics, the force elements and constitutive laws as well as kinematic and periodic constraints are described. Constitutive laws were defined based on physical considerations and examinations. The 20 parameters determining the force laws, were adjusted to fit the global force response per unit cell and the global and local kinematics gained from the experimental data (chapter 4). The level of non-affine deformations of a unit cell is shown to be an appropriate mesoscale criterion to quantify the mesh’s mechanical biocompatibility, independent of the underlying tissue properties.

Following the structure of this thesis, the complexity of the materials and structures to be characterized is stepwise decreased - from vaginal wall to mesh explants to dry meshes. Simultaneously, the mechanical descriptions and the applied theoretical approaches stepwise become more elaborate - with global scalar parameters used to assess the in vivo measurements to homogenized stiffness values for explants to mesoscale mechanistic representations of mesh filaments and junctions in dry meshes.
In vivo mechanical characterization
of the anterior vaginal wall
using the aspiration technique

**Motivation**  This study is aimed at in vivo measurements providing mechanical parameters of pelvic floor tissues and at finding correlations to the stage of POP.

**Methods**  A clinical study was performed using the aspiration technique, an in vivo technique, to characterize the mechanical behavior of the anterior vaginal wall. The new device and the corresponding data analysis procedures are presented. Measurements were performed on pre- and postmenopausal women with and without cystocele. Additionally, for women undergoing traditional prolapse repair, pre- and postoperative findings were compared.

**Results**  The results show that with respect to the evaluated mechanical parameters no significant differences can be assessed between the mechanical properties of the anterior vaginal wall of women with and without POP, however between the mechanical properties in pre- and postoperative state.

**Main conclusions**  Major challenges finding an appropriate reference configuration and reproducible boundary conditions are described and discussed in context with in vivo studies in the current literature. This study contributes to a more conscious and systematic future development of devices for an in vivo assessment of the pelvic floor mechanics.

*Parts of this chapter, including paragraphs of text, figures and tables are published in [94].*
2.1 Introduction

Pelvic organ prolapse (POP) is a change of the pelvic floor anatomy [3, 4] which is related to histological changes of pelvic floor tissues [95]. Histological changes might represent either the reason for the pelvic floor disorder [96, 97] or the result of a remodeling process caused by the pelvic floor disorder [98]. From a mechanical point of view, changes in both pelvic floor structural and material properties are linked with POP.

The role of the vaginal wall and of its mechanical properties in this context is a topic of discussion. POP can be described as a herniation of pelvic organs into the vagina. The vaginal wall represents a support to protect the structural integrity of the pelvic organs [53, 54, 59]. Hence, vaginal laxity, i.e. an increased compliance or inelasticity, might result in pelvic floor disorders or, in case of a native tissue repair, in a higher risk for recurrent POP. In fact, elastic fiber degradation in the vaginal wall has been shown to be associated with the appearance of POP [96, 97]. Interestingly, Gilchrist et al. [59] reported no correlation of ex vivo mechanical properties of the anterior vaginal wall and the rate of prolapse recurrence. Other groups consider the mechanical properties of the vaginal wall as representative for pelvic floor connective tissues in general without allocating the vaginal wall a particular supportive function [56–58].

In several studies, the mechanical behavior of pelvic floor structures [46–51] and tissues, i.e. vaginal wall [52–67] has been investigated in order to elucidate the complex chain of anatomical and histological events associated with POP. Ex vivo uniaxial tensile tests have been reported, using strips of prolapsed and less frequently non-prolapsed vaginal wall tissue excised during surgery from pre- and postmenopausal women [52–54, 56–60] or excised from female cadavers [55]. Correlations have been found with respect to the menopausal state, relating an increased stiffness to prolapsed postmenopausal tissue [56–58] and increased relaxation to non-prolapsed postmenopausal tissue [53], as compared to corresponding premenopausal tissue. Apart from [56], no correlations of ex vivo mechanical properties and POP have been found [54] or reported, often due to a non-sufficient sample size [60]. In fact, the limited availability of non-prolapsed human vaginal wall tissue restricts such ex vivo investigations [52, 55]. Another limitation of these studies is the apparent minor relevance of uniaxial tensile tests, as the complex in vivo boundary conditions resulting from the integration of the vaginal wall within the pelvic floor structures are not reproduced appropriately [52, 55].

Therefore, there is an increasing interest in an in vivo mechanical characterization of pelvic floor structures and tissues. Non-destructive in vivo studies are expected to represent valuable tools for POP diagnosis and prevention [63, 67]. Based on the mechanical properties of pelvic tissues, the risk for a recurrent prolapse shall be reduced through case specific treatment (native tissue or prosthetic mesh repair) or a more conscious selection of prosthetic
materials [52,54,55]. Moreover, a longterm follow-up observation of mechanical parameters should help to evaluate the surgical outcome [65,67].

Based on MRI, thus non-interventionally, structural properties, such as anterior compartment compliance [50] or strain dependent anterior vaginal wall position [51] have been correlated with the degree of anterior vaginal wall prolapse. Furthermore, for patients suffering from stress urinary incontinence, interventional, speculum-like, compressible and expandable devices are described for transvaginal application which are used to assess the active [47–49] and passive [46] vaginal closure forces. So far, comparable devices have not been applied to POP patients. However, tactile [67] and suction devices [61,63,66] have been proposed for a transvaginal evaluation of the local mechanical properties of vaginal wall tissue in women with and without POP. The in vivo compliance of vaginal wall tissue has been positively correlated with the degree of POP in [67] and in [63], whereas in [62] no such correlation has been observed. All devices have been able to discriminate between pre- versus postoperative states of the vaginal wall in patients treated by different surgical techniques, i.e. before and after mesh application [67], traditional prolapse surgery [62] or sacral colpopexy [65].

In this study, a suction device, called aspiration device, is used to evaluate in vivo mechanical parameters of the anterior vaginal wall. The new device and the corresponding data analysis procedure are described. Measurements were performed on pre- and postmenopausal women with and without cystocele. Moreover, for women undergoing traditional surgical cystocele repair, mechanical parameters are compared pre- and postoperatively. It is hypothesized that the assessed mechanical parameters allow a discrimination between the mechanical properties of the anterior vaginal wall of (i) women with and without cystocele and (ii) in pre- and postoperative state.
2.2 Background

Pelvic organ prolapse (POP) describes a failure of pelvic floor tissues [3]. As a consequence pelvic organs herniate into the vagina. A lateral cut through an intact female pelvic floor is shown in Figure 2.3 a.

![Lateral cut through the female pelvic floor](image1.png)

**Figure 2.1:** Lateral cut through the female pelvic floor. Figure adapted from www.fda.gov/MedicalDevices/ProductsandMedicalProcedures/ImplantsandProsthetics/UroGynSurgicalMesh/ucm262299.htm, 10/2012, with permission.

The most common types of POP are the cystocele (a prolapse of the bladder, Figure 2.2 a), the procidentia (a prolapse of the uterus, Figure 2.2 b) and the rectocele (a prolapse of the rectum, Figure 2.2 c).

![Pelvic organ prolapse](image2.png)

**Figure 2.2:** Pelvic organ prolapse, descensus of pelvic organs and herniation into the vagina, a Cystocele: prolapse of the bladder, b Procidentia: prolapse of the uterus, c Rectocele: prolapse of the rectum. Images printed with permission of Nucleus Medical Media.
The stage of the POP is quantified by means of the POP quantification (POP-Q) system [4] (Figure 2.3 b). The position of six vaginal landmarks relative to the hymenal ring and three geometric dimensions at the full extent of protrusion determine the stage of POP, as follows. The position of the hymenal ring is defined at 0. The position of the most distal point of the prolapse defines its stage. Stage 0 represents the intact pelvic floor, with Aa, Ba, Ap, Bp at -3. Stage I indicates that the most distal point is still proximal to the hymen, i.e. at <-1. Stage II marks a prolapse crossing the hymenal ring, with the most distal point at -1<0<+1. At Stage III the most distal point is at <-+2 and at stage IV it is at +2 or more distal, representing a complete eversion of the vagina [4].

**Figure 2.3:** POP quantification system (POP-Q). a) Six sites: Aa and Ba at the anterior vaginal wall, C cervix, D posterior fornix, Bp and Ap at the posterior vaginal wall, and three measures of length gh genital hiatus, pb perineal body and tvl total vaginal length used for pelvic organ support quantification. The position of all points are measured with respect to the hymen, negative values indicate positions proximal to the hymen, positivs values positions distal to the hymen. b) Total eversion of the vagina, Ba, C and Bp are at the same position (+8), Aa and Ap are maximally distal (+3). Figures reprinted from [4], with permission of the *American Journal of Obstetrics and Gynecology*
2.3 Methods

2.3.1 Study design

The present study is a prospective, non-randomised, controlled, non-blinded and interventional study. It is a collaboration between the Department of Obstetrics and Gynecology at the Zurich University Hospital and the Center of Mechanics at ETH Zurich. The trial has been approved by the cantonal (Zurich) ethics commission (StV 11/2009).

2.3.2 Study course

Voluntary pre- and postmenopausal women with (POP) and without (Control) cystocele which were hospitalized at the USZ for a gynecological intervention (native tissue repair of the anterior vaginal compartment in case of the POP group and hysterectomy in case of the control group) were invited to participate. Women were excluded from the study if they were younger than 18 years, pregnant or breastfeeding. Furthermore, they were excluded if they reported gynecological malignant tumors, carcinoma in the pelvic floor (anal or abdominal), local florid infections, or a medical history such as steroidal treatment or methotrexat treatment (rheumatism). Moreover, women who had already undergone a surgery for pelvic floor disorders were not admitted to this trial. All participants were provided written information about the aims of the present research and the protocol of this trial. Women’s informed consent was required for participation. As part of the anamnesis, women were asked to grade the subjective prolapse related labor by a Visual Analogue Scale (VAS), ranging from 0 (none) to 10 (worst). During subsequent clinical examination, participants were examined manually in dorsal lithotomy position with empty bladder by inserting a speculum. The stage of prolapse was quantified in line with the POP-Q standards [4]. Subsequently, the speculum was removed and women underwent an aspiration measurement and finally a perineal sonography of the vaginal canal. The surgery generally took place on the following day. Clinical parameters, such as age, body height, weight, body mass index (BMI), parity, birth procedure (vaginal birth, vacuum extraction, forceps delivery and cesarean) and the history of hormone treatment (in particular, peri- or postmenopausal hormone replacement therapy (HRT)) or previous surgeries (hernia, varices) were recorded. Pathologies possibly influencing clinical diagnoses, such as laxity of joints, collagene defects or connective tissue defects were documented. During post-operative consultation, women graded their remaining prolapse related labor (VAS). Furthermore, clinical examination, POP-Q assessment and another aspiration measurement were conducted.
2.3.3 Aspiration device

The employed aspiration technique is based on the pipette aspiration principle [99]. The tissue of interest is elevated by a defined vacuum pressure and the resulting tissue displacement is evaluated, quantifying the tissue compliance. The aspiration device has been used in several clinical applications, such as under sterile conditions on the uterine cervix during hysterectomy [100], in open surgeries on the liver [101–103] and with pregnant women to measure the cervix compliance [104]. No complications have been reported, no risks and no harm are to be expected for the participants.

Figure 2.4: Aspiration setup, a The pressure unit and the computer are installed on a trolley. The components of the pressure unit are integrated within the vacuum container (C). b Aspiration probe (AP), c Hydraulic diagram of the aspiration device with the aspiration probe (AP) containing an internal pressure sensor \(p_2\) and the pressure unit including a diaphragm gas pump (P), a vacuum container (C), an internal pressure sensor \(p_1\) and two valves \(V_1\) and \(V_2\). The drawn configuration is the default setting, where the AP is connected to the environment (tissue not loaded).

In this study, a new design specifically developed for vaginal applications has been used [105]. The aspiration setup consists of the measurement probe, a pressure unit and a computer (Figure 2.4 a, b). The aspiration probe (AP) was manufactured by Fiberoptic (Fiberoptic P & P AG, Spreitenbach, Switzerland). It is a slender tube with a modular tip, serving as a vacuum cavity (Figure 2.5). A camera objective and a pressure sensor \(p_2\) are mounted within the cavity. At its distal end, the tip has a lateral aperture shaped as a long-hole (13.5 mm x 8.5 mm). For in vivo measurements, the probe is applied transvaginally.
such that the aperture is placed on the vaginal wall of the patient (Figure 2.6). The pressure within the aspiration cavity, i.e. the load applied to the tissue, is provided by the pressure unit (Figure 2.4 c). Besides electronic components, the pressure unit includes a vacuum container (C), a diaphragm gas pump (P) (Type: NMP 830 KNDC, KNF Neuberger AG, Balter- swil, Switzerland), two valves (V₁, V₂) (Type: VDW21-5G-3-01F-Q, SMC Pneumatik AG, Weisslingen, Switzerland), an internal pressure sensor (p₁) (Type: XFGMC-3025KPGSRH, Pewatron AG, Zurich, Switzerland) and a tubing system. The vacuum container has a capacity of 10l evacuated by the pump to a controlled constant vacuum pressure. Controlled by the valves, the aspiration cavity is either connected to the vacuum container or to the environment, subjecting the tissue to the vacuum pressure or releasing it, respectively. The actuators, the pump and the two valves are computer-controlled, realizing a defined load history \( p(t) \) [N/mm\(^2\)] acting on the tissue. The respective software was programmed with Labview (National Instruments, Austin, Texas, USA). The integrated camera records a side view image sequence of the current tissue configuration at a frame rate of \( \sim 20 \text{s}^{-1} \) and allows a continuous visual control (Figure 2.7 a). The measurement output consists of a continuous pressure signal and a corresponding image sequence, which need further processing, in order to quantify the tissue compliance.

![Figure 2.5: Aspiration probe, a Modular tip with the aspiration opening and a scale, b Probe with the unmounted tip, exposing the aspiration pipe, the optical fiber providing light, the camera objective and the pressure sensor (p₂)](image)

### 2.3.4 Aspiration protocol

In this trial, the aspiration measurements were performed on the anterior vaginal wall, 5cm proximal from the hymenal ring (2cm proximal from POP-Q point Aa [4]). All measurements were done by the same examiner. The vacuum container was evacuated to -25mbar (relative to the atmospheric pressure), a pressure level, which in the preliminary study was qualified to be harmless for the patients and to be high enough to obtain a visible tissue deformation (see section 2.4). The tissue was loaded (-25mbar) and released (0mbar) cyclically by holding each level for 10s. One measurement consisted of ten such cycles, lasting 200s. One measurement
Figure 2.6: Transvaginal application of the aspiration probe (schematic sectional view). The lateral opening is attached to the vaginal wall. The tip cavity is evacuated through the aspiration pipe. The current vacuum pressure is observed by the internal pressure sensor \( (p_2) \) and controlled by the pressure unit. The camera records side view images of the intruding tissue at rest (dashed line) and at load (full line). The tissue elevation describes the difference between these two configurations.

was performed per patient. If possible an additional subsequent measurement was realized, in order to assess the reproducibility of the measurements. For the data analysis, only the first measurement was used. After the measurement, the modular parts, containing the tubing, the aspiration pipe and the tip were mechanically cleaned, disinfected (neodisher Septo MED, Chemische Fabrik Dr. Weigert GmbH & Co. KG, Hamburg, Germany) and sterilized within the autoclave. The body of the probe was cleaned manually and disinfected with ethanol.

2.3.5 Data processing

Programs for data analysis were written with Matlab (The Mathworks, Natick, Massachusetts, USA). The image sequence recorded by the camera documented the current, deformed tissue configuration dependent on the defined pressure history \( p(t) \). The tissue compliance was determined from the load dependent change of the tissue configuration. The current tissue configuration at timepoint \( t [s] \) subject to the current pressure \( p(t) \) was quantified by the two-dimensional, projected area \( A(p(t)) [\text{pixels}^2] \) enclosed by the contour of the tissue bubble. The steps for the extraction of the contour are illustrated in Figure 2.7 a-d. The grayscale (255: white, 0: black) image (Figure 2.7 a) was converted to a black and white (1:
white, 0: black) image (Figure 2.7 b). The conversion threshold value was manually defined as a grayscale value of 25. The pixel value gradient was calculated along radial lines starting in the center of the tissue bubble (Figure 2.7 c). A gradient of -1 marked the transition from white to black (-1=0-1) and thus a point of the contour (Figure 2.7 d).

For the calibration of the camera, two elliptically shaped calibration parts with different, known dimensions (vertical axis: 6mm or 2mm, respectively) (Figure 2.8 a) had been manufactured, mimicking the aspirated tissue bubble at two different states of deformation. The parts were constructed, such that they could be positioned uniquely within the aspiration opening, in the middle of the long axis of the long hole. As shown in Figure 2.8 b, the area enclosed by the contour of the calibration parts was evaluated as done for the tissue. The calibration factor was calculated as the ratio between the known area of the calibration part [mm$^2$] and the measured area enclosed by the contour [pixels$^2$], equal to 7.3e-4 mm$^2$/pixels$^2$.

![Figure 2.7: Contour extraction](image)

2.3.6 Mechanical parameters

In accordance with the literature [63], an intact integration of the vaginal wall within the pelvic floor and an intact microstructure were assumed to be associated with a compact, dense unloaded configuration and a stiff mechanical response. A stiffness-like mechanical parameter was evaluated based on the history of the current tissue configuration $A(p(t))$. The difference between any current tissue configuration $A(p(t))$ and a defined reference
2.3 Methods

Figure 2.8: Calibration of the camera, a Elliptically shaped calibration parts of two sizes (vertical axis: top: 6mm, bottom: 2mm) with positioning system, b The area enclosed by the contour (in pixels\(^2\)) was measured as done for the tissue and related to the known area of the calibration part (in mm\(^2\)).

configuration \( A_0 \) was called tissue elevation \( \Delta A(p(t)) = A(p(t)) - A_0 \), as illustrated in Figure 2.9 a. The cyclic tissue displacement was defined as the maximum tissue elevation per cycle \( i \Delta A = i A_{\text{max}} - i A_0 \) as shown in Figure 2.9 b, where the indices \( i, \text{max} \) and 0 denote the cycle number, the cyclic maximum and the cyclic initial tissue intrusion. It was observed, that the cyclic tissue displacement during the initial cycles was weakly reproducible, comparable to preconditioning effects, known for biological tissues. Thus, cycles 1-4 were not considered for parameter evaluation. The median cyclic tissue displacement of cycles 5-10 was called tissue displacement \( 5-10 \Delta A \), a parameter representing the local structural and material stiffness of the vaginal wall. (Proportions of structural and material contributions could not be distinguished.) A representative unloaded state of the vaginal wall was defined by the initial tissue configuration in cycle 5, called initial bubble \( 5 A_0 \) (Figure 2.9 b).

2.3.7 Grouping of participants

The participants were subdivided into age-matched groups and according to their stage of anterior vaginal wall prolapse (cystocele). A cystocele of stage 0 and I was considered as control and stage > I as POP. Age-matching was realized by subdividing each group into pre- and postmenopausal women. Perimenopausal women were considered as premenopausal. The level of estrogenization was primarily not considered in this study.

2.3.8 Statistics

Descriptive statistic data are presented as mean, standard deviation (SD) and range (min-max). Differences between pairs of expected values \( \mu_i \) and corresponding measured values \( y_i \), such as the values obtained from two subsequent measurements or from measurements
Figure 2.9: a Evaluation of the tissue elevation $\Delta A(p(t))$, Top: reference tissue configuration $A_0$, Bottom: current, deformed tissue configuration $A(p(t))$. The tissue elevation was defined as the areal difference $\Delta A(p(t)) = A(p(t)) - A_0$, b The evaluation of the mechanical parameters was based on the history of the load dependent tissue configuration. Only cycles 5-10 were considered. The cyclic tissue displacement $i\Delta A = iA_{max} - iA_0$ was defined as the maximum tissue displacement in each cycle. The initial bubble was defined as the initial unloaded tissue configuration in cycle 5.

before and after an intervention were assessed by the calculation of deviations: the signed deviation $dev_{signed} i = (y_i - \mu_i)/\|\mu_i\|$, indicating tendencies for directed differences and the unsigned deviation, indicating the relative scatter around the expected value $dev_{unsigned} i = \|y_i - \mu_i\|/\|\mu_i\|$. Wilcoxon ranksum tests were used to assess the statistical significance of these differences. Mann-Whitney U tests were used to test the significance of differences between the mechanical parameters and the quantified subjective POP related labor (VAS) in women with (POP) and without (Control) cystocele. All statistics were performed with Matlab (The MathWorks Inc., Natick, Massachusetts, US).
2.4 Preliminary study

**Motivation**  A preliminary study had been conducted as a preparation of the reported study, the *main study* [105].

**Methods**  Women had been recruited and included in the preliminary study according to the same criteria as in the main study. The study course (hospitalization, anamnesis, POP-Q examination, aspiration measurement) had been organized accordingly and had been conducted by the same examiner. The acquisition of the subjective prolapse related labor (VAS) had been omitted at this preliminary stage. The same probe and setup had been applied, except for the aspiration tip. A larger aspiration opening (19mm x 12mm) had been used, which has been replaced in the main study, in order to reduce the failure rate. In several cases, a large amount of tissue had intruded into the opening before load application and had impaired the measurement, an effect, which could be reduced by a smaller opening.

Moreover, in the preliminary study, a different measurement protocol had been applied, which had been aimed at exploring an optimized maximum vacuum pressure, in particular with regard to women’s comfort. The vacuum pressure was linearly increased according to a rate of 1.8mbar/s. A maximum vacuum pressure of 25mbar had been identified as an appropriate upper level. Only one such cycle was applied. Four to five measurements had been conducted per participant. The same procedures for data analysis as in the main study had been used and a similar mechanical parameter, related to the structural and material stiffness of the vaginal wall had been evaluated. The history of the tissue elevation \( \Delta A(p(t)) = A(p(t)) - A_0 \) with respect to the initial tissue configuration \( A_0 \) had been linearized according to \( y = \text{linfit}_{p=0}^{p=p_{\max}}(\Delta A(p(t)), t) = m \cdot t + y_0 \), where \( \text{linfit} \) stands for a linear regression, using least squares, \( m \) represents the slope of the resulting fitting function and \( y_0 \) its offset. The slope \( m \) provided a measure for the local vaginal wall compliance.

Participants had been grouped according to their stage of POP. The small number of participants recruited at this preliminary stage did not allow for proper age matching. However, for explorative reasons, they had been separated with respect to the trophic state of their vaginal wall. Atrophy is mainly linked with age and a lack of estrogen. Mann-Whitney U tests had been used to assess differences between the groups, with respect to the stiffness parameter.

**Results**  11 women had taken part in the preliminary study. The group sizes were not balanced, which impaired a proper statistical analysis. No significant differences were observed between these groups (Figure 2.10).

However, independent of the grouping, the results in Figure 2.10 a, and more clearly visualized in Figure 2.11 a, showed apparent differences with respect to the vaginal wall
compliance among the participants, motivating another separation into two groups: women with normally trophic vaginal walls, independent of the stage of POP, as well as with severely prolapsed (atrophic) vaginal walls versus women with atrophic vaginal walls (non or moderately prolapsed) (p=0.004).

Discussion and conclusions The major limitation of this preliminary study was the small number of participants and the lack of a proper control group. For the final discrimination, rather inhomogeneous groups were identified, apparently showing similar mechanical vaginal wall properties. These results already indicated an interplay and a probable compensation of different influencing factors, such as the state of estrogenization and age: We hypothesized, that an estrogenized vaginal wall and severe POP lead to a compliant mechanical response of the vaginal wall, whereas a lack of estrogene and age cause a stiff response. However, the final role of POP could not be assessed as only limited control data were available. The fact, that actual, quantifiable differences are present with respect to the mechanical behavior of the vaginal walls of women with different clinical profiles was the major motivation for the main study. A more elaborated test protocol and an increased number of participants, including a control group was expected to sharpen these differences.

Figure 2.10: a History of the tissue elevation for women with (POP)/ without (Control) cystocele and atrophic (-)/ normal (- -) vaginal walls. A separation of the curves is observed according to their global slope: curves with high slope can be associated with normal vaginal wall trophy, independent of the stage of POP, as well as with severe POP (atrophic vaginal wall) (*), b Distribution of the slope for the four study groups
2.4 Preliminary study

Figure 2.11: a Scattered plot of the *slope* for the four study groups illustrating the separation observed in Figure 2.10. Participants with severe POP are marked by *, b The observed separation into women with normally trophic vaginal walls, independent of the stage of POP (POP and Control), as well as with severely prolapsed (atrophic) vaginal walls (large POP: lPOP) versus women with atrophic vaginal walls, non (Control) or moderately prolapsed (small POP: sPOP), is statistically significant, marked by the index a (p=0.004).
2.5 Results

2.5.1 Study participants

48 women were recruited for this study. Their mean age was 58.18 years (SD 13.40, 38.20-83.70), their mean BMI 25.87 (SD 3.95, 18.10-35.30) and their mean parity 1.6 (SD 1.33, 0-5). The grouped patient data are summarized in Table 2.1. From the 6 postmenopausal controls, 2 received systemic and 1 received local hormone replacement therapy (HRT), from the 19 postmenopausal women with cystocele, 3 received systemic and 6 received local HRT. In 33 cases, two measurements could be performed. For the first (n=9) participants a subsequent measurement was omitted, in 5 cases a second measurement failed due to technical reasons (no images were recorded) and in 1 case the measurement procedure failed. For 14 women, both pre- and postoperative measurements were performed.

Table 2.1: Grouped data for the participants within the four study groups, n is the number of participants per group: premenopausal (PreMeno) women without (Control) cystocele, premenopausal (PreMeno) with (POP) cystocele, postmenopausal (PostMeno) women without (Control) cystocele, postmenopausal (PostMeno) women with (POP) cystocele. Descriptive statistic data are given as mean (standard deviation, minimum-maximum).

<table>
<thead>
<tr>
<th>Group</th>
<th>n</th>
<th>age</th>
<th>BMI</th>
<th>parity</th>
</tr>
</thead>
<tbody>
<tr>
<td>PreMeno-Control</td>
<td>17</td>
<td>46.6 (4.6, 38.2-55.8)</td>
<td>25.6 (3.7, 20.0-32.6)</td>
<td>1.3 (1.4, 0-4)</td>
</tr>
<tr>
<td>PreMeno-POP</td>
<td>6</td>
<td>47.0 (5.3, 40.3-56.0)</td>
<td>28.4 (2.7, 24.6-32.5)</td>
<td>3.0 (1.3, 2-5)</td>
</tr>
<tr>
<td>PostMeno-Control</td>
<td>6</td>
<td>61.0 (8.8, 52.8-72.9)</td>
<td>24.7 (5.0, 18.1-30.5)</td>
<td>1.2 (1.3, 0-3)</td>
</tr>
<tr>
<td>PostMeno-POP</td>
<td>19</td>
<td>71.2 (8.4, 55.5-83.7)</td>
<td>25.7 (4.1, 18.4-35.3)</td>
<td>1.6 (1.1, 0-4)</td>
</tr>
</tbody>
</table>

2.5.2 First versus second measurement

The parameter tissue displacement $5-10\Delta A$ reproduces well within the scatter (Figure 2.12 a). Values of the first and second measurement are similar ($p=0.416$, median($dev_{unsigned}$) = 21.2%, median($dev_{signed}$) = -2.0%). In contrast, for the initial bubble $5A_0$ the differences between the first and the second measurement per patient are statistically significant ($p=0.039$, median($dev_{unsigned}$) = 19.6%, median($dev_{signed}$) = 14.9%). The scatter plot (Figure 2.12 b) and the values for the unsigned deviation illustrate that $5A_0$ is larger for a second measurement as compared to a first measurement.
2.5 Results

Figure 2.12: Mechanical parameters of the second measurement versus the first measurement, a tissue displacement $\Delta A$, b initial bubble $A_0$, for the four study groups: premenopausal (Pre-Meno) women without (Control) cystocele (O), premenopausal (PreMeno) with (POP) cystocele (O), postmenopausal (PostMeno) women without (Control) cystocele (x), postmenopausal (PostMeno) women with (POP) cystocele (X), the dashed line marks equality of the parameters for both measurements.
2.5.3 POP versus control

With respect to the two mechanical parameters tissue displacement \((5-10\Delta A)\) and initial bubble \((5A_0)\), no statistically significant differences between women with (POP) and without (Control) cystocele can be assessed neither for premenopausal nor for postmenopausal women (Figure 2.13). According to Figure 2.14, the difference in subjective prolapse related labor, quantified by the VAS, is only statistically significant for premenopausal women \((p=0.041)\). No correlation of this subjective parameter can be seen with the two mechanical parameters (Figure 2.15).

\[
\begin{array}{c}
\text{a)} \\
\text{b)}
\end{array}
\]

**Figure 2.13:** Distribution of the mechanical parameters for the four study groups, \(n\) is the number of participants for each group: premenopausal (PreMeno) women without (Control) cystocele, premenopausal (PreMeno) with (POP) cystocele, postmenopausal (PostMeno) women without (Control) cystocele, postmenopausal (PostMeno) women with (POP) cystocele, \(a\) tissue displacement \(5-10\Delta A\), \(b\) initial bubble \(5A_0\)

**Figure 2.14:** Distribution of the subjective grade of women’s POP related labor (VAS), 0: none, 10: worst, for the four study groups, \(n\) is the number of participants for each group providing this grade (not assessable for all non-native speakers): premenopausal (PreMeno) women without (Control) cystocele, premenopausal (PreMeno) with (POP) cystocele, postmenopausal (PostMeno) women without (Control) cystocele, postmenopausal (PostMeno) women with (POP) cystocele, differences were only significant for the premenopausal group, \(C\): \(p = 0.0408\)
2.5 Results

Figure 2.15: Mechanical parameters versus the subjective grade of women’s POP related labor (VAS), 0: none, 10: worst, a tissue displacement $5-10\Delta A$, b initial bubble $5A_0$, for the four study groups: premenopausal (PreMeno) women without (Control) cystocele (o), premenopausal (Pre-Meno) with (POP) cystocele (O), postmenopausal (PostMeno) women without (Control) cystocele (x), postmenopausal (PostMeno) women with (POP) cystocele (X)
2.5.4 Pre- versus postoperative

Statistically significant differences between pre- and postoperative parameter values are observed for the two mechanical parameters \(tissue \text{ displacement} \Delta A\) (\(p=0.035\)) and the \(initial \text{ bubble} A_0\) (\(p=0.030\)) (Figure 2.16). Postoperatively, both \(\Delta A\) (median(\(dev_{\text{signed}}\)) = -25.8\%) and \(A_0\) (median(\(dev_{\text{signed}}\)) = -32.7\%) are smaller than their preoperative values (Figures 2.16 and 2.17).

Considering the outlying cases, where postoperative values are larger than preoperative values, no patterns within our data can be identified. These women are neither the same for the two parameters, nor is there any relation with the (remaining) stage of POP or the postoperative subjective grade of POP related labor (acquired by the VAS) (Figure 2.17, Tables 2.2, 2.3).

![Figure 2.16: Distribution of the mechanical parameters for pre (PreOP)- and postoperative (PostOP) measurements, a tissue displacement \(\Delta A\), b initial bubble \(A_0\), p-values < 0.05 are marked by the indices A,B, A: \(p=0.035\), B: \(p=0.030\)](image.png)

Table 2.2: Outlying cases: tissue displacement, women with larger postoperative than preoperative parameter value. The notation “nPOP” is used instead of “Control”, as these women have undergone prolapse repair.

<table>
<thead>
<tr>
<th>Participant</th>
<th>preOP POP stage</th>
<th>postOP POP stage</th>
<th>preOP VAS</th>
<th>postOP VAS</th>
</tr>
</thead>
<tbody>
<tr>
<td>P57</td>
<td>III/IV</td>
<td>II (POP)</td>
<td>6</td>
<td>2</td>
</tr>
<tr>
<td>P68</td>
<td>III/IV</td>
<td>II (POP)</td>
<td>5</td>
<td>2</td>
</tr>
<tr>
<td>P86</td>
<td>III/IV</td>
<td>0 (nPOP)</td>
<td>6</td>
<td>0</td>
</tr>
</tbody>
</table>
2.5 Results

Figure 2.17: Mechanical parameters of the postoperative measurement versus the preoperative measurement, a tissue displacement $5-10 \Delta A$, b initial bubble $5A_0$. Identification numbers of participants with higher postoperative than preoperative values are given for further examination (see Tables 2.2, 2.3), the dashed line marks equality of the parameters for both measurements.

Table 2.3: Outlying cases: initial bubble, women with larger postoperative than preoperative parameter value. The notation “nPOP” is used instead of “Control”, as these women have undergone prolapse repair.

<table>
<thead>
<tr>
<th>Participant</th>
<th>preOP POP stage</th>
<th>postOP POP stage</th>
<th>preOP VAS</th>
<th>postOP VAS</th>
</tr>
</thead>
<tbody>
<tr>
<td>P57</td>
<td>III/IV</td>
<td>II (POP)</td>
<td>6</td>
<td>2</td>
</tr>
<tr>
<td>P62</td>
<td>III/IV</td>
<td>II (POP)</td>
<td>4</td>
<td>0</td>
</tr>
<tr>
<td>P75</td>
<td>III/IV</td>
<td>II (POP)</td>
<td>k.A.</td>
<td>0</td>
</tr>
<tr>
<td>P86</td>
<td>III/IV</td>
<td>0 (nPOP)</td>
<td>6</td>
<td>0</td>
</tr>
<tr>
<td>P89</td>
<td>I (Control)</td>
<td>0 (nPOP)</td>
<td>3</td>
<td>0</td>
</tr>
</tbody>
</table>
2.6 Discussion

Using the present aspiration protocol, the evaluated mechanical parameters do not allow to discriminate between women with and without cystocele. The first hypothesis is thus refused. In contrast, differences between the pre- and postoperative state can be assessed confirming the second hypothesis. An interpretation of these outcomes is discussed in this section. The vaginal application of the aspiration device was shown to cause no discomfort, pain or complications in the participants. Furthermore, a reliable applicability of the device was proven by a very low failure rate. One major limitation of this study is the mismatch of the group sizes. Premenopausal women with cystocele are rare and postmenopausal controls are less frequently hospitalized in the Department of Obstetrics and Gynecology of the USZ.

2.6.1 Reproducibility of parameters in subsequent measurements

The tissue displacement was seen to reproduce within the scatter (21.2\%) in subsequent measurements. An intrapatient scatter of \(\sim 20\%\) is common for mechanical parameters evaluated from biological tissues. Conversely, the initial bubble was larger in a second measurement and thus dependent on the load history. Material and structural rearrangements might be a reason for this apparent softening. Moreover, suction of fluids might locally increase the tissue volume.

2.6.2 Limitations - reference configuration and boundary conditions

The reference configuration is the tissue configuration to which it is referred when calculating displacements or deformations. For soft biological tissues, commonly, an unloaded and reproducible configuration is chosen. With respect to the present aspiration experiments, both aspects are difficult to be controlled by appropriate boundary conditions.

Attaching the aspiration probe to the vaginal wall and closing the cavity requires a contact force, already preloading the tissue and thus influencing the non-aspirated initial tissue configuration and the suction induced tissue displacement. An appropriate contact force has to be found by the examiner, ensuring the vacuum and minimizing the preload on the tissue initially and throughout the whole measurement (200s). Variations within one measurement and between different measurements are unavoidable as the probe is handheld and the patient is not anesthetized. These variations impair the reproducibility of the reference configuration and the tissue response, seen by the large scatter (unsigned deviation) of the initial bubble and the tissue displacement (see Figure 2.12).

Due to the structure of the vaginal wall, it is difficult to find comparable boundary conditions for all groups, for all participants within one group or even for one patient in subsequent
measurements. The vaginal wall structure is strongly determined by the menopausal state or by substitute estrogenization. The estrogenized vaginal wall exhibits characteristic wrinkles, called rugae (circumferential) and columns (longitudinal) and is covered by a lubricating film. Even in the non-aspirated state, rugae intrude visibly into the aspiration opening and increase the initial tissue bubble. Moreover, their pattern is inhomogeneous, introducing an additional dependence on the measurement site. The lubrication reduces the friction between the aspiration probe and the vaginal wall and impairs the control of the kinematic boundary conditions. During cyclic loading, the tissue is not only deformed but additional rugae might slip into the opening increasing the tissue volume within the cavity until reaching a balanced state. A lack of estrogene (postmenopausal, non-estrogenized state) causes the vaginal wall to become atrophic, unwrinkled, thin and dry.

Any quantification of the non-aspirated tissue configuration, such as the parameter initial bubble, is thus influenced by different factors (contact force, menopausal state, measurement position, lubrication, load history) which can neither be differentiated nor be controlled. Due to these uncertainties related to a reference configuration, in this study, absolute displacements were calculated instead of relative deformations (usually normalized by the reference configuration).

### 2.6.3 Reported in vivo measurements on the vaginal wall

The protocols and the outcome of current in vivo studies were compared, in order to better understand the findings of this study and the significance of the outlined limitations (Table 2.4). This discussion in particular focusses on the procedural differences to [63], where a correlation between in vivo stiffness-related mechanical parameters of the vaginal wall and the stage of POP has been reported.

For suction devices, the size of the aperture determines the depth of tissue recruitment, which has been shown to be equal to $\sim 1-2$ times the radius of a circular aperture [103]. According to Song et al. [106], the postmenopausal vaginal wall thickness locally varies between 6mm (posterior, proximal anterior and lateral vaginal wall) and 12mm (distal anterior vaginal wall), while the premenopausal vaginal wall might be expected to be thicker. Thus, Epstein [63] (radius 5mm) should have addressed the full thickness of the lateral vaginal wall, whereas Werbrouck [61] (radius 3mm and 4mm) might have reached only part of the anterior distal vaginal wall and depending on the applied opening the full thickness of the proximal anterior and distal posterior vaginal wall. In our study (axes 6.75mm and 4.25mm), we expect to have addressed the full thickness distal anterior vaginal wall.

There are large differences between the forces applied in different studies, with the herein applied force being by far the smallest (Table 2.4). A reason for the actual need of high forces could be a high prestrain, realized by the application of a speculum or by an increased...
contact force. In our study, for vacuum pressures higher than ~25 mbar, tissue was observed to come into contact with the walls of the cavity, affecting the outcome of the measurement.

A non-controllable reference configuration is one of the major limitations of the present aspiration measurements. We aimed at a physiological reference configuration, closely reproducing the in vivo conditions. In contrast, Epstein et al. defined a standard protocol to set up an altered, but probably more reproducible reference configuration [63]. They used the lateral vaginal wall for the measurements, which is also structured by rugae, however, not by columns (only on the anterior and posterior vaginal wall), providing a less wrinkled measurement site. Moreover, they inserted a speculum, prestretching the vagina, and flattening the wrinkled structure. The measurement position was chosen as 2 cm proximal to the hymen, and thus more distal as compared to our protocol and closer to the speculum. The measurement site was “cleaned” and dried with a swab, removing the lubricating film and thus normalizing the friction behavior. The device was fixed with an adhesive pad and additionally held by the examiner resulting in a contact force. The measurement was displacement controlled, recording two aspiration pressure values when the tissue passed two optically controlled gates. According to the low initiating gate, an initial tissue intrusion was forced to be smaller than 1.5 mm, resulting in an estimated maximum initial bubble of 15 mm$^2$, lower as compared to our study. Mechanical parameters based on this newly created reference configuration apparently allowed to discriminate between women with and without POP. It is worth mentioning that the protocol in [63] reduced both anatomic (macroscale) and structural (mesoscale) influences on the boundary conditions and thus on the reference configuration.

The herein reported challenges with respect to rugae and an unloaded reference configuration have been assessed by using the recorded image sequence. With respect to the reported suction devices in [63] and [61], the lack of visual control might be one reason for the high reported failure rates.

Table 2.4: Comparison of different in vivo transvaginal measurement techniques based on the aspiration principle

<table>
<thead>
<tr>
<th>Author</th>
<th>aperture, radius [mm]</th>
<th>p$_{max}$ [mbar]</th>
<th>F$_{max}$ [N]</th>
<th>position</th>
<th>failure [%]</th>
<th>correlation POP</th>
<th>POP surgery</th>
</tr>
</thead>
<tbody>
<tr>
<td>Epstein [63]</td>
<td>5</td>
<td>200</td>
<td>1.57</td>
<td>lateral,-2cm</td>
<td>&lt;20%</td>
<td>yes</td>
<td>yes</td>
</tr>
<tr>
<td>Werbrouck [61]</td>
<td>3,4</td>
<td>300</td>
<td>0.85/1.51</td>
<td>Aa,Aa-2cm, Ap,Ap-2cm</td>
<td>&gt;50%</td>
<td>no</td>
<td>yes</td>
</tr>
<tr>
<td>Röhrnbauer</td>
<td>6.75 / 4.25</td>
<td>25</td>
<td>0.25</td>
<td>Aa-2cm</td>
<td>&lt;10%</td>
<td>no</td>
<td>yes</td>
</tr>
</tbody>
</table>
2.6.4 An interpretation attempt

In the context of the previous section, we might suppose, that differences in the mechanical behavior of the vaginal wall due to POP can more probably be assessed by normalizing the reference configuration and reducing influences of the anatomy and the vaginal wall structure, as done in [63] but not in our study. This motivates a short discussion about different competing influences and their possible impact on the mechanical parameters.

As indicated, the system vaginal wall can be seen at different length scales (Figure 2.18). At the macroscale, the vaginal wall is embedded within the pelvic floor structures, the anatomy, determining its global boundary conditions. Moreover, the vaginal wall is a wrinkled (rugae and columns) and a layered structure, representing the mesoscale. The microstructure is mainly characterized by the fibrous constituents, i.e. collagen and elastin.

Among other factors, POP, estrogenization and age influence the vaginal wall at all these length scales. POP is a change of the pelvic floor anatomy (macroscale), causing vaginal wall laxity and decreasing vaginal wall integrity [3,4]. Thus, POP is associated with an increased structural vaginal wall compliance. Moreover, POP is associated with alterations of the microstructure, i.e. of the collagen metabolism [3] and the content of normal elastin [96,97]. However, findings in the current literature diverge and we can only assume that POP related connective tissue degradation result in a lax and less load resistant vaginal wall. The level of estrogenization and age contrarily influence both the vaginal wall meso- and microstructure.

At the mesoscale, the two factors determine the appearance of rugae, which can be positively related to a decreased structural stiffness. At the microscale, they determine the collagen maturation and the content of elastin [3,95]. The combination of cross-linked collagen and a lack of elastin is assumed to result in a stiff material response.

In line with these simplified relations, we might construct different scenarios, illustrating the complexity of quantifying the in vivo mechanical properties of the vaginal wall. A POP related decrease in the structural and material stiffness of the vaginal wall, might be compensated by a poorly estrogenized and aged vaginal wall. In contrast, the measured stiffness of an intact vaginal wall might be reduced by the compliant structural response of the rugae. The relative contributions of each factor to the actually measured parameter value depend on the sensitivity of the aspiration measurement to phenomena at the different scales.

This interpretation attempt is based on literature findings and on the observations during our preliminary study (see section 2.4). It certainly contains simplifications. For example, microstructural changes due to tissue remodeling, in general a compensating mechanism, are not accounted for [98]. Additional important influencing factors, intrinsic and extrinsic [3] might be thought of, such as connective tissue alterations, neurologic abnormalities or pregnancy, childbirth, tissue trauma and an increased intra-abdominal pressure.
Based on this discussion, exemplarily, the influence of the participants hormonal state is examined. Therefore, the postmenopausal group is divided considering their estrogenization (HRT, n(on)HRT), accepting very small postmenopausal control groups (n=3). No significant differences between POP and control groups can be assessed (Figure 2.19). However, the tissue displacement seems to be dependent on the estrogenization for the postmenopausal control groups (p=0.10 for PostMeno-Control-HRT vs PostMeno-Control-nHRT). Its median is higher for the estrogenized group, which is in accordance with the assumption of a structurally compliant estrogenized vaginal wall. Moreover, for the postmenopausal groups without HRT (nHRT, POP and Control), the two mechanical parameters tissue displacement (p=0.217) and initial bubble (p=0.077) tend to be larger in women with cystocele (POP) compared to the allocated control group (Figure 2.19). As hypothesized in the previous paragraph, a POP related reduced vaginal integration might be a reason for an increased tissue displacement and an increased initial bubble.
2.6 Discussion

Figure 2.19: Distribution of the a tissue displacement $5−10\Delta A$ and b initial bubble $5A_0$ for further separated study groups, n is the number of participants for each group: premenopausal (PreMeno) women without (Control) cystocele, premenopausal (PreMeno) with (POP) cystocele, the postmenopausal (PostMeno) groups (with (POP) and without (Control) cystocele) are further divided into participants receiving hormone replacement therapy (HRT) and those without such treatment (nHRT)
2.7 Conclusions

The presented aspiration protocol for a transvaginal mechanical characterization of the vaginal wall does not allow to discriminate between women with and without cystocele. POP related processes might affect different structures at different length scales, the pelvic floor anatomy, the vaginal wall structure and the local tissue composition, in a complex and interdependent way. The current procedure is apparently not sensitive to one specific length scale. However, significant changes influencing all length scales, such as a surgical prolapse repair, are detected by the aspiration measurements.

The visual control, which is unique for this device, compared to other commercially available suction devices, increases patients’ safety and allows to decrease failure rates. Furthermore, it provides a deeper understanding of the challenges related to the mechanical behavior of the vaginal wall, including the definition of a reference configuration or a more sensitive differentiation between structural and material properties.

We believe that a characterization of the pelvic floor mechanics related to POP needs to include processes at different length scales and at the same time needs to be able to differentiate between them. In this context, the present study might contribute to a further improvement of currently available and the development of alternative in vivo techniques, which are still subject of fundamental research.
Towards a physiologically relevant mechanical characterization of prosthetic meshes

**Motivation** The following study proposes new protocols and data analysis procedures for a more thoughtful design of animal studies with respect to a physiologically relevant outcome.

**Methods** Uniaxial and biaxial ex vivo experimental protocols and corresponding mechanical parameters are proposed to characterize and categorize mesh explants from animal models. For biaxial tests, the inflation technique was applied. Tests were performed on explants from a rabbit abdominal wall model and corresponding native tissue (part I). An accordingly designed study was conducted using meshes embedded in an elastomeric matrix (part II).

**Results** The findings show that mechanical parameters describing the sample stiffness depend on the load case (uniaxial or biaxial), i.e. different results are obtained from the uniaxial as compared to the biaxial load case, and on the load level at which they are evaluated. Moreover, stiffnesses obtained from mesh elastomer composites are predictive for the corresponding explant stiffnesses.

**Main conclusions** These findings demonstrate that an evaluation of the mechanical biocompatibility of prosthetic meshes should be based on an experimental configuration (uniaxial or biaxial tension) which reproduces the expected in vivo conditions of mechanical loading and deformation. Non-biological model systems are recommended to be used for the development of the respective test setups and protocols as well as with respect to mesh design optimization.

*Parts of this chapter, including paragraphs of text, figures and tables are published in [5] and [107].*
3 Physiologically relevant mechanical characterization of prosthetic meshes

3.1 Introduction

From the mechanical point of view, prosthetic meshes are anisotropic, non-linear viscoelastic textiles [1, 79]. Following implantation the host develops a foreign body reaction and integrates compatible implants by deposition of fibrous tissue and neovascularization. The initial implant and the ingrown host tissue form an inhomogeneous laminate. A mechanical characterization of meshes requires methodologies to deal with such complex structures. Current experimental ex vivo studies, examine and characterize the mechanical behavior of explants, i.e. the original implant and the surrounding host tissue, harvested from animal models, as these samples mimic the ingrown in vivo state of the implant [43, 68–77]. However, extensive animal studies should be reduced for financial and ethical reasons. Another problem with animal studies is the fact that conclusions are often weakened by small sample sizes and large scatter [68, 73]. For this reason, several experimental studies on meshes before implantation, which we call dry meshes, are reported, which are obviously lacking the physiological property of being embedded in a matrix [68, 73, 78–92].

The most frequently used method for testing both explants and dry meshes is the uniaxial tensile test [68–73, 78–90]. Few biaxial setups, such as the ball burst test [74–76, 85, 88, 92] or the inflation-to-burst test [77] have been reported. With respect to the mechanical properties, the focus of most studies is still on parameters related to sample rupture (maximum force or elongation) [68–74, 76–79, 83–85, 87, 91, 92], though parameters describing the deformation behavior at moderate tensile forces are now known to be clinically more relevant [45, 69, 71–75, 78–85, 91].

Explant studies are primarily aimed at an evaluation of the mechanical biocompatibility of the corresponding meshes. Therefore, their mechanical behavior has to be investigated under physiologically relevant loading conditions [75], where “physiological” means having properties similar to the tissue to be replaced [45, 73, 75]. Uncertainties about physiological loading conditions and mechanical properties of the tissues and structures to be replaced are identified as a major challenge [68, 73]. A comparison of the results of these studies is often impaired by the fact that a standardized experimental protocol as well as standardized mechanical parameters are missing.

In contrast, dry mesh studies are mainly motivated by examining specific mechanical phenomena of meshes, such as anisotropy [81, 85, 86, 88, 89], or preconditioning effects in cyclic loading conditions [82, 86, 87, 92]. In some investigations, dry mesh tests are used to develop constitutive models for numerical simulations of meshes [89, 90]. Based on the results, different mesh types are ranked [73, 78, 80, 81, 85–89, 92], or grouped and compared with respect to different criteria, such as light-weight versus heavy-weight [91], prior versus after resorption [83] or new mesh types versus an established mesh type [79, 83, 85]. However, are the results of dry mesh studies also able to answer clinical questions with respect to
mechanical biocompatibility? Can mechanical tests on dry meshes mimic the mechanical in vivo behavior or at least reproduce the results gained from explant tests? So far, correlations between results for dry meshes and an expected clinical outcome are weak [73, 78–80, 83, 89, 92]. Several groups explicitly exclude direct correlations and stress the necessity for an additional mechanical characterization of mesh-tissue complexes [78, 82, 87, 89].

The present work uses uniaxial as well as biaxial experimental protocols, and proposes corresponding mechanical parameters to characterize and categorize meshes. In the first part of this work, explants obtained from a full thickness abdominal wall defect rabbit model are investigated. We hypothesize that mechanical parameters describing the stiffness of an explant depend (i) on the load case, so that different results are obtained from uniaxial as compared to biaxial tension, and (ii) on the level of mechanical loading at which they are evaluated (due to a pronounced non-linearity of the mechanical response). Verification of these hypotheses will confirm that the categorization of meshes according to the stiffness of the corresponding explants is to be carried out considering the specific load case (uniaxial or biaxial) and load level expected in vivo, i.e. under physiologically relevant conditions.

In the second part of this work, a non-biological model system is used, embedding the same meshes in dry condition within an elastomeric matrix. The resulting structures are called “composites” in the following. For these samples, we hypothesize that (i) uniaxial and biaxial stiffness parameters gained for composites are predictive for corresponding explant stiffnesses and (ii) explant stiffnesses cannot be predicted from uniaxial tensile data of dry meshes (without elastomer), which are most commonly available from the current literature. These hypotheses are tested through a comparison with the corresponding explant data.
3.2 Methods

The first part of this study, the explant study, was conducted at the Center for Surgical Technologies at the Katholieke Universiteit in Leuven. The second part, the composite study, was performed at the Center of Mechanics at ETH Zurich.

3.2.1 Materials

Experiments were done with two commercially available meshes, knitted from non-resorbable polypropylene fibers (Table 3.1). One was a heavy-weight construct (SPMM: Surgipro Polypropylene Monofilament Mesh; 96g/m², Covidien, Mechelen, Belgium) and the other was a so-called light-weight hybrid construct, containing polypropylene and polyglecaprone fibers (Gynemesh M; weight prior to resorption: 56g/m², after resorption: 32g/m², Ethicon Inc., Somerville, NJ, United States) [108]. The polyglecaprone fibers are added to improve the surgical handling properties of the material as well as because of the anti-inflammatory properties [73,109–111]. Resorption takes place within 90-120 days after implantation. In the composite study, Gynemesh M was employed in two configurations: in the standard state, containing both components and in the resorbed state, containing only polypropylene fibers. With respect to in-plane mechanical properties, SPMM is assumed to be quasi-isotropic [112], whereas Gynemesh M is orthotropic [109]. As a non-biological matrix material, the acrylic elastomer VHB (VHB 4910, thickness 1mm, 3M, St. Paul/Minnesota, USA) was used, an isotropic material with an adhesive surface and much lower stiffness compared to the present meshes [113]. All materials tested came from the same production lot and were provided sterile by the manufacturer.

For the explant study, rectangular mesh sheets were cut to size 6.0cm x 6.0cm for the biaxial tests and 5.0cm x 7.0cm for the uniaxial tests. Four explants were prepared per mesh type for each testing condition, biaxial and uniaxial, as described in the following section. For the composite study, materials were cut as circles of 5.0cm diameter for the biaxial tests and as strips of 2.0cm x 9.0cm for the uniaxial tests. Mesh VHB composites were prepared by embedding the mesh samples between two accordingly cut layers of VHB. Four composite samples per mesh type and four control (VHB only) samples were prepared for each testing condition, biaxial and uniaxial. Moreover, four samples of dry mesh per mesh type were prepared for uniaxial tension.

3.2.2 Implantation

For the biaxial tests, six New-Zealand white rabbits were randomly divided into two treatment groups, one for each material, and two control animals. Two meshes were implanted
3.2 Methods

Table 3.1: Physical properties of the used materials: SPMM and Gynemesh M (GM). Data were taken from [73]

<table>
<thead>
<tr>
<th>Material</th>
<th>Fiber material</th>
<th>Nature of fibers</th>
<th>Fiber-diameter [mm]</th>
<th>Pore-size [mm²]</th>
<th>Weight [g/m²]</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPMM</td>
<td>Polypropylene</td>
<td>Non-absorbable</td>
<td>0.155</td>
<td>0.825 x 0.657</td>
<td>96</td>
</tr>
<tr>
<td>GM</td>
<td>Polypropylene</td>
<td>Non-absorbable</td>
<td>0.068</td>
<td>3.8</td>
<td>56 prior and 32 after absorption</td>
</tr>
<tr>
<td></td>
<td>Polyglecaprone</td>
<td>Absorbable</td>
<td>0.127</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

in each animal of the treatment group, resulting in four test specimens per mesh type. Only one type of implant was used per animal (Figure 3.1). For the uniaxial tests, another three rabbits were included. Two animals underwent implantation of two meshes each, one SPMM implant and one Gynemesh M implant. Two test specimens were obtained per implanted mesh, resulting in four test specimens per mesh type. The third animal served as a control (Figure 3.1). Animals in the treatment groups were brought under general anesthesia and operated on in sterile conditions. Implantation was done as previously reported in detail [73] and is illustrated in Figure 3.2 (This figure is for illustration only. The shown implant dimensions do not correspond to the implant dimensions used in this study). Two longitudinal 2cm long full-thickness incisions were made on each side of the lateral anterior abdominal wall, parallel to the midline. The abdominal wall muscle layer was primarily closed by a continuous 4/0 polyglecaprone suture (Monocryl®, Ethicon) and overlaid by the implant of interest. The orthotropic Gynemesh M samples were implanted such that the stiffest material direction was aligned with the axis of the body. All implants were fixed tension-free with separate 4/0 Monocryl sutures placed with an inter-suture distance of 1cm. The subcutis and skin were closed with continuous 3/0 Monocryl. The animals within the control group were not operated on, as they should provide a native tissue reference to assess the mechanical biocompatibility of the meshes. They were housed in parallel with the implanted animals. Euthanasia was after 21 days by intravenous injection of 1mL of a mixture of embutramide 200mg, mebezonium 50mg and tetracaine hydrochloride 5mg (T61; Hoechst Marion Roussel, Brussels, Belgium).

Animals were treated in accordance with current national guidelines on animal welfare. The experiment was approved by the Ethics Committee for Animal Experimentation of the Faculty of Medicine of the K.U. Leuven.

3.2.3 Harvesting of samples

At obduction, the area of surgery, containing the initial implants together with the underlying abdominal wall muscles was excised. These specimens will be further referred to as “explants”. Explant harvesting was done differently for biaxial and uniaxial tests, as ex-
plained below. For biaxial explant preparation, rings of sandpaper with an inner diameter of 3.5cm and outer diameter of 5.5cm were glued on the explant measuring 6.0cm x 6.0cm, in situ (Figure 3.3a). The sandpaper-rings were applied to preserve the in vivo explant dimensions and to improve clamping of the tissue during mechanical testing [114]. To this end, the sandpaper was oriented such that its rough side pointed towards the explant. We applied superglue (Loctite 401, Henkel, Düsseldorf, Germany) between the sandpaper-ring and the explant, ensuring that the mechanical response of the explant within the circular region of measurement was not affected. The explants were harvested by cutting out the prepared abdominal wall section as shown in Figure 3.3a. A second ring of sandpaper with the same dimensions was glued on the abdominal side of the explant. For the uniaxial tests, two rectangular strips measuring 2.0cm x 7.0cm were cut from the 5.0cm x 7.0cm explants. Before harvesting them, the clamp distance of 5.5cm was marked with a tissue marker to allow future positioning of the samples within the testing machine. Biaxial and uniaxial control abdominal wall samples with similar dimensions were retrieved from the unoperated control animals. Immediately after harvesting, all explants were stored in a gauze soaked in normal saline, until testing. Testing always took place within two hours after euthanasia. Four samples were tested for each configuration and load case (Figure 3.1).

### 3.2.4 Biaxial testing

A custom-built device, which is a modified version of the inflation device earlier used for testing fetal membranes [114], was applied. Comparable devices are available at both labs in Leuven and in Zurich. Explant tests were performed in Leuven, corresponding composite tests were performed in Zurich. The inflation device consists of an aluminum cylinder
Figure 3.2: Schematic illustration of the implantation of mesh sheets. Two longitudinal, full-thickness incisions on each side of the lateral anterior abdominal wall were primarily closed by a continuous 4/0 polyglecaprone suture (Monocryl®, Ethicon) and overlaid by the implant of interest. This figure is adapted from [44]. The shown implant dimensions do not correspond to the implant dimensions used in this study.

with an inner diameter of 3.5cm. The circular specimens were clamped with the rings of sandpaper between the cylinder and a cover ring, such that for the explants the mesh was on the outer side, as shown in Figure 3.3b. The composite specimens were realized with a symmetric layup, with VHB layers on each side of the mesh. A schematic view of the setup is shown in Figures 3.4a and b. A peristaltic pump (Type 314VBM, Watson-Marlow Ltd., Zurich, Switzerland) pumped water into the cylinder at a fixed flow rate of 0.125mL/sec, applying a hydrostatic pressure on the (abdominal) inner side of the specimen and inflating it at a nearly constant rate of deformation until failure or the pressure limit of the experimental setup (∼2400mbar). The deformation rate was estimated at ∼1% s⁻¹, which was considered slow enough to exclude time dependent effects and to assume behavior corresponding to the long term mechanical response [86]. The hydrostatic pressure [N/cm²] was measured by a pressure sensor (digital manometer, LEX 1, -1 to 2 bar, accuracy 0.05%, Keller, Switzerland) positioned at the outlet of the cylinder at the level of the bottom plate. A 2/3” CCD camera (Point Grey, 1.4MP Color Grasshopper 1394b Camera, 2/3” CCD) recorded a side view image sequence of the inflating sample at a frame rate of 0.5 s⁻¹. The average duration of one test was 7-8 minutes. Routines for data analysis were programmed in Matlab (The MathWorks Inc., Natick, Massachusetts, US). The contour of the inflated sample was extracted from each image of the sequence. The highest point of the contour was referred to as the apex. The current apical position d [cm] (Figure 3.4b) was used for the calculation of the sample deformation. As inflating the sample leads to an inhomogeneous deformation pattern, the analysis of the local kinematics in the central (apex) region
Physiologically relevant mechanical characterization of prosthetic meshes

Figure 3.3: a Preparation of round explants for biaxial testing. Using clamps and a ring of sandpaper, in vivo dimensions were preserved. b Explant clamped for biaxial testing.

was based on a generic finite element analysis, using Abaqus (Dassault Systemes, Simulia, France), as documented in Appendix A. The in-plane stretch $\lambda_p$ [-] in the apex region was calculated as the ratio between the current length $l$ and the reference length $l_0$ of an arc at the center of the sample, $\lambda_p = l/l_0$. Simulating the experimental configuration, the arc length $l$ was found to correlate with the apical position $d$ according to a polynomial function of order four (Appendix A, Figure A.1 a,b). In this way, local equibiaxial stretch could be determined from the apical position $d$. As a measure of force acting on the specimen, the averaged membrane tension $T_m$ [N/cm] in the central region was calculated according to the static equilibrium, applied to a spherically inflated membrane, as $T_m = 0.5 \cdot p \cdot r$ [N/cm] (Lagrange model) with $p$ being the current pressure and $r$ the radius of curvature in the central region. Based on the finite element analysis, an inverse relationship between $d$ and $r$ was determined (Appendix A, Figure A.1 a,c). By characterizing mechanical loading in terms of membrane tension (not stress) the need to quantify the sample thickness was avoided, which is challenging to measure in a reliable way [115].

3.2.5 Uniaxial explant testing

The uniaxial tests on explants were performed in Leuven. A zwicki-Line Z0.5 uniaxial extensometer, for maximal test forces of 500N (Zwick/Roell GmbH & Co. KG, Ulm, Germany) was used with a 200N load cell, as described previously [73]. The orientation of the setup was vertical. The explants were inserted tension free between the clamps and were stretched to their physiological length of 5.5cm, according to the applied marks. Samples were further stretched up to a preforce of 0.1N, marking the actual start of the test and providing the
3.2 Methods

reference length $l_0$ for later calculations of nominal stretches. Without any preconditioning, the samples were stretched until failure. Preconditioning was omitted in order to provide equivalent loading conditions with respect to the biaxial inflation experiment, where this procedure was not applied. All tests were performed at the same nominal stretch rate as the biaxial tests, of $\sim 1\%s^{-1}$, corresponding to a velocity of $\sim 0.08mm/s$. The current distance between the clamps [mm] and the corresponding force [N] were recorded. The nominal stretch $\lambda [-]$ was calculated as the current distance between the clamps, equal to the current sample length $l$, divided by its reference length at the level of the preforce $l_0$, $\lambda = l/l_0$. The nominal membrane tension $Tm$ [N/cm] was calculated as the current force $F$ divided by the initial width of the sample $w_0$, $Tm = F/w_0$.

3.2.6 Uniaxial composite testing

Uniaxial tensile tests with composites and dry meshes were conducted in Zurich using a custom-made test setup described in detail in [116]. It consisted of a tensile test machine with two hydraulic actuators (242 Actuator, MTS Systems Corp., Eden Praire, MN, USA), each with 2.7kN capacity, mounted horizontally on a steel plate. 100N load cells (SMT S-Type, Interface Inc., Scottsdale, AZ, USA) were applied for force measurements. Custom-made clamps equipped with sandpaper at the clamping faces were directly attached to the load cells. For contactless measurement of the in-plane displacement field in the central region of the sample, a video extensometer system for triggered image acquisition (uniDAC FAST, Chemnitzer Werkstoffmechanik GmbH, Chemnitz, Germany) was installed. The system
3 Physiologically relevant mechanical characterization of prosthetic meshes

consisted of a 2/3” monochrome CCD camera (Pike F-100B, Allied Vision Technologies GmbH, Stadtroda, Germany) and a focusable 0.25x telecentric lens (NT55-349, Edmund Optics GmbH, Karlsruhe, Germany), mounted adjustable on a custom-made portal on top of the test setup. The specimens (mesh VHB composites and dry meshes) were mounted between the clamps by means of a supporting system in order to avoid slacking. A preforce of 0.01N determined the start of the test. Samples were stretched at $\sim 1\% s^{-1}$ up to an arbitrarily chosen, supra-physiological load limit of 49N, without any preconditioning cycles. Longitudinal and transversal stretch components, $\lambda_{\text{long}}$ and $\lambda_{\text{trans}}$, respectively, were evaluated locally from the recorded image sequence by using the software VEDDAC cam 3.2 (Chemnitzer Werkstoffmechanik GmbH). The nominal membrane tension was calculated as the current force $F$ [N] divided by the initial width $w_0$ [cm] of the sample, $Tm = F/w_0$ [N/cm]. Note that, alternatively, we could have calculated the current membrane tension $Tm_{\text{cur}} = F/w$ using the current width $w = \lambda_{\text{trans}} \cdot w_0$, which would be in accordance with the outcome of the inflation tests. However, we chose here to represent our results in terms of nominal membrane tension to allow a direct comparison with the uniaxial tests of the explant study, where the current width had not been measured (no optical strain analysis).

### 3.2.7 Mechanical parameters

In order to mechanically characterize the tested samples, tangent moduli/stiffnesses at defined reference levels of membrane tension were chosen as parameters. Reference levels of 1.1N/cm, 3.9N/cm and 7.3N/cm (explants only) were selected, representing the values of the mean membrane tension of abdominal wall tissue at stretch levels of 1.05, 1.1 and 1.15, under biaxial loading conditions (Figure 3.7a). These values of deformation were considered representative for a low (1.05), intermediate (1.1) and increased (1.15) physiological stretch of abdominal wall tissue [73, 108]. The tangent moduli were calculated as the slopes of the membrane tension versus stretch curve at these reference values. The extraction of tangent moduli at the reference level of membrane tension of 3.9N/cm is shown for all explants in Figures 3.5a-c for both the biaxial and the uniaxial test series.

### 3.2.8 Representation of data

All data were plotted in terms of membrane tension versus stretch (Figure 3.5). For clearer visualization and comparison between different sample configurations, average curves were calculated for all groups. In accordance with the analysis of load dependent tangent moduli, mean values were determined by taking the mean stretch at constant levels of membrane tension. The distributions of tangent moduli for all specimens were visualized by scatter plots. The corresponding data were reported in terms of mean and standard deviation.

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3.2 Methods

Figure 3.5: Results are presented as membrane tension versus stretch curves for biaxial (BI) and uniaxial (UNI) loading conditions. a SPMM explants, b Gynemesh M (GM) explants, c abdominal wall tissue (AW). Note that the horizontal axis in c differs from a and b. Local tangent stiffness values are extracted at the reference membrane tension of 3.9N/cm (marked by the dashed line).

3.2.9 Estimating the mechanical response of explants from composites data

In order to estimate the mechanical response of the explants from the composites data, a superposition approach was applied. It was assumed that (i) the layers of the present mesh VHB composites are perfectly adherent, such that the mesh and the matrix deform in an affine manner, and that (ii) the global force response is composed of the contributions of each component and an interaction term (mesh-matrix-interaction). For any composite (mesh-VHB or explant=mesh-AW), consisting of two components, part 1 and part 2, the overall membrane tension \( T_{m_{12}} \) can be calculated as \( T_{m_{12}} = T_{m_1} + T_{m_2} + T_{m_{inter12}} \), with the unknown mesh-matrix-interaction term \( T_{m_{inter12}} \). The explant was considered as a compos-
ite ($T_{\text{mesh-AW}}$) consisting of mesh ($T_{\text{mesh}}$) and abdominal wall ($T_{\text{AW}}$). Its behavior can be estimated from the measured data for the mesh VHB composite ($T_{\text{mesh-VHB}}$), for VHB alone ($T_{\text{VHB}}$) and for rabbits’ abdominal wall tissue ($T_{\text{AW}}$): $T_{\text{mesh-AW}} \approx T_{\text{mesh-VHB}} - T_{\text{VHB}} + T_{\text{AW}}$. $T_{\text{VHB}}$ and $T_{\text{AW}}$ represent the mean membrane tension values of all measured VHB or abdominal wall (AW) samples, respectively, and $T_{\text{mesh-VHB}}$ refers to each composite sample. This approach is graphically shown in Figure 3.6. Differences between this approximation and the response of the explant were expected to be due to the difference between the mesh-abdominal wall (AW)- and mesh-VHB-interaction terms, $err \approx T_{\text{inter-mesh-AW}} - T_{\text{inter-mesh-VHB}}$.

Figure 3.6: Superposition approach for the estimated a) biaxial (BI) and b) uniaxial (UNI) mechanical response of explants (exemplification for Gynemesh M resorbed), approximated membrane tensions for explants ($T_{\text{mesh-AW-approx}}$) are derived by the superposition of the membrane tension values of the corresponding composite ($T_{\text{mesh-VHB}}$), VHB ($T_{\text{VHB}}$) and rabbits’ abdominal wall ($T_{\text{AW}}$): $T_{\text{mesh-AW}} \approx T_{\text{mesh-VHB}} - T_{\text{VHB}} + T_{\text{AW}}$. Membrane tension values for VHB and AW correspond to mean values of all measured VHB and AW samples. For the uniaxial load case (b) and for stretches of <1.4, the mean mechanical response of VHB and AW is similar and very compliant. The approximated explant curve and the curve for the composite nearly coincide.

3.2.10 Statistics

A one-way analysis of variance (ANOVA) and post-hoc multiple comparisons using Tukey’s HSD test were performed, in order to assess differences between the materials with respect to the tangent moduli at each level of reference membrane tension, for each test series. For the explants, a mixed ANOVA was used, accounting for the influence of the specific animal, when taking more than one specimen out of the same rabbit. This random effect was shown to have negligible influence. P-values $< 0.05$ were considered statistically significant. The statistics were performed using R [117].
3.3 Results: explant study

For biaxial measurements, the findings were different for the three groups (Figure 3.7a). Uniaxial test results were similar for Gynemesh M and SPMM explants, which were both different from the controls (Figure 3.7b).

**Figure 3.7:** a Biaxial (BI) and b uniaxial (UNI) tests. Mean stretch values for all samples of each configuration (abdominal wall (AW) control samples, Gynemesh M (GM) explants, SPMM explants) are calculated for constant levels of membrane tension. The reference levels of membrane tension, 1.1N/cm, 3.9N/cm and 7.3N/cm, and the corresponding stretch levels for abdominal wall tissue 1.05, 1.1 and 1.15 under biaxial load are marked by the black dashed lines.

For biaxial measurements, at low (1.1N/cm) and moderate (3.9N/cm) membrane tension, Gynemesh M explants and native tissue (AW) are equally compliant (1.1N/cm: GM 59±44 N/cm, AW 40±23 N/cm, 3.9N/cm: GM 95±25 N/cm, AW 74±24 N/cm) and significantly less stiff than SPMM explants (1.1N/cm: 145±36 N/cm, 3.9N/cm: 191±15 N/cm) (Figure 3.8 a, b). The large scatter for Gynemesh M explants at 1.1N/cm is due to one sample providing very different results. At the higher (7.3N/cm) membrane tension, there are significant differences with respect to the tangent moduli for all groups (Figure 3.8 c). SPMM explants are stiffer than Gynemesh M explants and both are stiffer than native tissue (SPMM 183±22 N/cm, GM 115±15 N/cm, AW 68±12 N/cm).

At the reference levels of membrane tension (1.1N/cm, 3.9N/cm, 7.3N/cm), GM explants are subject to biaxial stretch levels of 1.03±0.02, 1.07±0.03 and 1.11±0.03 and SPMM explants are subject to biaxial stretch levels of 1.01±0.00, 1.02±0.00 and 1.04±0.00.

Uniaxial testing led to early failure of two control samples, which had to be excluded for parameter evaluation at the high membrane tension (7.3N/cm). Moreover, the variation of the tangent stiffness for the Gynemesh M samples doubled at this membrane tension compared to the moderate level (3.9N/cm). At low (1.1N/cm) membrane tension, both
explant types have similar tangent moduli (GM 21±5 N/cm, SPMM 24±5 N/cm) and are stiffer than native tissue (10±1 N/cm) (Figure 3.9 a, b). At the moderate level (3.9N/cm), Gynemesh M explants are stiffer (51±14 N/cm) than SPMM explants (35±3 N/cm) and native tissue (25±6 N/cm), with the two latter groups being at a similar level. At the high (7.3N/cm) membrane tension, differences between tangent stiffnesses of all types of samples vanish (SPMM 48±16 N/cm, GM 64±28 N/cm, AW 42±11 N/cm) (Figure 3.9 c). For uniaxial tests, the evaluation of stretches is omitted as it is prone to uncertainties, discussed in section 3.4.

Absolute biaxial tangent moduli are by a factor 2 to 7 larger than uniaxial tangent moduli, and even more (a factor 4 to 7) for low (1.1 N/cm) membrane tension.

**Figure 3.8:** Distribution of tangent moduli for biaxial (BI) tests on abdominal wall (AW) control samples, Gynemesh M (GM) explants and SPMM explants for different reference levels of membrane tension. P-values < 0.05 are indicated with corresponding letters A,B,C,D,E,F,G. a 1.1N/cm, A: p<1e-04, B: p<1e-03, b 3.9N/cm, C,D: p<1e-04, c 7.3N/cm, E: p<1e-04, F,G: p<1e-05.

**Figure 3.9:** Distribution of tangent moduli for uniaxial (UNI) tests on abdominal wall (AW) control samples, Gynemesh M (GM) explants and SPMM explants for different reference levels of membrane tension. P-values < 0.05 are indicated with corresponding letters H,I,J,K. a 1.1N/cm, H: p<1e-04, I: p<1e-04, b 3.9N/cm, J: p<0.001, K: p=0.013, c 7.3N/cm. *Two values are excluded due to sample failure.
3.4 Discussion: explant study

Our results demonstrate that the loading conditions (uniaxial versus biaxial) as well as the level of mechanical load chosen for parameter evaluation have an impact on the outcome. The relevant question becomes therefore: which protocol and which parameters should be considered when judging biomechanical performance?

3.4.1 On physiological relevance

Procedures for characterizing prosthetic meshes (and their corresponding explants) have to be chosen as a function of the specific medical application and the ultimate functional result aimed for. In other words, measurements have to be relevant for the specific clinical outcome. Animal models are used to “grow” explants for histological as well as mechanical characterization. The type of animal, the implantation site and the in vivo loading and boundary conditions determine the composition and the structure of the explant, hence its mechanical properties. In the present study, explants have been grown under biaxial loading conditions and at load levels typical for the abdominal wall of adult rabbits.

Both biaxial and uniaxial stress states can be considered as physiologically relevant, depending on the medical application of interest. However, as the animal model determines the configuration of the test samples, the experimental load case needs to be mechanically consistent with it. This is only fulfilled, in our case, by the inflation setup where loading conditions are similar to the in vivo conditions during incorporation of the implant in the host. Conversely, both loading and boundary conditions present during uniaxial testing differ from the in vivo state. Strips are cut, resulting in non-physiological free edges, which are a source of experimental artifacts [118], as further discussed in this section. Thus, in the present case, the biaxial procedure meets the requirements of physiological relevance and consistency with the applied animal models in a more appropriate way.

It should be noted, that physiological relevance is addressed only for the case of passive mechanical deformation due to internal pressure. Other relevant influences, such as active muscle contractions are not regarded herein.

Along the same lines, mechanical parameters should map the mechanical behavior in a physiological range of deformation and load. As biological tissues and structures as well as the implants show a non-linear relationship between tension and stretch, parameters depend on the specific range of load or deformation for which they are evaluated. Klinge et al. [115] used the law of Laplace to estimate different levels of membrane tension for the human abdominal wall resulting from physiologically relevant levels of intra-abdominal pressure, such as in rest, standing or coughing. A critical reference value of 16N/cm was chosen to be representative in order to design mesh implants for hernia repair. For prosthetic mesh
application in the female pelvic floor, Ozog [44] approximated the female pelvic cavity by an ellipsoid, spanning by defined landmarks. The distances between these landmarks provided patient specific parameters to estimate the membrane tension within the pelvic floor. The tension values obtained are one order of magnitude lower compared to the data generated by Klinge. The reference levels of membrane tension used in this study are in line with the reported values, and the corresponding parameters map a wide range of physiologically relevant load and deformation. In this sense they provide useful information for choosing an appropriate mesh in different clinical applications.

3.4.2 Limitations

The small sample size (n=4) represents a limitation of the present study. The power of the applied statistical tests could not be determined due to uncertainties in estimating a reference value of standard deviation within the groups. In particular for the biaxial data at the low level of membrane tension (1.1N/cm), it is seen that there is a large effect of single outlying values (Figure 3.8 a) on the corresponding standard deviation. Our results support the conclusion that different rankings with respect to the explant stiffness might be drawn from different loading conditions and/or load levels. When aiming at a quantitative comparison of different mesh types to serve as a basis for clinical decisions, larger number of samples should be tested to provide the corresponding statistical power.

With respect to the scatter of the tangent moduli, it is seen that uniaxial and biaxial results for native abdominal wall tissue (control) and GM explants show larger variability compared to the corresponding results for SPMM explants. Inter- and intra-animal variability, related to local variations of the thickness and composition of the abdominal wall, lead to significant scatter in the measured mechanical response. The mesh component adds a less scattered contribution to the overall mechanical response. Apparently, the stiffer the mesh, the less is the variability, due to the dominance of the mesh component.

There are issues and limitations associated with both uniaxial and biaxial testing. Uniaxial tensile tests were originally developed for metal testing in engineering applications. Commercial machines, equipment, software and protocols are readily available. Uniaxial tensile tests are currently widely used for characterizing mesh implants and explants. The biaxial inflation experiment is applied here for the first time to analyze explants, as the setup and control software for deformation and stress analysis is thus far only available in a few biomechanical laboratories [119, 120]. Protocols and reference experimental settings are not standardized so far. The recorded data, a sequence of images and the corresponding pressure values, are not immediately available for interpretation, but need to be translated in terms of stretch and membrane tension.

The chosen stretch rate of $\sim 1\%s^{-1}$ allows to examine the longterm response of the
explants, reducing the influence of time dependent effects. However, physiologically relevant
dynamic phenomena, such as high rate deformations due to coughing or sneezing cannot be
addressed with the current pump system.

Moreover, the present setup is not appropriate to investigate the rupture behavior of the
explants. Uncontrollable and non-physiological stress concentrations at the clamping site
would lead to early failure and would not allow to evaluate the structural strength of the
explants.

![Graph showing membrane tension versus thickness to diameter ratio t/d.](image)

**Figure 3.10:** Estimated membrane tension versus thickness to diameter ratio t/d. The Laplace
estimate (used in this study) (x) is plotted versus the membrane tension evaluated from an FEM
analysis (mean in-plane stress multiplied by the current thickness of the membrane)(+). For small
ratios t/d, both estimates are similar. The bending stiffness of the sample increases with the
thickness, leading to a linear stress distribution in thickness direction, with negative stress values
(compression) on the inner surface. The average in plane stress approaches zero (+). As the
bending effect cannot be captured by the Laplace assumption, the two solutions diverge. The
thickness of the explants used in this study are estimated as 2mm-2.5mm, locally even 3mm, which
is indicated by the black box (diameter: 35mm). The corresponding deviations are given.

The presented calculation of biaxial membrane tensions is based on the “membrane”
assumption that the thickness of the structure being tested is much smaller than the diameter
of the specimen. However, due to its relatively large thickness, the extracted biaxial values
of membrane tension may be overestimated by up to 20%, as was earlier demonstrated
by a corresponding finite element analysis. In Figure 3.10, the mean simulated inplane
stress multiplied by the current specimen thickness (modeled membrane tension) is plotted
versus the membrane tension obtained from the Laplace model. With increasing thickness
to diameter ratio, t/d, the specimen deformation becomes determined by bending, which is
not captured by the Laplace model. This effect could be reduced by increasing the diameter
of the specimen tested, though that will in turn lead to problems finding an animal model
which is large enough.
Both the biaxial inflation test and the uniaxial tensile test are techniques widely used for the characterization of synthetic materials that actually assume homogeneity. However, in reality the latter is untrue: explants are laminates consisting of inhomogeneous and anisotropic layers. The free edge effect is a phenomenon associated with the uniaxial procedure when applied to such specimens [118] causing stress concentrations and affecting sample stiffness and strength. It is a source of distinct experimental artifacts which is reflected, in our study, by an increased scatter of uniaxial tangent moduli at high tension levels. Uniaxial results are also prone to another problem. The start of the experiment and thus the reference length is determined by a preforce. For the explants under uniaxial load, the relationship between membrane tension and stretch is the typically knee-shaped, non-linear characteristic. The flat slope at the beginning combined with the noise of the force sensor renders the online detection of the preforce hardly reproducible. The start of the experiment, the reference length and the calculated deformation and thus the horizontal position of the tension versus stretch curve are uncertain and non-reproducible properties. For this reason, load dependent (in contrast to deformation dependent) parameters are evaluated in this study, and tangent moduli are extracted rather than an “initial” E-modulus. There are further challenges related to the definition of a preforce, which will be discussed in section 3.6.
3.5 Results: composite study

The findings for biaxial and uniaxial measurements are different for all groups. Biaxial and uniaxial membrane tensions of VHB are by an order of magnitude smaller than those of the mesh VHB composites (Figure 3.11). Biaxial membrane tension versus stretch curves are similar in shape for both mesh types (Figure 3a), whereas uniaxial curve shapes seem to be characteristic for each mesh type (Figure 3.11b).

![Figure 3.11: a Biaxial (BI) and b uniaxial (UNI) tests, mean membrane tension versus stretch curves for all samples of each configuration: Gynemesh M resorbed (GMres), Gynemesh M standard (GMst) and SPMM embedded in VHB. The reference levels of membrane tension, 1.1N/cm and 3.9N/cm, are marked as black dashed lines.](image)

3.5.1 Biaxial data

For the biaxial load case, at both levels of membrane tension (1.1N/cm, 3.9N/cm), SPMM composites are stiffer than Gynemesh M composites (1.1N/cm: GMres 19±2 N/cm, GMst 60±6 N/cm, SPMM 83±7 N/cm, 3.9N/cm: GMres 30±1 N/cm, GMst 50±2 N/cm, SPMM 95±9 N/cm). Gynemesh M composites in the non-resorbed (standard) state (GMst) are stiffer than in the resorbed state (GMres) (Figure 3.12a, c). Comparing the present values with the previous explant tests (Figure 3.12b, d), the same ranking of the meshes is observed. At both tension levels, tangent moduli for explants are nearly twice as large as for mesh VHB composites. However, the ratio of the mean tangent moduli between both mesh types (SPMM, GM) is close for mesh VHB composites and the corresponding explants (at 3.9N/cm: 2.4 versus 2.0). Thus, under biaxial loading conditions, SPMM samples are (more than) twice as stiff as GM samples, both when embedded in a VHB elastomeric matrix and when ingrown in abdominal wall tissue.
3.5.2 Uniaxial data (composites)

For the uniaxial load case, at the low (1.1N/cm) level of membrane tension, SPMM composites are stiffer than Gynemesh M composites (GMres 13±1 N/cm, GMst 18±1 N/cm, SPMM 30±1 N/cm). Contrarily, at the higher (3.9N/cm) level of membrane tension, SPMM composites are more compliant than Gynemesh M composites (GMres 46±1 N/cm, GMst 50±2 N/cm, SPMM 19±1 N/cm). At both membrane tension levels (1.1N/cm, 3.9N/cm), non-resorbed Gynemesh M composites (GMst) are stiffer than resorbed composites (GMres) (Figure 3.13a, c). Except for SPMM composites at the higher tension level (3.9N/cm), the tangent moduli of the mesh VHB composites and the corresponding explants have similar values (Figure 3.13b, d). SPMM composites are twice as stiff as Gynemesh M composites at the low level of membrane tension (1.1N/cm) (ratio between mean stiffnesses of SPMM and GM: 2.0), and less than half as stiff (ratio: 0.4) at the higher level (3.9N/cm). This load level dependent ranking of mesh types, is likewise observed for the means of the uniaxial explant data (1.1N/cm: 1.2, 3.9N/cm: 0.7). At the low membrane tension (1.1N/cm), differences between the explants lack statistical significance (Figure 3.13 b).
3.5 Results: composite study

Figure 3.13: Distribution of tangent moduli for uniaxial (UNI) tests on VHB mesh composites and explants a, c VHB mesh composite samples: Gynemesh M resorbed (GMres), Gynemesh M standard (GMst), SPMM embedded in VHB, b, d explants: GM and SPMM ingrown in rabbits’ abdominal wall. Reference levels of membrane tension: 1.1N/cm (a, b) and 3.9N/cm (c, d). P-values < 0.05 are indicated with corresponding letters g,h,i,j,k,l,K. a g,h,i: p<1e-04, c j: p=0.008, k,l: p<1e-04, d K: p=0.013

3.5.3 Predictions of explant response

For both mesh types (Gynemesh M, SPMM), the proposed superposition of biaxial membrane tensions of the embedded mesh, VHB elastomer and native abdominal wall tissue, provides excellent approximations for the respective explant behavior (Figure 3.14). For Gynemesh M, it is seen that the mechanical response of non-resorbed (GMst) and resorbed (GMres) composite samples are appropriate to estimate upper and lower bounds for the behavior of Gynemesh M explants, with a partly resorbed mesh (Figure 3.14a).

The approximated uniaxial mechanical response for both mesh types is close to the response of the respective explants, especially for low and moderate membrane tension levels up to 5 N/cm (Figure 3.15). In fact, the superposed membrane tension versus stretch curves are nearly identical with the curves of the according mesh VHB composites (compare Figure 3.11b), which are already close to the corresponding explant curves. For both VHB and native tissue, the uniaxial response without the mesh is similar and very compliant within the deformation range of interest (1.0-1.4). The explant response for Gynemesh M is estimated appropriately by upper and lower bounds from the non-resorbed (GMst) and resorbed (GMres) composites.
3 Physiologically relevant mechanical characterization of prosthetic meshes

3.5.4 Uniaxial data (dry meshes)

Finally, the uniaxial response is compared for mesh VHB composites and the corresponding dry mesh samples (Figure 3.16). For Gynemesh M, significant differences are observed with respect to the curve shapes (Figure 3.16 a). The initial stiffness is lower for dry samples, resulting in a sharper L-shape. For SPMM, the curve shapes are similar (Figure 3.16b). In all cases, intersection points are observed for corresponding composite and dry mesh curves. This leads to the apparent paradoxon, that for stretch levels larger than the intersection stretch, the tension required is larger for the dry mesh than for the composite.
3.5 Results: composite study

Figure 3.15: Uniaxial (UNI) superposition of membrane tensions, a Gynemesh M (GM), b SPMM, superposed uniaxial (UNI) membrane tension versus stretch curves (-AW-approx) and corresponding explant curves (-AW). All curves correspond to mean curves.

Figure 3.16: Uniaxial response of mesh VHB composites and dry mesh samples, a Gynemesh M resorbed (GMres) and standard (GMst), b SPMM. Nonphysical intersection points (x) between corresponding curves of mesh VHB composites and dry meshes are observed.
3.6 Discussion: composite study

The results demonstrate that replacing a biological tissue matrix by an elastomer in biaxial and uniaxial loading conditions leads to the same ranking of mesh materials with respect to their tangent stiffness. Relative differences in stiffness between the mesh types are very close to those observed for explants. In contrast to tests on explants, the scatter of the mechanical parameters is reduced distinctly, providing a statistically significant differentiation of meshes for both load cases, at all load levels.

Even more relevant is the finding that the mechanical response of explants can be quantitatively predicted using synthetic materials. The mechanical response of explants is well approximated by the superposition of contributions of the corresponding mesh VHB composites, VHB and native abdominal wall tissue. For the biaxial case, mesh and native tissue can be seen as arranged in parallel. For the uniaxial load case, the mechanical response of the mesh VHB composite is already close to the response of the respective explant. These results confirm our first hypothesis.

In this study, we constructed an idealized unstressed and undeformed reference configuration for the prosthetic meshes, uniformly embedded in a homogeneous matrix. In vivo, however, locally ingrown tissue components, inflammation and scar tissue formation might cause inhomogeneous inelastic mesh deformations, mesh contractions or wrinkles. The exact reference configuration is determined by more complex mesh-matrix-interactions and is hardly reproducible in a representative way. In this sense, the predictive capabilities of the proposed superposition procedure are better than initially expected.

3.6.1 The role of mesh-matrix interaction

Independent of the matrix properties (VHB or AW), mesh-matrix-interaction is identified to determine the uniaxial mechanical response of Gynemesh M composites and explants, which is due to the larger pore size of the mesh. In the absence of such interaction, i.e. for dry Gynemesh M samples, tension versus stretch curves cannot be predictive for the corresponding stiffness values of explants. On the other hand, for composites and explants containing the small porous SPMM, mesh-matrix-interaction plays a less important role. Their uniaxial response is very similar to the one of the corresponding dry mesh. Thus, our second hypothesis is confirmed for light-weight, large porous meshes, such as Gynemesh M, but not for heavy-weight, small porous meshes, such as SPMM. For inflation tests, dry meshes are not applicable for procedural reasons (a closed surface is necessary to create the inflation pressure).
3.6.2 Preforce related artifacts

The unphysical intersection between the uniaxial membrane tension versus stretch curves for mesh VHB composites and the respective dry meshes (Figure 3.16) is related to the introduction of a preforce to define the start of the tests. Its value is commonly determined considering the sensitivity and the noise of the force sensor. All data (tension and stretch) are set to start at this load level. Load versus deformation curves are thus shifted horizontally as illustrated in Figure 3.17 for an arbitrary preforce of 0.25N. The lower the initial slope of the curve, the larger is the horizontal shift. This results in disproportional horizontal curve shifting and nonphysical intersection points for curves with marked difference in initial stiffness, as shown in Figure 3.17. Similar effects can be expected in data previously reported in the literature, e.g. in [73, 78–85, 91].

![Figure 3.17](image-url)

**Figure 3.17:** Implications of a preforce, a) All data are set to start at the level of the preforce (e.g. at 0.25N) and b) are shifted horizontally to start at zero deformation. The curve with the lower initial stiffness (−) is shifted more than the curve with the higher initial stiffness (−−) resulting in the nonphysical intersection point.

3.6.3 Limitations

The validity of biaxial data of Gynemesh M composites is limited up to a tension level of \( \sim 5.0 \, \text{N/cm} \). VHB elastomer was chosen as matrix material for the composites due to its low stiffness. However, applying the inflation pressure leads to local penetrations of the VHB through the large mesh pores, becoming significant for tension values > 5.0 N/cm (Figure 3.18).
3 Physiologically relevant mechanical characterization of prosthetic meshes

Figure 3.18: Validity of biaxial Gynemesh M VHB composites data, VHB elastomer is penetrating through the large pores of the inflated Gynemesh M VHB composite (here shown for resorbed Gynemesh M). Protrusions start to be visible at 5N/cm and distinctly reduce the biaxial stiffness of the Gynemesh M VHB composites at higher levels of membrane tension.

3.6.4 Biaxial testing

The biaxial tensile test [121–123] and the ball burst test [75, 85, 88, 91, 92] represent alternative biaxial test setups which are more commonly applied to characterize either soft biological tissues [121, 122], dry meshes [85, 88, 91, 92], explants [75], or general textiles [123]. In this study, the inflation setup was applied in accordance with the explant study, in order to evaluate the predictive capabilities of mesh elastomer composites with respect to explants. The essential advantage of the inflation setup, which is of particular value for biological tissues and explants, is the simple and reliable control of mechanical boundary conditions. Sample geometry and clamp design for homogeneous load introduction are major challenges associated with biaxial tensile tests [124]. In case of the ball burst test, contact and friction between the ball and the tissue lead to unpredictable interactions.
3.7 Conclusions

The explant study shows that uni- and biaxial tests yield different results with respect to the mechanical behavior of mesh explants. As a consequence, the experimental loading conditions should be selected such that they reproduce the expected in vivo load and are consistent with the animal model used. The mechanical parameters should span the complete physiological range. Furthermore, results are dependent on the analyzed load level.

Bearing in mind the limitations due to the small sample size, and assuming that biaxial testing is more representative because it mimics physiological conditions, the following conclusions are drawn: in the low tension range, Gynemesh M explants are as compliant as native rabbits’ abdominal wall tissue, whereas in the high tension range, they are significantly stiffer. Conversely, SPMM explants are significantly stiffer in the entire tension range. This would mean that mechanically and for biaxial loading conditions, Gynemesh M would be preferable to SPMM. However, no direct quantitative recommendations can be derived for optimization of mesh materials to be used in biaxial loading conditions for the human abdominal wall or pelvic floor. There are currently no reference data for those specimens.

Uniaxial and biaxial tensile tests with dry meshes embedded in an elastomeric matrix lead to the same qualitative conclusions as the corresponding experiments with explants. Moreover, the stiffness of explants can be quantitatively determined based on data obtained from mesh elastomer composites. The proposed procedure allows to identify load case specific deformation mechanisms: The biaxial response of embedded and ingrown meshes is composed of the superposed response of the constituents. For the large porous mesh, the uniaxial response is dependent on mesh-matrix-interaction and less on the matrix material properties. Due to the latter, uniaxial tests using dry meshes are generally not appropriate for estimating the behavior of the ingrown mesh implant.

In contrast to biological tissues, mesh elastomer composites are easy to handle, inexpensive and the test results show moderate scatter. Tests with these materials represent valuable tools for a mechanical investigation and engineering optimization of prosthetic meshes, preceding validation in animal studies.
Uniaxial and biaxial mechanical characterization of a prosthetic mesh including different length scales

**Motivation**  This study is aimed at a comprehensive experimental analysis of the mechanical behavior of a dry prosthetic mesh including different length scales.

**Methods**  Uniaxial and biaxial protocols are presented to evaluate global macroscale mechanical phenomena of the dry mesh. Furthermore, procedures for local deformation analysis and evaluation of corresponding homogenized kinematic measures are described.

**Results**  The global mechanical response of the prosthetic mesh is characterized by anisotropy, a nonlinear force response, hysteresis and preconditioning effects. The local deformation analysis allows to identify mesh specific phenomena due to mesoscale mechanisms.

**Main conclusions**  The presented protocols and data analysis procedures are able to provide a comprehensive picture of the mechanical behavior of the dry mesh, including different mechanical phenomena at the macro- and at the mesoscale. In this way, this study might contribute new aspects to the definition of standardized protocols to be considered for comprehensive mesh product descriptions.

*Parts of this chapter, including paragraphs of text, figures and tables are published in [125].*


4.1 Introduction

Patients’ discomfort, mesh dislocation and erosion are severe complications associated with the transvaginal application of presently used surgical meshes [10]. However, there are indications, that an improved mechanical biocompatibility of such prosthetic meshes is able to positively influence the host response and a smooth integration of the implant [43–45]. The notion of mechanical biocompatibility of implants is rather general and not based on standards. Various aspects might contribute, such as different mesh phenomena in different loading conditions and at different length scales, all of which are to be evaluated in context with a physiological reference.

Experimental studies have been reported characterizing mesh explants from animal models, i.e. the original implant and the surrounding ingrown host tissue, in uniaxial [68,73] and less frequently biaxial [74, 75] loading conditions. These specimens mimic the in vivo state of the implant, however, the biological tissue component increases the scatter of the outcome and, due to mesh-tissue interactions, does not allow to identify mesh specific mesoscale mechanisms. Moreover, sample sizes are usually small, restricting a broad investigation of different mechanical phenomena.

Within the clinical community, several studies on prosthetic meshes before implantation, called dry meshes have been reported. Uniaxial stress [79,87–89,126] and biaxial stress [88, 91,92] loading conditions have been applied to evaluate the deformation (stiffness) [79,91,126] and rupture behavior (maximum elongation or load) [79,87,91,92] of different meshes. Global mesh phenomena, possibly influencing the clinical outcome, such as anisotropy [88, 89], preconditioning effects and hysteresis in cyclic loading conditions [87,92,126] have been investigated. These findings are based on the evaluation of global data, such as actuator force and displacement records. An evaluation of local deformation patterns, at the level of the pores (mesoscale), has not become a standard yet.

Besides, there is a large community, testing and modeling textiles, using advanced test setups and procedures for local deformation analysis [127–133]. In contrast to the mainly knitted prosthetic meshes, woven fabrics are focussed on. Commonly applied testing methods are the picture frame test [127, 128, 133–136] or the bias (45°) extension test [127,130,132,133,135–137], resulting in shear dominated loading conditions with respect to the warp (0°) and weft (90°) direction, the preferred material directions of orthotropy. Moreover, uniaxial [129,131,137] and biaxial [127,128] tensile tests, loading the textile in the warp and weft material directions have been reported. Structural stiffnesses in shear and tension as well as failure behavior are of interest. Unlike in the dry mesh studies, the characterization of textiles focusses on mesoscale phenomena, such as the alignment, interaction and slippage of yarns [127–133, 135–137]. For this reason, digital image correlation has been used to assess the full planar local deformation gradients from an image sequence of the specimens.
during deformation. The local deformation analysis is aimed at assessing the homogeneity of the deformation, evaluating the real material strain, which might differ from the globally imposed strain, observing mesoscale mechanisms and developing and validating physically based numerical models [127–131,136].

In this study, a prosthetic mesh, a knitted fabric, is characterized in uniaxial stress and uniaxial strain (biaxial stress) loading conditions, applied in the two preferred material directions of orthotropy and two off-axis directions. Test protocols are proposed to evaluate global macroscale mechanical phenomena of the mesh, such as anisotropy, the nonlinear force response, hysteresis and preconditioning effects in a way, comparable for all loading conditions. Furthermore, procedures for local deformation analysis and evaluation of corresponding homogenized kinematic measures are described allowing to identify mesh specific phenomena due to mesoscale mechanisms. It will be shown that the local deformation analysis helps to interpret the global outcome which might differ significantly from what is expected for general continua. The relevance of the findings should be discussed in two regards. (i) How do global and local data contribute to develop an appropriate and physically based model formulation for the prosthetic mesh? (ii) How can these findings be used to evaluate aspects of its mechanical biocompatibility?
4.2 Some words on length scales

In the following sections of this and the subsequent chapter, different length scales are used to describe and model the mechanical behavior of the mesh. These length scales include the macroscale, the mesoscale and the microscale. At each length scale, the kinematics can be seen from a global and local point of view, where the global kinematics is the homogenized local kinematics. These homogenized measures are in fact the transition to the next higher scale, and thus belong to both scales. Figure 4.1 shows the assignment of properties to the different scales. The complex knitting pattern of the filaments is associated with the microscale. Material and friction properties of the filaments are microscale properties. This thesis focuses on the meso- and the macroscale. Observations at the microscale are used to guide the definition of the model approach (chapter 5), which is located at the mesoscale. A system of ten rigid bodies connected by force elements abstracts one representative unit cell of the mesh in a physical way. The kinematics and forces for each rigid body are mesoscale properties and describe the mesoscale mechanics. The homogenized kinematics of each unit cell are seen as boundary conditions of the mesoscale kinematics and simultaneously as the macroscale kinematics. The force response per unit cell and the homogenized unit cell deformation describe the macroscale mechanics. Averaging (homogenizing) the unit cell deformation for the whole specimen, results in the global macroscale deformation.

The experimental study in chapter 4 characterizes the mesh at the macroscale. Experiments at the mesoscale are challenging and require a large number of observations as only a small and, in general, not representative part of the mesh can be examined during one experiment. A local analysis of the macroscale kinematics will be shown to provide a valuable basis to develop a model at the mesoscale, which in turn explains specific phenomena of the macroscale kinematics.
4.2 Some words on length scales

Figure 4.1: Notation of different length scales used in this study. The microscale is represented by the complex knitting pattern of the filaments. A multibody system abstracting the knitting pattern of one unit cell is referred to as the mesoscale. The homogenized kinematics and force of one unit cell describe the mechanics at the macroscale.
4 Mechanical characterization of a prosthetic mesh at different length scales

4.3 Methods

4.3.1 Material

In this study, the prosthetic mesh Gynemesh M, as described in chapter 3, section 3.2.1, was experimentally characterized (Figure 4.2 a). The mesh was used in its resorbed state, containing only polypropylene fibers (Figure 4.2 b). All materials tested came from the same production lot and were provided by the manufacturer.

![Figure 4.2: Gynemesh M. a A material coordinate system \((\vec{e}_x, \vec{e}_y, \vec{e}_z)\) was introduced, according to the preferred material directions of orthotropy. One pore is marked by the white rectangle. b One pore of Gynemesh M (black rectangle) in the resorbed state, i.e. without the polyglecaprone fibers.](image)

The mesh was regarded as a two-dimensional structure. The thickness direction was not considered. The in-plane material properties of the mesh are orthotropic [109], as indicated in Figure 4.2 a. A material coordinate system \((\vec{e}_x, \vec{e}_y, \vec{e}_z)\) was introduced: The direction of the blue lines, which serve for orientation only, is the stiffest material direction, called \(\vec{e}_y\), the orthogonal direction is the most compliant direction, called \(\vec{e}_x\), and \(\vec{e}_z = (0, 0, 1)^T\) is the out-of-plane direction. Tests were conducted in four material directions defined with respect to an inertial machine coordinate system \((\vec{e}_1, \vec{e}_2, \vec{e}_3)\) (Figure 4.3): The direction of load application was called \(\vec{e}_2\), the transverse direction \(\vec{e}_1\), the out-of-plane direction \(\vec{e}_3 = (0, 0, 1)^T\). The angle \(\alpha\) was introduced between \(\vec{e}_1\) and \(\vec{e}_x\), characterizing the loaded material directions: the two preferred material directions of orthotropy \(\alpha = 0^\circ, \alpha = 90^\circ\) and two off-axis directions \(\alpha = 33.5^\circ, \alpha = 56.5^\circ\), which were chosen with respect to the knitting pattern of the mesh (Figure 4.3 b). In the following sections, specimens are referred to by the corresponding loaded material direction \((\alpha)\), such as the \(0^\circ\)-material direction, meaning specimens loaded in the \(0^\circ\) material direction.
4.3 Methods

4.3.2 Loading conditions

Two types of uniaxial tensile tests, called uniaxial strain and uniaxial stress (Figure 4.4), were performed. The case of uniaxial strain is characterized by only one non-vanishing in-plane principle strain direction $E_{22}$, and a constrained transverse contraction, $E_{11} = 0$. The corresponding state of stress is biaxial, $T_{11} \neq 0, T_{22} \neq 0$. It is realized by a high aspect ratio $w/l$ (width/length) (Figure 4.4 a).

The case of uniaxial stress is characterized by only one non-vanishing in-plane principle stress direction $T_{22}$, and a free transverse contraction, $E_{11} \neq 0$. The corresponding state of strain is biaxial, $E_{11} \neq 0, E_{22} \neq 0$. It is realized by a small aspect ratio $w/l$ (Figure 4.4 b).

4.3.3 Specimens

For the two loading conditions and the four loaded material directions, eight different types of specimens were prepared. Approximate global specimen dimensions were chosen as $w = 100\, mm$, $l = 15\, mm$ ($w/l = 6.67$) for uniaxial strain experiments and $w = 20\, mm$, $l = 150\, mm$ ($w/l = 0.13$) for uniaxial stress experiments. However, as the mesh is a discrete structure, rather than the absolute dimensions, the number of discrete load bearing constituents, i.e. strands, was used to quantify the specimens’ dimensions. Therefore, a characteristic unit length $UL$ was introduced, representing the orientation dependent width of one mesh pore, containing two strands (Figure 4.5 a). For the off-axis directions, determining the $UL$ was not straightforward as the load carrying elements could not be followed from clamp to clamp.
4 Mechanical characterization of a prosthetic mesh at different length scales

Figure 4.4: Two types of uniaxial tensile tests. a) Uniaxial strain (biaxial stress) loading conditions, with in-plane strain components $E_{11} = 0$, $E_{22} \neq 0$, in-plane stress components $T_{11} \neq 0$, $T_{22} \neq 0$. b) Uniaxial stress (biaxial strain) loading conditions, with in-plane strain components $E_{11} \neq 0$, $E_{22} \neq 0$, in-plane stress components $T_{11} \neq 0$, $T_{22} = 0$.

The $UL$ was linearly interpolated between the limit values for the $0^\circ$ and $90^\circ$ directions, as shown in Figure 4.5 b. The size of the $UL$ and the specimen width in terms of $UL$s are given in Table 4.1 for each configuration. The width in terms of $UL$s was used to normalize the applied force.

<table>
<thead>
<tr>
<th>orientation $^\circ$</th>
<th>$UL$ [mm]</th>
<th>width $[UL]$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>2.80</td>
<td>36</td>
</tr>
<tr>
<td>90</td>
<td>4.30</td>
<td>23</td>
</tr>
<tr>
<td>33.5</td>
<td>3.36</td>
<td>28.87</td>
</tr>
<tr>
<td>56.5</td>
<td>3.74</td>
<td>26.74</td>
</tr>
</tbody>
</table>

4.3.4 Test setup

All experiments were performed on the custom-made test setup (Figure 4.6), described in chapter 3, section 3.2.6. For the uniaxial stress tests, the extra 20N calibration of the load cells (SMT S-Type, Interface Inc., Scottsdale, AZ, USA) was applied for force measurements. Custom-made clamps equipped with sandpaper at the clamping faces were directly attached to the load cells (Figures 4.6, 4.7 b). The specimens were mounted outside the setup as shown in Figure 4.7 a, by means of a supporting system for controlled positioning and in order to
avoid slacking. The contactless measurement of the in-plane displacement field in the central region of the sample was done as reported in chapter 3, section 3.2.6. A luminescent screen was placed below the specimen to improve the contrast and the brightness of the recorded images.

### 4.3.5 Test protocol

From preliminary tests with Gynemesh M, apart from anisotropy, different macroscale mechanical phenomena were identified: time dependent effects, such as creep and relaxation, inelastic behavior, such as load history dependent preconditioning effects and dissipative behavior (hysteresis). The herein applied test protocols were chosen to capture the nonlinear, anisotropic force response and preconditioning and hysteresis in cyclic load.

Cyclic tests were performed, straining the specimens at a constant nominal strain rate of 0.005s\(^{-1}\) up to a load case specific maximum force per UL, \(F_{\text{max}}\) (1.6N/UL for uniaxial strain, 0.6N/UL for uniaxial stress), and back. This strain rate was considered slow enough to provide the long term mechanical response. Specimens were preconditioned during ten such cycles. The test consisted of another three cycles (Table 4.2 for uniaxial strain, Table 4.3 for uniaxial stress). Three measurements were conducted per specimen type (material direction and loading conditions). For the uniaxial strain tests, one actuator was fixed and the other one was displaced. During the test, the whole length of the specimen remained within the field of the camera. For the uniaxial stress tests, both actuators were displaced symmetrically apart, such that the midregion of the specimen remained within the field of the camera. Force
data were acquired at a sample rate of 157.54Hz. Images were recorded at a rate of 1Hz.

Table 4.2: Test protocol for uniaxial strain loading conditions

<table>
<thead>
<tr>
<th>cycles</th>
<th>actuator velocity [mms$^{-1}$]</th>
<th>nominal strain rate [s$^{-1}$]</th>
<th>$F_{max}$ [N/UL]</th>
</tr>
</thead>
<tbody>
<tr>
<td>preconditioning</td>
<td>10</td>
<td>0.075</td>
<td>0.005</td>
</tr>
<tr>
<td>test</td>
<td>3</td>
<td>0.075</td>
<td>0.005</td>
</tr>
</tbody>
</table>

Table 4.3: Test protocol for uniaxial stress loading conditions

<table>
<thead>
<tr>
<th>cycles</th>
<th>actuator velocity [mms$^{-1}$]</th>
<th>nominal strain rate [s$^{-1}$]</th>
<th>$F_{max}$ [N/UL]</th>
</tr>
</thead>
<tbody>
<tr>
<td>preconditioning</td>
<td>10</td>
<td>0.75</td>
<td>0.005</td>
</tr>
<tr>
<td>test</td>
<td>3</td>
<td>0.75</td>
<td>0.005</td>
</tr>
</tbody>
</table>

For the analysis of the data, the preconditioning procedure was primarily seen as a preparation of the specimens, not as part of the test. The primary data analysis was focused on the three test cycles, which (after preconditioning) were perfectly reproducible and thus only the first one was considered. For the analysis of load history dependence, the first cycle of preconditioning was examined separately and according to the same procedures.

4.3.6 Local strain analysis

The recorded image sequence was used for a local strain analysis using VEDDAC, a software for digital image correlation (Chemnitzer Werkstoffmechanik GmbH, Chemnitz, Germany). The first sharp image, when the slightly slacking specimen was lifted to the focus of the
4.3 Methods

a) The clamping procedure for uniaxial strain specimens was performed outside of the machine: Top: The lower parts of the clamps, equipped with sandpaper were positioned within a support. The specimens were placed tension free on the clamps. Bottom: The upper parts of the clamps were fixed within the support and screwed to the lower parts. The support was used to mount these clamps within the clamping system installed at the load cells of the machine. b) The clamped specimen. Additional clamps were used to guarantee proper fixation over the whole specimen width.

Figure 4.7: Clamping of the specimens. a) The clamping procedure for uniaxial strain specimens was performed outside of the machine: Top: The lower parts of the clamps, equipped with sandpaper were positioned within a support. The specimens were placed tension free on the clamps. Bottom: The upper parts of the clamps were fixed within the support and screwed to the lower parts. The support was used to mount these clamps within the clamping system installed at the load cells of the machine. b) The clamped specimen. Additional clamps were used to guarantee proper fixation over the whole specimen width.

camera, was defined as the preliminary reference configuration (Figure 4.8 a). All subsequent images were defined as a deformed or current configuration. Note, that this preliminary reference configuration was not determined by any mechanical criterion (force or kinematics), but was dependent on the clamping of the specimen and on the rate of image acquisition. Measurement points were inserted on the images at the nodes of the mesh, as shown in Figure 4.8 a. The evaluation of the in-plane displacements of these points was based on a greyscale correlation algorithm (Figure 4.8 b). The output consisted of the history of the current in-plane position (1 and 2 component) over time for all measurement points, stored in respective data arrays.

In order to standardize (post-hoc) the start of each test by a reference configuration, a normalized preforce of 1/100 of $F_{max}$, i.e. 0.016N/$UL$ for uniaxial strain tests and 0.006N/$UL$ for uniaxial stress tests was determined. To avoid artifacts due to the noise of the force data, a moving polynomial function (degree 2) was fitted locally to the force versus time curve to smooth the curve and evaluate the starting index. The position arrays were interpolated to the sample rate of the force data. All raw data, the normalized force (and time) arrays and the position arrays for all measurement points, were cut at the starting index, i.e. at the preforce level.

The cut position data arrays were used to define, two directed line elements at the edges of each half pore for each timepoint $t$: $k\vec{G}_1$ and $k\vec{G}_2$ ($k\vec{G}_3 = (0, 0, 1)^T$) for the $k$th half pore in
the reference configuration and \( k\vec{g}_i(t) \) and \( k\vec{g}_2(t) = (0, 0, 1)^T \) for the \( k \)th half pore in the current configuration, respectively, as visualized in Figure 4.8. Note, that in practice, not the original images were used, but the interpolated and cut data. The dependence on \( t \) for the current configuration is omitted in the following. \((k\vec{G}_1, k\vec{G}_2, k\vec{G}_3)\) and \((k\vec{g}_1, k\vec{g}_2, k\vec{g}_3)\) form curvilinear bases in the reference and current configuration, with \( kG^{1/2} = k\vec{G}_1 \times k\vec{G}_2 \cdot k\vec{G}_3 \) and \( kg^{1/2} = k\vec{g}_1 \times k\vec{g}_2 \cdot k\vec{g}_3 \) and the two metrices \( kG_{ij} = k\vec{G}_i \cdot k\vec{G}_j = kG_{ji} \) and \( kg_{ij} = k\vec{g}_i \cdot k\vec{g}_j = kg_{ji} \).

Figure 4.8: Local strain analysis shown for the 0°-material direction in uniaxial strain loading conditions. a) Reference configuration with measurement points: Two directed line elements \( \vec{G}_1 \) and \( \vec{G}_2 \) (\( \vec{G}_3 = (0, 0, 1)^T \)) were defined at the edges of each half pore. Note: the shown image corresponds to the first sharp image of the sequence and was used as a preliminary reference configuration. The final reference configuration was determined by the preforce. b) Deformed configuration: By means of digital image correlation, the current positions of all measurement points were evaluated defining the displacement field (shown by the yellow/bright lines). The two directed line elements in the reference (\( \vec{G}_1 \) and \( \vec{G}_2 \) (\( \vec{G}_3 = (0, 0, 1)^T \))) and in the current configuration (\( \vec{g}_1 \) and \( \vec{g}_2 \) (\( \vec{g}_3 = (0, 0, 1)^T \))) were used to define the homogenized pore deformation (deformation gradient).

For each current configuration, i.e. timepoint \( t \), the change of the angle between these line elements (base vectors) and the change of length of these line elements with respect to the reference configuration, describe the complete local deformation gradient \( kF \), the local right Cauchy-Green deformation tensor \( kC \) and the local Green-Lagrange deformation tensor \( kE \) of the \( k \)th half pore [138]. These measures quantify the deformation of each half pore in a homogenized sense.

\[
kF = k\vec{g}_i \otimes k\vec{G}_i, \quad kC = kF^TkF, \quad kE = \frac{1}{2}(kC - I)
\]
\[ k \vec{G}_1 = k G^{-1/2}(k \vec{G}_2 \times k \vec{G}_3), \quad k \vec{G}_2 = k G^{-1/2}(k \vec{G}_3 \times k \vec{G}_1), \quad k \vec{G}_3 = k G^{-1/2}(k \vec{G}_1 \times k \vec{G}_2) \]

For the index i=1,2,3, the Einstein summation convention is to be used. Lower indices mark covariant base vectors and upper indices contravariant base vectors. \( I \) is the identity matrix. The components of the deformation tensors, e.g. \( E \), in the machine basis \((\vec{e}_1, \vec{e}_2, \vec{e}_3)\) were derived according to

\[ k E_{ij} = k E \cdot (\vec{e}_i \otimes \vec{e}_j) \]

with \( i,j=1,2,3 \).

From a macroscopic point of view, these are local measures of deformation, as they refer to each half pore. They were homogenized for each specimen by averaging component wise over all pores \( n_p \).

\[ F_{ij} = \frac{1}{n_p} \sum_{k=1}^{k=n_p} k F_{ij}, \quad C_{ij} = \frac{1}{n_p} \sum_{k=1}^{k=n_p} k C_{ij}, \quad E_{ij} = \frac{1}{n_p} \sum_{k=1}^{k=n_p} k E_{ij} \quad i,j = 1,2,3 \]

### 4.3.7 Representation of data

For each configuration (loading condition and material direction), the arrays of strain components and the corresponding normalized force arrays were interpolated to the same number of data points and averaged over the three specimens tested. The inter-specimen reproducibility was assessed by the standard deviation of the homogenized longitudinal strain \( E_{22} \) for the three specimens at maximum force (1.6\(N/UL \) for uniaxial strain, 0.6\(N/UL \) for uniaxial stress) normalized by the mean value.

The distribution of the two dimensional strain components over each specimen in the machine coordinate system were visualized by strain maps discretized on the level of the half pores. The standard deviation of the local longitudinal strain component \( (E_{22}) \) over each specimen is a measure for the inhomogeneity or intra-specimen variability of \( E_{22} \). The averaged standard deviation for the three specimens normalized by the averaged mean value was calculated for each loading configuration at maximum load (1.6\(N/UL \) for uniaxial strain, 0.6\(N/UL \) for uniaxial stress), to assess the intra-specimen variability of the deformation, specifically for each load case and material direction. The deformation mechanisms of the pores were evaluated and compared for all material directions (and loading conditions) in the local curvilinear coordinate system spanned by the vectors \( \vec{G}_1, \vec{G}_2 \) and \( \vec{G}_3 \) or \( \vec{g}_1, \vec{g}_2, \vec{g}_3 \), respectively. The lengths of and the angle between these directed line elements were averaged over all half pores of one specimen and for the three specimens of each configuration. These averaged, local deformation data were compared at the preforce level and at maximum load.

For a comparison of the angles \( \angle (\vec{G}_1, \vec{G}_2) \) or \( \angle (\vec{g}_1, \vec{g}_2) \), respectively, an auxiliary line
element $\vec{G}_1$ ($\vec{g}_1'$) was introduced for the $90^\circ$- and the $56.5^\circ$ material directions, as shown in Figure 4.9: The vectors $\vec{G}_1$ and $\vec{G}_2$ ($\vec{g}_1$ and $\vec{g}_2$) were defined, such that the resolution of strain was highest in the loading direction $\vec{e}_2$ for all material directions. Thus, the discretization of one pore was different for the $90^\circ$- and the $56.5^\circ$ material directions as compared to the $0^\circ$- and $33.5^\circ$ directions and had to be adapted for a comparison.

The mesoscale mechanisms evaluated through these procedures are discussed with respect to their clinical relevance. A more detailed discussion of the outcomes, mainly with respect to methodological aspects can be found in Appendix B.

**Figure 4.9:** Directed line elements for all material directions. $\vec{G}_1, \vec{G}_2$ (and $\vec{g}_1, \vec{g}_2$) were defined, such that the resolution was highest in the loading direction $\vec{e}_2$. In order to quantitatively compare the kinematic measure $\angle (\vec{G}_1, \vec{G}_2)$ ($\angle (\vec{g}_1, \vec{g}_2)$), for all material directions, the vector $\vec{G}_1'$ ($\vec{g}_1'$) was introduced, replacing $\vec{G}_1$ ($\vec{g}_1$), for the $90^\circ$ and the $56.5^\circ$ material directions. Note, that the fact that $(\vec{G}_1, \vec{G}_2, \vec{G}_3)$ and $(\vec{g}_1, \vec{g}_2, \vec{g}_3)$ are not right handed systems, does not influence the calculation of the in-plane angle.
4.4 Results

The experimental outcome is represented by macroscale data of force and deformation. Local data are derived at the level of each half pore. The homogenized local data for each specimen are referred to as global data.

4.4.1 Global force response per unit cell

For both loading conditions, uniaxial strain and uniaxial stress, and all material directions, a nonlinear global force response is observed (Figure 4.10). The initial stiffness, quantified as the secant modulus at $E_{22} = 0.01$, was by a factor of three larger for the case of uniaxial strain ($0^\circ$: 2.3N/UL, $90^\circ$: 1.4N/UL, $33.5^\circ$: 2.3N/UL, $56.5^\circ$: 2.0N/UL) compared to uniaxial stress ($0^\circ$: 0.6N/UL, $90^\circ$: 0.5N/UL, $56.5^\circ$: 0.6N/UL), except for the $33.5^\circ$ material direction (2.5 N/UL in uniaxial stress loading conditions), where the initial stiffnesses were similar.

![Graph](https://via.placeholder.com/150)

**Figure 4.10:** Global force response for all tested material directions. The force normalized by the number of ULs is shown versus the longitudinal Green-Lagrange strain component, homogenized for each specimen and averaged over the three specimens tested, for (a) uniaxial strain and (b) uniaxial stress loading conditions.

The different material directions show distinct anisotropy. For the case of uniaxial strain, the stiffest direction is the $0^\circ$ material direction, the most compliant the $90^\circ$ material direction and the off-axis directions are in between, with the $33.5^\circ$ direction being stiffer than the $56.5^\circ$ direction. This order would also be expected for an orthotropic continuum. However, for the case of uniaxial stress, the $33.5^\circ$ material direction is the stiffest direction. For all specimens, hysteresis in cyclic loading conditions is observed. Note that only the force response in longitudinal direction is shown. For specimens in off-axis material directions, also resultant reaction forces in transverse direction occur, which cannot be measured with the
current setup.

For the 0°- and the 90° direction, the data were reliably reproducible. The maximum relative deviation of longitudinal strain at the turning point was 4% for the 0° direction and 2% for the 90° direction. For the off-axis directions, this deviation was 6% (33.5°) -9% (56.5°), a phenomenon which is discussed in section 4.5.

4.4.2 Global unit cell kinematics

Although the transverse contraction is constrained for the case of uniaxial strain, small transverse deformations are seen for all material directions, almost vanishing for the 0° direction (Figure 4.11 a). At the maximum load, they remain smaller than 0.015 negative strain. For the case of uniaxial stress, large transverse contractions are observed reaching 0.4-0.45 negative strain at the turning point (Figure 4.11 b). For the 33.5° material direction, significantly smaller transverse deformations in the range of 0.2 are seen. Moreover, for this material direction, the transverse behavior is different for the loading and unloading path (kinematic hysteresis).

![Figure 4.11](image)

**Figure 4.11:** Global kinematic response for all tested material directions. The transverse Green-Lagrange strain component is shown versus the longitudinal Green-Lagrange strain component, homogenized for each specimen and averaged over the three specimens tested, for **a** uniaxial strain and **b** uniaxial stress loading conditions.

4.4.3 Local unit cell kinematics

Figures 4.12 and 4.13 show the local distributions of all in-plane strain components with respect to the machine coordinate system for uniaxial strain and uniaxial stress loading conditions. Here, local refers to the discretization of the specimen by half pores. The averaged standard deviation normalized by the averaged mean value for the three specimens
is shown in Figure 4.14. Inhomogeneous deformation is more pronounced in uniaxial strain (intra-specimen variabilities between 22-35%) compared to uniaxial stress loading conditions (intra-specimen variabilities between 14-26%). In both loading conditions, for the 0° material direction, the deformation is rather homogeneous, whereas in particular the off-axis specimens show local inhomogeneities for all strain components. For the 90° material direction, inhomogeneous patterns in longitudinal strain ($E_{22}$), ranging over several half pores, are observed, represented by transverse stripes. Longitudinal stripes are seen in the distribution of transverse strain ($E_{11}$) for the 33.5° direction in uniaxial stress loading conditions.

Table 4.4 summarizes the averaged lengths of and the angle between the directed line elements in the reference (at the preforce level) and the maximum deformed (at the turning point) configurations. For uniaxial strain loading conditions, the main kinematic mechanism is stretch of the line elements. Changes of the in-plane angle are small. For uniaxial stress loading conditions, except for the 33.5° material direction, the kinematics are determined by a distinct change in angle, a collapse of the pores. All off-axis directions are accompanied by distinct distortions, seen in the asymmetric stretch of the line elements. These deformation patterns, specific for each loading configuration, are due to structural mechanisms at the mesoscale, which are examined in more detail in chapter 5.

Figure 4.12: Local macroscale kinematic response in uniaxial strain loading conditions: The longitudinal component $E_{22}$, the transverse component $E_{11}$ and the shear component $E_{12}$. The results are shown for one representative specimen at the turning point (maximum load). The color of each triangle represents the homogenized strain component of one half pore.
Figure 4.13: Local macroscale kinematic response in uniaxial stress loading conditions: The longitudinal component $E_{22}$, the transverse component $E_{11}$ and the shear component $E_{12}$. The results are shown for one representative specimen at the turning point (maximum load). The color of each triangle represents the homogenized strain component of one half pore.

Figure 4.14: Mean variability (of three specimens) of the local longitudinal strain component $E_{22}$ normalized by the mean value, calculated at maximum force ($1.6N/UL$ for uniaxial strain, $0.6N/UL$ for uniaxial stress). This measure of inhomogeneity is higher for uniaxial strain than for uniaxial stress loading conditions. With respect to the material orientation, the level of inhomogeneity is lowest for the $0^\circ$ material direction, followed by the off-axis directions $33.5^\circ$ and $56.5^\circ$. The $90^\circ$ material direction shows the highest level of inhomogeneity.
Table 4.4: Preconditioned reference configurations (level of the preforce) and the configurations at maximum force (1.6N/UL for uniaxial strain, 0.6N/UL for uniaxial stress), norm (=length) of and angle between the two directed line elements $\vec{G}_1, \vec{G}_2$ and $\vec{g}_1, \vec{g}_2$, respectively.

<table>
<thead>
<tr>
<th>Reference configuration</th>
<th>Configuration at maximum load</th>
</tr>
</thead>
<tbody>
<tr>
<td>$</td>
<td></td>
</tr>
<tr>
<td>uniaxial strain</td>
<td></td>
</tr>
<tr>
<td>0$^\circ$</td>
<td>2.53</td>
</tr>
<tr>
<td>90$^\circ$</td>
<td>2.59</td>
</tr>
<tr>
<td>33.5$^\circ$</td>
<td>2.56</td>
</tr>
<tr>
<td>56.5$^\circ$</td>
<td>2.70</td>
</tr>
<tr>
<td>uniaxial stress</td>
<td></td>
</tr>
<tr>
<td>0$^\circ$</td>
<td>2.53</td>
</tr>
<tr>
<td>90$^\circ$</td>
<td>2.40</td>
</tr>
<tr>
<td>33.5$^\circ$</td>
<td>2.87</td>
</tr>
<tr>
<td>56.5$^\circ$</td>
<td>2.52</td>
</tr>
</tbody>
</table>

*a* for the 90$^\circ$ and the 56.5$^\circ$ material directions, $\vec{G}_1'$ ($\vec{g}_1'$) were used instead of $\vec{G}_1$ ($\vec{g}_1$).

4.4.4 Global force response per unit cell of the virgin mesh

In Figure 4.15, the global force response per unit cell is shown for the first cycle of preconditioning, i.e. for the mesh loaded from its virgin state. In uniaxial strain loading conditions, the global force response is qualitatively similar for the virgin and the preconditioned meshes (compare Figures 4.15 a and 4.10 a). For the latter, the curves are shifted horizontally to the left. Note, that the axes are scaled differently. For uniaxial stress loading conditions, extreme differences are seen between these two states, in particular for the 90$^\circ$ and 56.5$^\circ$ material directions, where the longitudinal strain $E_{22}$ at maximum load differs by a factor of 4-6 (compare Figures 4.15 b and 4.10 b). Moreover, for the 33.5$^\circ$ material direction, the curve shapes are distinctly different.

Using again Table 4.4 and comparing the data for the preconditioned reference configuration (first three data columns) to the data of the virgin (initial) configuration in Table 4.5, the following is observed: the same mesoscale mechanisms as described in section 4.4.3 for the deformation in cyclic load (stretch of the line elements in uniaxial strain, collapse of pores in uniaxial stress, distortion of pores for off-axis material directions), must have been activated during preconditioning and led to respective inelastic deformations.

Table 4.5: Initial configuration, norm (=length) of and angle between the two directed line elements $\vec{G}_{1\text{virgin}}, \vec{G}_{2\text{virgin}}$ in the virgin unloaded configuration.

| $||\vec{G}_{1\text{virgin}}||$ [mm] | $||\vec{G}_{2\text{virgin}}||$ [mm] | $\angle (\vec{G}_{1\text{virgin}}, \vec{G}_{2\text{virgin}})$ |
|----------------------------------|----------------------------------|----------------------------------|
| 2.60                             | 2.60                             | 64                               |
Figure 4.15: First cycle of preconditioning, global force response for all tested material directions. The force normalized by the number of ULs is shown versus the longitudinal Green-Lagrange strain component, homogenized for each specimen and averaged over the three specimens tested, for a uniaxial strain and b uniaxial stress loading conditions.
4.5 Discussion

An experimental study was performed, characterizing the mechanical behavior of the prosthesis, Gynemesh M, in the resorbed state, in eight different loading configurations (two loading conditions, four different material directions). The observed global macroscale mesh phenomena include a nonlinear force response, anisotropy, hysteresis and pronounced preconditioning effects. By means of the local deformation analysis, configuration-specific deformation patterns were identified, which are due to mesoscale mechanisms. Moreover, local structural phenomena were observed, which are also due to mesoscale mechanisms and are not seen for a homogeneous continuum (discussed in section 4.5.1).

From a theoretical mechanical point of view, a numerical simulation of prosthetic meshes requires the choice of an appropriate modeling approach. Do meshes behave as a continuum or as a structure? Which phenomena have to be included and which can be omitted without affecting the predictive capabilities of the model? A broad experimental basis, as provided by this study, helps to clarify these questions and offers data for fitting the respective model parameters (chapter 5).

The various mechanical phenomena identified, but also time-dependent effects, which are not treated here, are to be considered in mesh design optimization. Is an anisotropic and hysteretic behavior wanted or just a result of the manufacturing process? Which global, which local mechanical behavior is desired to improve the mechanical biocompatibility of prosthetic meshes? Large porous meshes are often preferred, as they appear more compliant compared to small porous meshes. However, the standard mechanical test, is uniaxial. Biaxial initial stiffnesses of the present mesh are by a factor of three larger than corresponding uniaxial values. A broad knowledge of the mechanical behavior in different loading conditions and at different length scales is necessary, in order to answer these questions.

4.5.1 Local phenomena due to mesoscale mechanisms

The mesh is assumed as an orthotropic structure based on periodically arranged patterns (pores), which are symmetric with respect to the preferred material directions \((0^\circ, 90^\circ)\). The evaluation of global, homogenized experimental data implicitly treats the mesh in a smeared, continuum-like sense. Local phenomena were observed, which require to additionally examine structural mechanisms at the mesoscale, as these might violate the homogeneity assumption, the orthotropy assumption and the periodicity assumption. These are global mesh distortions, local mesh distortions, in particular edge effects, size effects and structural anisotropy.

Parts of the virgin mesh were seen to be globally distorted, i.e. distortions included several pores. Global mesh distortions violate the orthotropy assumption. Strain in one of
the two preferred material directions of orthotropy (0° and 90°) causes unexpected shear deformations and non-desired reaction force components in transverse direction. Global distortions were mainly located at the edge regions of mesh sheets. With respect to self-tailored meshes in clinical applications, it is recommended that implants are cut from the mid region of the sheets.

Large intra-specimen variabilities of deformation were observed, ranging from 15%-35%. In [131], locally inhomogeneous deformations have been explained by local deviations from the periodic pattern. However, another reason has been identified: For all material directions except for the 0° direction, filaments end at the free edges, which are pulled into the specimen during load application. Such edge effects are seen as stripes pattern for the 90° material direction and as less systematic inhomogeneous deformation patterns for the off-axis directions (compare Figures 4.12 and 4.13). Inhomogeneous, mechanism-like deformations lead to a local mismatch of deformation between the implant and the underlying tissue. Consequences might include impairment of implant integration, relative movement, wrinkling and tissue injury representing severe mesh-related complications. Edge effects can be avoided either by only using implants cut in 0° material direction or by sealing the free edges, which is done for mesh kits.

Global mesh distortions as well as cutting the mesh in an off-axis direction (a direction other than 0° and 90°), impedes the structural reproducibility, as the number of load carrying strands per width is likely to vary between the mesh specimens. The common dimensions of mesh strips used for implantation or the dimensions of the anchoring arms in mesh kits are of a similar order of magnitude as the pores or ULs (10-20mm vs 3-4mm). Therefore, structural variations do matter and cause size effects, reflected as inter-specimen variability of the mechanical response, which, in this study, was up to 9%. Consequently, with respect to slender implant strips or the anchoring strips in mesh kits, the number of load carrying strands, rather than the global dimensions determine the mechanical response which has to be considered when tailoring a mesh or designing a mesh kit. In order to avoid such hardly controllable size effects, implant dimensions ideally should be much larger than one UL, in the range of 100ULs, or slender geometries should be cut in 0° or 90° material directions, by respecting the pore structure.

In particular mesh strips in 33.5° material direction offer a variety of additional phenomena. Their preconditioned and virgin force response was seen to be significantly stiffer compared to the other material directions (compare Figures 4.10 b and 4.15). This is due to mechanisms activated by the specific knitting pattern and can be seen as structure specific anisotropy. Load application in this direction, does not activate compliant collapse mechanisms, but rather stretch of the stiff strands (compare the angles \( \angle (\vec{G}_1, \vec{G}_2) \) and \( \angle (\vec{g}_1, \vec{g}_2) \) in Table 4.4). Several stitches are oriented transversely, internally restricting the globally free contraction. As a result, out-of-plane warping is observed. This unexpectedly
stiff response in the $33.5^\circ$ material direction is to be considered when using slender strips of seemingly arbitrary direction, such as in mesh kits. A different mechanical response of anchoring strips with different material directions might lead to mesh dislocation and loss of support. Locally, out-of-plane warping leads to a three-dimensional deformation mismatch between mesh and underlying tissue, which might be responsible for an impairment of implant integration, tissue injury or erosions.

Due to both size effects and structural anisotropies, for clinical use, it is recommended not to apply self-tailored meshes or mesh kits with “arbitrarily” oriented anchoring strips. An alternative design of mesh kits, where all anchoring strips have the same material orientation and a structure which is optimized for uniaxial stress loading conditions, might improve the mechanical biocompatibility of mesh kits at the macro- and at the mesoscale.

There is a need for an identification and even quantification of mesh specific mesoscale mechanisms to take them into account for clinical mesh application procedures as well as for mesh design optimization. The local strain analysis has been proven to be a useful tool to meet this need.

4.5.2 Preconditioning - a change of material?

For biological tissues, it is known that a preconditioning procedure is necessary to align the tissue microstructure and to obtain a reproducible mechanical response [139]. This process is reversible since it recovers with time. The preconditioned response is softer in the initial phase as the resistance of the fibrous tissue components is reduced. The preconditioned stress-free configuration (aligned) is different compared to the virgin reference configuration.

For the meshes, a comparable even though not the same behavior is observed. The knitting pattern is characterized by interlooping filaments. First load application leads to a relative movement and rearranging of the filaments, processes which are determined by the inter-filament friction. Unloading cannot reverse these processes. After the preconditioning cycles, the specimens are sagging between the clamps. The introduction of a preforce shifts the start of the test to the beginning of the recruitment of the structure. This results in an apparently stiffer initial force response as the filaments are already aligned and explains the different levels of deformation reached for virgin and preconditioned meshes. The preconditioned reference configuration (aligned) is significantly different compared to the virgin reference configuration (compare Tables 4.4 and 4.5). From a global point of view, the different reference lengths of the specimens (Table 4.6) clearly illustrate this effect.

The final experimental force deformation curve, i.e. the mechanical behavior is apparently dependent on the choice of an appropriate reference configuration (preforce) and the preconditioning procedure (material direction, cyclic force or deformation limit). The question is justified, whether the preconditioned mesh has to be seen as a different material.
Table 4.6: Reference lengths for the virgin and the preconditioned configuration. There are distinct differences between these reference lengths and compared to the initial specimen lengths of 15mm for uniaxial strain and 150mm for uniaxial stress loading conditions.

<table>
<thead>
<tr>
<th></th>
<th>virgin reference length [mm]</th>
<th>preconditioned reference length [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>uniaxial strain</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0°</td>
<td>15.5</td>
<td>16.5</td>
</tr>
<tr>
<td>90°</td>
<td>16.2</td>
<td>19.7</td>
</tr>
<tr>
<td>33.5°</td>
<td>15.6</td>
<td>16.8</td>
</tr>
<tr>
<td>56.5°</td>
<td>15.7</td>
<td>17.8</td>
</tr>
<tr>
<td><strong>uniaxial stress</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0°</td>
<td>152.5</td>
<td>162.0</td>
</tr>
<tr>
<td>90°</td>
<td>163.6</td>
<td>237.4</td>
</tr>
<tr>
<td>33.5°</td>
<td>152.0</td>
<td>163.3</td>
</tr>
<tr>
<td>56.5°</td>
<td>157.2</td>
<td>202.3</td>
</tr>
</tbody>
</table>

With respect to clinical mesh application, it has to be considered that after implantation, the mesh will be preconditioned according to the physiological conditions and might be subject to dramatic inelastic deformations. Loss of support function, dislocation and wrinkles might be direct consequences impairing the mechanical biocompatibility of these meshes. A physiologically relevant preconditioning procedure would have to be applied by the surgeon prior to implantation, but the definition of a corresponding procedure might be difficult. It has to be further considered that, after a hypothetic preconditioning procedure, the mesh would be less compliant than initially expected.

Preconditioning poses critical challenges also for modeling purposes. With respect to constitutive model formulations, a purely elastic model cannot map this preconditioning behavior. A simple hyperelastic material formulation strongly depends on the choice of the reference configuration as it determines the corresponding experimental curve which is to be fitted (Figure 4.10 or 4.15?). A structural model of a representative unit cell can account for different reference geometries in a physical way, as proposed in chapter 5. However, as long as the preconditioning behavior is not captured explicitly, such a model cannot be predictive.

### 4.5.3 Experimental uncertainties and accuracy of the measurements

The intra-specimen variability of the longitudinal strain component $E_{22}$ of 15-35% is attributed to edge effects and to local deviations from a perfectly periodic mesh structure. In fact, the evaluation of microscopy images of twenty mesh pores (see chapter 5, section 5.2.6) showed standard deviations of quantified angles and lengths of 5-10%, ranging up to 20%. The inter-specimen variability of $E_{22}$ of 2-9% mirrors the reproducibility of the samples’ structural geometry, of clamping procedures and the robustness of the applied test protocol. The high intra-compared to the inter-specimen variability of $E_{22}$ shows that (i) in a
smeared sense the results are reproducible, (ii) the specimens seem large enough to contain a representative distribution of local mesh distortions and (iii) artifacts due to edge effects are systematic.

The digital image correlation algorithm was seen to provide accurate results as long as the structure was “intact”. A structural collapse, as seen in particular for uniaxial stress loading conditions, impaired the quality of the correlation affecting mainly the transverse strain component. Moreover, as discussed in section 4.5.2 there is additional uncertainty in the horizontal position of the measurements curves, which depends on the choice of reference configuration, i.e. the preforce.

4.5.4 Challenges defining an appropriate test protocol

Common challenges in the mechanical characterization of soft materials, are the design of the clamps and the specimens’ geometry. These should be optimized for a homogeneous load introduction and in order to avoid stress concentrations [121, 124]. In contrast to biaxial tensile tests, uniaxial tensile tests, allow for a simple clamp design. Due to the discrete structure of the mesh, a rectangular specimen shape already results in a homogeneous load distribution, as long as the specimen edges are oriented in the preferred material directions (0° and 90°) - there is no use in dogbone shaped specimens.

However, quantifying anisotropy raises challenges. The specimen width and the cyclic limits (load or deformation) have to be chosen such that the results for different material directions are comparable. In the present study, we chose to load the structural elements, the strands of the mesh, equivalently for each material direction (directly valid for the 0° and 90° material directions and approximated for the off-axis directions). Load was defined as specific force per load carrying elements, i.e. the strands (UL), instead of force per width [mm] and a cyclic limit load was chosen instead of a limit deformation.

Tests were performed displacement controlled, with a constant actuator speed [mms⁻¹] or constant nominal strain rate (with respect to the clamped length). However, as it is shown in Table 4.6, there are distinct differences between the clamped lengths and the preconditioned reference lengths for the different material directions. A constant rate of real strain might be preferrable to a constant rate of nominal strain in order to exclude time dependent effects. This would require an online detection of the preforce and of the corresponding reference length, which is, in fact, realizable, but less robust. The flat initial slope of the force displacement curve and the noise of the force sensor make it difficult to reliably detect a small preforce value. Errors in its detection affect the reference length, the calculated and applied strain rate and thus the overall mechanical response (see also discussions on the preforce in chapter 3, sections 3.4 and 3.6).

The load history dependence of the mesh was characterized by a preconditioning pro-
4 Mechanical characterization of a prosthetic mesh at different length scales

procedure, defined by one single load limit. In order to quantify or model the load history
dependent inelastic behavior of the mesh structure, which would be needed for modeling a
“change in reference configuration”, different load limits would have to be applied. Moreover,
time dependent effects, such as structural creep and relaxation determine the dependence on
the strain rate and the rate of structural recovery and might be of interest to be investigated
separately.

Uniaxial strain loading conditions result in a biaxial state of stress. Biaxial tensile tests,
applying different ratios of strain for both loading directions, would provide additional valu-
able data. However, the design of the clamps and the specimen geometry are challenging
tasks. Simple cross-shaped specimens are apparently inappropriate due to the uniaxially
loaded slender “arms” with orthogonal material directions. Square samples require appro-
priate clamping for a homogenous load introduction and to avoid clamping effects, such
as restriction of transverse contraction and stress concentrations at the corners. Moreover,
defining a biaxial test protocol for a highly anisotropic material, requiring comparable strain
rates and cyclic limits in both loading directions, seems rather challenging.
4.6 Conclusions

In the present study, a set of experimental data is presented describing macroscale mechanical phenomena of the prosthetic mesh, such as the nonlinear force response, anisotropy, preconditioning effects and hysteresis. Moreover, local deformation patterns and local mesh phenomena are examined which are due to mesoscale mechanisms. Significant value was seen in the application of local strain analysis. Limitations with respect to assumptions on homogeneity, periodicity and orthotropy were identified. These data provide a comprehensive picture of the mechanical behavior of the prosthetic mesh, which can be used for the development of numerical models, but also for mesh design optimization and the improvement of clinical procedures towards a mechanically more biocompatible mesh application.

In order to quantitatively assess the mechanical biocompatibility of prosthetic meshes, reference data for native tissue are needed. Moreover, an ingrown tissue component will strongly influence the mechanical response of the mesh. Appropriate animal models are required to examine the mechanical behavior of mesh-tissue explants in similar loading conditions as in the present study. Even more relevant would be in vivo measurements, to investigate the post-operative kinematics of an implanted mesh at different length scales.
A physically based structural model for a prosthetic mesh at the mesoscale

Motivation  This study is aimed at a physically based model formulation for a prosthetic mesh to investigate the mechanical behavior at a lower scale, the mesoscale.

Methods  A structural model of a representative unit cell of a knitted prosthetic mesh is proposed based on the theory of multibody systems. The derivation of the respective system equations, the kinematics, the force elements and constitutive laws as well as kinematic and periodic constraints are described. Constitutive laws were defined based on physical considerations and examinations. The 20 parameters determining the force laws, were adjusted to fit the global force response per unit cell and the global and local kinematics gained from experimental data.

Results  A comparison between the experimental data and the model response show excellent descriptive capabilities of the model. The level of non-affine deformations of a unit cell is shown to be an appropriate mesoscale criterion to quantify the mesh’s mechanical biocompatibility, independent of the underlying tissue properties.

Main conclusions  The model represents a valuable tool to discuss aspects of the mesh’s mechanical biocompatibility at a lower scale and to deduce corresponding criteria.

*Parts of this chapter, including paragraphs of text, figures and tables are published in [140].
5.1 Introduction

There is a long history of structural models for textile fabrics, starting with the semi-analytical truss models for plain weaves by Kawabata [141–143] and followed by a large number of extended models accounting for different structural phenomena and using advanced computational techniques to solve these models in an efficient way [144–152].

The basic idea behind structural models is to use a simplified but physical mesoscale structure in order to capture more complex mechanical phenomena at the macroscale. Ideally, the mechanical properties of the mesoscale elements are directly measured in corresponding experiments [144] or deduced from manufacturer specifications [145–151]. Macroscale phenomena simulated by such models include geometric nonlinearity [144, 145, 147, 150], large Poisson ratios [146] or anisotropy [144–146]. The major challenge is the abstraction process, which has to be based on a thorough identification of the critical structures underlying the specific deformation mechanisms and structural features of interest. Weaves represent the simplest fabric structure and are therefore examined most frequently [144–151]. Respective structural models focus on yarn-yarn interactions, such as crimp interchange in tension or shear-locking [144–151]. There are only few structural models for knitted textiles, e.g. [152].

One major objective of structural models is understanding and quantifying the impact of mesoscale properties on the mechanical properties at the macroscale in order to deduce criteria for fabric design and optimization [144–152]. Moreover, multiscale approaches are aimed at physically based, as homogenized, constitutive model formulations to be implemented in finite element codes [144, 145, 148, 152]. Therefore, the incremental deformation gradient is applied on a structural unit cell model as kinematic boundary conditions. The model is solved according to minimum energy principles and the homogenized reaction forces at the boundaries are summarized in the incremental stress tensor.

Due to their complex interlooping structure and the associated challenges in finding an appropriate abstraction, knitted fabrics are more frequently modeled at the microscale [153–155], which was also approached within the scope of the present thesis (see Appendix C and [156]). Detailed models are associated with minor model assumptions, as the geometry and the assigned material properties for the filaments are based on real measured data. These models can be used to perform virtual experiments with perfectly controlled boundary conditions. However, such models are computationally expensive and require the solution of multi-contact problems, which often lack numerical convergence or require non-physical assumptions on the friction behavior and the inter-penetration of filaments [153,156]. Another challenge is the definition of a representative initial configuration, which might be found by an elaborate simulation of the manufacturing process and the corresponding solution of the unloaded equilibrium state [155–157].

In this study, a structural model of a representative unit cell of a knitted prosthetic mesh is
proposed based on the theory of multibody systems. The derivation of the respective system
equations, the kinematics, the force elements and constitutive laws as well as kinematic and
periodic constraints are described. Its geometrical structure and the constitutive laws are
based on physical measurements and considerations. The model is aimed at reproducing
the experimental data presented in chapter 4, in particular with respect to geometrical
non-linearity and anisotropy of the force response, and thus at providing insight into the
underlying mesoscale mechanisms. It will be shown that such a structural model has excellent
descriptive capabilities with respect to force response and meso- and macroscale kinematics.
Moreover, preconditioning will be interpreted as a change of the mesoscale structure and thus
of the material. The model will be shown to represent a valuable tool to assess mesoscale
aspects of the mesh’s mechanical biocompatibility.
5.2 Methods

The experimental study described in chapter 4 is located at the macroscale. The model, presented in the following sections lives at the mesoscale. Macro- and mesoscale communicate via homogenized mesoscale measures which correspond to local macroscale data. It is referred to section 4.2 for a more detailed description of the used notation with respect to different length scales.

5.2.1 Material properties

In this study, the prosthetic mesh, Gynemesh M, as described in chapter 3, section 3.2.1 was modeled (Figure 5.1). For the present modeling approach, the mesh was used in its resorbed state, containing only polypropylene fibers.

Gynemesh M is a knitted fabric. It consists of strands of stitches knitted from the polypropylene filaments. An additional continuous filament meanders through each strand and connects it to an adjacent strand at nodal points, forming the closed pores. Some filaments and the corresponding strands are coloured in blue, which is for orientation only. Thus, the direction of the blue lines marks the direction of the filaments, i.e. all filaments start and end at edges cut perpendicular to the blue lines.

5.2.2 Experimental basis

The experimental study on Gynemesh M (chapter 4) served as a basis for the development of the numerical model. The findings were used to choose the modeling approach and the phenomena to be captured. Moreover, these data helped to define the geometry and the kinematics, to determine the constitutive laws and fit parameter values and later to validate the model.

For each configuration (material direction and loading condition), one additional experiment had been conducted, where the samples had been recorded using a zoom lens (magnification $\sim 3.5x$) instead of the telecentric lens. The magnification allowed to observe single pores during deformation. However, this analysis was restricted to a small number (three) of pores, which were moreover, adjacent to each other and thus not independent. Therefore, these local findings were considered as qualitative rather than quantitative observations.

5.2.3 A structural model of a representative unit cell

A structural model was chosen, able to explain the configuration-specific deformation patterns and to account for some of the local mesh phenomena due to mesoscale mechanisms (see chapter 4, sections 4.4.3 and 4.5.1), which might, in general, not be captured by a
5.2 Methods

continuum model formulation. In contrast to a detailed three-dimensional model of the microstructure, the current approach was aimed at mapping the structural elements in an abstracted and numerically efficient way. We call the length scale of our model the *mesoscale*.

As Gynemesh M was assumed to consist of periodically arranged pores, only a representative pattern of the mesh, a unit cell, had to be modeled. A half pore was found to be the smallest unit cell of Gynemesh M (Figure 5.1 a,b) considering the four material directions of load application. For modeling reasons, the unit cell was chosen, such that, the most complex part, the node, was located in the middle of the unit cell. Unit cell models allow for the simulation of homogeneous load cases, such as the two types of uniaxial tensile tests of our experimental study. As the discretization of the mesh for the deformation analysis had also been done on the level of half pores, these data could directly be used, as shown in section 5.2.9.

![Figure 5.1: Identification of a unit cell of Gynemesh M.](image)

**Figure 5.1**: Identification of a unit cell of Gynemesh M. a Close-up of the mesh, a unit cell is marked within the (red) rectangle. b Microscope image of a unit cell overlaid by an abstracted system of ten bodies connected by line elements. c The abstracted unit cell geometry represented the geometry of the multibody system.

### 5.2.4 Modeled global mechanical phenomena

Preliminary tests and the previous experimental study, allowed to identify different mechanical phenomena, such as anisotropy, a nonlinear elastic force response, hysteresis, load history dependent inelasticity, and time dependent effects (creep, relaxation, dependence on the strain rate). The experiments in chapter 4, had been focussed on characterizing the non-linear force response, hystersis and anisotropy in two different loading conditions, uniaxial stress and uniaxial strain. Moreover, load history dependent inelasticity had been described by preconditioning effects. Creep and relaxation had not been examined. The present structural model was intended to capture in particular geometry based effects, such as the geometric nonlinearity and anisotropy. In order to account for the hysteretic behavior,
at each force level, the mean experimentally measured deformation of the loading and the unloading branch was calculated. The model was aimed at mapping this middle branch by regarding the upper and lower bounds represented by the two branches. Inelasticity was not modeled explicitly.

5.2.5 A static multibody system of a representative unit cell

The structural unit cell model was built based on the theory of multibody systems [158, 159]. The system equations for a general multibody system with \( n \) bodies are the projected Newton-Euler equations. These are the equations of linear and angular momentum projected into the space of generalized coordinates \( \vec{q} \).

\[
0 = \sum_{i=1}^{n} \left[ \left( \vec{p}_i \dot{\vec{p}}_i + \vec{r}_{PO} \times \dot{\vec{p}}_i \right) - \left( \vec{F}_i^A + \vec{M}_i^A \right) \right],
\]

or

\[
0 = M(\vec{q}, t) \ddot{\vec{q}} + \vec{g}(\vec{q}, \dot{\vec{q}}, t) - \vec{f}(\vec{q}, \dot{\vec{q}}, t, \tau) = 0
\]

- \( J_P^T \dot{\vec{p}}_i \) change of linear momentum for body \( i \)
- \( J_R^T (\vec{L}_O + \vec{r}_{PO} \times \dot{\vec{p}}) \) change of angular momentum for body \( i \)
- \( J_Q^T \vec{F}_i^A \) sum of active forces for body \( i \)
- \( J_R^T \vec{M}_i^A \) sum of active moments for body \( i \)
- \( M(\vec{q}, t) \ddot{\vec{q}} \) inertia forces for body \( i \)
- \( \vec{g}(\vec{q}, \dot{\vec{q}}, t) \) gyroscopic forces for body \( i \)
- \( \vec{f}(\vec{q}, \dot{\vec{q}}, t, \tau) \) active forces and moments for body \( i \)
- \( \vec{r}_{PO} \) position vector from the inertial origin \( O \) to the point \( P \) of the body \( i \)
- \( \vec{p}_i \) linear momentum of the body \( i \)
- \( \vec{L}_O \) angular momentum of the body \( i \), with respect to the inertial origin \( O \)
- \( J_P \) Jacobian of translation with respect to point \( P \)
- \( J_R \) Jacobian of rotation
- \( \vec{F}_i^A \) vector of active forces on the body \( i \)
- \( \vec{M}_i^A \) vector of active moments on the body \( i \) with respect to point \( Q \)
- \( M \) mass matrix

In line with the current literature [144, 147], inertia forces were disregarded, as the contribution of accelerated masses to the overall mechanical response of the mesh was assumed to be low compared to deformation related reaction forces. Friction was not considered. Furthermore, the experiments were conducted at a velocity, low enough to omit any time dependent
effects. For a static (unconstrained) system, the projected Newton-Euler equations reduce to an equilibrium of generalized active forces and moments, i.e. active forces and moments projected into the space of generalized coordinates.

\[ 0 = \sum_{i=1}^{n} \left[ J_Q^T \bar{F}_A^i + J_R^T \bar{M}_Q^i \right] \quad (5.1) \]

or

\[ 0 = -\dot{\bar{f}}(\bar{q}) \]

The model was implemented using Matlab (The MathWorks Inc., Natick, Massachusetts, US). In order to setup the corresponding model, the following steps were conducted, as described in detail in the subsequent sections.

- An abstracted mesoscale geometry of a representative unit cell was defined.

- The model was parameterized by a set of generalized coordinates, determining its kinematics.

- Force elements were introduced coupling the relative distances between the rigid bodies and the relative angles.

- Load case specific kinematic boundary conditions were applied by constraint equations.

- According to the representative unit cell approach, periodic boundary conditions were included.

### 5.2.6 Geometry

The unit cell geometry was directly based on the real knitted structure of one half pore (unit cell) of the mesh. The complex structure was abstracted by an arrangement of ten massless rigid bodies connected by (force) elements (Figure 5.1 b,c). There were no data available from manufacturers regarding an ideal unit cell geometry. Therefore, a set of 20 unit cells was recorded by a light microscope. The positions of the ten rigid bodies of the model were estimated for each microscope image. As both rigid bodies and (force) elements represented an abstraction or homogenization of structural mesh elements (stitches, node), which could rearrange during deformation, there was no unique assignment between elements of the model and the real structure (see also section 5.4.5). Averaged positions for all rigid bodies were evaluated from the 20 unit cells. The initial arrangement of the rigid bodies was chosen as symmetric with respect to two orthogonal axes parallel to the x- and y-axis of the inertial coordinate system (Figures 5.1, 5.2).
Table 5.1: Measured mean initial geometry represented in generalized coordinates, angles are given in $[^\circ]$, distances are given in [mm]

<table>
<thead>
<tr>
<th>$q_1$</th>
<th>$q_2$</th>
<th>$q_3$</th>
<th>$q_4$</th>
<th>$q_5$</th>
<th>$q_6$</th>
<th>$q_7$</th>
<th>$q_8$</th>
<th>$q_9$</th>
<th>$q_{10}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>32.78</td>
<td>20.14</td>
<td>65.91</td>
<td>16.60</td>
<td>9.12</td>
<td>14.35</td>
<td>71.52</td>
<td>16.78</td>
<td>1.05</td>
<td>0.38</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>$q_{11}$</th>
<th>$q_{12}$</th>
<th>$q_{13}$</th>
<th>$q_{14}$</th>
<th>$q_{15}$</th>
<th>$q_{16}$</th>
<th>$q_{17}$</th>
<th>$q_{18}$</th>
<th>$q_{19}$</th>
<th>$q_{20}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.95</td>
<td>0.41</td>
<td>1.14</td>
<td>0.41</td>
<td>1.11</td>
<td>0.38</td>
<td>-</td>
<td>-</td>
<td>0.45</td>
<td>69.27</td>
</tr>
</tbody>
</table>

Table 5.2: Idealized symmetric initial model geometry in generalized coordinates, angles are given in $[^\circ]$, distances are given in [mm]

<table>
<thead>
<tr>
<th>$q_1$</th>
<th>$q_2$</th>
<th>$q_3$</th>
<th>$q_4$</th>
<th>$q_5$</th>
<th>$q_6$</th>
<th>$q_7$</th>
<th>$q_8$</th>
<th>$q_9$</th>
<th>$q_{10}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>34.58</td>
<td>15.72</td>
<td>69.17</td>
<td>15.72</td>
<td>0</td>
<td>15.72</td>
<td>69.17</td>
<td>15.72</td>
<td>0.96</td>
<td>0.39</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>$q_{11}$</th>
<th>$q_{12}$</th>
<th>$q_{13}$</th>
<th>$q_{14}$</th>
<th>$q_{15}$</th>
<th>$q_{16}$</th>
<th>$q_{17}$</th>
<th>$q_{18}$</th>
<th>$q_{19}$</th>
<th>$q_{20}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.96</td>
<td>0.39</td>
<td>0.96</td>
<td>0.39</td>
<td>0.96</td>
<td>0.39</td>
<td>-</td>
<td>-</td>
<td>0.33</td>
<td>69.17</td>
</tr>
</tbody>
</table>

5.2.7 Kinematics

In the two-dimensional space, a system of ten unconstrained rigid bodies, results in 20 degrees of freedom (dofs). Therefore, a set of 20 generalized coordinates was introduced, describing the absolute position of the system with respect to the inertial system ($q_1$, $q_{17}$, $q_{18}$), the relative distances between the rigid bodies, i.e. the lengths of the (force) elements ($q_9$, $q_{10}$, $q_{11}$, $q_{12}$, $q_{13}$, $q_{14}$, $q_{15}$, $q_{16}$, $q_{19}$) and the relative angles between the elements connecting the bodies ($q_2$, $q_3$, $q_4$, $q_5$, $q_6$, $q_7$, $q_8$, $q_{20}$) (Figure 5.2). The measured mean initial geometry expressed in terms of this parameterization (Table 5.1) shows that the symmetry assumption is justified. An idealized initial geometry was chosen as shown in Table 5.2. Note, that despite of the symmetry assumption for the geometry, there is no symmetry assumption in the kinematics. In order to later transform force and moment vectors into the 20 dofs space of generalized coordinates (concept of generalized forces), respective Jacobians of translation and rotation were built for all bodies associated with force elements (all rigid bodies of the system). Therefore, the corresponding position vectors were represented in terms of generalized coordinates. Its total derivative with respect to time directly provided
5.2 Methods

Figure 5.2: Parameterization by a set of 20 generalized coordinates, represented by lengths (left) and angles (right).

the Jacobian of translation, as it is shown for body $A_1$.

$$\vec{r}_{OA_1} = \begin{pmatrix} q_{17} - \frac{1}{2} q_{19} \sin (q_5) - q_9 \cos (q_1) \\ q_{18} + \frac{1}{2} q_{19} \cos (q_5) - q_9 \sin (q_1) \\ 0 \end{pmatrix}$$

$$\dot{\vec{r}}_{OA_1} = \frac{\partial \vec{r}_{OA_1}}{\partial \dot{q}} \dot{q} + \frac{\partial \vec{r}_{OA_1}}{\partial t}$$

$$= \vec{J}^T_{A_1} \dot{q} + \nu_{A_1}$$

$$\vec{r}_{OA_1} = \begin{pmatrix} \dot{q}_{17} - \dot{q}_9 \cos (q_1) + q_9 \dot{q}_1 \sin (q_1) \\ \dot{q}_{18} + \frac{1}{2} \dot{q}_{19} - \dot{q}_9 \sin (q_1) - q_9 \dot{q}_1 \cos (q_1) \\ 0 \end{pmatrix}$$

$$= \begin{pmatrix} q_{9} \sin (q_1) & 0 & 0 & -\frac{1}{2} q_{9} \cos (q_5) & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \ -q_{9} \cos (q_1) & 0 & 0 & -\frac{1}{2} q_{9} \sin (q_5) & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \ \end{pmatrix} \dot{q}$$

The Jacobians of rotation were directly read from the angular velocities as follows, e.g. for body $F$.

$$\vec{\Omega}_F = \begin{pmatrix} 0 \\ 0 \\ \dot{q}_3 \end{pmatrix}$$

$$= \begin{pmatrix} 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \ \end{pmatrix} \dot{q}$$
5 A physically based mesoscale model for a prosthetic mesh

5.2.8 Force elements

Translational and rotational force elements were introduced to couple the motion of the rigid bodies (Figure 5.3). The concept of generalized forces was applied, which is exemplarily shown for the force element between the rigid bodies $A_1$ and $F$ (Figure 5.4). The force element is cut free and regarded separately from the system. It is defined in the one-dimensional space $\mathbb{R}$, whereas the system is defined in the three-dimensional space $\mathbb{R}^3$. In the one-dimensional space of the force element, the scalar $g(\vec{q})$ is introduced describing the current length of the force element. The reaction force at the force element is introduced as scalar contraction force $\lambda$. Its value is determined by the corresponding constitutive law or force law as $\lambda = \lambda(g(\vec{q}))$ (section 5.2.12). The corresponding reaction force of the three-dimensional system is defined as $\vec{n}\lambda(g(\vec{q}))$, where $\vec{n}$ is the normalized position vector connecting the rigid bodies and thus defining the direction of the reaction force and $\lambda$ is its absolute value.

With respect to Figure 5.4, the following kinematic property holds.

$$g(\vec{q}) = \vec{n}^T r_{A_1F}$$

(5.2)

Due to this definition, also negative distances are allowed (penetrations of rigid bodies). This concept was applied for both translational and rotational force elements.

According to equation 5.1, all forces and moments are projected into the space of generalized coordinates and summed for all rigid bodies of the system. For the shown force element $A_1F$, the contributions of bodies $A_1$ and $F$ are summed.

$$f_{AF} = (\vec{J}_F - \vec{J}_{A_1})^T \vec{n} \cdot \lambda(g(\vec{q}))$$

with the generalized force direction $\vec{w}_{F_{A_1}}$. 

Figure 5.3: Translational and rotational force elements, coupling the motion of all rigid bodies
The generalized force directions resulting from all active forces are summarized in a matrix $W(\vec{q})$. The current lengths and angles of all force elements are gathered in the vector $\vec{g}(\vec{q})$ and the absolute force and moment values in the vector $\vec{\lambda}(\vec{g}(\vec{q}))$. Thus, the projected Newton-Euler equations for the present (unconstrained) system were written as

$$-\vec{f}(\vec{q}) = -W(\vec{q})\vec{\lambda}(\vec{g}(\vec{q})) = 0$$

Recalling the derivation of the Jacobians (section 5.2.7) and the kinematic property of $\vec{g}$ (equation 5.2), the time derivative of $\vec{g}$

$$\dot{\vec{g}}(\vec{q}) = \frac{\partial \vec{g}(\vec{q})}{\partial \vec{q}} \dot{\vec{q}} + \frac{\partial \vec{g}(\vec{q})}{\partial t}$$

provides the important relationship

$$W^T(\vec{q}) = \frac{\partial \vec{g}(\vec{q})}{\partial \vec{q}}$$

Note that, e.g. for the force element $A_1F$, $\dot{\vec{g}} = \vec{n}^T \dot{\vec{r}}_{A_1F}$, as $\dot{\vec{n}}^T \perp \vec{r}_{A_1F}$.

The generalized force directions $W(\vec{q})$ are in accordance with the admissible virtual displacements $\delta \vec{q}$ and thus with the kinematics of the system. The concept of generalized forces allows a clear separation between the generalized force directions $W(\vec{q})$, i.e. the kinematic framework and the force laws $\vec{\lambda}(\vec{g}(\vec{q}))$, i.e. the constitutive model formulations for each force element.

Figure 5.4: Concept of generalized forces [158,159], exemplarily shown for the translational force element between bodies $A_1$ and $F$ (left) and the rotational element at body $E$. 
5.2.9 Kinematic boundary conditions - constraints

The experimental load cases were modeled by stepwise applying kinematic boundary conditions, represented by kinematic constraints, and solving the static system. For all of the simulated uniaxial tensile tests, the convention was introduced, that the direction of uniaxial displacement was equal to the inertial \(x\)-direction. The model was rotated correspondingly (Figure 5.5). According to chapter 4, the experimental data were represented by a sequence of deformation gradients described by the two directed line elements in the current and in the reference configuration, \(g_1\) and \(g_2\) (and \(g_3\)) and \(G_1\) and \(G_2\) (and \(G_3\)), respectively. As these vectors represented the local sample deformation at the level of the half pores (unit cell), they were directly used as kinematic constraints for the unit cell model. At each step, the current positions of the rigid bodies at the boundaries \(A_2, B_2, C_2\) and \(D_2\) were prescribed by \(g_1\) and \(g_2\). The boundary point \(A_2\) was chosen to be fixed, i.e. translational degrees of freedom in \(x\)- and \(y\)- direction were constrained. \(g_2\) was applied to point from \(A_2\) to \(C_2\) and \(g_1\) was positioned in such a way that \(g_1\) and \(g_2\) intersect at half of their lengths (Figure 5.5). The latter is an assumption and results in a globally point symmetric deformation with respect to the midpoint of the unit cell.

![Diagram](image)

Figure 5.5: Application of kinematic boundary conditions, constraints. Constraints were applied by prescribing the experimentally measured current unit cell configuration (represented by \(g_1\) and \(g_2\)). The convention was introduced that the direction of uniaxial displacement was equal to the inertial \(x\)-direction. For load application in different material directions, the model was rotated correspondingly.

For the case of uniaxial strain, both the inertial \(x\)- and \(y\)-components, were prescribed. Whereas for the case of uniaxial stress, only the components in loading direction, the inertial \(x\)-component was prescribed. The transverse contraction was free. The homogeneous
constraint equations for the case of uniaxial strain were written as follows.

\[
\vec{g}^c (\vec{q}) = \vec{0} = \begin{pmatrix}
(1 & 0 & 0) \cdot (\vec{r}_{OA2}(\vec{q}) - \vec{r}_{OA2}(\vec{q}_0)) \\
(1 & 0 & 0) \cdot (\vec{r}_{OA2}(\vec{q}) - \vec{r}_{OA2}(\vec{q}_0)) \\
(1 & 0 & 0) \cdot (\vec{r}_{OB2}(\vec{q}) - (\vec{r}_{OA2}(\vec{q}) + \frac{1}{2}\vec{g}_2 + \frac{1}{2}\vec{g}_1)) \\
(1 & 0 & 0) \cdot (\vec{r}_{OC2}(\vec{q}) - (\vec{r}_{OA2}(\vec{q}) + \vec{g}_2)) \\
(1 & 0 & 0) \cdot (\vec{r}_{OD2}(\vec{q}) - (\vec{r}_{OA2}(\vec{q}) + \frac{1}{2}\vec{g}_2 - \frac{1}{2}\vec{g}_1))
\end{pmatrix}
\]

For the case of uniaxial stress the constraint equation were reduced to

\[
\vec{g}^c (\vec{q}) = \vec{0} = \begin{pmatrix}
(1 & 0 & 0) \cdot (\vec{r}_{OA2}(\vec{q}) - \vec{r}_{OA2}(\vec{q}_0)) \\
(1 & 0 & 0) \cdot (\vec{r}_{OA2}(\vec{q}) - \vec{r}_{OA2}(\vec{q}_0)) \\
(1 & 0 & 0) \cdot (\vec{r}_{OB2}(\vec{q}) - (\vec{r}_{OA2}(\vec{q}) + \frac{1}{2}\vec{g}_2 + \frac{1}{2}\vec{g}_1)) \\
(1 & 0 & 0) \cdot (\vec{r}_{OC2}(\vec{q}) - (\vec{r}_{OA2}(\vec{q}) + \vec{g}_2)) \\
(1 & 0 & 0) \cdot (\vec{r}_{OD2}(\vec{q}) - (\vec{r}_{OA2}(\vec{q}) + \frac{1}{2}\vec{g}_2 - \frac{1}{2}\vec{g}_1))
\end{pmatrix}
\]

**Remark** For the 0° and 90° material directions, the vectors \(\vec{g}_1^c\) and \(\vec{g}_2^c\) were idealized to be symmetric.

Constraints result in constraint forces which have to be included into the projected Newton Euler equations. These constraint forces were in fact the unknown reaction forces, i.e. the force response which was to be compared to the experimental data. Following the principle of d’Alembert Lagrange, constraint forces \(\vec{f}_c\) do not contribute to the virtual work for all admissible virtual displacements \(\delta \vec{q}\).

\[
\vec{f}_c \delta \vec{q} = 0
\]

It can be shown [159] that this requirement is fulfilled if the constraint forces are of the following form.

\[
\vec{f}_c (\vec{q}) = W^c (\vec{q}) \vec{\lambda}^c
\]

with the generalized constraint force directions

\[
W^c (\vec{q}) = \frac{\partial \vec{g}^T}{\partial \vec{q}}
\]

Such constraints are said to be ideally bilateral. The constrained system was described by
the equilibrium of active and constraint forces, subject to contraints.

\[ -W(q) \bar{\lambda} - W^c(q) \bar{\lambda}^c = 0 \]

\[ -f(q) - f^c(q) = 0 \]

### 5.2.10 Periodic boundary conditions

In order to “tell” the unit cell model that it is a representative part of a periodic pattern, it was further constrained by periodic boundary conditions. From a physical point of view, periodic boundary conditions assure that there are no discontinuities (kinks and gaps) at the boundaries of one unit cell, when assembling them. In the present model, the angles of opposite outer force elements, e.g. elements \( A_1-A_2 \) and \( C_1-C_2 \) with respect to the inertial coordinate system were constrained to be equal, i.e. the elements had to be parallel (Figure 5.6). Expressed in generalized coordinates, the following relations were required.

\[ q_1 - q_{20} - q_6 = q_1 - q_3 - q_4 \] (5.3)

\[ q_1 + q_2 = q_1 - q_{20} + q_7 + q_8 \] (5.4)

![Figure 5.6: Periodic boundary conditions: opposite outer “arms” have to be parallel as indicated by the inclined \( \parallel \) symbols](image)

As the constraint forces realizing these constraint conditions were not of interest, in this case, a new set of generalized coordinates \( z \) was chosen, reducing the dimension of \( q \), \( \dim \bar{q} = 20 \), to \( \dim z = 18 \). \( \bar{q} \) and \( z \) were transformed by the matrix \( Q \).

\[ \bar{q} = Qz \] (5.5)
5.2 Methods

In detail, the transformation rules (summarized in \( Q \)) realizing the periodic boundary conditions (equations 5.3, 5.4) were formulated as follows.

\[
\begin{align*}
q_1 &= z_1 & q_2 &= z_2 & q_3 &= z_3 & q_4 &= -z_3 + z_{18} + z_5 \\
q_5 &= z_4 & q_6 &= z_5 & q_7 &= z_6 & q_8 &= z_2 + z_{20} - z_7 \\
q_9 &= z_7 & q_{10} &= z_8 & q_{11} &= z_9 \\
q_{12} &= z_{10} & q_{13} &= z_{11} & q_{14} &= z_{12} \\
q_{15} &= z_{13} & q_{16} &= z_{14} & q_{17} &= z_{15} \\
q_{18} &= z_{16} & q_{19} &= z_{17} & q_{20} &= z_{18}
\end{align*}
\]

The virtual displacements were described by

\[\delta \vec{q} = Q \delta \vec{z}\]

The virtual work was written as

\[0 = \delta \vec{q}^T \left( -\vec{f}(\vec{q}) - \vec{f}^c(\vec{q}) - \vec{f}^{cp}(\vec{q}) \right)\]

for all admissible \( \delta \vec{q} \) and with the periodic constraint forces \( \vec{f}^{cp} \)

\[0 = \delta \vec{z}^T \left( -Q^T \vec{f}(\vec{q}) - Q^T \vec{W}^c(\vec{q}) \vec{X}^c(\vec{q}) - Q^T \vec{W}^{cp}(\vec{q}) \vec{X}^{cp}(\vec{q}) \right)\]

for all admissible \( \delta \vec{z} \) and with the generalized periodic force directions \( \vec{W}^{cp} \) and the values of the periodic constraint forces \( \vec{X}^{cp} \).

It can be shown that in the space of the new, reduced set of generalized coordinates, the periodic constraint forces vanish [159]. The constrained projected Newton-Euler equations including periodic boundary conditions were now written as follows.

\[Q^T \left( -\vec{f}(\vec{q}) - \vec{f}^c(\vec{q}) \right) = 0\]

\[\vec{g}^c(\vec{q}) = 0\]

and \( \vec{q} = \vec{q}(\vec{z}) \)

5.2.11 Comments on the static determinateness

The present multi-body system is a 20 dof system. The two translational dofs in \( A_2 \) were constrained. Moreover, for the case of uniaxial strain another six dofs were constrained and for the case of uniaxial stress another three. Periodic boundary conditions reduced the set
of dofs by additional two dofs. The remaining ten or 13 dofs, respectively, were coupled by the force elements. In sum, 17 force elements were introduced to couple the positions of the rigid bodies, resulting in corresponding constraint forces. (In contrast, a mechanism would require constraining or coupling only these ten or 13 dofs. The kinematic boundary constraints would result in zero constraint forces.)

5.2.12 Force laws and parameters

Abstracting a three-dimensionally interlooping structure by one-dimensional force elements, required structural force laws instead of pure material laws. At the microscale, there are hard contacts between the filaments. However, at the level of the abstracted mesoscale, many continuously closing contacts and the smooth transition between structural and material response result in a regularized mechanical response. Therefore, mainly regularized force laws were implemented, relating the structural stiffness to the current structural configuration. The structural force laws and the parameters were not gained from corresponding experiments at the mesoscale. However, the force laws were physically based, as should be illustrated with the two following examples.

**Geometry inspired force laws, stitches** (Figure 5.7) The inner and outer “arms” of the unit cell model consist of strands of stitches, filament loops. The corresponding force laws were defined based on the following physical considerations.

Stitches collapse until the filaments are aligned and stretched. The mechanical response changes from a structural response to a stiffer material response. The transition is smooth.

A half ellipse, fixed at one end and simply supported at the other end was used to estimate the structural response of the stitches (Figure 5.8). When stretching the half ellipse along its long axis, it collapses and its structural stiffness is increased nonlinearly (Figure 5.9 a). In the limit, the structural stiffness is constant and equals the material stiffness of two filaments stretched in parallel.

Based on these considerations, a power law formulation was chosen to describe the stiffness $k_{itran}$ of the force element as a function of its current length $q_i$. A correction factor $c_{corr}$ was introduced to account for the difference between the chosen length of the force elements and the observed real length of a stitch and considering that inner and outer “arms” should be characterized by comparable force laws (Figure 5.8 a). An upper limit $l_u$ was defined determining the transition between structural and material deformation (Figure 5.9 b).

The function value of a power law function is smaller than the basis as long as the basis is smaller than one. For a basis larger than one, the function value rises quickly. With respect to a force law for the stitches, the basis one can mark the transition from structural to material response. The parameters play with this functionality (Table 5.3).
Together with the parameters of all remaining force elements, these parameters were adjusted, such that the global model response fits the experimental data per UL for different loading conditions, as shown later in section 5.2.15.

The structural stiffness for the stitches was defined as

\[
\begin{align*}
\text{if } (q_i + c_{corr}) \leq l_u, & \quad k_{\text{trans}} = c_4 \cdot \left[\left( (q_i + c_{corr}) + c_1 - 1 \right) \cdot c_2 + 1 \right]^{c_3} + c_5 \\
\text{else} & \quad k_{\text{trans}} = c_4 \cdot \left[\left( l_u + c_1 - 1 \right) \cdot c_2 + 1 \right]^{c_3} + c_5
\end{align*}
\]

for the inner arms \( i = 9, 11, 13, 15 \), \( c_{corr} = -0.18 \) [mm], for the outer arms \( i = 10, 12, 14, 16 \), \( c_{corr} = q_i \) [mm]

The force required to stretch the force element from its reference configuration \( q_{i,ref} \) to its current configuration \( q_i \) was derived by integration of the stiffness.

\[
-\lambda_{\text{trans}} = \int_{(q_{i,ref} + c_{corr})}^{(q_i + c_{corr})} k_{\text{trans}} dq_i
\]

The respective potential energy stored in this force element was derived by integration of the force.

\[
E_{\text{pot,trans}} = \int_{(q_{i,ref} + c_{corr})}^{(q_i + c_{corr})} -\lambda_{\text{trans}} dq_i
\]

Figure 5.7: Force elements for the inner and outer “arms”, representing the stitches. Due to symmetry, the force laws are only indicated for the upper half of the unit cell.
Table 5.3: Parameters for the force law of the inner and outer translational force elements, representing the stitches

<table>
<thead>
<tr>
<th>parameter</th>
<th>interpretation</th>
</tr>
</thead>
<tbody>
<tr>
<td>$c_1$</td>
<td>shifts the current configuration towards the transition range around the basis of one, [mm]</td>
</tr>
<tr>
<td>$c_2$</td>
<td>multiplies with the difference to one, [-]</td>
</tr>
<tr>
<td>$c_3$</td>
<td>exponent of the power law, [-]</td>
</tr>
<tr>
<td>$c_4$</td>
<td>factor of the power law, [N/mm$^2$]</td>
</tr>
<tr>
<td>$c_5$</td>
<td>offset of the power law, [N/mm]</td>
</tr>
<tr>
<td>$l_u$</td>
<td>at this length the transition is complete, the stiffness remains constant, [mm]</td>
</tr>
</tbody>
</table>

Figure 5.8: Abstraction of the stitches, a Microscope image overlaid by the force elements representing the stitches, b In order to find a physical force law, the stitch was abstracted by a half ellipse, fixed at the one end and simply supported at the other end. Note, that (i) the force elements for the inner “arms” are longer than the actual stitch as they range into the node, (ii) the outer “arms” only contain half of a stitch due to the boundaries of the unit cell.

Figure 5.9: Geometry inspired force law for the stitches. a An ellipse collapses due to force application along the axis. b Force law chosen for the force elements representing the stitches. The stiffness is related to the current configuration of the force element by a power law function. When the ellipse (stitch) is collapsed the stiffness is constant.
Kinematic inspired force laws, rotation of stitches, inner “arms” (Figure 5.10) There are physical bounds resisting the rotation of the stitches within the strands. Stitches meet at nodal points (hinges). They can rotate with low resistance until adjacent stitches come into contact or until they are restricted by the continuous filament (Figure 5.11). Further rotation and the required compression of stitches increases the structural rotational stiffness. The rotational stiffness is apparently dependent on the current configuration (angle between the stitches). There is an upper and a lower bound restricting the rotation of the stitches, which were estimated from the experimental observations using the zoom lens (see section 5.3.5 and the Appendix D). The rotational stiffness was modeled as a bilinear symmetric function of the current configuration. It is characterized by a low, constant value around the reference configuration and a distinct increase when exceeding the upper or lower bounds (Figure 5.12).

\[
\text{if } q_i \geq d_1 \text{ AND } q_i \leq d_2, \quad k_{i\text{rot}} = d_3 \quad i = 3, 7 \\
\text{elseif } q_i > d_2, \quad k_{i\text{rot}} = d_3 + d_4 \cdot (q_i - d_2) \\
\text{elseif } q_i < d_1, \quad k_{i\text{rot}} = d_3 + d_4 \cdot (d_1 - q_i)
\]

The force was derived by integration of the stiffness.

\[
-\lambda_{i\text{rot}} = \int_{q_i}^{q_{iref}} k_{i\text{rot}} dq_i
\]

The respective potential energy was derived by integration of the force.

\[
E_{pot\text{rot}} = \int_{q_i}^{q_{iref}} -\lambda_{i\text{rot}} dq_i
\]

The remaining force laws were defined in a similar way and are described briefly.

Figure 5.10: Force elements representing the rotational stiffness of the between the inner “arms”, i.e. the stitches gathering at the node

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Figure 5.11: Rotational behavior of the stitches at the node. a) Microscope image overlaid by the undeformed unit cell structure and the force element determining the rotational stiffness between the inner “arms”. b) Collapsed unit cell in uniaxial stress loading conditions. The decrease in angle is constrained by a lower bound, resulting from microstructural inter-filament contacts.

Table 5.4: Parameters for the force law of the rotational force element between the inner “arms”, i.e. the stitches gathering at the node

<table>
<thead>
<tr>
<th>parameter</th>
<th>interpretation</th>
</tr>
</thead>
<tbody>
<tr>
<td>$d_1$</td>
<td>lower bound, [°]</td>
</tr>
<tr>
<td>$d_2$</td>
<td>upper bound, [°]</td>
</tr>
<tr>
<td>$d_3$</td>
<td>constant stiffness for deformations close to the initial configuration, [Nm/°]</td>
</tr>
<tr>
<td>$d_4$</td>
<td>slope, defining the linear rise in stiffness after exceeding the bounds, [Nm/°²]</td>
</tr>
</tbody>
</table>

Relative rotation of the stitches, outer “arms” In particular for the case of uniaxial stress in the 0° material direction, the outer “arms” contribute to the collapse of the pores by flapping over (the relative angle enclosed between the inner and the outer arms changes its sign). The stiffness of the rotational force element (Figure 5.13) was chosen as constant and low enough to allow for the observed flapping behavior.

$$k_{rot} = constant = e_1, \quad i = 2, 4, 6, 8$$
5.2 Methods

Figure 5.12: Force law chosen for the rotational force elements between inner “arms”, i.e. the stitches gathering at the node. The stiffness is related to the current configuration of the force element by a bilinear symmetric function. The upper and the lower bound represent the configuration when the stitches come into contact and are compressed. As the structural stiffness during compression of the stitches is continuously increasing, the force law was designed as a regularized function rather than hard contact.

![Figure 5.12](image)

Figure 5.13: Force elements representing the rotational resistance between the stitches

Figure 5.14: Translational force element of the node
The abstracted node  The node in the middle of the unit cell is the most complex structure. It was abstracted by a combination of three force elements. A translational force element accounted for the distance between the bodies E and F (Figure 5.14). Similar to the rotational elements between the inner “arms”, the stiffness was modeled as a symmetric bilinear function of the current configuration with an upper and a lower bound. Moreover, a separation of the bodies by penetration of E and F was constrained by a hard contact (reduction of dofs) for a vanishing distance.

\[
\text{if } q_i \geq f_1 \text{ AND } q_i \leq f_2, \quad k_{i_{trans}} = f_3 \\
\text{elseif } q_i > f_2, \quad k_{i_{trans}} = f_3 + f_4 \cdot (q_i - f_2) \\
\text{elseif } q_i < f_1, \quad k_{i_{trans}} = f_3 + f_4 \cdot (f_1 - q_i) \\
\text{elseif } q_i = 0, \quad q_i \text{ eliminated}
\]

The second force element of the node was a rotational element resisting shear like deformations, in the sense of a rotation of the element between E and F (Figure 5.15). Again, the stiffness was modelled as a symmetric bilinear function with an upper and a lower bound.

\[
\text{if } q_i \geq g_1 \text{ AND } q_i \leq g_2, \quad k_{i_{rot}} = g_3, \quad i = 5 \\
\text{elseif } q_i > g_2, \quad k_{i_{rot}} = g_3 + g_4 \cdot (q_i - g_2) \\
\text{elseif } q_i < g_1, \quad k_{i_{rot}} = g_3 + g_4 \cdot (g_1 - q_i)
\]

The third force element of the node was a rotational force element restricting relative rotations of the two branches A_2A_1FB_1B_2 and D_2D_1EC_1C_2 (Figure 5.16). As
in reality, there is low resistance to such deformations, the stiffness of this element was chosen to have a small and constant value.

\[ k_{irot} = \text{constant} = h_1, \quad i = 20 \]

### 5.2.13 Solution

The system equations have been stated as

\[
Q^T \left( -\vec{f}(\vec{q}) - W^c\vec{\lambda}(\vec{q}) \right) = 0 \\
\vec{g}^c(\vec{q}) = 0
\]

and \( \vec{q} = \vec{q}(\vec{z}) \) The system was formulated as a static system. The uniaxial tensile test was modeled by stepwise applying new kinematic constraints (subsequent pairs of experimentally measured \( \vec{g}_1 \) and \( \vec{g}_2 \)). For each step \( k \), a solution \( (\vec{q}_k, \vec{f}_k) \) was calculated by means of a Newton-Raphson iteration.

### Linearization

Due to the parameterization and the force laws, the system equations were non-linear in \( \vec{q} \). For the Newton-Raphson algorithm, the system equations had to be linearized. The set of unknowns was summarized in a vector \( \vec{x} \).

\[
\vec{x} = \begin{pmatrix} \vec{q} \\ \vec{\lambda} \end{pmatrix}
\]

and the reduced set of parameters \( \vec{z}^* \)

\[
\vec{z}^* = \begin{pmatrix} \vec{z} \\ \vec{\lambda} \end{pmatrix}
\]

and similar to equation 5.5

\[
\vec{x} = Q^*\vec{z}^* \\
\Delta\vec{x} = Q^*\Delta\vec{z}^*
\]
The system equations were written as

\[
Q^T \left( -\vec{f} (\vec{x}) - \vec{f}^c (\vec{x}) \right) = 0 \\
\vec{g}^c (\vec{x}) = 0
\]

and linearized at the step (timepoint) \( k \)

\[
Q^T \left( -\vec{f} (\vec{x}_k) - \vec{f}^c (\vec{x}_k) - \frac{\partial \vec{f} (\vec{x})}{\partial \vec{x}} \bigg|_{\vec{x}_k} \Delta \vec{x} - \frac{\partial \vec{f}^c (\vec{x})}{\partial \vec{x}} \bigg|_{\vec{x}_k} \Delta \vec{x} \right) = 0 \\
\vec{g}^c (\vec{x}_k) + \left( \frac{\partial \vec{g}^c (\vec{x})}{\partial \vec{x}} \bigg|_{\vec{x}_k} \right) \Delta \vec{x} = 0
\]

\[
Q^T \left( -\vec{f} (\vec{x}_k) - \vec{f}^c (\vec{x}_k) - \frac{\partial \vec{f} (\vec{x})}{\partial \vec{x}} \bigg|_{\vec{x}_k} Q^* \Delta \vec{z}^* - \frac{\partial \vec{f}^c (\vec{x})}{\partial \vec{x}} \bigg|_{\vec{x}_k} Q^* \Delta \vec{z}^* \right) = 0 \\
\vec{g}^c (\vec{x}_k) + \left( \frac{\partial \vec{g}^c (\vec{x})}{\partial \vec{x}} \bigg|_{\vec{x}_k} \right) Q^* \Delta \vec{z}^* = 0
\]

These equations were summarized in matrix notation.

\[
\begin{pmatrix}
Q^T \frac{\partial \vec{f} (\vec{x})}{\partial \vec{x}} \bigg|_{\vec{x}_k} + Q^T \frac{\partial \vec{f}^c (\vec{x})}{\partial \vec{x}} \bigg|_{\vec{x}_k}
\end{pmatrix} Q^* \Delta \vec{z}^* = -\begin{pmatrix}
Q^T \vec{f} (\vec{x}_k) + Q^T \vec{f}^c (\vec{x}_k)
\end{pmatrix}
\]

This system is linear in \( \Delta \vec{z}^* = \begin{pmatrix} \Delta \vec{z} \\ \Delta \vec{\lambda}^c \end{pmatrix} \), and iteratively solved for \( \Delta \vec{z}^* \). The iterative solution for the generalized coordinates \( \vec{q}_{n+1} \) and the unknown reaction (constraint) forces \( \vec{f}^c_{n+1} \) are obtained with \( \Delta \vec{q} = Q \Delta \vec{z} \) and \( \Delta \vec{f}^c = W^c \Delta \vec{\lambda}^c \) as

\[
\vec{q}_{n+1} = Q \Delta \vec{z} + \vec{q}_n \\
\vec{f}^c_{n+1} = \Delta \vec{f}^c + \vec{f}^c_n
\]

For \( k = 1 \), the initial values \( (n = 1) \) are represented by the reference configuration \( \vec{q}_1 = \vec{q}_{ref} \) and \( \vec{f}^c_1 = \vec{0} \). For \( k > 1 \), the initial values are equal to the solution of the previous step.
Convergence criterion

For each step $k$, iterations were continued until the convergence criterion was fulfilled. Convergence was defined as

$$\| \vec{q}_{n+1} - \vec{q}_n \| < 0.001$$

In this case, $\Delta \vec{z}^*$ was the solution of step $k$ and $\vec{q}_{k+1} = \vec{q}_{n+1}$, $\vec{f}_{c_{k+1}} = \vec{f}_{c_{n+1}}$.

Error estimate

The error was estimated for each iteration $n$ of step $k$. It was defined as

$$e_n = \| -\vec{f}(\vec{q}_n) - W^c(\vec{q}_n) \vec{X}^c\|$$

5.2.14 Reference configuration

The experimental observations showed that there were distinct inelastic deformations comparing the virgin configuration and the preconditioned configuration (chapter 4, section 4.4.4). The initial configuration was equal for all loading conditions (with small differences at the level of the preforce), whereas the preconditioned configuration differed depending on the load history. As inelasticity was not included in the present model, different reference configurations were chosen for each load case. Starting from the initial geometry (Table 5.1), the experimentally measured load case specific (and inelastically deformed) reference configuration ($\vec{G}_1$, $\vec{G}_2$) was applied as kinematic boundary conditions. For some load cases ($90^\circ$ and $56.5^\circ$ material directions in uniaxial stress loading conditions), there were large differences between the virgin geometry and the preconditioned geometry. So, the corresponding kinematic boundary conditions were applied by interpolated steps. The resulting configuration in terms of generalized coordinates was used as load case specific, stress-free, preconditioned reference configuration. Correspondingly, the first cycle of preconditioning was simulated for each load case, using the respective not inelastically deformed reference configurations.

5.2.15 Fitting strategy

The constitutive behavior of the force elements was determined by a set of overall 20 parameters. These were adjusted to fit the experimentally measured global force response per unit cell and the global kinematic response. Moreover, the mesoscale kinematics gained from the magnified image sequences were used to further tune the parameters.

The $0^\circ$ material direction was the only direction where all filaments were fixed between the two clamps. For all remaining material directions, edge effects were expected resulting from free filament ends at the edges, which might slide into the specimen and locally influence the
deformation (chapter 4, section 4.5.1). Therefore, the parameters were preliminary adjusted for the two 0° load cases. Next, the two 90° load cases were taken into account. As there were different experimental limitations associated with the specimens loaded in off-axis directions (see chapter 4, section 4.5.1 and Appendix B), these were considered last.

Due to the structure of the model, for each load case, critical force elements, dominating the specific mechanical response were identified. This partial decoupling facilitated the fitting process. This study was not aimed at finding the perfect fit, but at understanding the structural properties leading to anisotropy and geometric nonlinearity. “Strategical” handfitting was therefore preferred to an optimization procedure. The goodness of the fit was quantified by calculating the mean deviation of the modeled force response \( f_{\text{model}} \) (loading branch) from the experimental force response per unit cell \( f_{\text{exp}} \) normalized by the latter, for all steps \( n \), called the mean normalized deviation, \( \text{mnDev} \).

\[
\text{mnDev} = \frac{1}{n} \sum_{i=1}^{n} \sqrt{\left( f_{\text{exp},i} - f_{\text{model},i} \right)^2 / f_{\text{exp},i}}
\]

### 5.2.16 Validation of the model

The readily fitted model was used to simulate the first cycle of preconditioning for all load cases. The results allowed to evaluate the predictive capabilities of the model, provided that the corresponding reference configuration is known.

Moreover, the local experimental data, the image sequences recorded with the zoom lens, were used to qualitatively validate the model kinematics by comparing the history of the 20 degrees of freedom. Therefore, the ten rigid bodies of the model were identified and marked on three unit cells on each sixth image of the sequence of the first test cycle. The corresponding mean current configuration in terms of the 20 degrees of freedom was evaluated for each marked image and interpolated to be compared to the model results.

**Remark** The markers were set manually in each image as the transparency of the filaments and the three-dimensional relative movement between the filaments did not allow for an automated digital image correlation. Moreover, as already mentioned in section 5.2.6, there was no unique assignment between the rigid bodies and real structural points. For the deformed structure, “finding” the locations of the rigid bodies was even more challenging, as it is discussed in section 5.4.5.

### 5.2.17 Non-affine deformations

According to continuum mechanics theory, the deformation gradient represents the deformation (changes in angle and lengths and rotations) of three infinitesimally small line elements
in the vicinity of a point. The unit cell approach used in this study, assumed that this point was represented by one unit cell and its kinematics was described by the measured deformation gradient, or equivalently by the line elements \( \vec{G}_1, \vec{G}_2 (\vec{G}_3) \) and \( \vec{g}_1, \vec{g}_2 (\vec{g}_3) \). Within one unit cell, the motion of the points (rigid bodies of the multibody system) is in general non-affine and dependent on the internal equilibrium of forces and moments. The size of one unit cell is not infinitesimally small but of the same order of magnitude as microstructural constituents of the body tissues. Thus, although the global deformation gradient of each mesh unit cell might be compatible with the deformation of the underlying body tissue (global mechanical biocompatibility), there might be a local deformation mismatch due to non-affine deformations.

The local deformation was quantified by the local deformation gradients calculated at the points (at the locations of the rigid bodies) using the corresponding connected line elements in the reference and the current configuration as shown in Figure 5.17. As the model was assumed to deform in a point symmetric way the local deformation gradients at the locations of the bodies \( A_1, F \) (2x) and \( B_1 \) provided the full local kinematics information. The local deformation gradients were calculated as

\[
F_{RB} = \tilde{b}_i \otimes \tilde{B}^i
\]

where the index \( RB \) indicates the point or rigid body \( (A_1, F, B_1) \). \( B_i \) are the line elements in the reference configuration and \( b_i \) are the corresponding line elements in the current configuration (compare Figure 5.17 to identify the respective line elements for each rigid body) [138]. Lower indices mark covariant base vectors and upper indices contravariant base vectors. For the index \( i=1,2,3 \), the Einstein summation convention is to be used. The mapping between the global deformation gradient \( F \) and each local deformation gradient \( F_{RB} \) was described by \( F_{\Delta RB} \)

\[
F_{RB} = F_{\Delta RB}F
\]

The norm of the tensor \( \|F_{\Delta RB}\| = F_{\Delta RB} \bullet F_{\Delta RB} \) (\( \bullet \) being the inner product) quantifies - although in a very abstract way - the overall difference between the non-affine local and the global (or an assumed affine) deformation at each point (at the location of each rigid body). This norm was averaged for all points \( (A_1, F \) (2x) and \( B_1 \)) and normalized by the norm of the global deformation gradient. Its absolute difference to one (one, meaning no affine deformations) represents a scalar measure for the relative amount of non-affine deformation, called sum of non-affinity.

\[
\text{sum of non-affinity} = \left| \frac{1}{n_{RB}} \sum_{RB} \|F_{\Delta RB}\|/\|F\| - 1 \right|
\]

with \( n_{RB} \) being the number of points (rigid bodies) for the evaluation. The sum of non-affinity was compared at a constant load level of 0.6N/UL for all loading configurations.
Thus, it describes the mismatch between an affine deformation and the non-affine model deformation at the load level of 0.6N/UL.

**Figure 5.17:** Identification of line elements for the calculation of local deformation gradients at the points (at the locations of the rigid bodies) $A_1$, $F$, $B_1$. Note that for the body $F$ two pairs of line elements are identified. As the unit cell is assumed to deform in a point symmetric manner, the local deformation gradients are evaluated only for one half of the unit cell.
5.3 Results

5.3.1 Parameters

The parameters, summarized in Table 5.5, were found to provide a good match for all load cases. The force response was seen to be less sensitive to small variations in stiffness-like parameters, such as $c_5$, $d_3$, $e_1$, $f_3$, $g_3$, and $h_1$. In contrast, parameters defining the transition to a nonlinear stiffening, such as $c_1$, $c_2$ or upper and lower bounds for specific degrees of freedom ($d_1$, $d_2$, $f_1$, $f_2$, $g_1$, $g_2$) determine the recruitment of force elements and, thus, the specific kinematics and the nonlinearity of the force response. Even small variations might change the model kinematics, e.g. stretch of elements instead of collapse of the pore, and shift the force response along the horizontal (deformation) axis. Parameters describing the change in stiffness, such as $c_3$, $c_4$, $d_4$, $f_4$, $g_4$ mainly influence the slope of the force response. Large variations additionally determine the recruitment of force elements.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$c_1$ [mm]</td>
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<td>$e_1$ [Nm/°]</td>
<td>0.04</td>
</tr>
<tr>
<td>$c_2$ [°]</td>
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<td>$f_1$ [mm]</td>
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</tr>
<tr>
<td>$c_3$ [°]</td>
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<td>$f_2$ [mm]</td>
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</tr>
<tr>
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<td>$f_3$ [N/mm]</td>
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</tr>
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<td>$c_5$ [N/mm]</td>
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<td>$f_4$ [N/mm²]</td>
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</tr>
<tr>
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</tr>
<tr>
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<td>$g_2$ [°]</td>
<td>55</td>
</tr>
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<td>$g_4$ [Nm/°²]</td>
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<td>$d_4$ [Nm/°²]</td>
<td>32</td>
<td>$h_1$ [Nm/°]</td>
<td>0.001</td>
</tr>
</tbody>
</table>

5.3.2 Homogenized unit cell force response

A comparison between the experimental force response per UL and the force response of the model is shown in Figure 5.18 for all examined load cases. The difference in initial stiffnesses for uniaxial stress and uniaxial strain loading conditions is mapped, even though values are slightly overestimated for uniaxial strain loading conditions. The simulated curve shapes are close to the experimental loading branches or mean branches for all load cases ($mnDev \sim 20\%$) except for the 90° material direction and the off-axis material directions in uniaxial strain loading conditions ($mnDev \geq 35\%$). The underlying mesoscale mechanisms resulting in the bulged curve shape seen for the 56.5° material direction (Figure 5.18 d) are explained in section 5.4.1.
With respect to anisotropy, the same order stiff to compliant, seen in the experiments, is obtained by the multibody system (chapter 4, section 4.4.1). Even the specifically stiff behavior of the 33.5° samples in uniaxial stress loading conditions is captured by the model. For uniaxial strain loading conditions, the modeled force response of the 33.5° samples is too stiff and close to the force response of the 0° samples.

Note that only the force response in longitudinal direction is shown. For specimens in off-axis material directions, also resultant reaction forces in transverse direction occur, which could be assessed by the model, but which cannot be measured by the experimental setup.

5.3.3 Homogenized unit cell kinematics

For uniaxial stress loading conditions, the transverse behavior is not constrained, but results from the kinematics of the structure (Figure 5.19). This kinematic response is captured in an excellent way for the 0° and 33.5° material directions. Note, that again the 33.5° material direction was seen to contract exceptionally less compared to the other material directions (chapter 4, section 4.4.2). For the 90° and 56.5° material directions, the transverse contraction is underestimated. The model kinematics strongly depend on the recruitment of the filaments and thus on the choice of the force laws and the corresponding parameters. The challenges finding an appropriate set of parameters are discussed in section 5.4.4.

5.3.4 Homogenized unit cell deformation energy

In Figure 5.20, the deformation energy stored in the model and split into a rotational and a translational contribution is compared to the work done by the experimentally measured forces (loading and mean branch). The analysis of energetic contributions represents a valuable tool to identify major mechanisms (translational, rotational) from a global point of view. As expected from the global force response (Figure 5.18), the energetic fit is appropriate for all load cases, with limitations for the 33.5° and 56.5° material directions in uniaxial strain loading conditions, where the initial stiffness was overestimated or the curve shape was different, respectively.

The 0° and 90° material directions in uniaxial stress loading conditions are dominated by rotational contributions and a delayed recruitment of translational force elements (as the upper/lower rotational bounds are exceeded). The corresponding cases in uniaxial strain loading conditions are almost exclusively determined by the translational force elements.

In contrast, for the off-axis types (33.5°, 56.5°) the energy is nearly equivalently composed of rotational and translational contributions. An exception is the 56.5° material direction in uniaxial strain loading conditions with a particularly high rotational contribution at global strain levels between $E_{xx} = 0.05$ and $E_{xx} = 0.1$ (bulging of the force response). The
energetic analysis is in accordance with the experimentally identified major load case specific deformation mechanisms (chapter 4, section 4.4.3).

5.3.5 Model validation - mesoscale kinematics

Figure 5.21 provides a visual comparison of the kinematics between the model and the experimentally identified (zoom lens) average unit cell configuration at the level of the preforce and at maximum force. The model kinematics are seen to be close to the observed unit cell kinematics. The simulated collapse of the unit cell accompanied by the flapping over of the outer “arms” for the 0° material direction in uniaxial stress loading conditions is in very good accordance with the experimental findings. Moreover, the extreme shear deformations for the off-axis samples, in particular in uniaxial stress loading conditions, are captured in a realistic looking way. In contrast, for the 90° material direction in uniaxial stress loading conditions, the collapse of the inner “arms” is underestimated, while the node contracts too much. Especially, for the cases of uniaxial strain, the nodal contraction seems by far overestimated.

In summary, the major kinematic mesoscale mechanisms for the 0° and 90° material direction in uniaxial stress loading conditions is a collapse of the unit cell (half pore) which is realized by a rotation of the inner “arms”, supported by flapping of the outer “arms” (Table 5.6). Uniaxial strain loading conditions, are characterized by an alignment and stretch of the inner and outer “arms”. The kinematics of all off-axis material directions, are represented by shear-like deformations of the node. These modeled mesoscale observations are in accordance with the local experimental findings at the macroscale (chapter 4, section 4.4.3). A detailed comparison of the experimentally observed and simulated histories of the major degrees of freedom for each load case and material direction can be found in Appendix D.

5.3.6 Model validation - homogenized force response of the virgin unit cell

Figure 5.22 shows a surprisingly good match between the experimental virgin force response (first cycle of preconditioning) and the response of the model, considering the respective reference configuration. Mean normalized deviations ($mnDev$) are in the range of 12%-28%, for all loading configurations, except for the 33.5° material direction in uniaxial stress and uniaxial strain loading conditions ($mnDev = 65\%-76\%$). Again, the initial stiffness is overestimated for the 33.5° material direction in uniaxial strain loading conditions. Moreover, the bulged curve shape (compare Figure 5.18 d) is seen for the cases of 56.5° in uniaxial strain and 33.5° in uniaxial stress loading conditions. For the latter, this effect is also seen in the experimental data - even though less pronounced (compare chapter 4, section 4.5.1
Table 5.6: Major kinematic mechanisms

| uniaxial strain | 0°   | $q_{19}$, $q_{2,4,6,8}$, $q_{9,11,13,15}$, $q_{10,12,14,16}$ | contraction of the node, straightening of the outer “arms”, stretch of the stitches |
| 90°   | $q_{19}$, $q_{2,4,6,8}$, $q_{9,11,13,15}$, $q_{10,12,14,16}$ | contraction of the node, straightening of the outer “arms”, stretch of the stitches |
| 33.5° | $q_{5}$, $q_{19}$, $q_{9,11,13,15}$, $q_{10,12,14,16}$ | shear-like deformation of the node, contraction of the node, stretch of the stitches |
| 56.5° | $q_{5}$, $q_{19}$, $q_{9,11,13,15}$, $q_{10,12,14,16}$ | shear-like deformation of the node, contraction and opening of the node, stretch of the stitches |

| uniaxial stress | 0°   | $q_{3,7}$, $q_{2,4,6,8}$ | collapse of inner “arms”, flapping of outer “arms” |
| 90°   | $q_{3,7}$, $q_{19}$ | collapse of inner “arms”, contraction of the node |
| 33.5° | $q_{5}$, $q_{19}$, $q_{9,11,13,15}$, $q_{10,12,14,16}$ | shear-like deformation of the node, opening of the node, stretch of the stitches |
| 56.5° | $q_{5}$, $q_{19}$ | shear-like deformation of the node, contraction of the node |

and Appendix B, Figure B.5). The underlying mechanisms at the mesoscale are discussed in 5.4.1.

Comparing the simulated reference configurations of the virgin state to the preconditioned state (Figure 5.23), the observed specific mesoscale mechanisms (section 5.3.5), rotation of the inner and outer “arms”, alignment of the “arms”, distortion of the node, are activated and inelastically realized during the preconditioning procedure (compare also 4, section 4.4.4).
5.3 Results

Uniaxial strain

$\alpha = 0^\circ$

$c)$

$f)$

$\alpha = 90^\circ$

$e)$

$g)$

$\alpha = 33.5^\circ$

$d)$

$h)$

$\alpha = 56.5^\circ$

$\alpha = 0^\circ$

$\alpha = 90^\circ$

$\alpha = 33.5^\circ$

$\alpha = 56.5^\circ$

$\text{mnDev} = 0.19$

$\text{mnDev} = 0.18^*$

$\text{mnDev} = 0.35$

$\text{mnDev} = 0.47$

$\text{mnDev} = 0.12$

$\text{mnDev} = 0.19$

$\text{mnDev} = 0.15$

Figure 5.18: Homogenized unit cell force response, force per $UL$ versus longitudinal Green-Lagrange strain, model response (grey) versus experimental cyclic response (black) and mean branch of loading and unloading (- -), a-d uniaxial strain and e-h uniaxial stress loading conditions. The goodness of the model fit with respect to the loading branch is given by the mean normalized deviation $\text{mnDev}$. * The $0^\circ$ material direction in uniaxial stress loading condition is compared to the mean branch.
Figure 5.19: Homogenized unit cell kinematic response, transverse Green-Lagrange strain versus longitudinal Green-Lagrange strain, model response (grey) versus experimental cyclic response (black), a-d uniaxial strain and e-h uniaxial stress loading conditions.
Figure 5.20: Homogenized unit cell energy response, deformation energy / work done by applied forces versus longitudinal Green-Lagrange strain. The potential energy of the model (grey) is split into a translational and a rotational contribution, their sum (−) is seen to be equal to the work done by the reaction forces (+). The experimental work done by the applied (measured) forces is shown for the loading branch (black −) and mean branch (black - -) for a-d uniaxial strain and e-h uniaxial stress loading conditions
Figure 5.21: Mesoscale kinematics in the reference configuration (top) and at maximum load (bottom). The images recorded by the zoom lens are overlaid by the experimentally obtained average unit cell configuration (light, cyan −) and the model configuration (dark, red - -) for a-d uniaxial strain and e-h uniaxial stress loading conditions.
5.3 Results

Figure 5.22: Homogenized unit cell force response of the first cycle of preconditioning, force per $UL$ versus longitudinal Green-Lagrange strain, model response (grey) versus experimental cyclic response (black) and mean branch of loading and unloading (- -), a-d uniaxial strain and e-h uniaxial stress loading conditions. The goodness of the model fit with respect to the loading branch is given by the mean normalized deviation $mnDev$. 

<table>
<thead>
<tr>
<th>Angle</th>
<th>Uniaxial Strain</th>
<th>Uniaxial Stress</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\alpha=0°$</td>
<td>$mnDev = 0.23$</td>
<td>$mnDev = 0.18*$</td>
</tr>
<tr>
<td>$\alpha=90°$</td>
<td>$mnDev = 0.24$</td>
<td>$mnDev = 0.23$</td>
</tr>
<tr>
<td>$\alpha=33.5°$</td>
<td>$mnDev = 0.76$</td>
<td>$mnDev = 0.65$</td>
</tr>
<tr>
<td>$\alpha=56.5°$</td>
<td>$mnDev = 0.28$</td>
<td>$mnDev = 0.12$</td>
</tr>
</tbody>
</table>

$mnDev$ = mean normalized deviation
Figure 5.23: Mesoscale reference configurations, the virgin reference configuration (---) is overlaid by the preconditioned reference configuration (--) for a-d uniaxial strain and e-h uniaxial stress loading conditions.
5.3.7 Aspects of mechanical biocompatibility - non-affine deformations

The difference between the mesoscale deformation of the multibody system and an assumed affine deformation can be analyzed visually in Figure 5.24 for a constant load level of 0.6N/UL. Almost for all cases, the mismatch is largest at the node. Moreover, for all cases of uniaxial stress (Figure 5.24 a-d), the rotation and alignment of the outer arms significantly contributes to non-affinity. Quantitatively, it is confirmed, that the *sum of non-affinity* is larger for uniaxial stress loading conditions than for uniaxial strain (Figure 5.25). For the latter, all material directions are close. For uniaxial stress loading conditions in 0° material direction, the *sum of non-affinity* is remarkably high, although this value is underestimated: After the “flapping” point, the local deformation gradients at \( A_1 \) and \( B_1 \) could not be calculated as this deformation is not in accordance with general continuum mechanics. The respective local deformation gradients at the “flapping” point are taken instead.

![Uniaxial strain](image1)

**Figure 5.24:** Mesoscale kinematics at the constant load level of 0.6N/UL. The model configuration (- -) is overlaid by a configuration resulting from a homogeneous deformation (-). Due to non-affine model deformations, the configurations do not coincide. Visual differences between these configurations are a qualitative measure for the non-affinity of the unit cell deformation. **a-d** uniaxial strain, **e-h** uniaxial stress loading conditions
Figure 5.25: Sum of non-affinity for all examined loading configurations at the constant load level of 0.6N/UL. The sum of non-affinity is significantly higher for uniaxial stress than for uniaxial strain loading conditions. It is highest for the 0° material direction, followed by the off-axis directions. * Note, that for the 0° material direction in uniaxial stress loading direction, the local deformation gradients for the points A₁ and B₁ could not be evaluated after the “flapping” of the outer “arms”. The respective local deformation gradients at the “flapping” point are taken instead, which is supposed to significantly underestimate the sum of non-affinity.
5.4 Discussion

A structural model of a representative unit cell of the prosthetic mesh Gynemesh M was developed, based on multibody theory. Its geometric structure and kinematics were deduced from and in accordance with physical observations. The force laws were derived from physical considerations. One set of parameters was presented, allowing to appropriately map experimental data from eight loading configurations (two loading conditions, four material directions) with respect to the force response and the homogenized kinematics of the unit cell. Moreover, the mesoscale unit cell kinematics were captured in a representative way.

The model provides insight into the structural mechanisms at the mesoscale, which are responsible for a nonlinear and anisotropic force response. Deformation patterns, specific for different loading configurations were identified, such as rotation-, stretch- or shear-dominated unit cell deformations.

The model allows to explain the distinct differences of the force response for the virgin and the preconditioned mesh by a change of reference configuration. It allows to conclude that the preconditioned mesh is in fact a different material with a different mesostructure.

The model represents a basis to discuss aspects of mechanical biocompatibility by assessing local non-affine deformation patterns within one unit cell. Without knowing the constitutive behavior of native tissue, but assuming that it behaves as a homogeneous, isotropic continuum, criteria for mesh design optimization towards an improved mechanical biocompatibility can be deduced.

5.4.1 Experimentally observed mesoscale-based mechanisms

Referring to chapter 4, section 4.5.1, in uniaxial stress loading conditions, the mechanical behavior in 33.5° material direction (stiff initial response and restricted transverse contraction) was significantly different compared to other material directions, which is unusual for a general orthotropic continuum. The present mesoscale model is able to predict this behavior. Figure 5.20 g shows, that, compared to the other material directions in uniaxial stress, the translational energy contribution is almost equal to the rotational energy contribution. The major deformation mechanisms are stretch and shear-like deformations of the nodal elements, represented by the change of the coordinates \( q_5 \) and \( q_{19} \) (exceeding the upper bound), as shown in Figure 5.26 b. Moreover, translational energy is stored in the stitches aligned in the direction of load application, \( F - B_2, E - D_2 \), which are deformed less, due to their high stiffness. Moreover, Figure 5.21 g shows that, at maximum load, the stitches \( A_1 - A_2 \) and \( C_1 - C_2 \) are oriented in transverse direction restricting the contraction.

For the first cycle of preconditioning of the 33.5° samples in uniaxial stress loading conditions, i.e. for the virgin sample the simulated force response follows a bulged, convex shape.
In Appendix B, an interpretation attempt is presented for the experimental data, showing a comparable behavior. The model response is explained by the rotation of the nodal translational element towards the direction of load application. As $q_5$ decreases towards $-33.5^\circ$, such that $EF$ is oriented orthogonal to the loading direction the “lever” of force application (projection of $EF$ into the inertial $y$-direction) increases, resulting in a compliant response. For $q_5 < -33.5^\circ$, the “lever” decreases until reaching the lower bound of this force element ($-55^\circ$), resulting in firstly moderate and then distinct stiffening. The modeled preconditioned reference configuration is already “inelastically” deformed, such that $q_5$ is smaller than $-33.5^\circ$. The same effect is seen for the $56.5^\circ$ material direction in uniaxial strain loading conditions. In an abstracted sense, this behavior is in accordance with the experimental observation, as explained in Appendix B.

Other experimentally observed local phenomena, such as mesh distortions and edge effects (stripes pattern) cannot be modeled by a characteristic unit cell model, as they violate the underlying homogeneity and periodicity assumption.

### 33.5° material direction in uniaxial stress loading conditions

![Figure 5.26](image)

Figure 5.26: Major kinematic contributions for the $33.5^\circ$ material direction in uniaxial stress loading conditions, represented by the generalized coordinates a $q_5$, b $q_{19}$.

### 5.4.2 Preconditioning -

**a change of structural geometry, a change of material**

Defining an appropriate preconditioning procedure for the meshes was identified as one of the major experimental challenges (chapter 4, section 4.5.2). Preconditioning also represents one of the major modeling challenges. The present approach uses the knowledge about the virgin and preconditioned reference configurations, which might differ significantly (Figure 5.23), to model the respective mechanical response. However, for any new loading condition, such
as equibiaxial stress, the preconditioned reference configuration is not known. Even for cyclic uniaxial stress tests at a different maximum load, the preconditioned reference configuration is not known. Thus, the present model has predictive capabilities only provided that a reproducible (which is usually the preconditioned) reference configuration is known. The preconditioned mesh, in fact, has to be seen as a different material with a different underlying mesostructure.

5.4.3 Non-affine deformations and mechanical biocompatibility

With respect to the mechanical biocompatibility of the mesh, there are three major aspects which can be learned from examining the non-affine deformation within the unit cell and which can be considered for mesh design optimization. Uniaxial stress loading conditions activate different mechanisms than uniaxial strain loading conditions. The collapse of the pores, seen in uniaxial stress is related to a higher level of non-affinity than the alignment and stretch of strands in uniaxial strain. The node deforms in a highly non-affine manner represented by a distinct contraction and shear, which might cause local tissue damage. The flapping behavior of the outer “arms” for the 0° material direction, in uniaxial stress loading conditions, leads to a significant deformation mismatch between mesh and underlying tissue, as this behavior is not seen for any continuum.

The calculated scalar parameter *sum of non-affinity* is proposed as it summarizes the differences in local and global deformation related to the amount of global deformation. This measure is abstract and does not differentiate between rotational and translational deformation mismatches, which might affect tissues in a different way. Moreover, as the deformation gradient is a measure taken from continuum mechanics theory, it cannot be applied where this theory is not valid, as seen for the “flapping” behavior. However, assuming that body tissues behave as continua, such deformation patterns should be avoided in general and might be treated separately in mesh design optimization.

The evaluation of non-affine deformations is based on the mesoscale model. Its kinematics are in good agreement with the experimentally observed and abstracted mesoscale kinematics. However, due to the abstraction of the complex microstructure by simplified mesoscale elements, both the findings of the model and of the local experimental analysis DO capture main kinematic mechanisms, but DO NOT provide information on the detailed kinematics of the filaments. Moreover, as intrinsic for models, also this model is based on assumptions and does not perfectly match the experimental observations. These limitations have to be kept in mind for a more detailed evaluation of the mesh’s mechanical biocompatibility.
5.4.4 The fitting compromise

The chosen set of parameters is not a unique one. According to our strategy, the 0° material direction was the preferred direction throughout the fitting process and is thus mapped best. Alternative parameter sets might be found, providing a better fit for other material directions. Apart from the parameters determining the force laws, variations of the model geometry might have an additional influence on the force and kinematic response.

The current set of parameters was found by manual fitting. A smart optimization algorithm could be applied in order to find an improved match. However, there are sources of uncertainties, questioning the need for an improved fit: Inter-specimen variabilities of the experimentally measured deformations were seen in the range of 2%-9% (chapter 4, section 4.5.3). There is further uncertainty in the exact horizontal position of the measured curves, depending on the chosen level of preforce. Moreover, dissipative effects (hysteresis) are not included in the current model, however, both, the loading and the unloading branch are relevant descriptions of the mechanical behavior of the mesh. Which one should be modeled? Another limitation is the calculation of the specific force per \( UL \) for the off-axis directions. As the number of unit cells (and strands) is not discrete, it was estimated by an interpolation approach (chapter 4, section 4.3.3). Normalizing the measured force by this estimated value introduces further uncertainty.

5.4.5 Experimental mesoscale kinematics - interpretation of the rigid body positions

There is no unique assignment of the rigid bodies to structural fix points. During deformation, filaments move relatively to each other or are constrained by other filaments. Thus, the micro-structure might change significantly, which is in particular seen for the node deformed in off-axis directions. As it is demonstrated in Figure 5.27, there is obviously some room for interpretation with respect to the abstraction of the deformed unit cell. At this stage, the experimental examination of the mesoscale kinematics provides valuable but only qualitative findings. A more detailed, quantitative analysis would require a large number of specimens to be tested (due to the high intra-specimen variability of the deformation), intelligent, automated procedures to identify characteristic points within the complex microstructure and improved methods for the digital image correlation to follow these characteristic points in a reliable way.

5.4.6 Finding an appropriate modeling approach

In general, three different approaches are followed in the current literature to model textile fabrics, mainly weaves: the detailed microscale approach [153–155], the structural mesoscale
5.4 Discussion

a) c)

Figure 5.27: Abstraction of the deformed unit cell for the 33.5° (top) and the 56.5° (bottom) material directions in uniaxial stress loading conditions. a Within the node, two strands of stitches are interwoven by a continuous filament (dark, red). During deformation, the strands of stitches are rearranged relative to the continuous filament. The marker (circle) indicates the node between two stitches. b In order to find abstracted positions for the bodies E and F, the position of the markers are projected to the tangents on the continuous filament of the corresponding strand. c The abstracted unit cell (light, cyan) with the rigid body positions (circles). It is seen that for the off-axis directions, the rigid body positions are not assigned to structural fix points.

approach [141–152] and the continuum macroscale approach [89,90], all of which offer benefits but also have drawbacks.

Within the framework of the present thesis, a microscale model of a mesh unit cell has been worked on based on CT data (Appendix C and [156]). As the geometry is mapped in a realistic way, there are ideally no assumptions on structural or material properties. Moreover, three-dimensional effects, which cannot be evaluated with our present experimental setup (and are in general difficult to be seen from experiments) can be examined through simulations. However, such finite element models are computationally expensive. In our case, the solution of the multi-contact model led to convergence problems and the need for non-physical assumptions on friction coefficients and filament interpenetration. Comparing the outcome to the modeling efforts, the microscale approach is inefficient and thus not recommended for knitted fabrics.

The mesoscale approach, followed in this study, abstracts the real geometry in a physical way. Therefore, mainly geometry related phenomena, such as nonlinearity of the force response and anisotropy are captured, even in a predictive way. Such models can be implemented very efficiently. The major challenge is finding an appropriate level of abstraction, and thus a compromise between simplicity and physical mapping. As the force elements represent structures, such as stitches or nodes, appropriate structural force laws, which are in general nonlinear have to be defined. The respective parameters, in our case 20, can often
not directly be transferred from real physical properties and thus require a rather extensive fitting procedure.

With respect to a continuum model approach it has to be considered that fabrics are not continua. It has to be analyzed, if their homogenized mechanical behavior allows for a corresponding model assumption. Continuum models can be implemented within a FEM code and thus can be used to simulate inhomogeneous load cases in a very efficient way, which is not straightforward for unit cell models at the micro- or the mesoscale. However, most continuum approaches are not based on an underlying microstructure. A large experimental data set is needed to fit the corresponding model parameters, in a way that the model could have predictive capabilities. Based on our experimental observations (chapter 4), the continuum assumption is questionable for the present knitted fabric.

5.5 Conclusions

The presented structural model allows to understand and simulate mesoscale mechanisms leading to a globally anisotropic and nonlinear force response. Further, it provides deeper understanding of the experimentally observed phenomena, which cannot be explained by using a continuum approach.

Still, there is room for improvement. Aiming at a predictive model formulation, referring in particular to a representative reference configuration, inelasticity has to be included in the force laws. Moreover, viscoelasticity and dissipative effects might represent important mechanical phenomena to be taken into account for a more comprehensive assessment of the meshes’ mechanical biocompatibility. Regarding the identification of the constitutive parameters of the force elements, experiments on mesoscale structures, such as strands of stitches, would represent a more physical but rather challenging approach.

Unit cell models allow to simulate homogenous load cases. For simulations of inhomogeneous, multiaxial loading conditions in complex structures, the implementation of this model in a finite element code is needed. Alternatively, the knowledge gained from the present model can be used to design a corresponding physically based continuum model formulation. The structural model -if predictive- can be used to perform virtual experiments increasing the “experimental” database to fit a more sophisticated continuum model formulation with an increased number of parameters.

So far, the mesoscale model has been used to assess the level of non-affine deformations, resulting in valuable findings to discuss aspects of mechanical biocompatibility without the need for assumptions with respect to the underlying tissue. Even recommendations for mesh design optimization can be deduced. In the future, the model can be extended by matrix elements with different levels of affinity, e.g. at the unit cell boundaries or at all rigid bodies,
5.5 Conclusions

simulating the underlying tissue at different stages of integration. The corresponding mesh-matrix interactions could provide additional knowledge with respect to the consequences of local deformation mismatches, which might be the reasons for severe clinical complications.
Conclusions and outlook

This study was organized to start from in vivo observations to stepwise reduce compliance with physiological conditions. In turn, the mechanical characterization was stepwise refined towards more detailed theoretical descriptions. With respect to boundary conditions and materials the focus was shifted from physiological to controllable. The investigated length scales changed from homogenized, macroscopic to detailed, mesoscopic. Clinical experimental protocols were defined with respect to patients’ needs and feasibility, protocols in the lab with respect the mechanical phenomena of interest. Mechanical descriptions and models ranged from empirical kinematic parameters to theoretical constitutive formulations. The objectives of each step were different and contributions were made in different ways.

6.1 Contributions of the present work

**In vivo mechanical characterization of the vaginal wall** The aspiration measurements represent the clinically most relevant study as it focused on a patient specific, mechanical characterization of pelvic floor tissues under physiological boundary conditions. However, the present setup and protocol for transvaginal mechanical characterization of the vaginal wall did not allow to discriminate between women with and without cystocele. The definition of a reproducible reference configuration and according boundary conditions were identified as major challenges, which apply to in vivo studies in general. POP related processes and other influencing factors, such as age and estrogenization, affect different structures at different length scales (anatomy, vaginal wall structure and tissue composition), in a complex and interdependent way. The current procedure is apparently not sensitive to one specific length scale. However, significant changes influencing all length scales, such as a surgical prolapse repair, were detected, proving the general feasibility of the technique. Based on the present results and discussions, we believe that a characterization of the pelvic floor mechanics related to POP needs to consider processes at different length scales and at the same time needs to be able to differentiate between them. In this context, the present study might contribute to an improvement of currently available and the development of alternative in
vivo techniques, which are still subject of fundamental research.

**Ex vivo characterization of mesh explants**

The experimental study on explants was mainly focused on physiologically relevant testing and boundary conditions, which is directly related to the relevance of the mechanical data and parameters extracted. Explants from animal models are highly relevant study objects, as they mimic the ingrown in vivo conditions of the meshes, including the complex host response. The findings show that uniaxial and biaxial tests yield different results with respect to the mechanical behavior of mesh explants. As a consequence, the experimental loading conditions have to be selected such that they reproduce the expected in vivo load and are consistent with the animal model used, which in the present case, is true for the biaxial inflation measurements. The mechanical parameters should span the physiological range, as they are dependent on the analyzed load level.

Given the large scatter of the results, the small sample size was a limitation of this study. Despite this restriction and rating the level of mechanical biocompatibility only with respect to the biaxial stiffness, we might conclude that the large porous mesh is to be preferred for the tissue considered, as corresponding explants are similarly stiff as native tissue. However, no direct quantitative recommendations can be derived for the selection or optimization of mesh materials to be used in biaxial loading conditions for the human pelvic floor, where both loading conditions and levels of deformation might be significantly different.

The presented methodologies provide a simple and reliable protocol and corresponding data analysis procedures, which can be used by biomedical engineers to compare different implant materials. A corresponding categorization of different meshes is expected to be physiologically relevant - also for humans.

**Ex vivo characterization of mesh VHB composites**

Mesh elastomer composites have been shown to represent an appropriate non-biological model system to predict the mechanical behavior of explants. Uniaxial and biaxial tensile tests with these specimens lead to the same qualitative conclusions as the corresponding experiments with explants. Moreover, the stiffness of explants was quantitatively determined based on data obtained from mesh elastomer composites. Obviously, the influence of the complex host response cannot be captured by this model system. Nevertheless, the load case specific kinematic mechanisms were analyzed and were seen to appropriately mimic the explant behavior: The biaxial response of embedded and ingrown meshes is composed of the superposed response of the constituents. For the large porous mesh, the uniaxial response is dependent on mesh-matrix-interaction and less on the matrix material properties. Due to the latter, uniaxial tests using dry meshes are generally not appropriate for estimating the behavior of the ingrown mesh
implant.

In contrast to biological tissues, mesh elastomer composites are easy to handle, inexpensive and the test results show moderate scatter. This model system allows to evaluate appropriate test protocols and parameters and to examine major mechanical phenomena prior to animal studies, which can then be designed in a more sensitive and efficient way. Furthermore, tests with these materials represent valuable tools for a mechanical investigation and engineering optimization of prosthetic meshes, preceding validation in animal studies.

**Mechanical characterization of a dry prosthetic mesh at different length scales** The major contribution of the present experimental study on the dry mesh is the inclusion of a lower scale, the mesoscale, which so far has not been considered for prosthetic meshes. The variety of tested loading configurations allowed to examine macroscale mechanical phenomena, such as the nonlinear force response, anisotropy, hysteresis and load history dependent inelasticity (preconditioning). The local analysis of the homogenized unit cell deformation revealed local mechanical phenomena which are due to mesoscale mechanisms.

The findings provide a database for the development of physically meaningful numerical models for mesh materials, including the choice of an appropriate modeling approach, as well as a large basis of force and kinematic data for parameter adjusting. Moreover, in general, a broad knowledge of global and local mesh phenomena allows to question whether these phenomena are desired or a result of the manufacturing process and thus allows to incite a more active mesh design optimization process. Both the design of mesh kits and their clinical application should be more based on the mechanical properties of the mesh itself, which can be imagined from the following observations: Uniaxially and biaxially loaded regions, i.e. strips and sheets, respond with a different initial stiffness (factor of three). The response of strips significantly depends on the loaded material direction, which might not be the same for all anchoring strips (compare Figure 1.1). Load history dependent inelastic deformations might require an appropriate preconditioning procedure before mesh insertion. The resulting initial stiffnesses are higher after such a procedure. Mesh kit design and application should consider these aspects in order to avoid unexpected clinical outcomes. An increased knowledge of all these different mesh phenomena can help improving clinical procedures towards a mechanically more biocompatible mesh application.

**A physically based mesoscale model for a prosthetic mesh** The presented structural model showed excellent descriptive capabilities with respect to the complete set of experimental data. Thus, it allows to understand and simulate mesoscale mechanisms leading to the global anisotropic and nonlinear force response. Further, it provides deeper understanding of the experimentally observed phenomena, which could not be explained by using a continuum approach.
The observed preconditioning effects represent the major restriction for the model to have appropriate predictive capabilities without a known reference configuration. Inelasticity would have to be included in the force laws, requiring further experiments. An examination of the corresponding virgin and preconditioned reference configurations lead to the conclusion that cyclic load in fact changes the mesh’s microstructure and thus the material in a homogenized sense.

Due to its descriptive capabilities, the model was seen to be an appropriate tool to quantify the level of non-affinity within the unit cell, which can represent a criterion for the mechanical biocompatibility of the mesh. With respect to an underlying continuum matrix, non-affine mesh deformations result in a non desired mismatch of deformation and in mesh-matrix interactions. The evaluation of non-affine deformation patterns allows to deduce recommendations for mesh design optimization with respect to properties of mesoscale elements, i.e. the nodes and stitches.

6.2 Outlook

In vivo mechanical characterization of the vaginal wall A deeper understanding of the pathogenesis of POP might help to identify critical pelvic floor structures and tissues and the corresponding length scales to be investigated. Transvaginal measurements are particularly attractive as the vaginal canal allows non-invasive access to pelvic floor structures. However, the role and the relevance of the vaginal wall in POP is not yet clear. An improved transvaginal measurement technique should aim at a normalization of the reference configuration and the according boundary conditions with respect to the length scale of interest, i.e. exclude influences from other length scales.

A separate investigation of phenomena at different scales might help to clarify the complex interplay of diverse influencing factors. 3D imaging of pelvises of women with and without POP during defined loading maneuvers, e.g. valsalva, combined with procedures for optical strain measurements are expected to help understanding the macroscopic kinematics of pelvic floor compartments dependent on the stage of POP. Moreover, the histological analysis of the biopsies taken during surgery and their correlation with clinical and mechanical data, is supposed to shed further light on the role of the vaginal wall microstructure.

Ex vivo characterization of explants and mesh VHB composites The inflation technique is proposed to supplement or even replace uniaxial tensile tests in future animal studies as it provides data that are physiologically more relevant, as more consistent with the animal models commonly used. Having in mind the findings of the present study, standardized protocols for mesh explants are strongly recommended to consider physiologically relevant loading conditions.
A categorization of different mesh types and an assessment of their mechanical biocompatibility with respect to the in vivo loading conditions will not be possible until a large database, including increased sample sizes and a larger variety of different mesh types, is established. It should be noted that mechanical biocompatibility is a broad concept. Here, only the homogenized, structural stiffness is looked at, but other criteria are expected to contribute. Moreover, again, this database could only provide quantitative data with respect to the animal model used. A direct transfer of conclusions to mesh applications in humans is not recommended until respective reference data are available.

Improvements of the current setup will allow for a more flexible definition of physiologically relevant protocols, including preconditioning procedures and a wider range of strain rates. Moreover, digital image correlation for a local strain analysis in the central region of the sample will allow to omit the isotropy and homogeneity assumption and rather allow to investigate the influence of these phenomena. Tests on mesh VHB composites should represent the material of choice for these future developments.

**Mechanical characterization of a dry prosthetic mesh at different length scales** The presented experimental protocols and procedures for data analysis might contribute to the discussion on standardized protocols to evaluate relevant mechanical mesh properties for premarket approval. Additionally to commonly proposed properties, such as the deformation dependent stiffness and the strength, parameters describing anisotropy, hysteresis and inelastic deformations might be thought of. Moreover, time dependent behavior, which is not treated here, seems worth to be quantified. In any case, it is strongly recommended to include mesoscale based parameters, such as the intra-specimen inhomogeneity of deformation, a description of edge effects or locking behavior in specific loading directions.

**A physically based mesoscale model for a prosthetic mesh** In general, unit cell models allow to simulate homogenous load cases. However, the presented model formulation can be implemented in a finite element code to simulate inhomogeneous, multiaxial loading conditions in complex geometries. Alternatively, the knowledge gained from the present model seems valuable to be used to design a corresponding physically based continuum model formulation. Working with the structural model and performing virtual experiments will increase the “experimental” database to fit a more sophisticated model formulation with an larger number of parameters.

Apart from that, the descriptive capabilities of the present model seem further helpful to simulate the influence of a matrix. According elements can be implemented with different assigned material properties. Constraining these elements at the unit cell boundaries or at all rigid bodies could simulate different stages of integration. The corresponding mesh-matrix interactions will provide additional knowledge with respect to the consequences of
local deformation mismatches, which might be the reasons for severe complications.

Concluding, it can be said that POP is a complex clinical indication, acting at different length scales of the pelvic floor. Understanding its pathogenesis as well as appropriate treatment has to consider this multiscale property and the same is true for corresponding “mechanical descriptions”. In vivo measurements allow for physiological boundary conditions and are at the same time affected by their complex multiscale character. This challenge was not solved in the present study. However, a stepwise reduction of the system complexity, has been shown to allow for relevant contributions with respect to an evaluation of quantitative criteria for an improved mechanical biocompatibility of prosthetic meshes.

The combined experimental and numerical approach to characterize the prosthetic mesh at different length scales exemplarily shows a way for a detailed analysis of its mechanical properties. These protocols and procedures can be applied to any kind of prosthetic mesh and might contribute to the discussion on a standardized mesh description. From a more general perspective, the presented approach can be used to model other microstructure based materials, such as knitted fabrics or biological membranes.

6.3 Are currently available prosthetic meshes suitable for prolapse repair?

Prosthetic meshes have initially been designed for hernia repair and were later used for prolapse repair, often without premarket approval for this specific application.

In hernia repair, prosthetic meshes close holes or support weak regions of the abdominal wall. Physiological loading conditions are biaxial with moderate to high levels of membrane tension (16 N/cm [115]) and deformation. Prosthetic meshes inserted for prolapse repair are subject to locally varying uniaxial (strips) or biaxial (sheets) loading conditions with large ranges of deformation. However, estimated physiological levels of membrane tension are one order of magnitude lower as compared to corresponding values for the abdominal wall [44]. Moreover, as the mesh is draped within the three-dimensional structure of the pelvic floor it is subject to three-dimensional structural deformations, such as relocation or folding. Meshes are anchored within the bony structures of the pelvis, they replace ligament structures and they are attached to and integrated within the connective tissue. Thus, the design of mesh kits should consider (i) the different physiological loading conditions of the pelvic floor compared to abdominal wall, and (ii) the multiscale character of the pelvic floor mechanics - a challenging task.

According to this hypothesis, there is no reason why meshes succeeding in hernia repair should be appropriate for prolapse repair. Explant studies using a rabbit abdominal wall model do not provide sufficient evidence on mechanical biocompatibility to qualify meshes
to be applied in human pelvic floors. Assuming that the present complications are, among other aspects, related to the mechanical properties of meshes, criteria have to be evaluated to assess their mechanical biocompatibility with respect to their specific application in pelvic organ prolapse repair. The capability to globally reconstruct the anatomy and a structural uniaxial stiffness, which is “similar” to native tissue, - which seem to be the current criteria - are certainly not sufficient.

On the contrary, the present experimental investigations on the dry mesh (chapter 4) give rise to formulate concrete arguments why NOT use the present large-porous meshes, in general, for prolapse repair. The initial stiffness of the preconditioned mesh is by a factor of three larger in biaxial as compared to uniaxial loading conditions. With respect to mesh kits this means that initial load application leads to significant lengthening of the anchoring strips and a delayed biaxial deformation of the sheets. From the experimental observations, it can be assumed that a not negligible part of the deformation is inelastic. As the anchoring strips are cut in different material directions, different arms are deformed differently. Consequences might be unpredictable mesh dislocation and loss of support. Additionally, it has to be considered that the preconditioned mechanical response of the inelastically deformed mesh is much stiffer compared to the virgin response and might thus differ significantly from the response of the surrounding native tissue.

Apart from these concerns with respect to the meshes’ macroscopic deformation, deformation mismatches at the mesoscale might lead to complications at the tissue level. Edge effects were seen to cause systematic inhomogeneous deformation patterns, particularly for the 90° material direction. For the mechanical response of strips loaded in 33.5° material direction, locking and out of plane warping were observed. Moreover, significant non-affine deformations were evaluated including flapping of stitches and distinct contractions and shear deformations of the nodal structures. All these local deformation patterns cannot be expected for a continuum and thus for an underlying native tissue. A mismatch of deformation results in relative movement and stress concentrations (normal tensile and compressive as well as shear) at the mesoscale. Consequences might include impairment of mesh integration, local mesh dislocation and wrinkling or tissue injury by local shearing or compressing of tissue components. Scar tissue formation, mesh contraction or erosion might result from these local, destructive events.
Numerical investigations of the inflation experiment

Inflating a sample leads to an inhomogeneous deformation pattern, as there is a region of equibiaxial stress in the center of the sample and regions of uniaxial strain close to the clamps. As for the current setup no facilities or procedures for optical strain measurements were available, a numerical model approach was used to analyse the local kinematics of an inflating sample.

A generic axisymmetric finite element model was setup as shown in Figure A.1 a, using Abaqus (Dassault Systemes, Simulia, France). A hyperelastic material model (Ogden type) was applied, - the type of model and the constitutive parameters were chosen arbitrarily as only kinematic relations, not the response in terms of stresses, were examined. The load case was defined as a linearly increasing pressure acting on the inner surface of the sample. The in-plane stretch $\lambda_p$ in the apex region was calculated as the ratio between the current length $l$ and the reference length $l_0$ of an arc at the center of the sample, $\lambda_p = l/l_0$. The arc length $l$ was extracted at several steps of the simulated inflation process and was plotted versus the apical displacement $d$. Fitting a function to these data, the arc length $l$ was found to correlate with the apical position $d$ according to a polynomial of order four (Figure A.1 a,b). In this way, local equibiaxial stretch could be determined from the apical position $d$.

A analog procedure was applied to determine an inverse relationship between the radius of curvature $r$ in the central region and the apical displacement $d$ (Figure A.1 a,c). The central radius of curvature $r$ was needed to calculate the averaged membrane tension $Tm$ [N/cm] in the central region, according to the static equilibrium, applied to a spherically inflated membrane, as $Tm = 0.5 \cdot p \cdot r[N/cm]$ with $p$ being the current pressure.
A Numerical investigations of the inflation experiment

Figure A.1: a FEM model of the inflated sample. Axissymmetric model, thickness of the sample=2mm, hyperelastic material model. The contour plot shows the element strain solution, d: apical position, l: current length of an arc at the center of the sample, r: radius of curvature in the central region, b Kinematic relationship between the arc length l and the apical position d. FEM solution (+) versus order 4 fit (−), c Kinematic relationship between the central radius of curvature r and the apical position d. FEM solution (+) versus rational fit (−).
Local phenomena due to mesoscale mechanisms - further methodological aspects

This appendix provides a detailed description of local mesh phenomena due to mesoscale mechanisms. The focus is rather on methodological and mechanical aspects, which should provide further background to the discussion on mainly clinical aspects in chapter 4, section 4.5.

**Global mesh distortions** The mesh is assumed as an orthotropic structure based on periodically arranged patterns (pores), which are symmetric with respect to the preferred material directions ($0^\circ$, $90^\circ$). Global deviations from this perfectly periodic knitting pattern, such as *global mesh distortions* introduced during the manufacturing process, result in unexpected global phenomena, such as shear deformations caused by load application in $0^\circ$ material direction (Figure B.1).

![Global mesh distortion](image)

**Figure B.1:** Global mesh distortion: Shear deformation for the $0^\circ$ material direction, **a** Shear strain $E_{12}$ versus longitudinal strain $E_{22}$ for the three specimens tested in uniaxial strain loading conditions in the $0^\circ$ material direction. For one specimen (---), unexpected shear deformation is seen. **b, c** Undeformed close ups the mesh. For the specimen in **b** (---), global mesh distortions are represented by a deviation of the $\vec{c}_2$-material direction (--) from the $\vec{c}_1$-machine direction (---). For the specimen in **c** (--), these two directions nearly coincide.
Local phenomena due to mesoscale mechanisms - further methodological aspects

Local mesh distortions  In [131], locally inhomogeneous deformations have been explained by local deviations from the periodic pattern. The local strain analysis has been reported useful in general to assess the level of homogeneity of the deformation [127,128,131], and thus to evaluate the validity of homogenized global mechanical measures of deformation. However, the added value of this analysis strongly depends on its resolution. In this study, the resolution was chosen equal to the size of one half pore, which is the smallest representative pattern, considering the four material directions examined. Thus, the seen inhomogeneities of 15%-22% for the 0° material directions (see also following discussion on edge effects) can mainly be attributed to deviations from periodicity.

Due to the fact, that in uniaxial strain loading conditions the full kinematics are prescribed, local deviations from periodicity can much less be compensated by rearrangements as compared to uniaxial stress loading conditions, were the tranverse behavior is not constrained. The intra-specimen variability of the longitudinal strain component $E_{22}$ is therefore larger for uniaxial strain loading conditions (22% compared to 15% in uniaxial stress, 4.14).

Edge effects  The 0° direction, is the only material direction, with no filaments ending freely at the edges. In contrast, for the 90° direction, all filaments end at the free edges. During load application, free filaments are pulled into the specimen (partly rearranging the locally inhomogeneous mesh structure), only restricted by the inter-filament friction. These edge effects result in local structural deformations and lead to the stripes pattern seen for the 90° specimens in uniaxial strain and uniaxial stress loading conditions (Figure B.2). Edge effects are the reason for the outlying high intra-specimen variability of deformation (up to 35% for the 90° material direction in uniaxial strain). Similar but less systematic edge effects are present in the off-axis specimens.

![Figure B.2: Stripes pattern in the longitudinal strain $E_{22}$ distribution for specimens, loaded in the 90° material direction in a uniaxial strain and b uniaxial stress loading conditions](image)

Size effects  For the off-axis material directions, the inter-specimen reproducibility was seen to be less compared to the 0° and 90° material directions (6%-9% inter-specimen variability of longitudinal strain at maximum force, Figure B.3). The specimen dimensions are not
determined by a discrete number of ULs or load carrying strands (in particular not realizable for the 56.5° direction), but by global dimensions (in [mm]). A globally distorted mesh structure impairs the reproducibility of the structural geometry of these specimens, leading to local variances in the number of closed pores per width (although global dimensions are equal), as shown in Figure B.4. As the ULs are nearly of the same order of magnitude as the global dimensions of the specimens, this structural variation causes size effects, represented by an outlying mechanical response. Only a significantly increased specimen size, in the range of 100ULs, would help to overcome these size effects. It is noted, that not only closed pores but also open pores contribute to load transmission, even though at a lower level, which was accounted for by the interpolation approach to define the off-axis ULs.

**Figure B.3:** Outliers due to size effects, caused by differences in the structural geometry. Normalized force per UL versus longitudinal strain $E_{22}$ for a uniaxial strain, 56.5° material direction, b uniaxial stress, 33.5° material direction, c uniaxial stress, 56.5° material direction

**Figure B.4:** Variations of the structural geometry: For the off-axis specimens, here shown for the 56.5° direction, the width is determined as a global dimension. The number of load carrying strands per width, varies along the length of the specimen. The rectangle (–) includes five closed pores, whereas within the dashed (– -) rectangle, there are only four closed pores, the fifth one is incomplete (arrow). The distribution of five and four pore cross sections is different for all specimens.

**Structural anisotropy** In uniaxial stress loading conditions, 33.5° specimens show several structure dependent phenomena. Compared to the other material directions, the global
force response is the stiffest (recall Figure 4.10 b). In Figure B.5 a, it is seen that during load application, stitches are oriented transversely, *internally restricting the globally free contraction*. Instead, out-of-plane warping is observed, which cannot be quantified by our two-dimensional optical system, but is seen as a longitudinal stripes pattern in the distribution of the transverse strain component (Figure B.5 b). Moreover, stretch in the $33.5^\circ$ material direction does not activate compliant collapse mechanisms, but rather nodal shear and stretch of the stiff strands (compare the angles $\angle (\vec{G}_1^*, \vec{G}_2^*)$ and $\angle (\vec{g}_1^*, \vec{g}_2^*)$ in Table 4.4).

![33.5° uniaxial stress](image)

**Figure B.5:** Mesoscale phenomena for the $33.5^\circ$ material direction in uniaxial stress loading conditions. a) Transversely oriented stitches restrict the transverse contraction and result in a stiff mechanical response. One such stitch is exemplarily marked by the arrow. b) Out of plane warping is the reason for the longitudinal stripe pattern, seen for the distribution of the transverse strain $E_{11}$. c) Not only different levels of longitudinal strain $E_{22}$ at maximum load, but also different curve shapes are observed comparing the virgin and the preconditioned force response, i.e. normalized force per $UL$ versus longitudinal strain $E_{22}$. d) Hysteresis is also observed for the kinematic response, i.e. transverse ($E_{11}$) versus longitudinal ($E_{22}$) strain.

Distinct differences in the curve shapes of the global force response of specimens loaded from the virgin or the preconditioned state have been observed (Figure B.5 c). The initial response is similarly stiff, which is not the case for the other material directions. As the preconditioned curve follows the typical knee shape, the virgin curve flattens towards a more compliant, almost linear response. Comparing the unloaded reference configurations in the virgin and preconditioned state in Figure B.6, the following explanation seems obvious. An inelastic rearrangement of the strands of stitches within the nodal structure results in an initially stiff response of the virgin sample, which is followed by an elastic, rather compliant structural collapse of the stitches (convex, bulged curve shape). In contrast, the deformation of the inelastically preconditioned sample starts with this structural collapse phase, which is then followed by a transition to material stretch (nonlinear force response).

Another phenomenon observed is the hysteretic kinematics, meaning that the deformation pattern ($E_{11}$ versus $E_{22}$) is different for the loading and unloading branch, which is interpreted as the consequence of friction during structural rearrangements (Figure B.5 d).
Figure B.6: Interpretation of the different curve shapes for the virgin response compared to the preconditioned response for the 33.5° material direction in uniaxial stress loading conditions, comparing a the virgin to b the preconditioned reference configuration. The strand of stitches slips/is pulled “through” the node during preconditioning. Due to friction, this rearrangement is not reversed during unloading.

Modeling aspects  Continuum model formulations and representative unit cell approaches are both based on a homogeneity or periodicity assumption. The observed local inhomogeneities cannot be mapped and might limit the validity of such models. However, structure dependent phenomena as discussed for the 33.5° material direction in uniaxial stress loading conditions, can be captured and quantitatively explained by a structural model at the mesoscale (chapter 5).
A microscale model of a prosthetic mesh based on CT data

This study was conducted by Stan Banaszak as his master project supervised by the author. A detailed documentation of this work can be found in the corresponding master thesis [156].

C.1 Motivation

This project was aimed at a detailed, microstructural unit cell model of the prosthetic mesh Gynemesh M for numerical simulations using FEM.

C.2 Methods

C.2.1 Extraction of a geometric unit cell model

A piece of mesh (resorbed state) containing several unit cells was scanned with a micro CT by Scanco Medical (Scanco Medical AG, Brütisellen, Switzerland). In a first step, the CT images were automatically segmented in order to create a three-dimensional surface model, using the software package Amira 4.0 (VSG - Visualization Sciences Group, Burlington, Massachusetts, USA). Geomagic Studio 9.0 (Geomagic GmbH, Morrisville, North Carolina, USA) was applied to extract one unit cell and manually process the model, e.g. by separating filaments at contact points or repairing surface defects. The geometry model was parameterized using Matlab (The Mathworks, Natick, Massachusetts, USA) by defining splines to describe the path of the filaments’ midline. In order to adjust the geometry to fulfill periodicity requirements, mainly concerning the inclinations of the filaments at the boundaries, the spline model was imported to the CAD software UGS/NX 7.5 (Siemens, Munich, Germany). This software allowed for simple geometric manipulations. The resulting parameterized geometry was assumed to represent one specific unit cell geometry, however not a
representative one.

### C.2.2 Approaching a representative unit cell geometry model

In order to gain a representative unit cell geometry, the knitting process was modelled, using Ansys (Ansys Inc., Canonsburg, Pennsylvania, USA). The properties of the polypropylene filaments, the filament diameter, equal to 0.089mm and the filament E-Modulus, equal to 9168.4 N/mm$^2$, were provided by Johnson&Johnson, gained from internal tests. Starting from a straight filament, meshed with beam elements, the nodes of subsequent cross sections were stepwise constrained to the positions given by the initial geometry model. The resulting prestressed model was fixed at the boundaries and periodicity conditions were applied. Contacts were defined between the filaments. The idea was to release all geometric constraints related to the simulation of the knitting process and allow the model to rearrange itself. A resulting unit cell configuration would be based on the minimum of deformation energy. However, this complex multi-contact problem could not be solved due to convergence problems. Several measures have been taken, such as changing contact parameters, contact search regions and even building up a new mesh, using solid type elements. None of these helped to solve the problems. Thus, the non-representative geometric model was used, instead, for preliminary simulations.

### C.2.3 Preliminary simulations

The simulations were intended to map the experiments reported in chapter 4 of this thesis. Only uniaxial strain loading conditions were looked at. Kinematic boundary conditions were applied by stepwise prescribing nodal displacements according to the respective loaded material direction. Displacements in transverse direction were constrained. Unloading was only regarded in a few investigative simulations. The application of periodic boundary conditions required coupling of nodal degrees of freedom which impaired the convergence of the model. As periodic constraints have only minor influence for uniaxial strain loading conditions, they were omitted at this preliminary stage.

### C.3 Results

Preliminary simulation results show good agreement of the force response with experimental data for uniaxial strain loading conditions in 0° and 90° material directions (Figure C.1). With respect to the kinematics, the model provides realistic looking results (Figures C.2, and C.3). A more detailed quantitative analysis of the kinematics, similar to the examinations in Appendix D is documented in [156]. Apart from these findings, the outcome at this
stage is poor.

![Diagram](image_url)

**Figure C.1:** Force response per $UL$ versus longitudinal engineering strain for uniaxial strain loading conditions in **a** $0^\circ$ and **b** $90^\circ$ material directions. The simulated force response (grey) is within the limits provided the experimental data (black). There is some uncertainty with respect to the horizontal position of the experimental curve, which depends on the defined level of the preforce. Note: The experimental curves correspond to the preconditioned curves.
Figure C.2: Kinematics of the 0° material direction in uniaxial strain loading conditions at a simulated strain of 0.50

Figure C.3: Kinematics of the 90° material direction in uniaxial strain loading conditions at a simulated strain of 0.91
C.4 Discussion

The present microscale model maps the orthotropic force response in uniaxal strain loading conditions. Preliminary simulations allow to analyse the detailed filament kinematics, also in thickness direction, which is not possible with the present experimental dataset. Many limitations were identified associated with this detailed modeling approach. Due to the complex unit cell geometry, the development of a geometry model is a time-consuming task. Working with an according FEM model is even more extensive as the model contains a large number of degrees of freedom (∼800 for the beam model, ∼200000 for the solid model) and the solution procedure is nonlinear, due to both large structural deformation and the implementation of contacts. The latter was seen to cause convergence problems, which could not be solved in this study. Consequently, we did not succeed setting up a representative geometry model.

Moreover, difficulties occurred defining periodic coupling conditions, which are particularly important for uniaxial stress loading conditions. These were not simulated in this study. Additionally, the definition of displacement boundary conditions to simulate off-axis load cases turned out to be challenging, as either displacement or coupling conditions have to be defined in a coordinate system rotated with respect to the inertial system, which was not straightforward using a commercial software.

The scanned geometry is a virgin geometry. The preconditioned reference configuration is unknown. At the present stage, preliminary simulations showed, that the implementation of friction allowed to model hysteresis in cyclic load. However, only very small remaining deformation (global inelasticity) was observed in the model response, which apparently did not map the experimentally observed preconditioning effects.

C.5 Conclusions

So far, only minor added value was seen in using a model at the microscale. In general, this model approach was shown to be ineffective for knitted fabrics, due to significant technical efforts and the problems which still have to be solved in order to come up with a stable descriptive or possibly predictive model. Therefore, this approach was not continued after this project.
In order to validate the model kinematics in more detail, the mesoscale kinematics of the model were compared to the local experimental observations (zoom objective). Therefore, the respective histories of the relevant degrees of freedom were analysed, which were specific for each loading configuration (loading conditions and material direction). Note, that due to the small number of unit cells observed and due to their interdependence, the experimental data are not fully representative.

In the following figures, upper and lower bounds, implemented in the force laws are marked for orientation. This analysis helps to understand the specific kinematic mechanisms in more detail.
Figure D.1: Major kinematic contributions for the $0^\circ$ material direction in uniaxial strain loading conditions, represented by the generalized coordinates $a\ q_{19}$, $b\ q_2, q_4, q_6, q_8$, $c\ q_9, q_{11}, q_{13}, q_{15}$, $d\ q_{10}, q_{12}, q_{14}, q_{16}$.
90° material direction in uniaxial strain loading conditions

Figure D.2: Major kinematic contributions for the 90° material direction in uniaxial strain loading conditions, represented by the generalized coordinates \( q_{19}, q_2, q_4, q_6, q_8, q_9, q_{11}, q_{13}, q_{15} \), \( q_{10}, q_{12}, q_{14}, q_{16} \).
Figure D.3: Major kinematic contributions for the 33.5° material direction in uniaxial strain loading conditions, represented by the generalized coordinates \( a \ q_5, \ b \ q_{19}, \ c \ q_9, \ q_{11}, \ q_{13}, \ q_{15}, \ d \ q_{10}, \ q_{12}, \ q_{14}, \ q_{16} \).
56.5° material direction in uniaxial strain loading conditions

Figure D.4: Major kinematic contributions for the 33.5° material direction in uniaxial strain loading conditions, represented by the generalized coordinates a $q_5$, b $q_{19}$, c $q_9$, $q_{11}$, $q_{13}$, $q_{15}$, d $q_{10}$, $q_{12}$, $q_{14}$, $q_{16}$.
0° material direction in uniaxial stress loading conditions

Figure D.5: Major kinematic contributions for the 0° material direction in uniaxial stress loading conditions, represented by the generalized coordinates a $q_3$, $q_7$, b $q_2$, $q_4$, $q_6$, $q_8$, c $q_9$, $q_{11}$, $q_{13}$, $q_{15}$, d $q_{10}$, $q_{12}$, $q_{14}$, $q_{16}$. 
90° material direction in uniaxial stress loading conditions

![Graph showing major kinematic contributions for 90° material direction.](image)

**Figure D.6:** Major kinematic contributions for the 90° material direction in uniaxial stress loading conditions, represented by the generalized coordinates $a \ q_3$, $q_7$, $b \ q_{19}$.

33.5° material direction in uniaxial stress loading conditions

![Graph showing major kinematic contributions for 33.5° material direction.](image)

**Figure D.7:** Major kinematic contributions for the 33.5° material direction in uniaxial stress loading conditions, represented by the generalized coordinates $a \ q_5$, $b \ q_{19}$. 
56.5° material direction in uniaxial stress loading conditions

Figure D.8: Major kinematic contributions for the 56.5° material direction in uniaxial stress loading conditions, represented by the generalized coordinates \( a \ q_5 \), \( b \ q_{19} \).
References


[125] B Röhrnbauer and E Mazza. Uniaxial and biaxial mechanical characterization of a prosthetic mesh at different length scales. *submitted*.


List of publications

Journal publications


submitted

B Röhrnbauer, M Bajka, C Betschart, D Perucchini, E Mazza, and D Scheiner. Mechanical characterization of the anterior vaginal wall using the novel aspiration technique in vivo.

B Röhrnbauer and E Mazza. Uniaxial and biaxial mechanical characterization of a prosthetic mesh at different length scales.

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**Posters**


**Talks**
