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Biomechanical Impact of a Subsequent Childbirth on the Female Pelvic Floor

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Master in Bioengineering

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Abstract

Every year, millions of childbirths occur globally, yet the rate of maternal morbidity and mortality remains unacceptably high. Multiple studies have demonstrated that pelvic floor injuries occurring during vaginal delivery can be regarded as a major contributor to the development of pelvic floor dysfunction. Consequently, these disorders have been extensively researched lately, especially in relation to the effects of pregnancy and childbirth. Although recent studies on the impact of a first birth on pelvic floor lesions have increased exponentially, there is still a dearth of information about subsequent deliveries. Epidemiological statistics indicate that the likelihood of pelvic floor dysfunction rises with the number of deliveries and that subsequent deliveries are linked to a higher risk of dysfunction. Therefore, it is crucial to investigate how later deliveries affect these lesions. Thus, this research aims to implement a mechanical model developed in the context of the finite element method to analyze the biomechanical influence of the accumulated damage on the pelvic floor musculature for subsequent deliveries. For that, computer simulations of a first vaginal delivery and second vaginal deliveries were performed, where different muscle recovery rates and fetal head sizes were tested.

In this study, a validated finite element model comprising the pelvic floor muscles and a fetal head was utilized to simulate vaginal births. The muscles were defined using a transversely isotropic viscohyperelastic constitutive model with cumulative damage and nonlocal averaging, and the fetal head was considered a rigid structure. The simulations of vaginal birth were led with the fetus in the vertex presentation and occipito-anterior position. Therefore, a first simulation was carried out with a 50th percentile head. Then two second simulations, one considering a 50% recovery of the accumulated damage in the muscle fibers, and the other considering no recovery of the muscle. These second simulations were executed using a 50th percentile head and a 5th percentile head so that the influence of head size on damage accumulation could also be assessed. Muscle stretch, maximum principal stresses, and damages were assessed in all simulations.

The computer model indicated that the muscle fibers are more damaged at the pubovisceral muscle's origin. Hence, this is the most likely location for a rupture to occur, which is consistent with previous studies. Apart from that, the number of damaged elements is greater for the simulations of the second birth, being higher for the simulations where there was no muscle recovery. Therefore, the first simulation only had one noteworthy damage element, the second simulation with recovery presented two elements with high damage, and the second simulation with no recovery showed one high damage element and one critical damage element. This evidence is in line with the literature, which indicates that subsequent childbirths are correlated with a higher risk of developing pelvic floor dysfunction. Lastly, a smaller head results in less stretch, stresses, and accumulated damage in the muscles.

In conclusion, this investigation complements earlier work by including a new study topic and provides a step forward in accomplishing more accurate results in the biomechanics of childbirth.

Resumo

Todos os anos ocorrem milhões de partos a nível mundial, mas as taxas de morbidade e mortalidade materna continuam a ser inaceitavelmente elevadas. Vários estudos demonstram que as lesões do pavimento pélvico que ocorrem durante o parto vaginal podem ser consideradas como um dos principais fatores que contribuem para o desenvolvimento da disfunção do pavimento pélvico. Consequentemente, estas perturbações têm sido objeto de uma investigação extensiva nos últimos tempos, especialmente em relação aos efeitos da gravidez e do parto. Embora estudos recentes sobre o impacto de um primeiro parto nas lesões do pavimento pélvico tenham aumentado exponencialmente, ainda existe uma escassez de informação sobre partos subsequentes. As estatísticas epidemiológicas indicam que a probabilidade de disfunção do pavimento pélvico aumenta com o número de partos, e que os partos subsequentes estão associados a um maior risco de disfunção. Assim, é crucial investigar como os partos posteriores afetam estas lesões. Portanto, esta investigação tem como objetivo implementar um modelo mecânico desenvolvido no contexto do método dos elementos finitos para analisar a influência biomecânica do dano acumulado na musculatura do pavimento pélvico em partos subsequentes. Para tal, foram realizadas simulações computacionais de um primeiro parto vaginal e de um segundo parto vaginal, onde foram testadas diferentes taxas de recuperação muscular e tamanhos de cabeça fetal.

Neste estudo, foi utilizado um modelo validado de elementos finitos, que inclui os músculos do pavimento pélvico e uma cabeça fetal, para simular partos vaginais. Os músculos foram definidos usando um modelo constitutivo visco-hiperelástico transversalmente isotrópico com dano cumulativo e média não local, e a cabeça do feto foi considerada uma estrutura rígida. As simulações do parto vaginal foram realizadas com o feto na apresentação vértex e na posição occipito-anterior. Portanto, uma primeira simulação foi realizada com uma cabeça de percentil 50. Em seguida, duas segundas simulações, uma considerando uma recuperação de 50% do dano acumulado nas fibras musculares, e outra não considerando recuperação do músculo. Estas segundas simulações foram executadas utilizando uma cabeça de percentil 50 e uma cabeça de percentil 5, para que a influência do tamanho da cabeça na acumulação de danos também pudesse ser avaliada. O alongamento do músculo, as tensões principais máximas, e os danos foram avaliados em todas as simulações.

O modelo computacional indicou que as fibras musculares são mais danificadas na origem do músculo pubovisceral. Portanto, este é o local mais provável para a ocorrência de uma rutura, o que é consistente com estudos anteriores. Para além disso, o número de elementos danificados é maior nas simulações do segundo parto, sendo superior nas simulações em que não houve recuperação muscular. Assim, a primeira simulação teve apenas um elemento de dano notável, a segunda simulação com recuperação apresentou dois elementos com dano elevado, e a segunda simulação sem recuperação identificou um elemento com dano elevado e um elemento com dano crítico. Esta evidência está de acordo com a literatura, que indica que os partos subsequentes estão correlacionados com um maior risco de desenvolvimento de disfunção do pavimento pélvico. Por fim, uma cabeça mais pequena resulta em menor alongamento, tensões e danos acumulados nos músculos.

Concluindo, esta investigação complementa trabalhos anteriores ao incluir um novo tópico de estudo e fornece um passo em frente na obtenção de resultados mais precisos sobre a biomecânica do parto.

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“Há é que ter paciência, dar tempo ao tempo. Já devíamos ter aprendido, e de uma vez para sempre, que o destino tem de fazer muitos rodeios para chegar a qualquer parte.”

José Saramago

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Acronyms and Abbreviations

FEM	Finite Element Method
FI	Fecal Incontinence
HGO	Holzapfel-Gasser-Ogden
IAP	Intra Abdominal Pressure
ICS	International Continence Society
IUGA	International Urogynecological Association
LAM	Levator Ani Muscles
LOA	Left Occipito-Anterior
LOP	Left Occipito-Posterior
LOT	Left Occipito-Transverse
ML	Machine Learning
MUI	Mixed Urinary Incontinence
OA	Occipito-Anterior
OP	Occipito-Posterior
PFD	Pelvic Floor Dysfunction
PFM	Pelvic Floor Muscles
POP	Pelvic Organ Prolapse
ROA	Right Occipito-Anterior
ROP	Right Occipito-Posterior
ROT	Right Occipito-Transverse
STATEV	State Variables
SUI	Stress Urinary Incontinence
UI	Urinary Incontinence
UMAT	User Material
USA	United States of America
UUI	Urgency Urinary Incontinence

Chapter 1

Introduction

1.1 Context

On a global scale, millions of childbirths take place annually. Nevertheless, the level of maternal morbidity and mortality remains unacceptably elevated. For instance, every day in 2020, around 800 women passed away from preventable factors associated with pregnancy and childbirth, which translates to a rate of approximately one maternal death every two minutes ([World Health Organization, 2023](#)). Biomechanical complexities constitute a significant contributor to these numbers ([Hemmerich et al., 2019](#)), so numerous studies have been undertaken to investigate the mechanisms of labor using computational models ([Oliveira et al., 2016b](#)).

The biomechanical impact of vaginal birth on the female pelvic floor is an essential aspect of the reproductive health of women that demands attention and understanding. Vaginal birth, while a natural and essential process, can exert considerable forces on the pelvic floor muscles (PFM) and connective tissues, which may lead to structural changes.

Female pelvic floor dysfunction (PFD), encompassing conditions such as urinary incontinence (UI), fecal incontinence (FI), and pelvic organ prolapse (POP), significantly impacts a considerable proportion of the female population. Nearly one-quarter of all women and one-third of older women report experiencing at least one PFD throughout their lifetime in the United States of America (USA). This issue represents a major contributor to both physical and psychological morbidity, emerging as a critical global health challenge ([Horak et al., 2014](#)).

There are several risk factors for the development of PFD, including vaginal parity, advanced maternal age, birth-related maternal injuries, method of delivery, persistent chronic straining, obesity, high-impact exercise, ethnicity, as well as racial background, collagen irregularities, hysterectomy, elevated intra-abdominal pressure, and genetics ([Easley et al., 2017](#); [Torrisi et al., 2012](#); [Romeikienė and Bartkevičienė, 2021](#); [Teixeira et al., 2018](#)).

Regrettably, the availability of extensive longitudinal and prospective studies on the accurate incidence of PFD remains quite limited. Nonetheless, there is unanimous acknowledgment that PFD poses a substantial health challenge for aging women ([Bozkurt et al., 2014](#)).

Data reveals that the enduring consequences of these disorders post-childbirth prompt over 300,000 women annually in the USA to undergo surgery (DeLancey et al., 2024). However, it is important to note that surgical interventions frequently manifest high recurrence rates, primarily due to a constrained understanding of the involved physiological loading conditions (Easley et al., 2017).

Moreover, with the escalating global population of older women, there arises a more significant societal burden manifested through substantial healthcare expenses and an overall decline in quality of life (Horak et al., 2014). The prevalence of PFD among American women is projected to witness a 50% increase by the year 2050 in comparison to the statistics reported in 2010 (Wu et al., 2011). As the aging population expands, the need for medical services addressing these conditions is anticipated to rise steadily. Accordingly, the economic burden associated with PFD will continue to grow (Gonzalez and Anger, 2022).

In the recent research panorama, the correlation between PFD and factors such as vaginal childbirth and pregnancy has become a subject of investigation (Memon and Handa, 2013). Multiple researches have indicated a higher prevalence of pelvic floor disorders in women who have experienced childbirth compared to those who have not. Current literature suggests that pelvic floor injuries sustained during vaginal delivery constitute one important contributing reason to the onset of PFD (Oliveira et al., 2016b). While not commonly emphasized in discussions about motherhood, it is estimated that a range of 13% to 41% of women undergoing vaginal delivery experience some extent of pelvic muscle injury, particularly in the context of their first labor experience (Vila Pouca et al., 2022a).

The application of mathematical models indicates that the primary cause of pelvic floor trauma is attributed to the delivery of the first child. However, epidemiological data reveals that subsequent childbirths are correlated with a higher risk of POP (Horak et al., 2014). In comparison to nulliparous women, those with one child are four times more likely, while women with two children had an 8.4 times higher probability of encountering POP (Mant et al., 1997a). Likewise, women who underwent one delivery exhibited a relative risk of 1.9, while those with two deliveries had a relative risk of 2.3 of developing stress UI (Rortveit et al., 2003).

That being said, in the biomedical research domain, employing biomechanics is beneficial to comprehend the influence of modified pelvic loading conditions on the emergence of PFD. Biomechanics employs principles from mechanical engineering to comprehend how biological systems undergo shape alterations in response to external forces (Clark-Patterson et al., 2022). Therefore, the progression of research in this field, more precisely related to the study of the female pelvic cavity, culminates in a more in-depth understanding of the possible associated pathologies. This, in turn, contributes to the enhancement of existing treatments.

Consequently, in contemporary times, several computational modeling tools have surged that strategically employ mathematical models to evaluate and comprehend the phenomena behind complex problems with a wide range of application domains, such as medicine. So, biomechanical models have taken center stage in several studies, providing a platform to estimate mechanical variations in the maternal pelvic floor during childbirth, analyze fetal descent dynamics, and ex-

plore the influence of fetal head modeling on the mechanical dynamics of the maternal pelvic floor (Oliveira et al., 2016b). This approach also serves as a substitute for animal experimentation, which has historically involved distressing procedures causing injury, impairment, or mortality in animals.

Accordingly, progress in this domain leads to numerous innovations and advancements in the medical field, enhancing the overall quality of life. Specifically, in the context of childbirth, it allows for the comparison of maternal and newborn outcomes to reduce morbidity and mortality.

1.2 Motivation

The biomechanical aspects of the female reproductive system remain significantly underexplored, and women have been expected to pay a lifelong price for the birth of a baby (Ashton-Miller and DeLancey, 2009). Consequently, to make well-informed and suitable choices, it is imperative to have a comprehensive understanding of the dynamic structure of the female pelvis. Fortunately, over the past 20 years, researchers worldwide have been contributing to the field of reproductive mechanics. This led to the realization that childbirth is one of the preponderant factors in the potential development of PFD. Therefore, a series of studies have been conducted to assess its impact on the female pelvis, but most only consider women who have never given birth before. Thus, the focus, until now, has been chiefly on the biomechanical impact of the first birth on the PFM and how this may contribute to consequent injuries. In this sense, although the literature portrays the effect of a first birth on PFM, there are still very few studies on the effect of a subsequent birth using computational models.

As mentioned before, some studies report that the prominence of PFD increases with the number of births. Accordingly, this work aims to contribute to a better knowledge of the impact of the first birth on injuries suffered in subsequent births.

1.3 Objectives

This dissertation, conducted as the final milestone of the Master's degree in Bioengineering, specialized in Biomedical Engineering, at the Faculty of Engineering of the University of Porto, seeks to explore the biomechanical impact of a first vaginal childbirth on injuries sustained during subsequent births. For this purpose, a numerical simulation based on the Finite Element Method (FEM) was carried out using Abaqus® software. The model employed for the simulation represented the pelvic floor muscles and the fetal head. Therefore, simulations were performed for the first and second delivery processes, and the effect of the recovery percentage in a second birth was also evaluated. This methodology aimed to estimate biomechanical changes (stretch, stresses, damage accumulation) in tissues, which might help explain known phenomena associated with childbirth. It is also important to mention that transversely isotropic visco-hyperelastic properties with cumulative damage were attributed to the PFM via user material (UMAT) subroutines and the Fortran programming language.

Hence, the objectives of this work are the following:

- Modelling of the fetus head and the PFM of the woman.
- Application of a constitutive model with cumulative damage – change of PFM properties.
- Numerical simulations of first and second vaginal births, with the second simulation incorporating the damage resulting from the first birth.
- Simulation of a second birth with the introduction of a recovery percentage.
- Simulation of the second births with a smaller fetal head.
- Analysis of the effects induced in the PFM of the woman (stretch, maximum principal stresses, and damage).

1.4 Structure of Dissertation

In order to achieve the stipulated objectives, this dissertation is structured in different chapters. Therefore, in addition to the introduction, this document contains eight more chapters. Chapter 2 presents a description of the anatomy of the female pelvis. Chapter 3 focuses on some pathologies associated with the pelvic cavity. In Chapter 4, the process of labor is described, in particular, the implications for the fetus and the maternal pelvis. Chapter 5 tackles the topic of FEM, as well as some concepts about solid mechanics, hyperelasticity, viscoelasticity, and damage models. Besides, the software used in this work, Abaqus®, is introduced. Some models of vaginal birth already developed are mentioned in Chapter 6. Chapter 7 displays the finite element model used in this study, the modifications implemented, and the methodology adopted. The results of the numerical simulations performed and an analysis and discussion of the results obtained are included in Chapter 8. Finally, the ultimate conclusions and suggestions for future work are delineated in Chapter 9.

1.5 Achievements

This work has been submitted to and accepted by CMBBE 2024: The 19th International Symposium on Computer Methods in Biomechanics and Biomedical Engineering, an international conference dedicated to biomechanics and biomedical engineering advancements. The accepted paper is titled "Exploring the Biomechanical Consequences of Subsequent Childbirth on the Female Pelvic Floor" and will be presented at the symposium in Vancouver, Canada, on July 30, 2024. Detailed information about the paper can be found in Appendix A.1.

Chapter 2

Pelvic Anatomy

2.1 Introduction

An in-depth comprehension of pelvic anatomy is of great significance in diagnosing PFD and guiding subsequent treatments ([Fritsch, 2007](#); [Corton, 2009](#)). Indeed, aside from the brain and cranial base, the pelvic anatomy stands out as one of the most complex in the human body, which makes the assessment of the pelvic floor and associated conditions even more complicated ([Scotti et al., 2003](#)).

2.2 Female Pelvis

The female pelvic floor plays a multifaceted role, including aspects of pleasure, sexuality, parturition, urination, urinary continence, defecation, fecal continence, and maintaining the position of pelvic organs. To fulfill these diverse functions, the pelvic floor relies on an intact anatomical structure composed of muscles, connective tissue, and nerves. Additionally, impairments in pelvic floor function and continence can result from not only direct anatomical injuries, such as those occurring during vaginal delivery, but also from neural control dysfunction, observed in conditions like neurologic disease, diabetic neuropathy, and cognitive disorders ([Jundt et al., 2015](#)).

Positioned between the thighs inferiorly and the lumbar region of the abdomen superiorly, the pelvis is the central part of the human body and is responsible for supporting the weight of the upper body. The human pelvis is a complex anatomical region, which includes the bony pelvis, the pelvic cavity, the pelvic floor, and the perineum ([Eggleton and Cunha, 2023](#)).

2.2.1 Bony Pelvis

The three main functions of the bony pelvis are force transmission (it transmits the weight of the trunk, upper limbs, and head to the lower limbs through the hip joints), protection (it safeguards the pelvic viscera and vessels contained within its boundaries), and muscle attachment (the external surface of the bony pelvis serves as an anchor point for several powerful muscles and reproductive organs) ([Eggleton and Cunha, 2023](#); [Bazira, 2021](#)).

The bony pelvis has an irregular yet complete ring shape. Essentially, the bony pelvis is an important connecting link between the spine and the lower limbs. Therefore, the bony pelvis results from the fusion of the ilium, ischium, and pubis (forming the hip bones), along with the sacrum and coccyx. Its anterior boundary is marked by the pubic bones, while the sacrum articulates with the coccyx inferiorly, and the ilium and ischium define its lateral limits (Bazira, 2021; Boaviagem et al., 2019). Throughout the extent of the pelvic surface, various points serve as origins and insertions for soft tissues, ligaments, and muscles. These structures play a significant role in influencing and being influenced by pelvic movements (Boaviagem et al., 2019).

As can be observed in Figure 2.1, the bony pelvis can be subdivided into two main parts: the anterior portion, known as the pelvic girdle, comprising the pubis, ischium and ilium, and the posterior segment, represented by the pelvic spine, which includes the coccyx and sacrum. The innominate bones, formed by the fusion of the pubic, ischial, and iliac bones on each side, contribute to the structure of the pelvic girdle. This results in two innominate bones, one on the right and the other on the left side (Chaudhry et al., 2023).

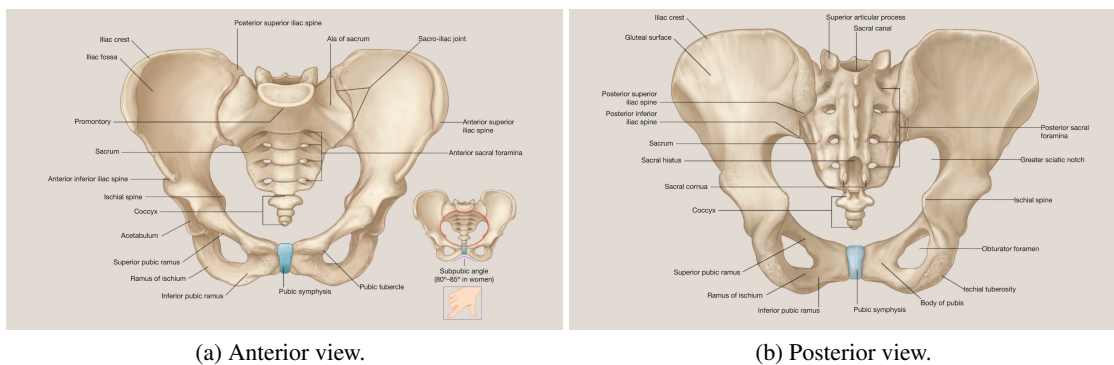


Figure 2.1: Schematic representation of the bony female pelvis (adapted from Bazira (2021)).

The pelvic bones form four pelvic joints. The two hip bones are connected anteriorly at the pubic symphysis, which results in a robust and nearly impenetrable bond. Posterolaterally, on each side, the medial aspect of the hip bone articulates with the corresponding lateral aspect of the sacrum, forming a synovial joint known as the sacroiliac joint. In turn, the inferior end of the sacrum articulates with the superior surface of the coccyx, creating the sacrococcygeal joint. It becomes important to mention that both the pubic symphysis and the sacrococcygeal articulation are secondary cartilaginous joints (Bazira, 2021).

It can be deduced that the foundational framework of the human pelvis was passed down from quadrupedal predecessors. However, the transition to bipedalism triggered substantial alterations in the pelvic girdle's muscular and skeletal structures. Additionally, the subsequent growth in adult and neonatal brain size led to further adaptations to the pelvic anatomy (DeSilva and Rosenberg, 2017).

Previous publications have suggested that the human pelvis exhibits notable evolutionary adaptability, particularly compared to other primates (DeSilva and Rosenberg, 2017). Research

indicates that populations facing socioeconomic challenges, such as poverty, gender inequality, and malnutrition, may experience a decrease in both height and pelvic size among women. Conversely, obesity contributes to the birth of disproportionately large newborns. The interplay of these factors could lead to adverse consequences, complicating and endangering childbirth conditions, for both opposite situations (Wells, 2017).

2.2.1.1 Shape of the Bony Pelvis

The anatomical configuration of the birth canals of women shows notable diversity in their shapes, more than other anatomical features. This variability is evident within and among human populations, and distinct pelvic shapes have been correlated with differences in the process of labor (Betti, 2021).

The most predominant approach for characterizing variations of the bony pelvis involves the classification method proposed by Caldwell and Moloy (1938). This system takes into account the configuration of the birth canal entrance through radiographic images. The female bony pelvis, based on this classification, is generally categorized into four primary types: gynecoid, anthropoid, android, and platypelloid, as can be observed in Figure 2.2. However, it is essential to note that several pelvises exhibit mixed types, representing a combination of characteristics rather than conforming to a singular pure type (Eggleton and Cunha, 2023).

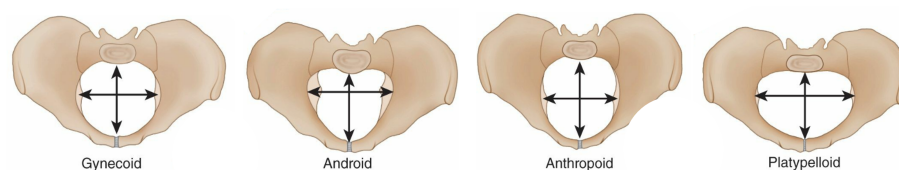


Figure 2.2: Different shapes of bony pelvis described by Caldwell & Moloy (adapted from Posner et al. (2013)).

The gynecoid conformation is characterized by an ovoid to a nearly circular inlet, a wide pubic arch, and a spacious pelvic outlet, possessing the biggest internal dimensions. In contrast, the android shape exhibits a heart-shaped inlet with convergent sidewalls, resulting in smaller internal dimensions than the gynecoid pelvis, often called the "classical" male pelvis. The anthropoid form features an ovoid-shaped inlet and outlet, with greater dimensions from front to rear than from side to side. On the other hand, the platypelloid structure is relatively uncommon in both male and female individuals, characterized by an exceptionally wide and shallow pelvic inlet (Eggleton and Cunha, 2023; Wobser et al., 2018).

It is essential to comprehend the morphological changes of the birth canal in obstetric practice, as the shape of the birth canal is involved in several aspects of the labor process, including the probability of complications and the position and rotation of the fetus (Betti, 2021).

Several sources suggest that the gynecoid pelvic type is particularly well-suited for childbirth due to its broad, elliptical cranial opening. This characteristic allows for a more substantial coupling, accommodating the pressure exerted by the fetal head (Wobser et al., 2018). Additionally,

authors have consistently identified the gynecoid form as the most prevalent pelvic type in their samples, considering it representative of the "normal" female pelvic shape (Betti, 2021).

In fact, some researchers have verified that, generally, European women exhibit a transversally oval pelvic inlet, which aligns with an android, gynaecoid, or platypelloid shape. In turn, in several East Asian and sub-Saharan African populations, the average pelvic inlet tends to be rounder and frequently oval in the anterior-posterior direction, which corresponds to a higher prevalence of the anthropoid pelvic shape (Betti, 2021; Kurki, 2013; Betti and Manica, 2018). That being said, in the context of obstetrics, the portrayal of "normal" delivery is based on a particular pelvic morphology prevalent among European women. Accordingly, evidence suggests that this model inadequately represents the diversity among women, particularly those of non-white ethnicities. Consequently, the normalization of a model rooted in white-centric female anatomy and childbirth experiences is empirically unfounded, experimentally inadequate, and potentially threatening, imposing disadvantages and discriminatory practices on women of non-European ancestry (Betti, 2021).

Moreover, the maternal mortality rates among black women in Britain compared to their white counterparts are five times higher. Similarly, women of Asian descent face twice the risk of maternal death (Knight et al., 2017). Various factors contribute to these inequities, including differences in socio-economic status, disparities in access to and quality of prenatal care, environmental factors, and underlying structural racism. Consequently, achieving inclusivity in maternal care demands comprehensive understanding and anticipation of human diversity, culminating in developing protocols that cater to the needs of women from diverse ethnic backgrounds (Betti, 2021).

2.2.2 Pelvic Cavity

The area enclosed by the pelvic bones is referred to as the pelvic cavity, characterized by a funnel-shaped structure (Chaudhry et al., 2023; McEvoy and Tetrokalashvili, 2023). Superiorly, the pelvic cavity connects to the abdominal cavity. Inferiorly, its boundary is defined by the musculofascial pelvic diaphragm, suspended above the pelvic outlet, thus forming the pelvic floor. Posteriorly, it is enclosed by the coccyx and the lower portion of the sacrum, with the upper segment of the sacrum serving as a roof over the posterior portion of the cavity (Parente, 2008).

The human pelvic cavity is partitioned into two distinct sections, namely the greater pelvis and the lesser pelvis, as illustrated in Figure 2.3. The greater (false) pelvis forms part of the abdomen and consists of the space above the iliopectineal line, including the two iliac fossae and the region between them. In contrast, the lesser (true) pelvis is situated below the iliopectineal line. It is delimited anteriorly by the pubic bones, posteriorly by the sacrum and coccyx, and laterally by the ischium and a small portion of the ilium. The greater pelvis protects the lower abdominal organs, while the lesser pelvis acts as the skeletal structure for both the pelvic cavity and the perineum compartments of the trunk (Chaudhry et al., 2023; Parente, 2008).

In conclusion, the pelvic cavity accommodates various vital organs and tissues, including the urinary bladder, pelvic colon, internal reproductive organs, and rectum, as well as a multitude of muscles, arteries, veins, nerves, and the pelvic connective tissue (Chaudhry et al., 2023).

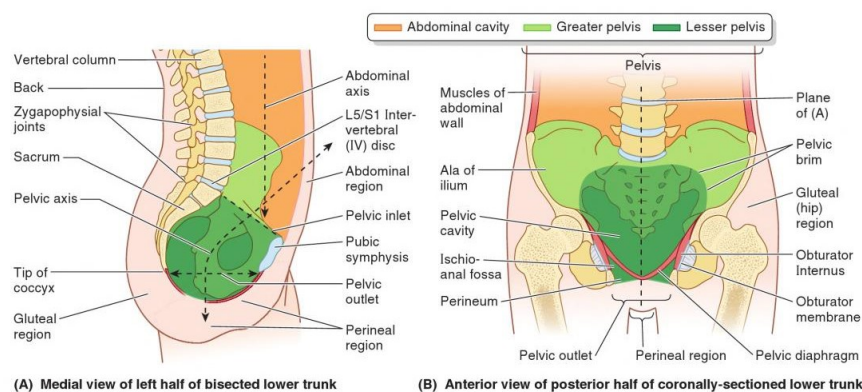


Figure 2.3: Pelvis and perineum (adapted from [Moore et al. \(2013\)](#)).

2.2.3 Pelvic Floor

All of the muscles, connective tissue, and organs that fill the pelvic cavity are collectively referred to as the pelvic floor. The pelvic floor comprises the urogenital triangle, the urogenital diaphragm, and the pelvic diaphragm. Besides, the pelvic floor also includes the levator hiatus, which allows the urethra, vagina, and rectum to pass through, and it extends from the symphysis pubis to the coccyx. Furthermore, it divides the pelvic cavity from the perineum ([Chaudhry et al., 2023](#); [Silva et al., 2017](#)).

Moreover, its main functions involve delimiting the pelvis and abdominal cavity, providing structural support for visceral organs, and controlling the rectal and urogenital openings. The female pelvic floor provides support for various organs, including the bladder, uterus, vagina, pelvic colon, rectum, and anus, as demonstrated in Figure 2.4. Additionally, its muscular fibers act as sphincters to maintain continence for the rectum and urethra ([Chaudhry et al., 2023](#)).

Nevertheless, the female pelvic floor remains an understudied area of the body from a biomechanical viewpoint. Its anatomic components are required to prevent POP and incontinence daily during the movements and increases in intra-abdominal pressure (IAP) linked with everyday athletic activity. However, they must also allow waste to be expelled by feces and urine ([Ashton-Miller and DeLancey, 2007](#)).

That being said, to explain pelvic disorders, it is fundamental to study the mechanical characteristics of the female pelvic floor. These disorders can result from the inability of the tissues to retain the pelvic organs in their regular placements due to a decrease in elasticity. The inadequate mechanical properties of the supporting structures, such as deteriorated muscles or ligaments or stiffness changes, and perturbations within the pelvic fascia may be a contributor factor ([Silva et al., 2017](#)).

2.2.3.1 Pelvic Floor Muscles

The pelvic diaphragm comprises the coccygeus and levator ani muscles (LAM), which interconnect to form the structural framework of the pelvic floor. The coccygeus muscle, which

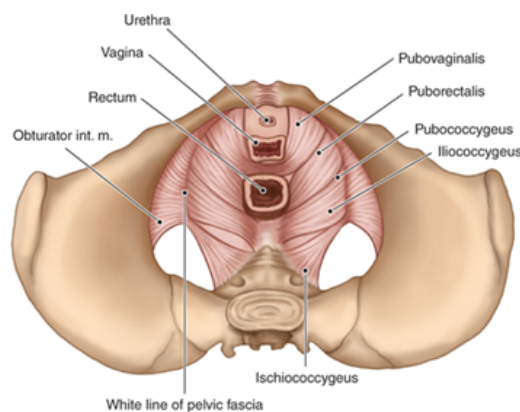


Figure 2.4: Female pelvic floor (adapted from [Posner et al. \(2013\)](#)).

extends from the side of the sacrococcygeal junction to the tip of the ischial spine, is a small, thin muscle. In turn, LAM is one of the key anatomical structures supporting the pelvic organs. It is a bigger and wider muscle that comprises three components (the puborectal and pubococcygeal – which collectively form the pubovisceral – and iliococcygeal muscles). LAM is a movable, thin, gutter-shaped muscle sheet that extends from one side of the pelvic walls to the other. It originates from each side of the walls along the condensation of the obturator fascia, known as the tendinous arch of the LAM. While fibers from the ischiococcygeus attach to the sacrum and coccyx, the remaining sections of the muscle converge at the midline. Additionally, fibers from the iliococcygeus intersect partly by fibrous tissue and create the iliococcygeal raphe posterior to the anorectal junction ([Standring, 2015](#); [Silva et al., 2017](#); [Bazira, 2021](#)).

When the LAM experiences pathological weakening or momentary inactivation, the pressure on one side of a pelvic organ becomes larger compared to the other. This imbalance can lead to the descent of the organ, resulting in genital prolapse ([Janda et al., 2003](#)).

The rectum, urethra, and vagina, in females, are transmitted by the levator hiatus, a natural space that exists anteriorly between the medial borders of the two levator ani muscles ([Bazira, 2021](#)).

The urogenital diaphragm consists of two muscles: the deep transverse perineal and the external urethral sphincter ([Bolla et al., 2023](#)).

2.2.4 Perineum

The perineum, situated beneath the pelvic floor, represents the lowest region of the trunk. It is further divided into two distinct areas: the anal region (anal triangle), located posteriorly, and the urogenital region (urogenital triangle), positioned anteriorly ([Bazira, 2021](#)).

The perineum is characterized by its diamond-like shape when the lower limbs are abducted and occupies a position below the LAM. This region contains several significant anatomical structures, including the external genitalia, perineal muscles, and anal canal ([Drake et al., 2009](#); [Moore et al., 2013](#)).

In anatomical terms, the perineal region, delineated by the narrow area between the proximal portions of the thighs, corresponds to the surface of the perineum. Within this region, several muscles are identified, including the ischiocavernosus, bulbospongiosus, deep, and superficial transverse perineal muscles, as can be observed in Figure 2.5 (Drake et al., 2009; Moore et al., 2013).

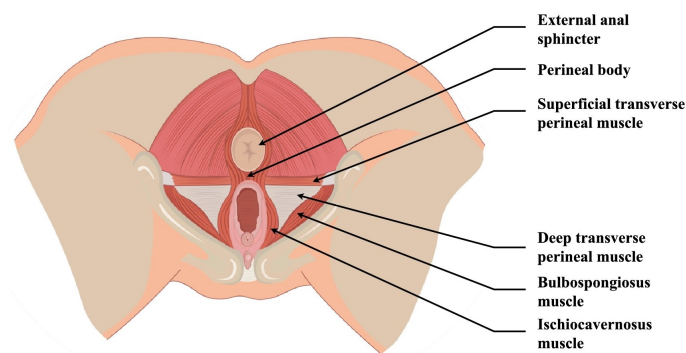


Figure 2.5: Schematic representation of the anatomy of the perineum (adapted from Moura et al. (2024b)).

In the perineal region, a fundamental connective tissue known as the perineal body plays a central role, serving as a point of attachment for both the PFM and adjacent perineal structures. Within this region, the ischiocavernosus muscle securely attaches to the ischial tuberosity, while the bulbospongiosus muscle extends posteriorly to connect with the perineal body and anteriorly to the surface of the bulb and the perineal membrane. Additionally, the superficial transverse perineal muscles align along the posterior edge of the inferior perineal membrane, anchoring to the ischial tuberosities and extending inward toward the perineal body (Drake et al., 2009; Moore et al., 2013; Moura et al., 2024b).

The osseofibrous structures that define the borders of the perineum are the pubic symphysis anteriorly, the ischiopubic rami anterolaterally, the ischial tuberosities laterally, the sacrotuberous ligaments posterolaterally and the inferior most sacrum and coccyx posteriorly (Moore et al., 2013).

Chapter 3

Pelvic Floor Dysfunction

3.1 Introduction

Many women view PFD as a minor health issue that arises naturally after childbirth. The well-being of the newborn was historically given first attention, overshadowing the health of the mother. However, these days, attention is drawn to the role of the mother and the possible risk of physical and psychological challenges following childbirth. Consequently, it is increasingly acknowledged that many postpartum issues stem from PFD ([Schütze et al., 2023](#)).

PFD is a condition that includes various symptoms and anatomical abnormalities associated with improper PFM function ([Grimes and Stratton, 2023](#)). Accordingly, the term PFD encompasses a wide range of clinical situations, such as urinary and fecal incontinence, overactive bladder, POP, and sexual disorders ([Bozkurt et al., 2014](#)). Nonetheless, UI, FI, and POP stand out as the three most prevalent and well-defined conditions encountered in clinical settings.

Moreover, PFD following childbirth imposes significant and enduring consequences on women, demanding surgical intervention for over 300,000 individuals annually. This number constitutes approximately 10% of the 3 million women who undergo vaginal delivery each year ([DeLancey et al., 2024](#)).

While it is evident that pelvic floor damage related to childbirth occurs, it is imperative to determine the prevalence and severity of such trauma to justify modifications in clinical protocols. In fact, there are nulliparous women who experience PFD and some women who have undergone vaginal delivery who do not. Throughout childbirth, the puborectalis-pubococcygeus complex of the PFM can undergo excessive stretching, leading to mechanical and neural impairment. Given that vaginal delivery stands as a primary risk factor for PFD development, further investigation is needed to elucidate the extent of PFD ([Oliveira et al., 2016b](#)).

Over the past decade, there has been a growing interest in this topic. However, findings presented by various authors often contradict one another, leading to controversial conclusions. For instance, the prevalence of stress UI associated with pregnancy and delivery ranges widely from 3% to 67%, while postpartum FI rates vary from 6.8% to 18%. This inconsistency may stem from differences in the populations studied, variations in obstetric management of labor, retrospective

study designs prevalent in most literature, and the frequent utilization of unvalidated tools for evaluating PFD (Torrisi et al., 2012).

Additionally, the economic impact of these conditions comprises direct expenses, such as regular healthcare, medical consultations, and treatments, and indirect costs stemming from decreased productivity. In 2006, the direct yearly expenses linked to outpatient care for PFD in the USA amounted to an estimated US\$ 412 million. Therefore, with the escalating prevalence of these disorders, it is expected that these costs will further escalate in the years to come (Memon and Handa, 2013).

Lastly, the symptoms associated with these conditions can have notable effects on the quality of life of individuals, ranging from slight discomfort and embarrassment to severe disruptions in daily functioning and overall well-being. Thus, PFD is recognized as a substantial public health concern due to its widespread occurrence and adverse impact on individual quality of life and healthcare resources.

3.2 Risk Factors

Various circumstances can compromise the functional attributes of connective tissues, increasing the likelihood of pelvic dysfunction. Thus, risk factors can be genetic/environmental, such as age, obesity, chronic elevation of IAP, diabetes, ethnicity, racial background, collagen irregularities, connective tissue disorders, and neurological conditions, or obstetric, including pregnancy, childbirth, duration of labor stages, episiotomy, epidural analgesia. Other contributors may be high-impact exercise and low socioeconomic status. However, there is a common denominator across the majority of PFD symptoms: a vaginal birthing process (Easley et al., 2017; Torrisi et al., 2012; Romeikienė and Bartkevičienė, 2021; Teixeira et al., 2018; Parente, 2008; Bozkurt et al., 2014; Memon and Handa, 2013).

As the biological origins of PFD are yet unknown, investigating the relationship between childbirth and PFD could provide valuable insights into their pathophysiology, paving the way for the creation of preventive measures (Memon and Handa, 2013).

Vaginal birth not only causes substantial stretching of the LAM vertically but also significantly stretches the pudendal nerves in the same direction. It is worth noting that any stretch of a somatic nerve exceeding 12% has been documented to cause some level of permanent nerve damage. Throughout the vaginal birthing process, the descent of the perineum might cause lifelong damage to the pudendal nerve due to its extreme stretching. Apart from directly affecting the muscles and nerves of the pelvic floor, vaginal birth also causes symptomatic and concealed damage to the anal sphincteric mechanism (Parente, 2008).

Several studies have shown a greater occurrence of PFD in women who have given birth compared to those who have not, at the same age. According to reports, the prevalence of UI was three times higher among primiparous women as compared to their age-matched nulliparous peers. This parity-related effect is most prominent among younger women aged 20 to 34 and diminishes in

those over 65. Furthermore, the likelihood of developing POP also increases linearly with increasing parity. A study conducted in Britain revealed that women with one child were four times more susceptible to POP, while those with two children were 8.4 times more prone compared to women who had not given birth. Similarly, the prevalence of FI escalates with each additional childbirth (Memon and Handa, 2013). Moreover, the prevalence of PFD rises notably with advancing age. Roughly 50% of women aged 80 years old suffer from at least one PFD condition, compared to 10% of women between the ages of 20 and 39 (Bozkurt et al., 2014).

The mode of delivery is a controversial factor in the development of PFD. It is believed that vaginal birth, particularly during the second stage of labor, may harm pelvic support tissues such as muscles, connective tissues, and neurological structures, which may contribute to the development of PFD. Research has shown that the pelvic floor can become partially denervated, especially during the first pregnancy. In most women delivering vaginally, the risk of PFD increases with the severity of the injury. However, despite these investigations, the current literature presents contradicting findings. Therefore, there is insufficient evidence to conclusively establish vaginal delivery as the sole cause of PFD (Bozkurt et al., 2014; Schütze et al., 2023).

Finally, in the USA, the lifetime probability of having a single surgery for prolapse or incontinence by the age of 80 is 11%, with a reoperation rate of 30% (Memon and Handa, 2013). It is predicted that within the next 40 years, the prevalence of symptomatic PFD will rise by 56% due to the projected rapid population growth. Moreover, the number of women who will undergo surgery for UI or POP in the USA is expected to increase by approximately 50% from 2010 through 2050 (Wu et al., 2011).

3.3 Urinary Incontinence

Maintaining continence relies on the support mechanisms of the pelvic floor. UI, as defined by the International Continence Society (ICS), refers to the involuntary loss of urine resulting from bladder pressure exceeding urethral pressure (D'Ancona et al., 2019). UI can be categorized into stress UI (SUI), urgency UI (UUI), or mixed UI (MUI) (Preda and Moreira, 2019). SUI is characterized by involuntary urine leakage from the urethral opening during physical exertion, sneezing, or coughing. UUI entails leakage associated with a sudden, compelling urge to void. In turn, MUI represents a combination of SUI and UUI (D'Ancona et al., 2019).

UI affects over 200 million individuals globally, and its prevalence is higher in women (Preda and Moreira, 2019). In Western nations, the reported rates of UI vary from 15 to 69%. While the prevalence of these disorders escalates with increasing age and parity, they are also prevalent among younger and nulliparous women (Stansfield et al., 2021). Patel et al. (2022) revealed that over 60% of adult women residing in the USA encounter some form of UI, with over 20% experiencing moderate or more severe UI. Additionally, a systematic analysis found that three months after giving birth, the pooled prevalence of any UI was 36% (Thom and Rortveit, 2010). In Portugal, epidemiological data is limited. However, an epidemiological study conducted in the

Portuguese population of adults over 40 years in 2008 revealed a prevalence of UI in women to be 21.4% (Preda and Moreira, 2019).

Female UI is frequently attributed to the repercussions of pregnancy and childbirth. The process of childbirth can lead to pelvic floor laxity due to the stretching and weakening of muscles and connective tissues. Spontaneous lacerations and episiotomies during delivery can also contribute to tissue damage. UI development can happen through various mechanisms, including diminished pelvic floor support and compromised urethral pressure maintenance resulting from ligament tears and nerve stretching, particularly of the pudendal nerve (Grimes and Stratton, 2023).

Although UI is not considered life-threatening, it can significantly impact the quality of life of a woman by greatly disrupting her social, professional, and personal activities. The stigma surrounding urinary leakage remains prevalent and beyond its social implications, often coexisting with conditions like depression and anxiety and significantly affecting interpersonal relationships (Preda and Moreira, 2019).

3.4 Fecal Incontinence

Fecal incontinence is characterized by the involuntary loss of mucous, liquid, or solid stool, posing a social and hygienic concern. Recent reviews of cross-sectional studies across different populations globally suggest a prevalence of FI in the adult population ranging from 11% to 15% (Macmillan et al., 2004). Chronic diarrhea, parity, UI, high body mass index, and past anorectal surgery are all risk factors for FI in women, with obstetric trauma being the most common cause. Additionally, older women are at risk of experiencing coexisting UI and FI (Gerten et al., 2008).

Despite its significant prevalence and traumatic nature, the mechanism by which delivery affects FI is not fully understood. The risk of developing fecal or flatus incontinence increases with laceration of the external anal sphincter during vaginal birth. During parturition, the LAM and vaginal wall stretch, the rectovaginal septum tears, and the pudendal nerves are compressed against the pelvic sidewall. The risk of sphincter injury is higher in individuals with perineal tears, and some patients experienced FI symptoms after a third-degree tear (including the sphincter muscle). Similarly, episiotomies are associated with incontinence and can also increase the risk of sphincter injury (Patel et al., 2006).

As a result, Ryhammer et al. (1995) found that the probabilities of flatus incontinence increased 6.6-fold following the third vaginal delivery compared to the first or second. Additionally, Pollack et al. (2004) discovered that women who had more than one childbirth throughout the follow-up period were considerably more likely to experience FI than those who had only one vaginal delivery. A study conducted thirty years after delivery revealed similar prevalence rates of flatus incontinence and FI among women who experienced anal sphincter disruption during their index delivery, those who underwent vaginal deliveries with episiotomy, and those who had cesarean deliveries (Patel et al., 2006).

3.5 Pelvic Organ Prolapse

According to the International Urogynecological Association (IUGA) and ICS, female POP is characterized by deviations from normal sensation, structure, or function concerning the position of the pelvic organs (Haylen et al., 2010). This condition occurs as a result of the weakening of the vaginal walls, which are usually supported by connective tissues and pelvic musculature, leading to the descent of pelvic organs into the vaginal cavity. POP can be considered a form of herniation, where the bulge may include various pelvic organs, such as the bladder, bowel, or uterus. Consequently, it can cause symptoms of pelvic pressure, vaginal bulge, urinary and bowel dysfunction, and sexual dysfunction, particularly among elderly patients (Smith et al., 2014).

Various types of POP can be distinguished, and women may exhibit a combination of them:

- Uterine/cervical prolapse: It involves the descent of the uterus or uterine cervix from its normal position.
- Vaginal vault prolapse: This type occurs when the vaginal vault, often the site of a hysterectomy scar, descends.
- Anterior vaginal wall prolapse (cystocele): It manifests as a bulge caused by the bladder protruding into the vaginal space. It is commonly known as a "falling out" bladder appearance and is the most frequently encountered form of POP.
- Posterior vaginal wall prolapse: This condition has two subcategories. Rectocele is a bulge caused by a defect of the rectum into the vaginal cavity, while enterocele corresponds to a bulge caused by the small intestines or even sigmoid colon pressing into the vaginal canal, resulting in tearing, stretching, or a combination of the two.
- Apical prolapse: It corresponds to the descent of the vaginal canal apex.
- Procidentia: The most advanced stage of uterine prolapse, where the anterior, posterior, and apical compartments prolapse simultaneously. Patients with procidentia are at risk of prolapsed contents becoming incarcerated, leading to complications such as urinary retention, bowel necrosis, or uterine swelling, which need emergency surgical intervention (Haylen et al., 2010; Smith et al., 2014).

The etiology of POP entails interactions between muscle dysfunction and connective tissues supporting the pelvic organs. Typically, intact LAM uphold the closure of the pelvic hiatus, maintaining equilibrium in pressure across the vaginal walls. However, damage to the LAM may induce a hiatal opening, creating a pressure gradient and protrusion of the vaginal wall. This disturbance in the load distribution between muscle and connective tissue can contribute to prolapse if the tissue resilience is surpassed. Ultimately, injury to PFM may initiate several events culminating in POP (DeLancey et al., 2024).

Every year, over 200,000 women in the USA undergo inpatient surgery for POP. An analysis of women aged 50 to 79 revealed that 41% of women with a uterus exhibited some form of POP at baseline (Hendrix et al., 2002). Similarly, in a cross-sectional study of 487 Swedish women aged 20 to 59, Samuelsson et al. (1999) found that 31% of all women and 44% of parous women had some type of POP. Mant et al. (1997b) found that, compared to nulliparous women, those

with one child were four times more likely, and those with two children were 8.4 times more likely to experience POP requiring hospitalization. Moreover, it is worth noting that less than 1% of prolapse cases occurred in nulliparous women in this study. However, the population-based incidence of POP remains unclear due to a scarcity of prospective or longitudinal studies, particularly those focusing on demographic groups beyond women admitted to hospitals with symptomatic prolapse ([Patel et al., 2006](#)).

Finally, a considerable time gap exists, spanning decades, between childbirth and the onset of POP. Consequently, given that symptoms tend to manifest later in life, it is understandable that the majority of existing studies on POP center around postmenopausal women or those nearing menopause ([Mirskaya et al., 2019](#)).

Chapter 4

Childbirth

4.1 Introduction

Labor represents the physiological process of a fetus being released from the uterus into the outer world ([Hacker et al., 2009](#)). Traditionally, labor and childbirth occurred at home, with the mother receiving assistance from another woman and being supported by her family. However, during the 20th century, particularly after World War II, childbirth was institutionalized and shifted from the home to the hospital, motivated by the imperative to mitigate the prevailing rates of maternal and newborn mortality, and, as a result, it became medicalized ([Brüggemann et al., 2005](#)).

With that in mind, normal labor is characterized by a sequence of significant physiological adjustments in the mother's body, facilitating the passage of the fetus through the birth canal. Labor corresponds to the process of the cervix gradually thinning and dilating due to regular contractions of the uterus, lasting 30 to 60 seconds and happening at least once every three minutes ([Hacker et al., 2009](#)).

Consequently, understanding the dynamic interaction between the fetal head and maternal pelvis during labor requires knowledge of their respective properties. The fetal head undergoes a series of adaptations involving molding, flexion, and subsequent internal rotation as it traverses the maternal pelvis during the stages of labor ([Hacker et al., 2009](#)).

As previously stated, giving birth vaginally stands as a notable risk factor for the development of PFD. Despite potential latency in manifestation, labor-related injuries can occur during the early phase of the second stage of labor. Throughout childbirth, the puborectalis-pubococcygeus complex may undergo excessive stretching, resulting in mechanical and nervous damage ([Oliveira et al., 2016b](#)).

4.2 Anatomical Structure of the Fetus

Several fetal characteristics affect the course of labor and delivery ([Gabbe et al., 2017](#)):

- The **fetal size** estimation can be determined clinically using abdominal palpation or ultrasound, but these procedures are vulnerable to a substantial degree of error.

- The term **fetal lie** refers to the orientation of the longitudinal axis of the fetus in relation to the uterine longitudinal axis. Fetal lying can be longitudinal, transverse, or oblique, as illustrated in Figure 4.1. In a singleton pregnancy, only fetuses positioned in a longitudinal lie are suitable for vaginal delivery.
- **Presentation** denotes the specific fetal part positioned directly over the pelvic inlet. The presentation may be cephalic (vertex) or breech in a fetus with a longitudinal lie. Different presentations exist, determined by the parts overlaying the pelvic inlet.
- **Attitude** describes the posture of the fetal head with reference to the fetal spine (the degree of flexion and/or extension of the fetal head). Head flexion plays a crucial role in facilitating fetal head engagement within the maternal pelvis.
- **Fetal position** corresponds to the alignment of the fetal presenting part in relation to the maternal pelvis, and it can be measured most precisely on vaginal examination. Various positions of a cephalic presentation are delineated in Figure 4.2.
- **Station** serves as a metric for assessing the descent of the bony fetal presenting part through the birth canal. The conventional classification system, ranging from -5 to +5, is based on a quantitative measurement in centimeters of the distance between the leading bony edge and the ischial spines. The midpoint (0 station) is delineated as the plane of the maternal ischial spines.

Anomalies in any of these fetal variables can influence the progression of labor and the mode of delivery. For instance, occipito-posterior (OP) presentation is widely recognized for its association with prolonged labor, operative vaginal delivery, and an elevated likelihood of cesarean section (Gabbe et al., 2017).

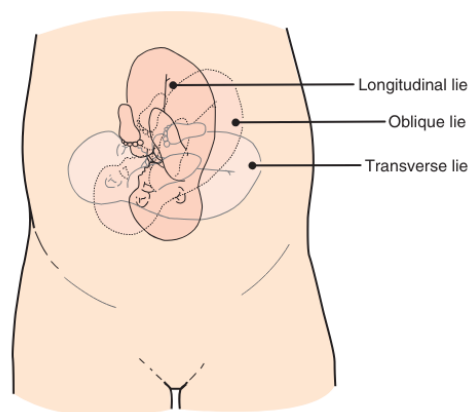


Figure 4.1: Examples of fetal lie (adapted from Gabbe et al. (2017)).

4.2.1 Fetal Head Diameters

The head represents the largest and least compressible body component of the fetus, being greatly important from an obstetric perspective. Comprising a base and a vault, known as the cranium, the fetal skull detains cranial bones that are thin, weakly ossified, and interconnected

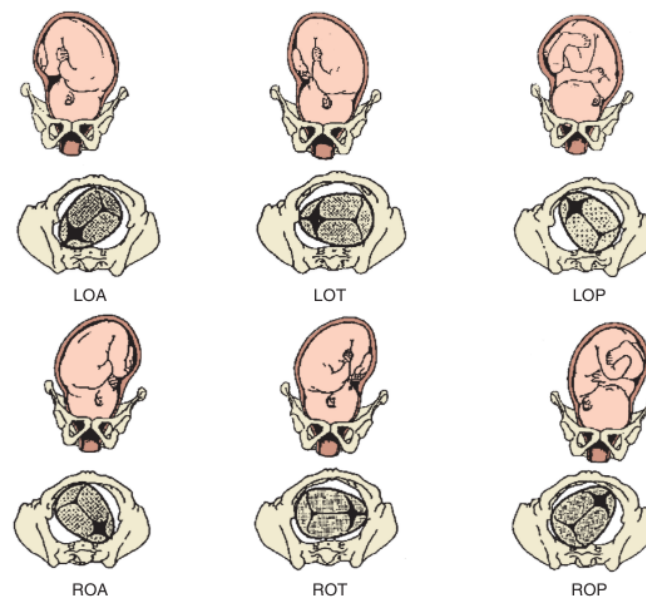


Figure 4.2: Positions and presentations of the fetus during labour: left occipito-anterior (LOA); left occipito-posterior (LOP); left occipito-transverse (LOT); right occipito-anterior (ROA); right occipito-transverse (ROT); right occipito-posterior (ROP) (adapted from [Gabbe et al. \(2017\)](#)).

solely by membranes upon birth. This enables the molding process, where cranial bones can overlap when compressed and alter shape to fit the maternal pelvis, a process called molding ([Hacker et al., 2009](#)).

As depicted in Figure 4.3, different fetal skull diameters are significant. The anteroposterior diameter facing the maternal pelvis is contingent upon the degree of flexion or extension of the head, a critical factor due to the varying lengths of these diameters. Below are the average measurements for the anteroposterior diameters, regarded as typical for a full-term fetus ([Hacker et al., 2009](#)):

- Suboccipitobregmatic (9.5 cm): This dimension represents the presenting anteroposterior diameter when the head is properly flexed, as in the occipito-transverse/occipito-anterior (OA) positions. It is defined by a line extending from the undersurface of the occipital bone at the neck junction to the center of the anterior fontanelle.
- Occipitofrontal (11 cm): This diameter denotes the presenting anteroposterior diameter when the head is deflexed, such as in an OP presentation. It spans from the external occipital protuberance to the glabella.
- Supraoccipitomenta (13.5 cm): This measurement represents the presenting anteroposterior diameter observed in a brow presentation and stands as the longest anteroposterior diameter of the head, extending from the vertex to the chin.
- Submentobregmatic (9.5 cm): This dimension represents the presenting anteroposterior diameter observed in face presentations. It extends from the junction of the neck and lower jaw to the center of the anterior fontanelle.

Moreover, the fetal skull has two transverse diameters. The biparietal diameter (9.5 cm) represents the largest transverse diameter, connecting both parietal bones. In turn, the bitemporal diameter (8 cm) corresponds to the shortest transverse diameter, extending between the temporal bones. The average circumference of the fetal head at full term, measured in the occipitofrontal plane, is 34.5 cm (Hacker et al., 2009).

Lastly, it is noteworthy that when the fetal chin is ideally flexed onto the chest, the suboccipitobregmatic diameter becomes apparent at the pelvic inlet and represents the smallest feasible presenting diameter in cephalic presentation. While the head is deflexing, the diameter presented to the pelvic inlet gradually enlarges even before encountering malpresentations such as brow and face, potentially contributing to a lack of progress in labor (Gabbe et al., 2017).

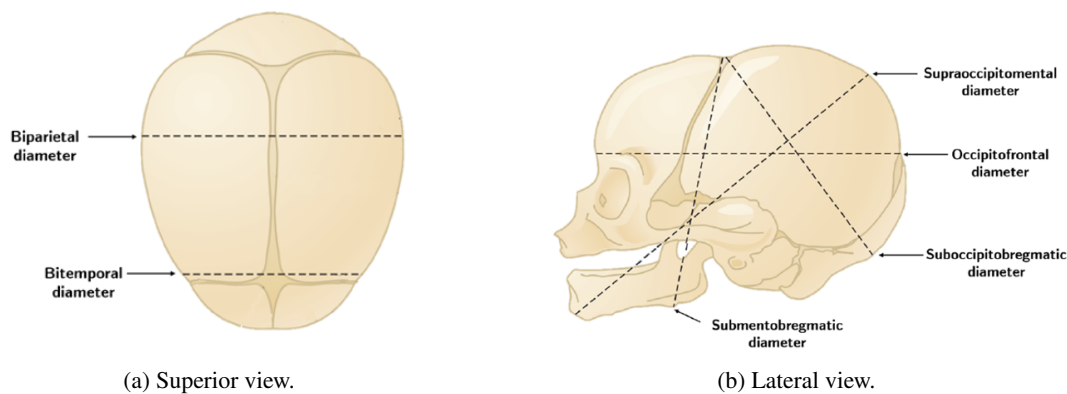


Figure 4.3: Diameters of the fetal skull (adapted from Hacker et al. (2009)).

4.3 Maternal Pelvis

The child's passage consists of the bony pelvis and the opposition offered by the surrounding soft tissues. Thus, precise measurements of the many characteristics of the female pelvis have been conducted both directly on cadavers and by radiographic imaging on living women. These methods led to the separation of the actual pelvis into multiple planes, which the fetus must navigate while passing down the birth canal (Gabbe et al., 2017).

4.3.1 Maternal Pelvic Diameters

With descriptive clarity in mind, the pelvis is categorized into four planes: the pelvic inlet, the plane of greatest diameter, the plane of least diameter, and the pelvic outlet (Figure 4.4). Each of these planes corresponds to the available space at various levels within the pelvis. They are imaginary, flat surfaces spanning the pelvis at different heights. Every plane but the plane of maximum diameter holds clinical significance (Hacker et al., 2009). Therefore, in the obstetrics field, it is crucial to analyze the various diameters of the female pelvis to assess its capacity for childbirth (Gabbe et al., 2017).

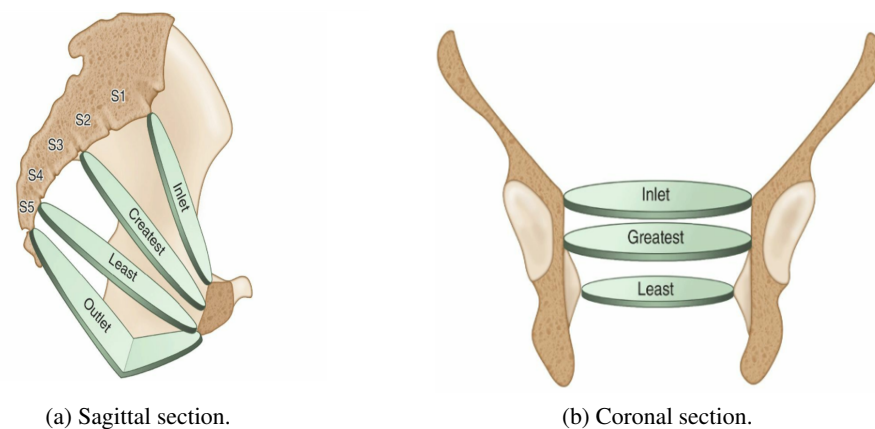


Figure 4.4: Pelvic planes (adapted from [Posner et al. \(2013\)](#)).

4.3.1.1 Pelvic Inlet

Concerning the pelvic inlet, there are five important diameters to be considered, which are clearly seen in Figure 4.5. Two measurements are used to describe the anteroposterior diameter. The true conjugate, often referred to as the true or anatomic diameter, spans from the midpoint of the sacral promontory to the superior aspect of the pubic symphysis. The obstetric conjugate measures the space available to the fetus, starting from the middle of the sacral promontory and extending to the nearest point on the curved posterior surface of the symphysis pubis. In turn, the transverse diameter refers to the greatest distance between the iliopectineal lines, being the largest measurement. The distance from the sacroiliac joint to the iliopectineal eminence on the opposite side is recognized as the oblique diameter. Lastly, the posterior sagittal diameter spans from the point where the anteroposterior and transverse diameters connect to the midpoint of the sacral promontory ([Hacker et al., 2009](#)).

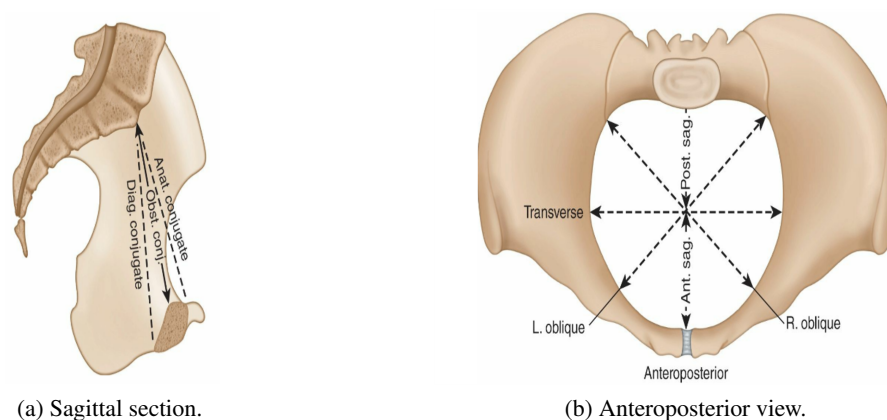


Figure 4.5: Pelvic inlet diameters (adapted from [Posner et al. \(2013\)](#)).

4.3.1.2 Plane of Greatest Diameter

This plane consists of two significant diameters. The anteroposterior diameter extends from the midpoint of the posterior surface of the pubis to the place where the second and third sacral vertebrae meet. The transverse diameter is the widest distance between the lateral edges of the plane ([Hacker et al., 2009](#)).

4.3.1.3 Plane of Least Diameter (Midplane)

This plane is the most significant of the pelvis, and it features three important diameters (Figure 4.6). The anteroposterior diameter goes from the lower border of the pubis to the junction of the fourth and fifth sacral vertebrae. The transverse (bispinous) diameter spans between the ischial spines, constituting the limiting element of the midpelvis. The posterior sagittal diameter stretches from the midpoint of the bispinous diameter to the junction of the fourth and fifth sacral vertebrae ([Hacker et al., 2009](#); [Gabbe et al., 2017](#)).

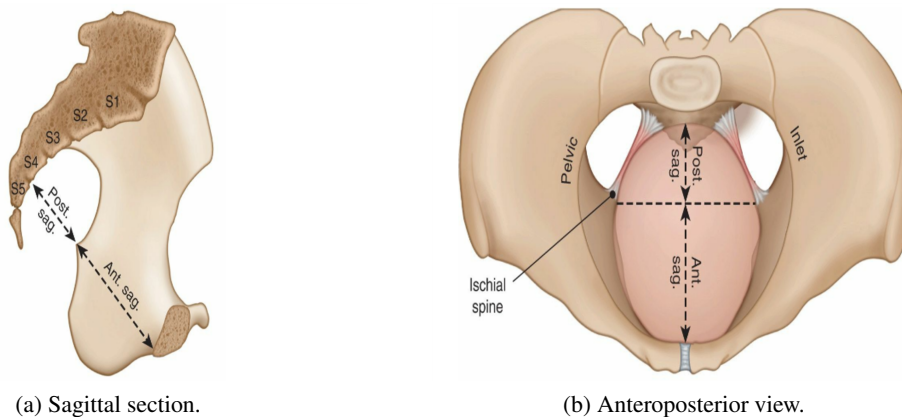


Figure 4.6: Plane of least dimensions (adapted from [Posner et al. \(2013\)](#)).

4.3.1.4 Pelvic Outlet

Despite not being clinically significant, the pelvic outlet has four key diameters, as expressed in Figure 4.7. The anatomic anteroposterior diameter is measured from the inferior edge of the pubis to the tip of the coccyx. In contrast, the obstetric anteroposterior diameter extends from the inferior margin of the pubis to the sacrococcygeal junction. The distance between the inner surfaces of the ischial tuberosities corresponds to the transverse or bituberous diameter, and the posterior sagittal diameter runs from the middle of the transverse diameter to the sacrococcygeal joint ([Hacker et al., 2009](#); [Gabbe et al., 2017](#)).

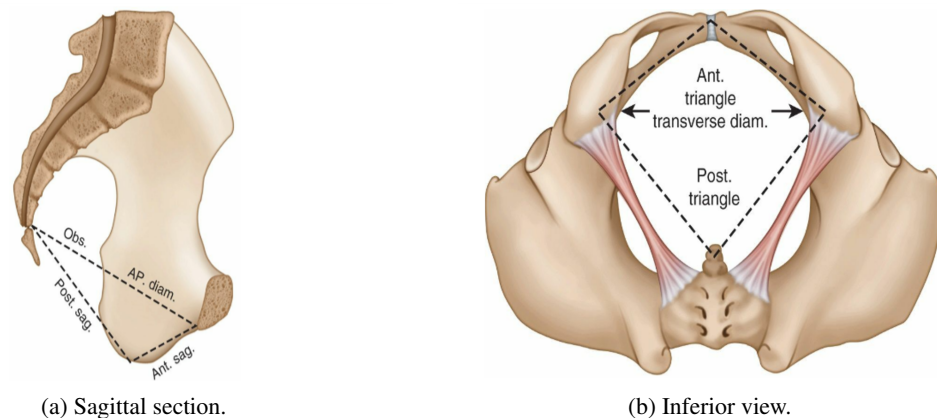


Figure 4.7: Pelvic outlet diameters(adapted from [Posner et al. \(2013\)](#)).

4.3.2 Classification of the Pelvis

The anatomical structure of women's birth canals exhibits significant morphological variability. As detailed in Section 2.2.1.1, the pelvic shape can be categorized into four fundamental categories based on the overall bony architecture. This classification identifies the following types: gynecoid, android, anthropoid, and platypelloid ([Caldwell and Moloy, 1938](#)).

The gynecoid pelvic shape is observed in approximately 50% of women, whereas the android shape is present in less than 30%. The anthropoid shape is found in about 20% of women, while the platypelloid shape occurs in only 3% of women ([Hacker et al., 2009](#)).

Therefore, in light of current knowledge, this classification distinguishes between the female pelvis with favorable characteristics for vaginal delivery (gynecoid, anthropoid) and those that are less favorable (android, platypelloid) in order to assess whether a given fetus will be able to safely navigate through a specific pelvis ([Hacker et al., 2009](#)).

4.4 Normal Delivery and Birth

In 2022, approximately 67% of birth deliveries in the USA were vaginal ([Osterman et al., 2024](#)).

Vaginal childbirth, often termed natural or normal childbirth, involves a sequence of events characterized by painful, regular uterine contractions that gradually cause the cervix to efface and dilate, allowing the fetus to descend through the pelvis. This process culminates in the spontaneous vaginal birth of the fetus, followed by the expulsion of the placenta and membranes. This sequential pattern is typically divided into three stages ([Hacker et al., 2009](#); [Posner et al., 2013](#)). This division can be seen in Figure 4.8.

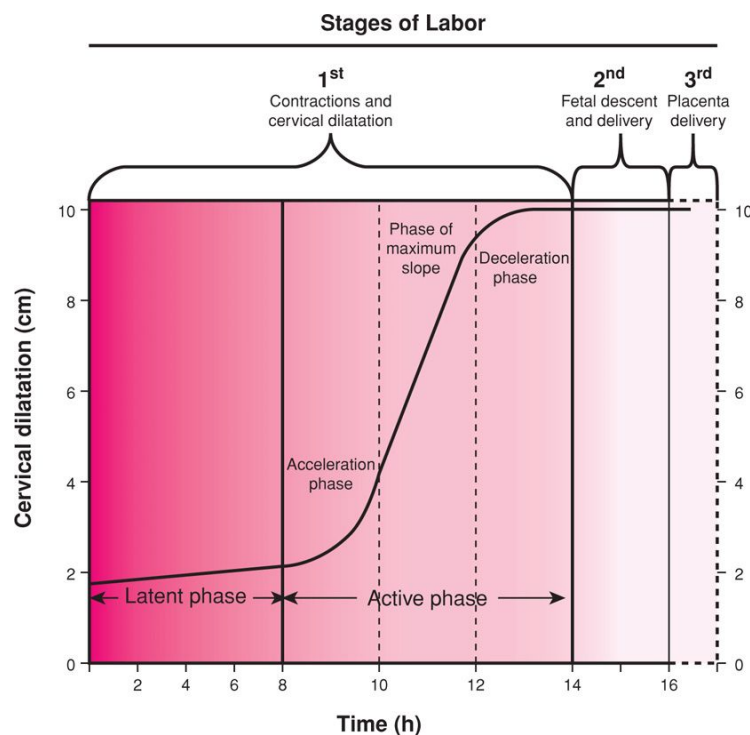


Figure 4.8: Average dilation curve for labor in nulliparous women (adapted from [Bloom et al. \(2014\)](#)).

4.4.1 First Stage of Labor

Stage 1 usually begins with contractions and continues until the cervix is dilated to 10 centimeters. The first stage of labor consists of two phases: the latent period, characterized by cervical effacement and early dilation, and the active phase, marked by more rapid cervical dilation. While cervical softening and initial effacement may begin before labor, the entire cervical length is retracted into the lower uterine segment during the first stage. The duration of this stage varies with parity; primiparous women typically experience a longer first stage than multiparous women. During the active phase, the minimum dilation rate is similar for primiparous and multiparous women, at approximately 1.0 and 1.2 cm/hour, respectively ([Hacker et al., 2009](#)).

4.4.2 Second Stage of Labor

At full dilation of the cervix, stage 2 starts. The “pushing” stage, as it is known, lasts until the baby is born.

The mother often feels a need to bear down with every contraction during the start of the second stage. Together, the uterine contractile force and this abdominal pressure force the fetus out. In order to assess the course of labor, fetal descent must be closely observed during this stage. Therefore, the progress of the presenting component through the birth canal is the unit of measurement for the descent. The evaluation of descent in cephalic presentations is complicated by the possibility that the fetal head shape changes throughout labor. Although the timings may

vary depending on numerous factors, primigravid women generally require 30 minutes to 3 hours for the second stage, and multigravida women take 5 to 30 minutes (Hacker et al., 2009).

Furthermore, in nulliparous women, the diagnosis of a prolonged second stage should be considered if the duration exceeds 3 hours with regional anesthesia or 2 hours without regional anesthesia (Rouse et al., 2009). Conversely, the second stage of labor is deemed precipitous if it lasts for less than 30 minutes (Quiñones et al., 2018).

4.4.3 Third Stage of Labor

The third stage of labor begins with the delivery of the child and concludes with the expulsion of the placenta. This stage is typically brief. Factors such as preterm delivery, preeclampsia, labor augmentation, nulliparity, maternal age over 35 years, and a second-stage labor length exceeding two hours have all been linked to a prolonged third stage (Gabbe et al., 2017).

4.5 Cardinal Movements in Labor

The positional changes of the fetal head as it passes the birth canal are called cardinal movements. Due to the asymmetry in the shapes of the fetal head and the maternal bony pelvis, these rotations are necessary for the fetus to navigate the birth canal successfully. Seven distinct cardinal movements can be identified, and they are present in Figure 4.9 (Gabbe et al., 2017):

1. **Engagement:** The passage of the widest diameter of the presenting part to a level below the plane of the pelvic inlet is known as engagement. Clinically speaking, palpation of the presenting area both vaginally and abdominally can verify engagement. Engagement in a cephalic presentation is reached when the presenting part is at zero station during a vaginal examination. This is a crucial clinical prognostic indication, as it shows that the maternal bony pelvis is sufficiently large to allow the descent of the fetal head, at least at the level of the pelvic inlet. The fetal head engages in nulliparous women by 36 weeks gestation on average; however, in multiparas, engagement can happen later in gestation or even during childbirth.
2. **Descent:** It corresponds to the downward movement of the presenting part through the pelvis. The process is not continuous, with the greatest rates of descent occurring during the late active phase and the second stage of labor.
3. **Flexion:** The fetal head flexes passively as it descends, influenced by the shape of the bony pelvis and the resistance from the soft tissues of the pelvic floor. Only during labor is complete flexion typically possible to present the suboccipitobregmatic diameter, the smallest diameter for optimal passage through the pelvis.
4. **Internal Rotation:** It refers to the rotation of the presenting part from its initial position at the pelvic inlet to an anteroposterior position as it progresses through the pelvis. It is a passive movement brought about by the PFM and pelvis shape. The widest part of the fetus

can pass the pelvis at its widest dimension because the occiput of the fetus rotates toward the symphysis pubis or, less often, into the sacral hollow as the head descends.

5. **Extension:** Extension happens when the fetus has reached the level of the introitus. At this point, the base of the occiput meets the inferior margin of the symphysis pubis. The birth canal curves upward, and the fetal head is delivered by expanding and rotating around the symphysis pubis - this motion is caused by the downward force of uterine contractions and the upward force of the PFM.
6. **External Rotation:** It refers to the return of the fetal head to its correct anatomical position in relation to the fetal torso. This passive movement occurs as the forces exerted on the fetal head during labor are released. Depending on the fetus's orientation, the head typically rotates to either side.
7. **Expulsion:** Delivering the remainder of the fetus is established as expulsion. Delivered similarly to the head, the anterior shoulder rotates under the symphysis pubis before the posterior shoulder passes over the perineal body. Usually, the rest of the body is delivered without any problems after the shoulder.

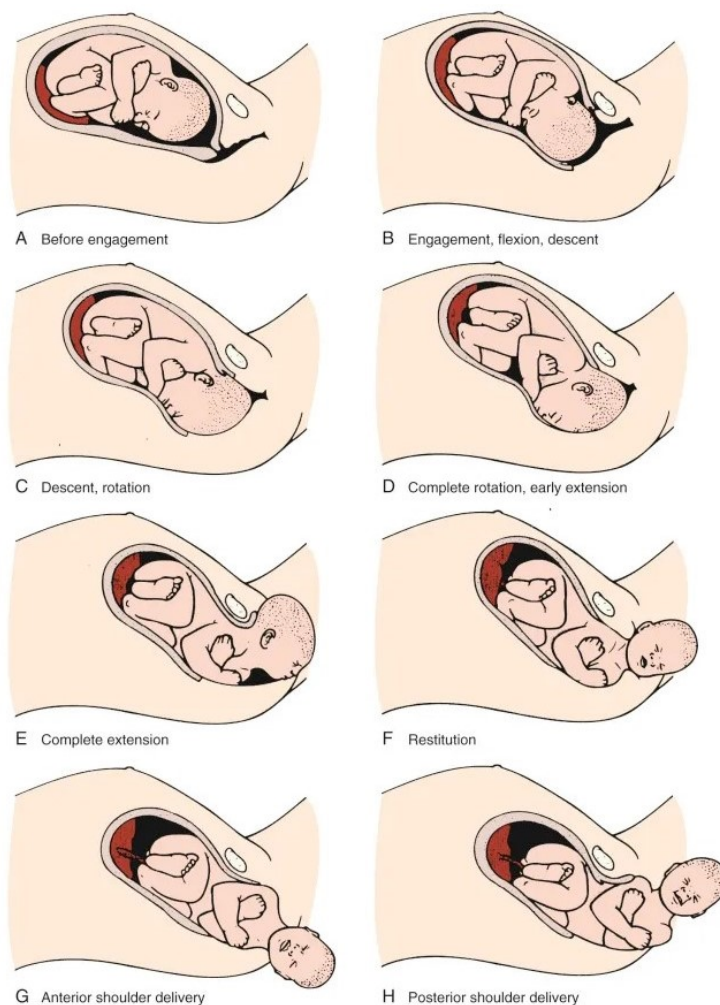


Figure 4.9: Cardinal movements of labor (adapted from [Gabbe et al. \(2017\)](#)).

Chapter 5

Computational Solid Mechanics

5.1 Introduction

Utilizing the most recent advances in computational power, computational bioengineering addresses several biomedical issues. Unbound by ethical constraints that may limit experimental research, it serves as a potent and valuable tool, contributing significantly to innovations and developments in the medical field.

Consequently, computational models are essential resources that allow a quantitative investigation of the physiology of labor. Therefore, childbirth simulations can transcend the limits of traditional physical models or invasive examinations, offering a better perception of the mechanisms of vaginal delivery and the related maternal and fetal injuries ([Moura et al., 2024b](#)).

Keeping that in mind, particularly from a mechanical standpoint, it is essential to consider the concepts of continuum mechanics, as they offer mathematical frameworks for modeling the behavior of matter. This approach has been extensively utilized in the investigation of biological tissues.

Moreover, it is crucial to consider the FEM for analyzing and determining the effects of mechanical stresses, strains, and interactions among different components within complex geometries. FEM simplifies these geometries by breaking them down into a finite number of elements with simpler shapes. As a result, this method is frequently employed to model and simulate biological components.

5.2 Preliminary Concepts

5.2.1 Configuration and Motion of Continuum Bodies

A continuum approach is commonly employed to describe macroscopic systems, resulting in the continuum theory. A fundamental assumption of this theory is that a body, denoted as B , can be understood as having a continuous distribution of matter in space and time, and it is believed to have a composition of a (continuous) set of particles, represented by $P \in B$. Therefore, in

the macroscopic study of a body, mass and volume are treated as continuous functions of these continuum particles, defining what is known as a continuum body (Holzapfel, 2002).

A continuum body B with a particle $P \in B$ immersed in three-dimensional Euclidean space at a specific time t is illustrated in Figure 5.1. Thus, the configuration $\Phi(B)$ of the body B is the disposition of the continuum body B , which transitions through space over time (Holzapfel, 2002).

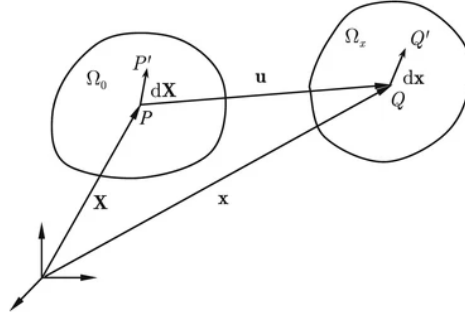


Figure 5.1: General motion of a deformable body (adapted from Kim (2015a)).

A mapping function Φ between the starting and current particle locations can thus be used to characterize the motion mathematically. Stated otherwise, the vector field $\Phi(B)$ defining the new position $\mathbf{x}$ of a particle $\mathbf{X}$ for a fixed t is related to the motion of the body B :

$$\mathbf{x} = \Phi(\mathbf{X}) \quad (5.1)$$

When considering a particle $\mathbf{X}$ occupying the location $(\mathbf{x}, t)$, its position, associated with the place $\mathbf{x}$ at time t , is given by:

$$\mathbf{X} = \Phi^{-1}(\mathbf{x}) \quad (5.2)$$

5.2.2 Material and Spatial Descriptions

In finite deformation analysis, it is crucial to differentiate between the coordinate systems used to describe the behavior of a body. Key quantities, such as density, can be expressed in terms of the pre- or post-deformation positions of the body. The former approach is known as the material, or Lagrangian, description, while the latter is referred to as the spatial, or Eulerian, description. A material description focuses on the behavior of specific material particles, whereas a spatial description concerns the behavior at specific spatial locations. Regardless of the chosen description, the governing equations must ultimately address the current state of the body, necessitating a primary formulation in spatial terms (Bonet and Wood, 2008).

5.2.3 Deformation Gradient

The motion of a body typically modifies its orientation, position, and shape. Thus, a continuum body is deformable if it can alter its shape. Referring to Equation 5.1, for a given point P in the

undeformed configuration, there is a unique corresponding point Q in the deformed configuration. This mapping relationship can be expressed as (Kim, 2015b):

$$\mathbf{x} = \Phi(\mathbf{X}, t) = \mathbf{X} + \mathbf{u}(\mathbf{X}, t) \quad (5.3)$$

where the displacement of point P is denoted by $\mathbf{u}(\mathbf{X}, t)$. In the two geometries shown in figure 5.1, nearby points P' and Q' at infinitesimal distances from P and Q are indicated by vectors $d\mathbf{X}$ and $d\mathbf{x}$, respectively. It is worth noting that the vector $d\mathbf{X}$ deforms to $d\mathbf{x}$. Therefore, the connection between differential elements $d\mathbf{X}$ and $d\mathbf{x}$ can be stated as follows, assuming continuous mapping:

$$d\mathbf{x} = \frac{\partial \mathbf{x}}{\partial \mathbf{X}} d\mathbf{X} \Rightarrow d\mathbf{x} = \mathbf{F} d\mathbf{X} \quad (5.4)$$

where $\mathbf{F}$ is known as the deformation gradient and is explicitly written as follows:

$$F_{ij} = \frac{\partial x_i}{\partial X_j} \quad (5.5)$$

By utilizing the relationship described in equation 5.3, it is possible to express the deformation gradient as:

$$\mathbf{F} = \mathbf{1} + \frac{\partial \mathbf{u}}{\partial \mathbf{X}} = \mathbf{1} + \nabla_0 \mathbf{u} \quad (5.6)$$

where the term $\frac{\partial \mathbf{u}}{\partial \mathbf{X}}$ corresponds to the displacement gradient. As a notational convenience, $\nabla_0 = \frac{\partial}{\partial \mathbf{X}}$ denotes the gradient operator at the undeformed geometry, such that $(\nabla_0 \mathbf{u})_{ij} = \frac{\partial u_i}{\partial X_j}$. If $\mathbf{F} = \mathbf{1}$, then $d\mathbf{x} = d\mathbf{X}$, indicating that no deformation has occurred. Even though an infinitesimal volume in the undeformed geometry can change its size, it cannot shrink to a point (a zero volume). Mathematically, this implies that the determinant of the deformation gradient has to be positive:

$$\det(\mathbf{F}) \equiv J > 0 \quad (5.7)$$

where J corresponds to the Jacobian. This property is crucial for ensuring a valid mapping of $\Phi(\mathbf{X}, t)$ during large deformations (Kim, 2015b).

5.3 Constitutive Equations

When a material is deformed, the resulting stresses must be expressed in terms of some measure of this deformation, such as the strain. Depending on the specific material under study, these relationships - known as constitutive equations - may be dependent or independent of time. Constitutive equations are required to adhere to specific physical rules in general. For instance, the equations must clearly be objective, meaning they are not affected by changes in the frame of reference (Bonet and Wood, 2008).

5.3.1 Hyperelastic Materials

Elastic materials are those for which the constitutive behavior is merely a function of the state of deformation at the moment. Under such circumstances, any stress measurement at a particle depends on the current deformation gradient $\mathbf{F}$ associated with that particle. Specifically, if the work done by the stresses during a deformation process depends only on the initial state at time t_0 and the final configuration at time t , the material behavior is described as path-independent and is termed hyperelastic (Bonet and Wood, 2008).

Therefore, in a hyperelastic material, the strain energy density is a function of the strain, and the stress can be calculated by differentiating this strain energy density with regard to the strain. This model is unaffected by the deformation history, meaning the material exhibits the same deformation under identical final loads. Examples of hyperelastic materials include rubber-like substances and human tissues (Kim, 2015b).

It is assumed that there is a per unit undeformed volume defined Helmholtz free-energy function Ψ in a hyperelastic material. The Helmholtz free-energy function is also known as the strain-energy function or stored-energy function in the specific situation when $\Psi = \Psi(\mathbf{F})$ is a function of $\mathbf{F}$ or some strain tensor. The first Piola-Kirchhoff stress tensor can then be derived as follows (Holzapfel, 2002):

$$\mathbf{P} = \frac{\partial \Psi(\mathbf{F})}{\partial \mathbf{F}} \quad (5.8)$$

Considering the symmetric Cauchy stress tensor, $\boldsymbol{\sigma} = J^{-1} \mathbf{P} \mathbf{F}^T = \boldsymbol{\sigma}^T$:

$$\boldsymbol{\sigma} = J^{-1} \mathbf{F} \left(\frac{\partial \Psi(\mathbf{F})}{\partial \mathbf{F}} \right)^T \quad (5.9)$$

It is assumed that physical experiments can provide Ψ , which characterizes a certain material.

Several comparable representations of the strain-energy function exist. It is recognized that under rigid body rotations, Ψ must be invariant, that is, independent of the rotational component of $\mathbf{F} = \mathbf{R}\mathbf{U}$. Hence, a hyperelastic material relies on the stretching component of $\mathbf{F}$, namely the symmetric right stretch tensor $\mathbf{U}$.

As $\mathbf{C} = \mathbf{U}^2$ and $\mathbf{E} = (\mathbf{U}^2 - \mathbf{I})/2$ provide the right Cauchy-Green tensor and the Green-Lagrange strain tensor, respectively, Ψ can be written as a function of the six components of the symmetric material tensors $\mathbf{C}$, $\mathbf{E}$:

$$\Psi(\mathbf{F}) = \Psi(\mathbf{C}) = \Psi(\mathbf{E}) \quad (5.10)$$

then, it is feasible to simplify the constitutive equations for hyperelastic materials under finite strains. Consequently, it can be inferred that:

$$\left(\frac{\partial \Psi(\mathbf{F})}{\partial \mathbf{F}} \right)^T = 2 \frac{\partial \Psi(\mathbf{C})}{\partial \mathbf{C}} \mathbf{F}^T \quad (5.11)$$

this leads to a crucial simplified form of the constitutive equation for hyperelastic materials,

namely:

$$\boldsymbol{\sigma} = J^{-1} \mathbf{F} \left(\frac{\partial \Psi(\mathbf{F})}{\partial \mathbf{F}} \right)^T = 2J^{-1} \mathbf{F} \frac{\partial \Psi(\mathbf{C})}{\partial \mathbf{C}} \mathbf{F}^T \quad (5.12)$$

Alternative expressions can be derived for the Piola-Kirchhoff stress tensors $\mathbf{P}$ (non-symmetric) and $\mathbf{S}$ (symmetric):

$$\mathbf{P} = 2\mathbf{F} \frac{\partial \Psi(\mathbf{C})}{\partial \mathbf{C}} \quad (5.13)$$

$$\mathbf{S} = 2 \frac{\partial \Psi(\mathbf{C})}{\partial \mathbf{C}} = \frac{\partial \Psi(\mathbf{E})}{\partial \mathbf{E}} \quad (5.14)$$

5.3.1.1 Isotropic Hyperelastic Materials

Isotropy is predicated on the concept that the response of a material in a stress-strain experiment remains consistent regardless of direction (Holzapfel, 2002). This uniformity in properties across all directions means that the constitutive relation for isotropic materials is independent of the coordinate system. Consequently, invariants of strain are used to define the strain energy density function. When the undeformed state is chosen as the reference frame, the three invariants of the right Cauchy-Green deformation tensor $\mathbf{C}$ are defined as follows (Kim, 2015b):

$$I_1 = \text{tr}(\mathbf{C}) \quad (5.15)$$

$$I_2 = \frac{1}{2} (\text{tr}(\mathbf{C})^2 - \text{tr}(\mathbf{C}^2)) \quad (5.16)$$

$$I_3 = \det(\mathbf{C}) \quad (5.17)$$

The invariants remain consistent across different coordinate systems and are the same for both the right and left Cauchy-Green deformation tensors.

An example of a material exhibiting (approximately) isotropic behavior with versatile applications is rubber (Holzapfel, 2002).

5.3.1.2 Incompressible Hyperelastic Materials

Incompressible materials can tolerate finite strains without a volume change. The incompressibility constraint characterizes them, and they maintain the volume constant during a motion (Holzapfel, 2002; Bonet and Wood, 2008):

$$J = 1 \quad (5.18)$$

With this condition, the strain energy function can be expressed for incompressible hyperelastic materials as:

$$\Psi = \Psi(F) - p(J - 1) \quad (5.19)$$

The scalar p acts as an indeterminate Lagrange multiplier, which can be identified as hydrostatic pressure. This scalar p can only be determined from the equilibrium equations and the boundary conditions.

Distortion and volumetric components of the deformation of a material can be separated. It has already been noted that the three invariants can be used to describe the dependency of Ψ on the Cauchy-Green tensor $\mathbf{C}$ in the situation of isotropy. As square root of I_3 indicates the material's volume change, an incompressible material has $I_3 = 1$. Thus, the only independent deformation variables are the two major invariants, I_1 and I_2 (Holzapfel, 2002; Bonet and Wood, 2008).

5.3.1.3 Transversely Isotropic Hyperelastic Materials

Many materials consist of a matrix material combined with one or more types of fibers and, therefore, are known as composite materials or fiber-reinforced composites. These materials are heterogeneous due to their varying compositions throughout their structure. Notably, they qualify as composite materials only when the fibers are continuously embedded within the matrix. These composites are anisotropic, meaning they possess significant directional characteristics.

A material with a single fiber family as reinforcement has a single preferred direction. Usually, this kind of composite material has a far higher stiffness along the fiber direction than in the orthogonal directions to the fibers. It is the most basic example of an anisotropic material, considered transversely isotropic in this preferred direction. Along directions orthogonal to this preferred direction, the material response is isotropic.

Therefore, the first three $\mathbf{C}$ invariants, I_1 , I_2 , and I_3 , can be used to represent an isotropic hyperelastic material. In the anisotropic analysis, two pseudo-invariants, I_4 and I_5 , appear and contribute to the free energy.

$$I_4(\mathbf{C}, \mathbf{a}_0) = \mathbf{a}_0 \cdot \mathbf{C} \mathbf{a}_0 = \lambda^2 \quad (5.20)$$

$$I_5(\mathbf{C}, \mathbf{a}_0) = \mathbf{a}_0 \cdot \mathbf{C}^2 \mathbf{a}_0 \quad (5.21)$$

where λ indicates fiber stretch and $\mathbf{a}_0$ relates to fiber direction. They define the properties of the fiber family and how they interact with other materials (Holzapfel, 2002).

Finally, biological tissues are heterogeneous composite materials composed of cells and extracellular matrix components in bioengineering. For example, muscle tissue comprises collagen fibers coated by an extracellular matrix that includes collagens, laminins, fibronectin, and proteoglycans (Grzelkowska-Kowalczyk, 2016).

5.3.1.4 Hyperelastic Constitutive Models for Quasi-Incompressible Materials

The PFM of the mother significantly deform during the second stage of labor. Various constitutive models can be used to describe the behavior of materials experiencing such extensive deformations accurately. This section discusses the relevant models for the present study.

- **Neo-Hookean**

The Neo-Hookean constitutive model is characterized by the mathematical expression:

$$\Psi(\mathbf{C}) = c(I_1 - 3) + \Psi_{\text{vol}}(J) \quad (5.22)$$

where the material constant $c = \frac{\mu}{2}$, μ is a Lamé parameter, and $\Psi_{\text{vol}}(J)$ is the volumetric contribution.

This strain energy function offers a mathematically simple and reliable constitutive model for capturing the nonlinear deformation behavior of isotropic rubber-like materials (Holzapfel, 2002). Thus, it arose from molecular chain statistics considerations, in which vulcanized rubber is viewed as a three-dimensional network of long-chain molecules linked at a few places (Kim, 2015b).

- **Holzapfel-Gasser-Ogden**

The Holzapfel-Gasser-Ogden (HGO) model was originally developed to describe the behavior of arteries, explicitly accounting for the distribution of collagen fiber orientations (Holzapfel et al., 2000; Holzapfel and Gasser, 2001).

The isochoric strain energy function $\bar{\Psi}$ is divided into two components: $\bar{\Psi}_{\text{iso}}$, which is associated with isotropic deformations, and $\bar{\Psi}_{\text{aniso}}$, which is related to anisotropic deformations. Hence, the strain energy function is expressed as:

$$\bar{\Psi}(\bar{\mathbf{C}}, \mathbf{a}_{01}, \mathbf{a}_{02}) = \bar{\Psi}_{\text{iso}}(\bar{\mathbf{C}}) + \bar{\Psi}_{\text{aniso}}(\bar{\mathbf{C}}, \mathbf{a}_{01}, \mathbf{a}_{02}) \quad (5.23)$$

where the families of collagenous fibers are characterized by two direction vectors, $\mathbf{a}_{01}$ and $\mathbf{a}_{02}$.

For the formulation of the model, it is essential to include two tensors, $\mathbf{A}_1$ and $\mathbf{A}_2$, which are defined as the tensor products $\mathbf{a}_{0i} \otimes \mathbf{a}_{0i}$ (for $i = 1, 2$). Thus, with these tensors, it becomes feasible to define two invariants:

$$\bar{I}_4(\bar{\mathbf{C}}, \mathbf{a}_{01}) = \bar{\mathbf{C}} : \mathbf{A}_1 \quad (5.24)$$

$$\bar{I}_6(\bar{\mathbf{C}}, \mathbf{a}_{02}) = \bar{\mathbf{C}} : \mathbf{A}_2 \quad (5.25)$$

The invariants $\bar{I}_4$ and $\bar{I}_6$ represent the squares of the stretches in the directions of $\mathbf{a}_{01}$ and $\mathbf{a}_{02}$, respectively. Thus, they serve as stretch measures for the two families of fibers, providing a clear physical interpretation. For simplicity, the reduced form of the strain energy function is:

$$\bar{\Psi}(\bar{\mathbf{C}}, \mathbf{A}_1, \mathbf{A}_2) = \bar{\Psi}_{\text{iso}}(\bar{I}_1) + \bar{\Psi}_{\text{aniso}}(\bar{I}_4, \bar{I}_6) \quad (5.26)$$

Finally, the two contributions, $\bar{\Psi}_{\text{iso}}$ and $\bar{\Psi}_{\text{aniso}}$, to the function $\bar{\Psi}$ must be specified. The isotropic contribution is based on the Neo-Hookean model:

$$\bar{\Psi}_{\text{iso}}(\bar{I}_1) = \frac{c}{2}(\bar{I}_1 - 3) \quad (5.27)$$

where $c > 0$ is a stress-like material parameter.

The strain energy function describing the energy stored in the collagen fibers can be expressed as:

$$\bar{\Psi}_{\text{aniso}}(\bar{I}_4, \bar{I}_6) = \frac{k_1}{2k_2} \sum_{i=4,6} (\exp[k_2 \langle E_\alpha \rangle^2] - 1) \quad (5.28)$$

with:

$$\langle E_\alpha \rangle = \kappa(I_1 - 3) + (1 - 3\kappa)(I_i - 1), \quad i = 4, 6 \quad (5.29)$$

where $k_1 > 0$ is a stress-like material parameter and $k_2 > 0$ is a dimensionless parameter. The Macaulay brackets remove the effect of fiber compression.

The model supposes that the collagen fibers in each family are distributed around a mean preferred direction. The level of dispersion in the fiber orientations is described by the parameter κ ($0 \leq \kappa \leq \frac{1}{3}$). Therefore, fibers have no dispersion and are precisely aligned when $\kappa = 0$, and fiber distribution is random at $\kappa = \frac{1}{3}$.

The first term in the expression of the strain energy function denotes the contribution of the non-collagenous isotropic ground material. In contrast, the second term corresponds to the contributions from the various families of collagen fibers, incorporating the effects of dispersion. A fundamental assumption of the model is that collagen fibers can solely withstand tension, as they would buckle under compressive loading. Consequently, the anisotropic contribution in the strain energy function arises only when the strain of the fibers is positive, or equivalently, when $E_\alpha > 0$.

5.3.2 Viscoelasticity

In an elastic material, the stress is proportional to the deformation the material has undergone. This suggests that an applied force causes the material to deform. So, when the force is removed, the substance resumes its original shape. Conversely, this does not happen in viscous materials. Usually connected to liquids, viscosity is defined as flow resistance. While a high-viscosity material flows more slowly, a low-viscosity material flows quickly. The stress in these materials relates exactly to the rate of deformation (Lakes, 2009).

Linear elasticity, according to Hooke's law, applies to most solid materials. Unlike elastic materials, a viscous fluid under shear stress follows the following:

$$\sigma = \eta \frac{d\varepsilon}{dt} \quad (5.30)$$

where η is the viscosity.

That said, a viscoelastic material has both viscous and elastic properties, and because of the time-dependent relationship between stress and strain, a material's reaction to force or deformation may vary over time.

Stress-strain curves are frequently used to show the stiffness and strength of materials. When the material is linearly elastic, the curve is a straight line with a slope equal to the elastic modulus. Conversely, as Figure 5.2 shows, a linearly viscoelastic material produces a curved stress-strain

plot because, under constant strain rate deformation, both time and strain increase. Time is thus a factor for the viscoelastic material (Lakes, 2009).

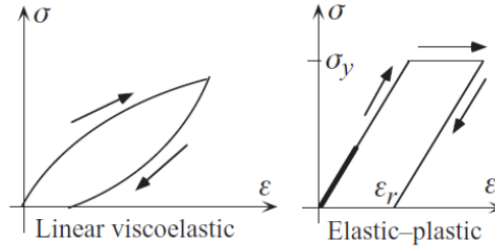


Figure 5.2: Stress–strain plots for deformation at constant strain rate followed by un-loading (adapted from Lakes (2009)).

Viscoelasticity causes variable paths for loading and unloading, resulting in the formation of a hysteresis loop. The hysteresis loop area represents the energy lost per unit volume over the entire deformation cycle (Meyers and Chawla, 2008).

As the influence of time can be isolated from any nonlinearity, it is better to apply a step stress or a step strain in time rather than a ramp, comparable to a constant rate of strain, to test and characterize viscoelastic materials. Consequently, step stress correlates with creeping, while step strain results in stress relaxation (Lakes, 2009).

Stress relaxation happens when an immediate strain is carried out in a viscoelastic solid and maintained constantly so that the associated stresses in the body decrease over time. This phenomenon is utilized as a tool to characterize the viscoelastic behavior of a material.

On the contrary, if the body is rapidly stressed and the stress is kept constant subsequently, viscoelastic materials experience a strain rise that varies with time. Accordingly, this process is known as viscoelastic creep, which relates to the tendency of a material to deform progressively under the effect of a steady mechanical force. Once a steady level is applied to the body and held, strain continues to build in response. Higher loads are frequently related to faster creep (Fung, 2013).

• Maxwell model

Mechanical models are employed to describe their behavior in the study of viscoelastic materials. These models typically comprise a combination of linear springs, characterized by a spring constant μ , and dashpots, defined by a viscosity coefficient η . The linear spring responds to an applied load by producing an immediate deformation directly proportional to the load. In turn, the dashpot generates a velocity proportional to the applied load at any given moment (Fung, 2013).

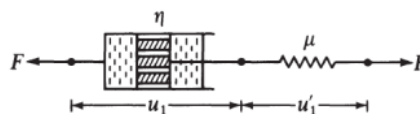


Figure 5.3: Maxwell model of viscoelastic material (adapted from Fung (2013)).

Each Maxwell element can be modeled as a spring connected in series with a dashpot. The spring represents the elastic, solid-like behavior, while the dashpot characterizes the viscous behavior, dependent on a relaxation time. Consequently, in the Maxwell model, as illustrated in Figure 5.3, the same force is transmitted through both the spring and the dashpot. When a force is suddenly applied at $t = 0$, the spring experiences an instantaneous deformation. However, the initial deflection of the dashpot is zero because there is no time for viscous deformation to occur.

Hence, when a load is applied suddenly, it causes an immediate deflection in the elastic spring, which is then followed by the gradual creep of the dashpot. If a sudden deformation is imposed, the spring reacts instantaneously, and stress relaxation in the dashpot follows (Fung, 2013).

5.3.2.1 Biological Tissues

Biomechanics finds particular utility for viscoelastic models since biological tissues exhibit viscoelastic properties, meaning that their loading history influences the mechanical state of these materials. The capacity of tissues to reduce stress in response to sustained deformation enables them to endure significant deformations without injury.

Thus, the relevance of viscoelastic behavior in biological tissues is particularly significant for the tissues in the female pelvis. During the second stage of labor, for instance, the fetus descends into the birth canal, which causes the vagina and surrounding pelvic tissues to deform, thereby increasing the applied force. With each successive contraction, these pelvic tissues exhibit viscoelastic properties, softening and behaving like a viscous material. This phenomenon allows the fetal head to dilate the vagina further, facilitating the progression of labor. Over time, and without injury, the tissues can return to their original shape (Abramowitch and Easley, 2016).

Moreover, incorporating a viscous component into the model is a crucial aspect of this study. Vila Pouca et al. (2020, 2022b) have developed visco-hyperelastic constitutive models with applications to biological tissues.

5.3.2.2 Viscoelastic Constitutive Model

The generalized Maxwell model for viscoelasticity is the most pertinent to the present investigation (Holzapfel et al., 2000).

Therefore, in this section, a comprehensive three-dimensional viscoelastic model for finite strains and minor deviations from equilibrium, as developed by Holzapfel et al. (2000), will be discussed. Notably, this model is not limited to isotropic conditions.

In Holzapfel et al. (2000), a decoupled formulation of the Helmholtz free-energy function Ψ is proposed, using the multiplicative decomposition of the deformation gradient into dilational and volume-preserving components. However, this approach employs multiplicative decomposition of the deformation gradient into elastic (rate-independent) and permanent (viscous) components. A viscoelastic material is a substance that exhibits free energy at any given position and time.

The variation of Ψ throughout an isothermal elastic process from the reference to the current configuration is expressed as:

$$\Psi(\mathbf{C}, \Gamma_1, \dots, \Gamma_m) = \Psi_{\text{vol}}^{\infty}(J) + \Psi_{\text{iso}}^{\infty}(\bar{\mathbf{C}}) + \sum_{\alpha=1}^m \gamma_{\alpha}(\bar{\mathbf{C}}, \Gamma_{\alpha}) \quad (5.31)$$

The first two terms in Equation 5.31 represent strain energy functions per unit reference volume, defining the equilibrium state of the solid. These terms describe the volumetric and isochoric elastic responses as $t \rightarrow \infty$, respectively. The superscript $(\bullet)^{\infty}$ denotes functions that illustrate the hyperelastic behavior in processes occurring at sufficiently slow rates.

The third term in the referenced equation relates to the dissipative potential, which accounts for the viscoelastic contribution, thereby extending the decoupled strain energy function into the viscoelastic regime. The scalar-valued functions γ_{α} ($\alpha = 1, \dots, m$) represent the configurational free energy of the viscoelastic solid and characterize the non-equilibrium state, describing the behavior of relaxation and creep.

The changes in the system are induced solely by isochoric deformations, ensuring that the volumetric response remains entirely elastic. Consequently, the configurational free energy depends on the modified right Cauchy-Green tensor $\bar{\mathbf{C}}$ and a set of strain-like internal variables denoted by Γ_{α} . This tensor characterizes the material's relaxation and/or creep behavior. The viscoelastic behavior is described by $\alpha = 1, \dots, m$ viscoelastic processes, each associated with a specific relaxation time $\tau_{\alpha} \in (0, \infty)$.

Equation 5.31 is the foundation for deriving the constitutive equations that describe viscoelastic behavior at finite strains. By starting from the decoupled free energy, a simple computation results in an additive split of $\mathbf{S}$.

$$\mathbf{S} = 2 \frac{\partial \Psi(\mathbf{C}, \Gamma_1, \dots, \Gamma_m)}{\partial \mathbf{C}} = \mathbf{S}_{\text{vol}}^{\infty} + \mathbf{S}_{\text{iso}} \quad (5.32)$$

where the last term corresponds to:

$$\mathbf{S}_{\text{iso}} = \mathbf{S}_{\text{iso}}^{\infty} + \sum_{\alpha=1}^m \mathbf{Q}_{\alpha} \quad (5.33)$$

Based on Equations 5.32 and 5.33, the following equation can be written:

$$\mathbf{S} = \mathbf{S}_{\text{vol}}^{\infty} + \mathbf{S}_{\text{iso}}^{\infty} + \sum_{\alpha=1}^m \mathbf{Q}_{\alpha} \quad (5.34)$$

The values $\mathbf{S}_{\text{iso}}^{\infty}$ and $\mathbf{S}_{\text{vol}}^{\infty}$ can be derived from constitutive models. In Equation 5.33, additional internal tensor variables were introduced, which can be interpreted as non-equilibrium stresses. The isochoric second Piola-Kirchhoff stress is split into an equilibrium part and a non-equilibrium part, characterized by the elastic response of the system and the viscoelastic response, respectively.

The isochoric non-equilibrium stress tensors $\mathbf{Q}_\alpha$ are defined as:

$$\mathbf{Q}_\alpha = J^{-\frac{2}{3}} \mathbb{P} : \bar{\mathbf{Q}}_\alpha \quad (5.35)$$

where $\mathbb{P}$ is the fourth-order projection tensor, and $\bar{\mathbf{Q}}_\alpha$ can be expressed as:

$$\bar{\mathbf{Q}}_\alpha = 2 \frac{\partial \gamma_\alpha(\bar{\mathbf{C}}, \Gamma_\alpha)}{\partial \bar{\mathbf{C}}} \quad (5.36)$$

Equations of the following form must be established to describe the development of the viscoelastic process:

$$\dot{\mathbf{Q}}_\alpha + \frac{\mathbf{Q}_\alpha}{\tau_\alpha} = \dot{\mathbf{S}}_{\text{iso } \alpha} \quad (5.37)$$

The material time derivative is indicated by the superposed dot. The isochoric second Piola-Kirchhoff stresses described in Equation 5.37 by the tensors $\mathbf{S}_{\text{iso } \alpha}$ correspond to the strain energies $\Psi_{\text{iso } \alpha}(\bar{\mathbf{C}})$ of the system.

Convolutional integrals allow one to get the solution of evolution Equation 5.37, which is true for some semi-open time interval $t \in (0, T]$:

$$\mathbf{Q}_\alpha = \exp\left(-\frac{T}{\tau_\alpha}\right) \mathbf{Q}_\alpha^{0+} + \int_{t=0+}^{t=T} \exp\left[-\frac{(T-t)}{\tau_\alpha}\right] \dot{\mathbf{S}}_{\text{iso } \alpha}(t) dt \quad (5.38)$$

5.3.3 Damage Model

The constitutive model employed for the simulations in this study incorporates a fatigue damage material model, originally developed by Vila Pouca et al. (2022b). This adopted constitutive model has previously demonstrated success in the simulation of childbirth. This section will provide a detailed explanation of the model.

5.3.3.1 Cumulative Damage Model Without Permanent Deformation

The constitutive material model utilized in this study is a quasi-incompressible, transversely isotropic visco-hyperelastic-damage model. That being said, the model applied to the PFM in this project incorporates several components: the Holzapfel-Gasser-Ogden model (Holzapfel et al., 2000) to represent the transversely isotropic hyperelastic behavior; the generalized Maxwell model, as described by Holzapfel and Gasser (Holzapfel and Gasser, 2001), to capture the viscoelastic behavior; and a fatigue damage model, developed based on experimental evidence, to account for damage accumulation due to submaximal repetitive loadings. Additionally, a nonlocal damage averaging of the integral type is included to reduce the mesh dependency on the damage contours (Vila Pouca et al., 2022b). Therefore, the integral-type nonlocal averaging method considers the microstructure to prevent the damage from concentrating in one area (Ferreira et al., 2017).

The objective is to simulate the behavior of soft tissue, which is nearly incompressible due to its high water content. The strain energy function is typically expressed in a decoupled form, separating the volumetric and isochoric contributions. This approach allows for the application

of a penalty function to the volumetric component, ensuring the material remains incompressible. Furthermore, the isochoric component can be further separated to incorporate the specific characteristics of the different constituents of the soft tissue. Specifically, this model considers a single family of muscle fibers embedded within a ground substance (or ground matrix) material. It is worth noting that a family of fibers comprises a group of fibers that share a preferred direction. Consequently, the complete strain energy function can be defined in a decoupled form as follows:

$$\Psi = \Psi_{\text{vol}} + \left[(1 - \widehat{D}_m) \cdot (\bar{\Psi}_m + \Psi_m^{\text{vsc}}) \right] + (1 - \widehat{D}_f) \cdot (\bar{\Psi}_f + \Psi_f^{\text{vsc}}) \quad (5.39)$$

where Ψ_{vol} represents the volumetric contribution, $\bar{\Psi}_m$ denotes the isochoric contribution of the ground matrix, and $\bar{\Psi}_f$ signifies the isochoric contribution of the family of fibers. Additionally, Ψ_*^{vsc} ($*$ = m, f) account for the viscous effects, and $\widehat{D}_*$ ($0 < \widehat{D}_* < 1$) ($*$ = m, f) represent the nonlocal continuous damage variables. The viscous effects and the damage variables influence each isochoric component of the strain energy. The indices m and f correspond to the contributions from the ground matrix and the fibers, respectively.

Each contribution can be defined as:

$$\Psi_{\text{vol}} = K \left(\frac{J^2 - 1}{2} - \ln(J) \right) \quad (5.40)$$

$$\bar{\Psi}_m = C_{10} (\bar{I}_1 - 3) \quad (5.41)$$

$$\bar{\Psi}_f = \frac{k_{11}}{2k_{12}} \left\{ \exp \left[k_{12} (\bar{I}_4 - 1)^2 \right] - 1 \right\} \quad (5.42)$$

$$\Psi_*^{\text{vsc}} = \sum_{\alpha=1}^m \Upsilon_{\alpha}^* \quad * = m, f \quad (5.43)$$

In this context, $J = \det \mathbf{F}$, known as the Jacobian, represents the local rate of change in the deformed configuration volume v relative to the reference configuration volume V . The symmetric right Cauchy-Green deformation tensor is defined as $\mathbf{C} = \mathbf{F}^T \mathbf{F}$. The deformation gradient $\mathbf{F}$ can be decomposed into a volumetric (dilatational) part $\mathbf{F}_{\text{vol}} = J^{1/3}$ and a volume-preserving (distortional) part $\bar{\mathbf{F}} = J^{-1/3} \mathbf{F}$, where $\det \bar{\mathbf{F}} = 1$. Consequently, the isochoric right Cauchy-Green tensor is $\bar{\mathbf{C}} = \bar{\mathbf{F}}^T \bar{\mathbf{F}}$, with its first invariant given by $\bar{I}_1 = \text{tr} \bar{\mathbf{C}}$. The stretch $\bar{\lambda}_f$ in the direction $\mathbf{m}_0$ of the fiber family is related to the pseudo-invariant $\bar{I}_4$ such that $\bar{\lambda}_f^2 = \bar{I}_4 = \mathbf{m}_0 \cdot \bar{\mathbf{C}} \cdot \mathbf{m}_0$. The scalar value functions Υ_{α}^* ($*$ = m, f) serve as dissipative potentials and determine the viscoelastic contribution of each process α , which corresponds to the number of parallel elements in the Maxwell model. As mentioned earlier, each Maxwell element consists of a spring in series with a dashpot, where the spring represents the solid behavior, characterized by a free energy parameter β_{α^*} , and the dashpot represents the viscous behavior, being a function of a relaxation time τ_{α^*} .

The parameters of the transversely isotropic visco-hyperelastic model are as follows: K for the volumetric contribution, C_{10} for the ground matrix contribution, k_{11} and k_{12} for the fibers' contribution, and β_{α^*} and τ_{α^*} ($*$ = m, f) for the viscoelastic contribution.

The continuous damage model was developed based on the concept of accumulated equivalent strain, Ξ_t^{acc} , as defined by Peña (2014):

$$\Xi_t^{acc} = \int_0^t \left| \frac{d\bar{\Psi}}{ds} \right| ds \quad (5.44)$$

This enables continuous damage accumulation within the deformation process, characterized by a straightforward equation for damage evolution:

$$\frac{d\Xi_t^{acc}}{dt} = \left| \frac{d\bar{\Psi}}{dt} \right| \quad (5.45)$$

The damage propagation and accumulation were characterized by an adaptation of the Lemaitre and Plumtree nonlinear cumulative expression (Milella, 2012). In line with prior research (Vila Pouca et al., 2020), the ratio between the number of cycles and the number of cycles to fatigue failure was substituted with the ratio between the accumulated equivalent strain and its maximum value. Additionally, a parameter was introduced to facilitate the easy adjustment of the weight of the damage variable:

$$D_* = D_{\max*} \left\{ 1 - \left(1 - \left(\frac{\Xi_{t*}^{acc}}{\Xi_{\max*}^{acc}} \right)^{\frac{1}{C_*}} \right) \right\} \Big|_{*=m,f} \quad (5.46)$$

The parameters governing this damage evolution include the maximum equivalent strain $\Xi_{\max*}^{acc}$ ($*$ = m, f), the exponent C_* ($*$ = m, f), and the weighting parameter $D_{\max*}$ ($0 < D_{\max*} \leq 1$) ($*$ = m, f). The criteria for the evolution can be delineated as:

$$\dot{\Xi}_{t*}^{acc} = \begin{cases} \frac{d\bar{\Psi}_*}{dt}, & \dot{\Xi}_{t*}^{acc} \geq 0 \\ 0, & \dot{\Xi}_{t*}^{acc} < 0 \end{cases} \Big|_{*=m,f} \quad (5.47)$$

This damage model is categorized as a "local" model due to its dependency on spatial discretization. In essence, the mesh structure influences the solution, and with mesh refinement, these effects become more pronounced. To address this challenge, nonlocal theory approaches can be employed to mitigate or eliminate mesh dependency. So, Vila Pouca et al. (2022b) decided to utilize an integral-type averaging scheme integrated with the material law. Through the application of nonlocal averaging of the integral type, the nonlocal damage variable $\widehat{D}_*$ ($*$ = m, f) can be expressed using a spatially weighted averaging integral as follows:

$$\widehat{D}_* = \int_V \chi(x, \eta) D_* dV(\eta) \Big|_{*=m,f} \quad (5.48)$$

where $\chi(x, \eta)$ is a weighted average operator, D_* ($*$ = m, f) represents the local damage, and $\widehat{D}_*$ ($*$ = m, f) denotes the measurement of nonlocal damage at a specific material point x , which has been averaged over a finite volume V containing a set of points. The dimension of the nonlocal volume is determined by a constitutive parameter denoted as the characteristic length L_c , as seen

in Figure 5.4.

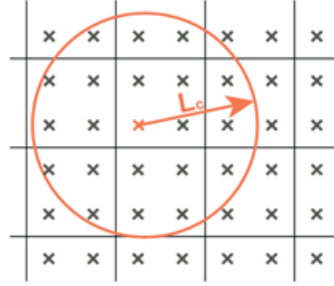


Figure 5.4: Characteristic length (adapted from Vila Pouca et al. (2022b)).

The nonlocal damage at a material point x arises from an averaging process of the local quantities of material points within the nonlocal averaging volume.

That said, the averaging operator can be defined as:

$$\chi(x, \eta) = \frac{\varpi(x, \eta)}{\Omega_r(x)} \quad (5.49)$$

where $\varpi(x, \eta)$ is the weighting function, and $\Omega_r(x)$ represents the representative volume, defined as:

$$\varpi(x, \eta) = \begin{cases} (1 - \frac{\|x - \eta\|^2}{L_c^2})^2 & \|x - \eta\| \leq L_c \\ 0 & \|x - \eta\| > L_c \end{cases} \quad (5.50)$$

$$\Omega_r(x) = \int_V \varpi(x, \eta) dV(\eta) \quad (5.51)$$

The only parameter in the nonlocal implementation is the characteristic length. This parameter requires careful selection as it can significantly influence the outcomes. Opting for a value that is too small results in the non-local solution converging toward the local solution, potentially retaining mesh dependency. Conversely, selecting an excessively large value leads to overhomogenization, thereby diminishing the significance of the results.

The constitutive parameters were determined through a custom micro-genetic algorithm, as outlined in Vila Pouca et al. (2020). No constraints were placed on the domains of the ground matrix and muscle fibers, except for the nonlocal damage parameter, the characteristic length L_c . The calibration process involved replicating laboratory experiments in computer simulations. The parameter set was iteratively adjusted (randomly by the algorithm), and the computational force-displacement curve was compared with the experimental counterpart until the least error was attained. The material parameters yielding the closest approximation between the experimental and computational force-displacement curves were utilized in the simulations. Since these material models are phenomenological, their properties lack physical interpretation and cannot be directly observed (Vila Pouca et al., 2022b).

5.4 Finite Element Method

Many physical events in engineering and scientific domains can be represented in partial differential equations. However, employing classical analytical techniques to solve these equations for arbitrary shapes often proves exceedingly challenging. Thus, the FEM is a vital numerical tool, offering an approximate solution to these equations. From an engineering perspective, the FEM is a method for tackling engineering problems like stress analysis, heat transfer, fluid flow, and electromagnetics using computer simulation (Fish and Belytschko, 2007).

Solving a set of differential equations analytically is often highly difficult. Consequently, a practical approach involves reformulating the problem through discretization, which consists of replacing the continuous problem with an approximate discrete problem. Therefore, this process facilitates the transformation of a continuous model into an approximate discrete model (Oliveira, 1990).

The fundamental concept of the FEM is to divide the body into finite elements, frequently just termed elements, connected by nodes, as illustrated in Figure 5.5, and produce an approximate solution of the variable that is being researched. The elements in FEM analysis can vary in shape, depending on the nature and scale of the problem being addressed. These shapes include triangular, quadrilateral, tetrahedral, cubic, and others, determined by the spatial arrangement of nodes. The resulting configuration is called the finite element mesh, and the process of its creation is known as mesh generation. The behavior of each element is described in relation to a finite set of degrees of freedom. In order to produce a precise solution, a large number of nodes is usually required. Hence, a high-performance computer analysis is necessary. Consequently, the computational effort increases significantly, presenting a crucial consideration in FEM applications (Fish and Belytschko, 2007).

Therefore, FEM is a great tool to assist in elucidating the mechanics connected with vaginal delivery. However, some precautions are necessary to obtain correct data.

5.4.1 Abaqus[®] Finite Element Software

Abaqus[®] represents a software tailored for finite element analysis and computer-aided engineering. It is based on the FEM and can tackle issues ranging from simple linear analysis to the most demanding nonlinear simulations. In Abaqus[®]/CAE (Complete Abaqus[®] Environment), the model of the physical problem is defined, and an Abaqus[®] input file is created. Abaqus[®]/CAE comprises distinct modules, each tasked with specific functions within the modeling process, such as geometrical definition, material property specification, and mesh generation. Subsequently, the simulation, which generally is conducted as a background process, is the stage in which Abaqus[®]/Standard or Abaqus[®]/Explicit solve the numerical problem stated in the input file. The fundamental difference between Abaqus[®]/Explicit and Abaqus[®]/Standard resides in the type of problems they are suitable for. Abaqus[®]/Standard is a general-purpose finite-element solver designed to simulate true static and structural dynamic events (Simulia, 2024c). Conversely, Abaqus[®]/Explicit is an explicit-dynamic finite-element solver ideal for the simulation of brief

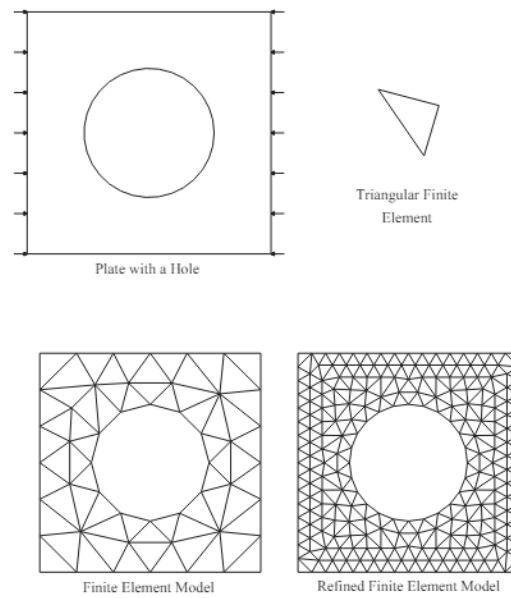


Figure 5.5: Geometry and finite element meshes (adapted from [Fish and Belytschko \(2007\)](#)).

transient, quasi-static, and dynamic events ([Simulia, 2024b](#)). When the simulation is completed or ongoing, it becomes feasible to assess the results interactively using Abaqus®/Viewer. This component provides diverse visualization options.

The software provides an extensive library of elements and the most common materials already defined. However, this solution is also customizable since the users can build their own material models for complex situations, including them as user-defined material subroutines (UMAT) ([Simulia, 2009b](#)).

More complicated constitutive models are commonly defined in a UMAT, which is responsible for computing the Cauchy stress tensor and the Jacobian matrix in the spatial description. These variables must be determined within the subroutine. The former is defined in the software as STRESS and the latter as DDSDD ([Simulia, 2009b](#)).

In this work, the simulations were carried out using Abaqus®/Standard. Thus, Figure 5.6 provides a basic flowchart of data and actions from the beginning of an Abaqus®/Standard analysis to the end of a phase, therefore helping better to comprehend the overall structure and operation of the software.

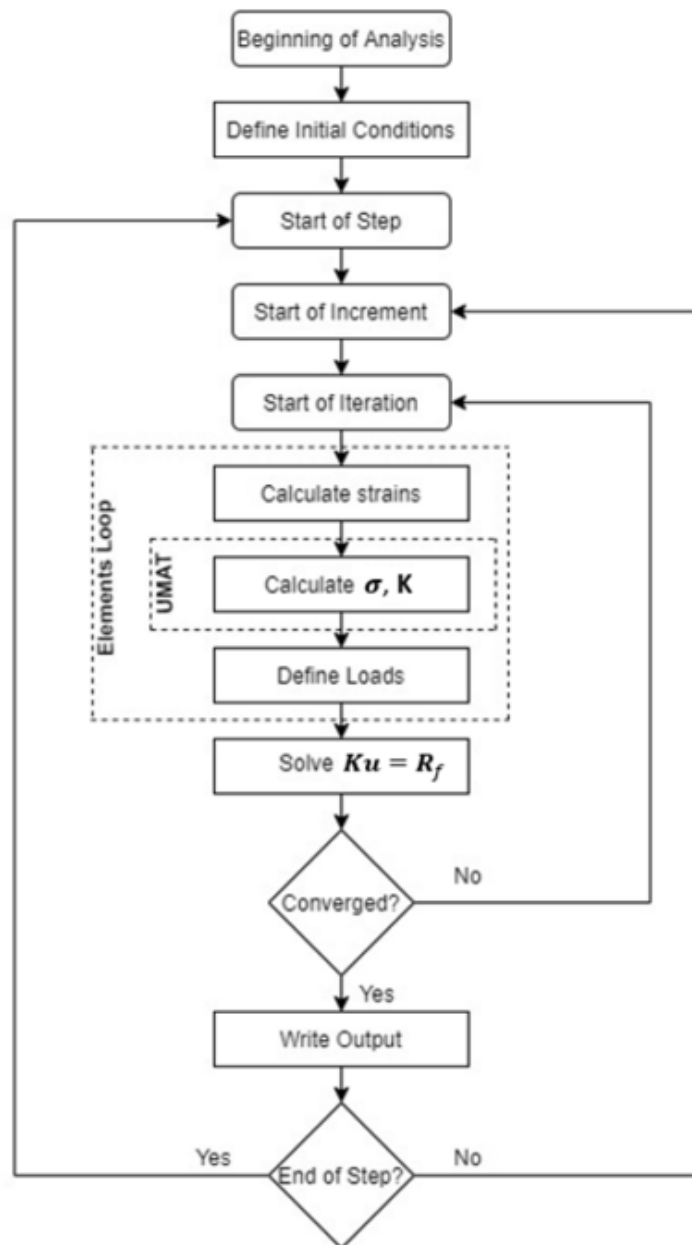


Figure 5.6: Abaqus®/Standard flowchart (adapted from [Pereira \(2020\)](#)).

Chapter 6

Childbirth Biomechanics

6.1 Introduction

Biomechanics, a specialized branch of bioengineering, utilizes engineering principles and mechanical techniques to analyze biological systems. Thus, by examining the interaction between physical forces and living tissues, biomechanics seeks to understand the mechanics of living organisms. This knowledge is crucial for comprehending normal physiological functions, anticipating the effects of changes, and suggesting techniques for artificial intervention ([Moura, 2020](#)).

Computational biomechanics, particularly the application of finite element modeling, has proven especially significant in medicine by evaluating the impact of mechanical stresses and strains. In the context of the female pelvis, this approach has enhanced the understanding of the complex phenomena occurring during vaginal childbirth ([Alperin, 2016](#)).

Childbirth simulations offer significant advantages over traditional physical models and invasive evaluations by providing a valuable understanding of the mechanisms of vaginal delivery and the associated maternal and fetal injuries. Moreover, clinical practice recognizes that various maternal, fetal, and obstetric factors influence the childbirth process in complex, interrelated ways that are challenging to quantify *in vivo*. Hence, simulations enable a profound comprehension of these interactions and can guide future adjustments. Additionally, a biomechanical understanding of the birth process is essential for developing preventative strategies ([Howard and Makhoul, 2016](#)). In this sense, computer models help to improve clinical practice, enhancing women's health and quality of life and lowering medical costs ([Oliveira et al., 2016b](#)).

6.2 Previous Vaginal Birth Models

Computational models offer potential resources for quantitatively investigating the physiology of labor in ways that were previously inconceivable.

In 2004, [Lien et al.](#) developed a three-dimensional computational model to estimate LAM stretch during vaginal delivery, marking one of the first attempts to analyze the mechanical process

of childbirth. Since then, significant advancements have been made in the analysis of vaginal delivery and the physiology of the female pelvic floor.

Parente et al. (2008) built a finite element model that simulated the pelvic bones, PFM, and fetus, to analyze the stretch and deformations of the pelvic floor generated by the passage of the fetus during vaginal birth. The computer simulations demonstrated that the PFM are subjected to substantial deformations during the passage of the fetal head. Additionally, significant progress was made in understanding the influence of the material properties of the PFM during vaginal delivery. In 2009, Parente et al. adopted a modified version of the incompressible transversely isotropic hyperelastic constitutive model established by Humphrey and Yin (1987) to characterize the PFM. The results underscored the significance of accurately defining material parameters and the necessity of a suitable constitutive model. As the fetus significantly influences the dynamics of vaginal delivery, Parente et al. (2009a) also investigated the effect of an OP malposition of the fetus on PFM stretch compared to the normal occipito-anterior (OA) position. Their study concluded that the OP malposition results in substantially higher stretch values for the PFM, thereby increasing the risk of stretch-related injuries. In 2010, Parente et al. assessed the impact of pelvic muscle activation during vaginal delivery. Their findings indicated that the activation of PFM correlated with heightened resistance to the descending fetus, implying that muscle contraction could impede fetal descent and potentially prolong labor. To measure the damage inflicted by the fetal head to the pelvic floor muscles, Parente et al. (2010a) studied the influence of fetal head flexion during vaginal delivery. The results imply that the fetal head flexion may aid birth and protect the pelvic floor.

Jing et al. (2012) created a subject-specific finite element model of the human pelvic floor and fetal head using *in vivo* magnetic resonance imaging data. The PFM were described by an anisotropic visco-hyperelastic constitutive model with material parameters derived from biaxial tests. They simulated a brief second stage of labor to analyze the pathomechanics of LAM injury. Results indicated that the pubovisceral muscle and the muscle proximal to the perineal body experience the highest strain levels.

Additionally, Lepage et al. (2015) developed a finite element model of the pelvic cavity of a pregnant woman to simulate childbirth and assess urogenital prolapse. Their findings revealed that uterosacral ligaments incur significant deformations, suggesting a potential association with urogenital prolapse. Furthermore, Jean Dit Gautier et al. (2018) employed a finite element model to examine the effects of pregnancy on the uterosacral ligament and PFM. Said model facilitated the evaluation of pelvic structures across various gestational stages and postpartum periods. Changes in the length of uterosacral ligaments and the thickness of the puborectal portion of the LAM were analyzed during and after pregnancy.

Still connected to the biomechanics of labor and the injuries on the PFM, Oliveira et al. (2016b) examined the PFM damage during vaginal delivery by increasing the complexity of the constitutive model used to define the mechanical behavior of the PFM. In other investigations, the structural damage model employed had already been utilized to characterize the damage process in biological soft tissues. The model indicated that even an uncomplicated vaginal delivery causes harm to

the PFM, obtaining more than 10% of damaged fibers. This analysis established that the puborectalis component of the LAM is the most prone to injury. These computer models can also be used to evaluate the effect of different obstetrical interventions on the PFM. Henceforth, [Oliveira et al. \(2016a\)](#) conducted a biomechanical investigation into the effects of episiotomy during childbirth. For that, a normal vaginal delivery, both without and with episiotomy, was simulated, with the fetus in vertex presentation and OA position. The findings showcased that a mediolateral episiotomy decreases muscle stress and confines damage to a localized area.

More recently, [Vila Pouca et al. \(2017\)](#) delved into the influence of introducing a viscous component, characteristic of biological soft tissues, on the PFM, thereby enabling a rate-dependent analysis. Simulations were done for varied labor lengths to explore the differences between precipitated, normal, and prolonged second stages of labor. The addition of viscoelasticity introduced a stiffness component, resulting in higher forces compared to an elastic response. The results revealed that lower efforts in pelvic tissues were observed as the delivery duration increased progressively. Moreover, it was demonstrated that resting stages during labor allowed tissues to relax and partially regain initial properties, reducing forces and stresses. In [Vila Pouca et al. \(2018\)](#), the model was enhanced by incorporating an anisotropic visco-hyperelastic constitutive model that accounted for tissue damage. This advancement enabled the study of reaction forces, the loss of pelvic floor integrity, and the effects of different labor durations. Results demonstrated that reaction forces varied with duration, with higher force levels correlated with higher stretch rates. The pubovisceral muscle emerged as the most affected area of the LAM. Moreover, the relaxation properties of the tissue contributed to a reduction in damage levels, supporting the theory of delayed pushing during the second stage of labor. Additionally, [Vila Pouca et al. \(2019\)](#) introduced a chemo-mechanical constitutive model to investigate uterine contractions during fetus expulsion without a predetermined trajectory. Finally, in 2022, [Vila Pouca et al.](#) implemented a transversely isotropic visco-hyperelastic model for the LAM, incorporating cumulative damage with nonlocal averaging, for the second stage of labor. The aim was to investigate various labor management practices, comparing spontaneous versus directed pushing. Consequently, the results indicated that the origin of the pubovisceral muscle accumulates the highest damage, making it the most susceptible site for tear development. Moreover, the three maternal pushes per contraction approach resulted in less damage and appeared to be the optimal approach.

Concerning the research question of fetal head molding, [Moura et al. \(2021\)](#) conducted a numerical investigation on the biomechanics of fetal head molding, focusing on measuring specific diameters and calculating the corresponding molding index.

Additionally, regarding birthing positions, [Borges et al. \(2021\)](#) explored the effects of various maternal postures on the PFM. The analysis concluded that in non-flexible sacrum positions, the maternal pelvis and PFM experience elevated stresses, which are biomechanically disadvantageous. Further, in the study published in 2022, [Moura et al.](#) aimed to advance understanding by investigating the consequences of a persistent OP malposition on both mother and fetus, particularly when the mother adopts a flexible-sacrum posture and the fetal head is deformable, thus enabling molding. This study marked the first instance where a finite element model compris-

ing maternal pelvic bones and a deformable fetal head with viscous properties was employed to simulate vaginal delivery while considering various fetal positions.

Considering adding the perineum to the PFM model, [Routzong et al. \(2019\)](#) created the first model encompassing both the pubovisceral muscle and superficial perineal structures. This study quantified the stretch of the pubovisceral muscle and perineal body, ultimately concluding that superficial perineal structures play a pivotal role in female birth injuries and should be incorporated into computational simulations of vaginal delivery. In 2021, [Cechova et al.](#) developed a finite element model incorporating the perineum and PFM. Their study aimed to assess the points and timing of maximum stresses on the LAM and perineal body resulting from the expulsion of the fetal head. In the recent study published in 2024, [Moura et al.](#) conducted a comprehensive analysis of perineal injuries during childbirth, using a finite element model of the female pelvis that integrated both the pelvic floor and perineum. The primary objectives were to accurately represent the dimensions of the levator hiatus, gain a deeper understanding of the impact of incorporating perineal structures in the analysis, and identify the regions most susceptible to injury during delivery. This investigation placed particular emphasis on elucidating the mechanisms underlying often overlooked perineal injuries.

Additionally, [Moura et al. \(2024a\)](#) extended the research by developing a machine learning (ML) framework to predict stresses on crucial regions of the PFM during vaginal delivery. This premise was achieved by training ML algorithms on FEM simulation data.

Indeed, the development of these models has been driven by specific research questions, often simplifying factors not directly pertinent to each issue. Material properties such as viscoelastic effects were commonly overlooked until recently. Furthermore, cyclic accumulation of damage, a phenomenon that has just been shown to happen in collagenous and muscular tissues, was only incorporated into models by [Oliveira et al. \(2016a,b\)](#) and [Vila Pouca et al. \(2022a,b\)](#).

Although the literature on the biomechanical impact of subsequent births is scarce, [Campos \(2023\)](#) investigated this topic in her master's thesis. Accordingly, a simplified model of the fetal head (a sphere) and the PFM (a cone) was developed, and a continuous damage model was implemented using an UMAT subroutine. [Campos](#) found a slight reduction in the maximum principal stress and reaction forces values in a second simulation.

Therefore, it is notable that none of the prior published research aimed to assess the damage resulting from subsequent births on a woman's pelvic floor. Thus, the critical distinction between earlier projects and the current study lies in the incorporation of the ability to cyclically accumulate damage within the finite element model, specifically investigating the impact of both first and second vaginal deliveries. Therefore, this project represents a novel contribution to the field.

Chapter 7

Materials and Methods

7.1 Damage Model Evaluation

In order to evaluate the damage model that was chosen for the development of this dissertation, it was decided to test it on a simpler geometry before moving on to the vaginal delivery simulations. Thus, a numerical simulation of a thin perforated plate undergoing cyclic stretching along its longitudinal axis was conducted (Oliveira et al., 2016b; Calvo et al., 2007).

Therefore, the plate was subjected to a periodic triangular displacement profile, which caused it to alternate between tensile and compressive strain. The strain rate for the uniaxial test (applied in direction x) was 14 mm/s and 19 mm/s, while for the biaxial test (applied in directions x and y), it was 10 mm/s. This information is illustrated in Figure 7.1. The uniaxial test lasted 4 seconds, while the biaxial test ran for 2 seconds.

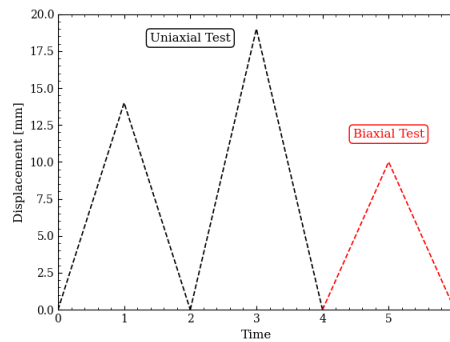


Figure 7.1: Cycles of the uni and biaxial tests.

Given the symmetry of the problem, the finite element simulation only considered one-fourth of the plate. Figure 7.2 illustrates the geometry, boundary conditions, and C3D8H element-based finite element mesh coupled with the element and integration points defined. Lastly, the fibers were defined by the x direction ($n_0 = \{1, 0, 0\}$) (Oliveira et al., 2016b).

Hence, to clearly demonstrate the performance of the constitutive model, the transversely

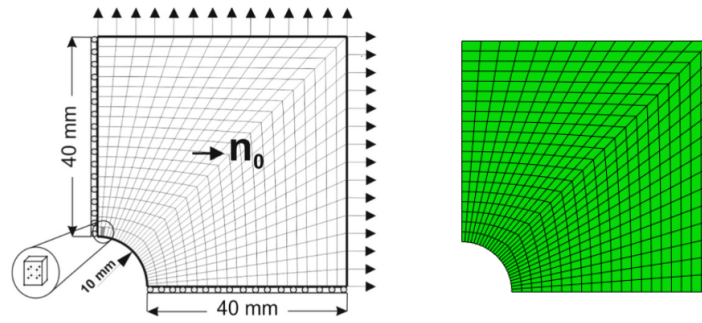


Figure 7.2: Representation of the geometry, fiber direction, boundary conditions, and finite element mesh of the perforated square plate (adapted from Oliveira et al. (2016b)).

isotropic viscohyperelastic model, which incorporates cumulative damage with nonlocal averaging, as developed by Vila Pouca et al. (2022b) and discussed in Section 5.3.3.1, was implemented. This execution was achieved by using a UMAT subroutine developed in FORTRAN, allowing the incorporation of both the viscoelastic component and cumulative damage into Abaqus®.

The material and damage parameters for the thin plate were the same as used for the PFM in the simulations carried out later for the vaginal childbirth. They will be detailed in Section 7.2.1.1.

The first simulation was therefore carried out in Abaqus®. The damage variables resulting from this first simulation were introduced directly into a second simulation. For that, the UMAT subroutine was initialized using an SDVINI subroutine, which is used to define initial solution-dependent state variable fields. So, two second simulations were carried out: one with all the damage variables from the first simulation, and another in which the matrix damage variables were initialized to zero (total matrix damage recovery), so that only the fiber damage variables were initialized with the values from the end of the first simulation.

To evaluate the performance of the model in the plate, the following parameters were analyzed:

- Impact of different characteristic length values in the local and nonlocal damage results.
- Maximum principal stresses as a function of the normalized path for the three cycles of each simulation.
- The distribution of the damage state variables in the extracellular matrix and the fibers throughout the simulations.
- Damage values of the fibers and matrix as a function of the normalized path at the end of each simulation.
- Stresses as a function of strain for a specific node throughout the uniaxial and biaxial tests.

It is worth mentioning that for the analysis of results, a path and a specific node were also defined, as can be seen in Figure 7.3. The path is represented by the red curve, and the node by the yellow dot.

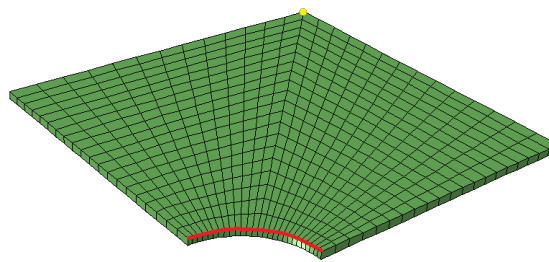


Figure 7.3: Path (red curve) and node (yellow dot) defined on the mesh of finite elements of the thin perforated plate.

7.2 Vaginal Birth Simulations

The present study involved adapting a validated finite element model of the pelvic floor muscles and the fetal head to conduct simulations of the first and subsequent normal childbirths. The primary goal of such an approach is to study how a second birth might influence the damage caused to the pelvic floor of a woman. Predicting delivery outcomes can be fundamental in preventing difficulties for both the mother and fetus.

7.2.1 Anatomic Modeling of the Female Pelvic Floor Muscles

The geometric model of the PFM developed by [Moura et al. \(2024b\)](#) was employed for the simulations in this research. This model includes the LAM and the coccygeus muscles. The magnetic resonance images of a 30-year-old healthy woman were used to segment the pelvic bones and the visible portions of the LAM (pubococcygeus and puborectalis muscles). The geometries were created using Rhinoceros® software. Additionally, the iliococcygeus and coccygeus muscles were added based on recommendations from obstetricians and anatomical facts from the literature ([Drake et al., 2009](#); [Netter, 2014](#)) since they were not visible in the utilized images.

Due to the constraints of this engineering approach in accounting for geometric variability, it was opted to use the standard geometries of a gynecoid pelvis. Therefore, the pelvic diameters roughly correspond to the average values documented in the literature ([Keller et al., 2003](#)). Thus, the dimensions of the muscles also approximate the average exterior measurements because these dimensions are determined mainly by their insertions. Figure 7.4 displays a visual representation of the geometric model of the PFM.

After the geometrical model was exported to Abaqus® software, a finite element mesh was created. This mesh, composed of hexahedral solid elements, included 5712 nodes and 3618 C3D8H elements for the PFM. Therefore, the mesh comprises 8-node hexahedral elements with a hybrid formulation and constant pressure. Hexahedral elements were selected for their superior accuracy compared to other three-dimensional finite elements. Moreover, hybrid elements were employed in this scenario due to the belief that they are the optimal choice when dealing with nearly incompressible materials. The mixed formulation provides a solution to numerical issues caused by

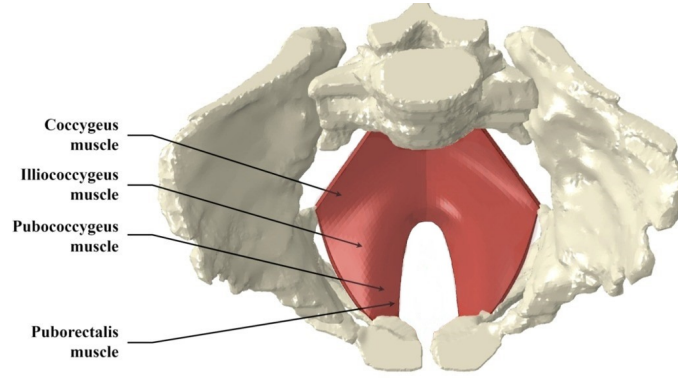


Figure 7.4: Geometric model of the PFM: coccygeus, iliococcygeus, pubococcygeus and puborectalis muscles (adapted from [Moura et al. \(2024b\)](#)).

incompressibility, such as volume strain "locking" ([Vila Pouca et al., 2022b](#)). The final mesh for the PFM can be observed in Figure 7.7.

Moreover, the PFM were modeled with varying thicknesses. The puborectalis and pubococcygeus muscles were assigned a thickness of 6.5 mm ([Majida et al., 2010](#)), while the iliococcygeus and coccygeus muscles were set to be 3.6 mm thick ([Azzam et al., 2019](#)).

7.2.1.1 Mechanical Characterization

As mentioned before, a transversely isotropic viscohyperelastic model with cumulative damage and nonlocal averaging, as seen in [Vila Pouca et al. \(2022b\)](#), was used to characterize the PFM. The constitutive parameters were determined using a custom-developed microgenetic algorithm, which involved calibrating experimental results with numerical data. Thus, it is essential to mention that the experimental data resulted from fatigue tests performed on the ovine ishiococcygeous muscle, which is the most homologous structure to the origin of the pubovisceral muscle in humans ([Vila Pouca et al., 2020](#)). Therefore, the material parameters used to characterize the PFM are listed in Table 7.1. Since this work only included the passive behavior of the muscles, it was assumed for all the simulations that the muscles are completely relaxed due to anesthesia, which is observed in more than 60% of women who give birth vaginally, in conformity with the Centers for Disease Control ([Osterman and Martin, 2011](#)).

Table 7.1: PFM constitutive model parameters, obtained from [Vila Pouca et al. \(2022b\)](#).

Hyperelastic	Viscoelastic	Local damage	Nonlocal damage
$C_{10} = 0.14 \text{ MPa}$	$\beta_{1_m} = \beta_{1_{f1}} = 5.08 \text{ (—)}$	$D_{max_m} = 0.9 \text{ (—)}$	$L_c = 0.001 \text{ mm}$
$\kappa_{11} = 0.05 \text{ MPa}$	$\tau_{1_m} = \tau_{1_{f1}} = 33.59 \text{ s}$	$\Xi_{max_m}^{acc} = 4.07 \text{ MPa}$	
$\kappa_{12} = 0.23 \text{ (—)}$	$\beta_{2_m} = \beta_{2_{f1}} = 0.79 \text{ (—)}$	$C_m = 1.65 \text{ (—)}$	
	$\tau_{2_m} = \tau_{2_{f1}} = 44.14 \text{ s}$	$D_{max_{f1}} = 1.0 \text{ (—)}$	
		$\Xi_{max_{f1}}^{acc} = 2.25 \text{ MPa}$	
		$C_{f1} = 0.27 \text{ (—)}$	

The implemented constitutive model requires the input of the orientation of the muscle fibers. Thus, it is presumed that the direction of the muscle fibers aligns with the orientation of the maximal principal stress lines when the structure is being deformed (Silva et al., 2015).

7.2.2 Anatomic Modeling of the Fetal Head

Only the fetal head, which is the most voluminous structure of the fetus, was modeled, following the approach used in prior numerical simulations of vaginal birth. Hence, the fetal head was modeled as a rigid body with anatomical dimensions consistent with literature values for a full-term fetus. The specific diameters used for the 50th percentile fetal head were: suboccipitobregmatic diameter of 10 cm, suboccipitofrontal diameter of 10.5 cm, occipitofrontal diameter of 12.0 cm, mentovertical diameter of 13.0 cm, and submentobregmatic diameter of 11.5 cm (Parente et al., 2008). For a better visualization of the diameters, they are presented in Figure 7.5. Since the results will be analyzed only on the PFM and not on the fetal head, the head was defined as a simpler numerical model - a shell with a thickness of 1 mm.

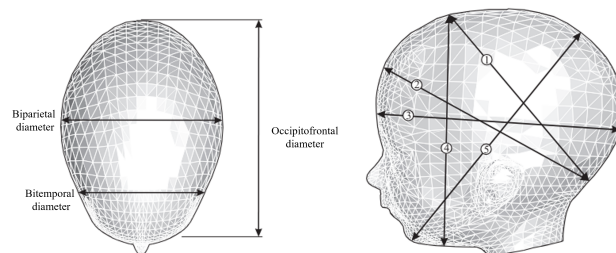


Figure 7.5: Principal diameters of the fetal head. 1: suboccipitobregmatic diameter; 2: suboccipitofrontal diameter; 3: occipitofrontal diameter; 4: mentovertical diameter; 5: submentobregmatic diameter (adapted from Parente (2008)).

Subsequently, the geometrical model was exported to Abaqus[®] software to create a finite element mesh. The mesh for the fetal head consisted of 6310 nodes and 4128 elements. Triangular shell elements with reduced integration (S3R) were used to model the head. The final mesh for the fetal head is displayed in Figure 7.6. To control the movements of the fetal head, a reference node was assigned to its center.

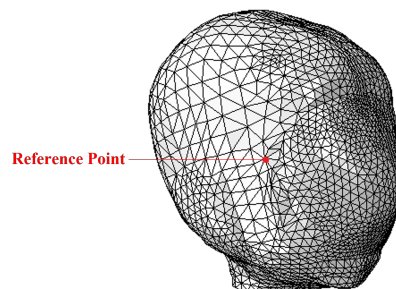


Figure 7.6: Finite element mesh of the fetus head with the identification of the reference point (red dot) used to control its movement.

For the second part of this study, the simulations of second vaginal births were carried out with a smaller head. Such process will be detailed in Section 7.2.4.4.

7.2.2.1 Mechanical Characterization

As previously stated, the fetal head was considered rigid in the simulations conducted throughout this study. Nevertheless, linear elastic properties have been assigned to the fetal head since Abaqus® requires properties to be specified. However, these will not be relevant because the head will be regarded as a rigid structure in the simulations.

This decision to consider the fetal head rigid was made to simplify the simulations, as accounting for a deformable head significantly increases computational demands. By adopting this simplification, the convergence of the simulations was facilitated (Moura et al., 2024b).

7.2.3 Interactions and Boundary Conditions

Concerning the boundary conditions, the nodes on the extremities of the PFM were fixed. Additionally, a curve, referred to as a path, was defined on the most inferior part of the pelvic floor mesh. This specific group of nodes will be crucial for later assessing maximum principal stresses, stretch, and damage values. These mentioned boundary conditions are illustrated in Figure 7.7.

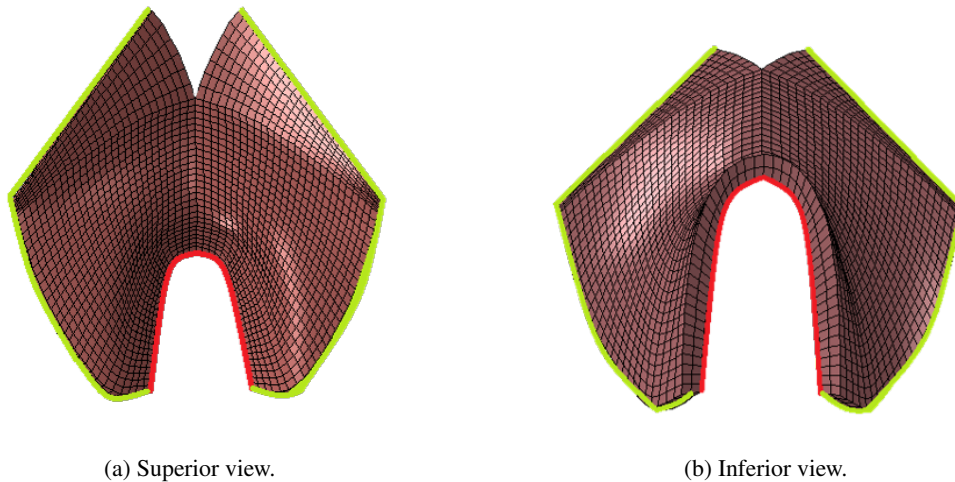


Figure 7.7: Boundary conditions and path defined on the mesh of finite elements. The fixed nodes are shown by the light green lines, and the path is shown by the red line.

It is also worth mentioning that in order to assess some results, such as the maximum principal stresses and damage values, the length of the path has been normalized so that one end corresponds to 0 and the other to 1, as shown in Figure 7.8.

During the simulation process, the interactions between the PFM and the fetal head were modeled using the standard Abaqus® contact algorithm. Kinematic contact conditions were enforced by establishing contact constraints between these anatomical structures, utilizing the Augmented Lagrange method to facilitate surface interactions (Moura et al., 2024b). This approach uses

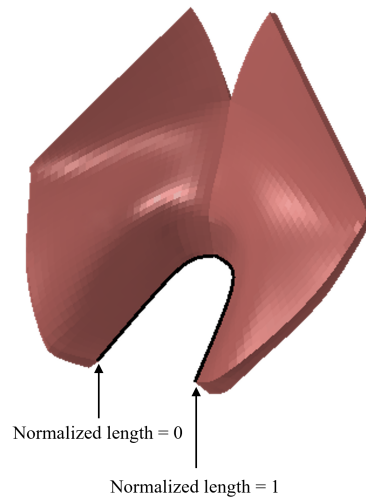


Figure 7.8: Normalized path length, presented in black.

the linear penalty method and an iterative methodology for augmentation, effectively reducing the penetration distance. This concept is only applicable to hard pressure-overclosure relationships (Simulia, 2024a).

7.2.4 Simulation of the Vaginal Birth

The simulations were conducted using Abaqus[®] software to replicate the second stage of labor, with the fetus in the OA position and vertex presentation (Moura et al., 2024b).

The simulations were conducted by enforcing a trajectory to the fetal head. Therefore, the descent and extension of the fetal head were controlled by the reference node in the fetal head, while the other degrees of freedom were left free. The trajectory of the fetal head was based on the cardinal movements of labor, as explained in Section 4.5, and on the Curve of Carus, as defined by Jing et al. (2012). This curve demonstrates the anteroposterior movement relative to the vertical displacement of the fetal head and can be observed in Figure 7.9.

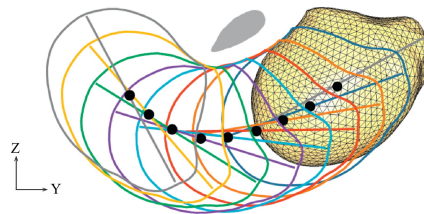


Figure 7.9: Curve of Carus (black dots) for prescribing the displacement and rotation of the fetal head during the second stage (adapted from Jing et al. (2012)).

In the current simulations, fetal descent was divided into nine distinct steps, with the vertical movement of the fetal head reaching up to 80 mm, which corresponds to the average length of the birth canal. The steps and associated vertical displacement are expressed in Table 7.2. It can be

seen that steps 4, 5, and 6 exhibit the same vertical displacement. This action was taken in order to release the x and z degrees of freedom of the fetal head.

Although the pelvic bones were not included in the simulation, the boundary conditions applied to the PFM, particularly the fixed nodes near the pubic symphysis insertions, were sufficient to constrain the movement of the head throughout descent (Moura et al., 2024b). The position of the fetal head at various stages of descent is illustrated in Figure 7.10.

Table 7.2: Vertical displacement of the fetal head (mm) for each step.

Step	1	2	3	4	5	6	7	8	9
Vertical displacement of the fetal head (mm)	15	20	25	35	35	35	50	65	80

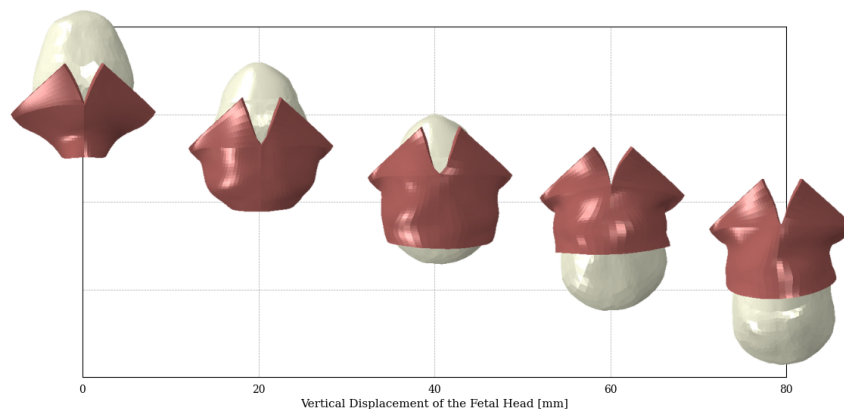


Figure 7.10: Fetal head evolution from the starting position (0 mm of fetal descent) in respect to the PFM.

Several numerical challenges were found and resolved in order to effectively execute the realistic passage of the fetal head through the PFM. Therefore, the Stabilize method in Abaqus[®] was employed to tackle these difficulties, in steps 8 and 9. Said method employs the dissipated energy fraction and assumes that the problem is initially stable, but instabilities may arise during the process. While the model remains stable, viscosity's effects, including viscous energy's dissipation, are minimal. Therefore, the inclusion of extra artificial damping does not produce any impact. When a specific area becomes unstable, the velocities in that area increase. As a result, some of the stored energy in the region is released and dissipated by the applied damping. Abaqus[®]/Standard computes and prints the damping factor, derived from the solution of the initial increment of a step, to the message file. Typically, the first increment of the step remains steady and does not require the introduction of damping. The damping factor is then determined so that the extrapolated dissipated energy for the step is a small proportion of the extrapolated strain energy (Simulia, 2009a).

With the average dimensions for the fetal head (50th percentile), three distinct simulations were conducted: the first one, to replicate an average second stage of labor in a nulliparous woman, and two second vaginal births (with and without recovery rate), to mimic subsequent births.

In order to test another parameter, the simulations for second births were also performed with a smaller fetus head.

7.2.4.1 Simulation of a First Vaginal Birth

The simulation for the first birth was conducted as mentioned in the Section 7.2.4. After its end, using a Python script, it was possible to extract the damage state variables (STATEV) for the matrix and fibers, obtained in this first simulation.

7.2.4.2 Simulation of a Second Vaginal Birth without Recovery

The damage variables obtained in the first simulation were introduced at the start of the second simulation. For that, the UMAT subroutine had to be initialized with the state variables of interest using an SDVINI subroutine, which allowed state variables dependent on assigned solutions to be initialized. For this simulation, the state variables were introduced exactly as they were taken from the first simulation.

After trying to simulate the second delivery, as mentioned, the computer simulation showed great difficulties in progressing, probably due to the damage introduced right from the start. In this sense, the variables corresponding to the matrix were initialized to zero, while those corresponding to the fibers maintained their value. In this approach, the simulation managed to progress until the end.

7.2.4.3 Simulation of a Second Vaginal Birth with Recovery

However, considering that PFM have the ability to recover between births, for this simulation, the damage state variables were set to half the value at the end of the first simulation. Therefore, a recovery percentage of 50% was considered. This 50% recovery rate was chosen due to the lack of literature on recovery percentages and the complexity of quantifying such recovery.

As verified for the second simulation without recovery, the damage state variables of the matrix were initialized at zero, with those of the fibers remaining at 50% of the value of the first simulation.

7.2.4.4 Simulations of Second Vaginal Births with a Smaller Head

Two more simulations of a second delivery were carried out, this time with a smaller fetal head. In line with what was mentioned earlier, one of the simulations was obtained for a recovery rate of 50% and the other for the PFM without any recovery. Once again, the matrix damage state variables were initialized to zero for both of these simulations.

An automated mesh morphing algorithm was employed to achieve a smaller head size, designed to adjust fetal head diameters without needing to identify the specific axes of variation. Notably, the automation for the occipitofrontal and supraoccipitomenta diameters is identical (Figure 4.3b), as the literature indicates these diameters change in two directions. Similarly, the

automation for the biparietal and bitemporal diameters (Figure 4.3a) is the same since both these diameters are known to vary in only one direction (Bernardo, 2022).

Therefore, a fetal head in the 5th percentile was selected for these simulations. Thus, the diameters of the head are as follows: occipitofrontal diameter of 10.56 cm, supraoccipital diameter of 12.48 cm, biparietal diameter of 8.84 cm, and bitemporal diameter of 7.35 cm (Bernardo, 2022).

It becomes important to mention that the mechanical characterization of the fetal head is the same as described in Section 7.2.2.1.

7.2.5 Biomechanical Model Evaluation

The following parameters were investigated for every simulation in order to evaluate the biomechanical model:

- The stretch of the PFM: the stretch ratio is defined as the ratio between the current length and the original length. The path shown in Figure 7.7 was selected for this analysis, as it was shown to exhibit the highest stretches in previous studies. This approach allows for the measurement of length variations throughout the simulation, thereby enabling the determination of stretch values corresponding to fetal descent.
- The maximum principal stresses: they were quantitatively assessed along the defined path (Figure 7.7) at the moment of maximum stretch.
- The damage values: the (nonlocal) damage values for the extracellular matrix and the fibers were computed for the path at the moment of maximum stretch.
- The distribution of the damage variable in the muscle fibers, denoted as $\widehat{D}_{f1}$, color-coded as follows: red for $1 \geq \widehat{D}_{f1} \geq 0.8$, orange for $0.8 > \widehat{D}_{f1} \geq 0.45$, yellow for $0.45 > \widehat{D}_{f1} \geq 0.35$, green for $0.35 > \widehat{D}_{f1} \geq 0.1$, and blue for $\widehat{D}_{f1} < 0.1$. Hence, it was established a qualitative risk scale, such that red areas indicate "critical" risk, where muscle fiber mechanical properties are affected up to 100% (complete tear); orange areas denote "high" risk; yellow areas signify "noteworthy" risk; green areas represent "moderate" risk; and blue areas indicate "low" risk (Vila Pouca et al., 2022b). This distribution was analyzed throughout and at the end of each simulation.
- The damage evolution at the critical location, which is the element of maximum expected damage.
- The maximum damage at the critical location.

Chapter 8

Results and Discussion

8.1 Damage Model Evaluation

Initially, simulations were performed on a perforated thin plate to validate the chosen constitutive model.

- **Influence of different characteristic length values:**

Different characteristic length values were tested in order to see how they would impact the damage distribution for the extracellular matrix and fibers in the local and nonlocal contexts.

As previously mentioned in Section 5.3.3.1, the selection of the characteristic length parameter is essential, as it strongly influences the results. A small value results in the nonlocal solution converging toward the local solution, while selecting an excessively large value leads to over-homogenization, thereby diminishing the significance of the results.

Various values of L_c were therefore tested: 0.00001 mm, 0.001 mm and 10 mm. As can be seen in Figure 8.1, for a value of 0.001 mm, the local and nonlocal solutions end up being close, but for a value of 10 mm, the spread of local and nonlocal damage is quite different, both for the matrix and the fibers.

Since the model developed by Vila Pouca et al. (2022b) had already validated the value of 0.001 mm for L_c , and taking into account the results obtained here, it was decided to use this value in all the simulations carried out in this work.

- **Maximum principal stresses:**

Moreover, the maximum principal stresses were measured along the defined path at the hole of the plate (Figure 7.3) at the moments of the simulations when the three maximum stretches occurred. Therefore, there were two stretches for the uniaxial test and one for the biaxial test. This evaluation was performed for the three simulations mentioned earlier: the first simulation, the second simulation, and the second simulation, where the matrix damage values were initialized at zero. The resulting plots are depicted in Figure 8.2. Each full line corresponds to the displacement imposed in the corresponding cycle.

Hence, for the three displacement peaks that occur at 14 mm and 19 mm for the uniaxial test and at 10 mm for the biaxial test, the maximum principal stresses obtained are always greater

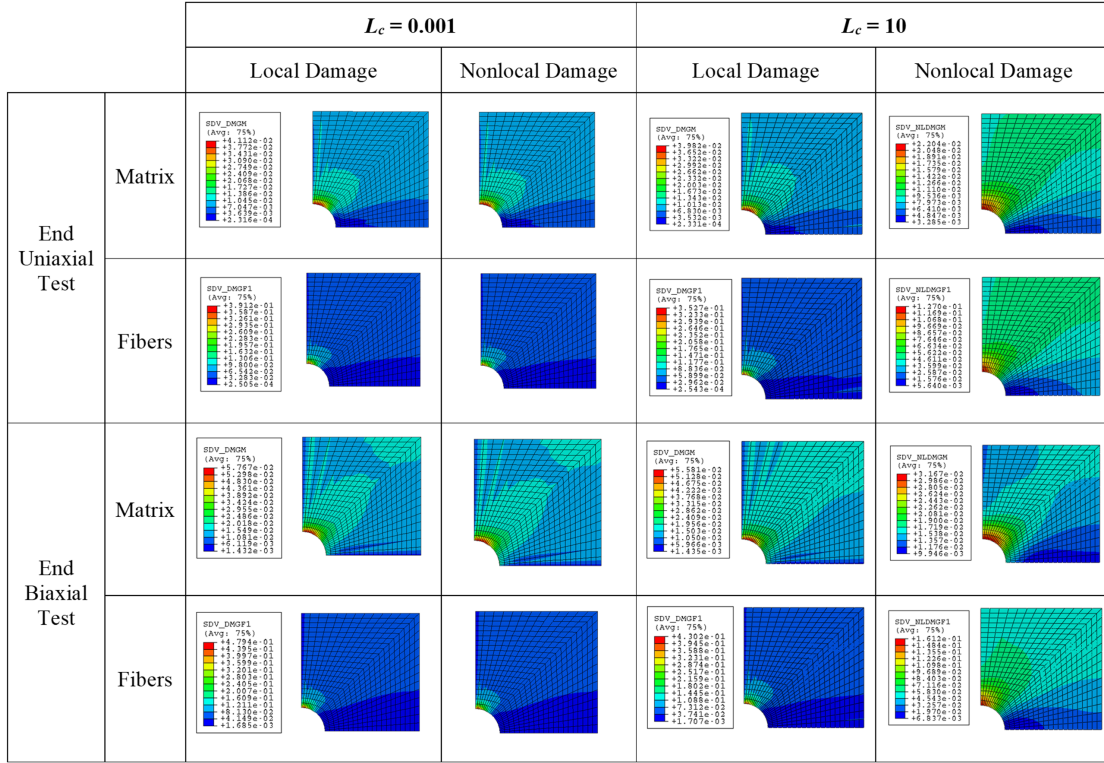
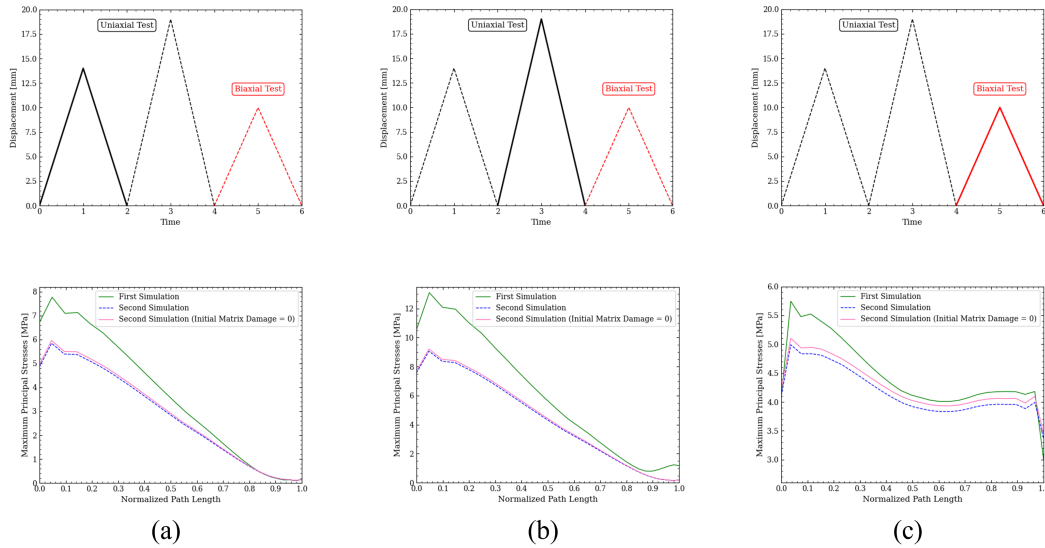
Figure 8.1: Evaluation of different L_c values on local and nonlocal damage.

Figure 8.2: Maximum principal stresses (in MPa) measured on the defined path, for the 3 different simulations, at displacements of (a) 14 mm, (b) 19 mm and (c) 10 mm, achieved during the two cycles of the uniaxial test and the one cycle of the biaxial test, respectively.

for the first simulation. This result is expected since the plate is already damaged after the first simulation. Therefore, lower stress values indicate that in order to obtain the same elongation of

the fibers, a lower force is required.

- **Damage:**

The damage process in plates with a hole is limited to the regions next to the perforation, where stress concentrations manifest at the early stages of loading. The Figures 8.3 - 8.5 display the matrix and fiber damage contour in the deformed configuration during both uniaxial and biaxial tests across the three simulations, allowing for the observation of this physical event. As the load increases, the damage propagates from these areas of strain energy concentration.

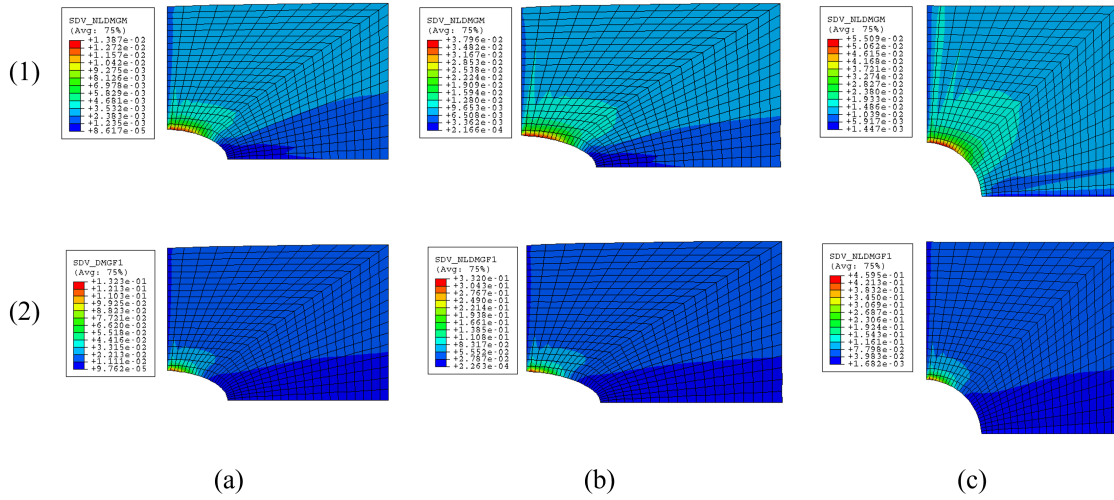


Figure 8.3: First simulation: extracellular matrix (1) and fiber (2) damage contour plot in the deformed configuration at displacements of (a) 14 mm, (b) 19 mm and (c) 10 mm, achieved during the two cycles of the uniaxial test and the one cycle of the biaxial test, respectively.

Regarding the uniaxial test, the peak damage values for the extracellular matrix resulted in 0.0139, 0.0599, and 0.0122 at 14 mm for the first, second and second simulation where the matrix damage was initially zero, respectively. For the displacement of 19 mm, the peak matrix damage values obtained were 0.0380, 0.0842, and 0.0362 for the first, second and second simulation where the matrix damage was initially zero, respectively. Concerning the damage of the fibers, the peak values were 0.132, 0.402, and 0.401 at 14 mm, for the first, second, and second simulation where the matrix damage was initially zero, respectively. Lastly, at 19 mm displacement, the fibers damage values obtained were 0.332, 0.559, and 0.556 for the first, second, and second simulation where the matrix damage was initially zero, respectively.

When the biaxial test was conducted, the tissue's anisotropic properties caused extra damage in the matrix and fibers. This damage was concentrated in places with high strain energy values, as shown in Figures 8.3 - 8.5. Furthermore, it is evident that the progression of damage varies in the two directions. These findings indicate that the constitutive model effectively simulates damage accumulation and accurately predicts failure modes and zones.

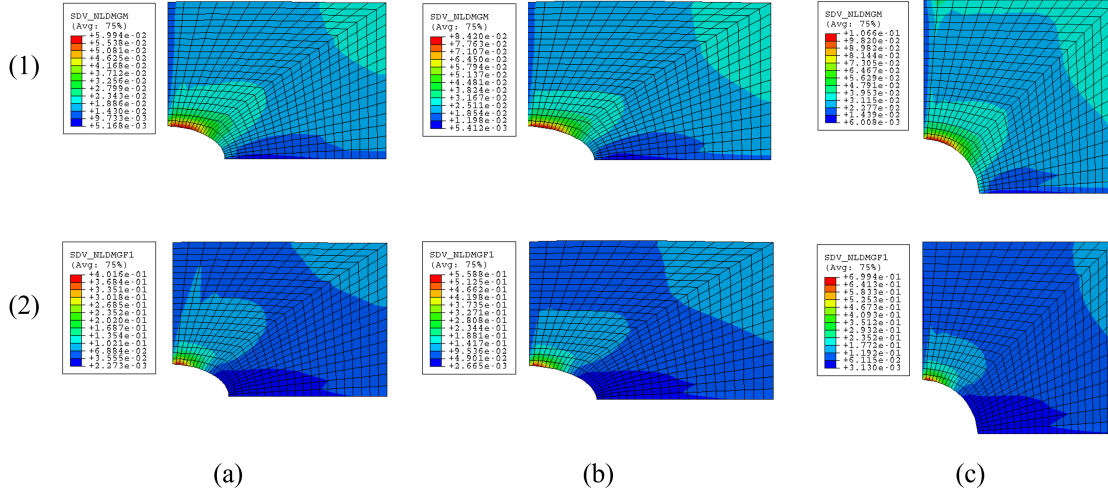


Figure 8.4: Second simulation: extracellular matrix (1) and fiber (2) damage contour plot in the deformed configuration at displacements of (a) 14 mm, (b) 19 mm and (c) 10 mm, achieved during the two cycles of the uniaxial test and the one cycle of the biaxial test, respectively.

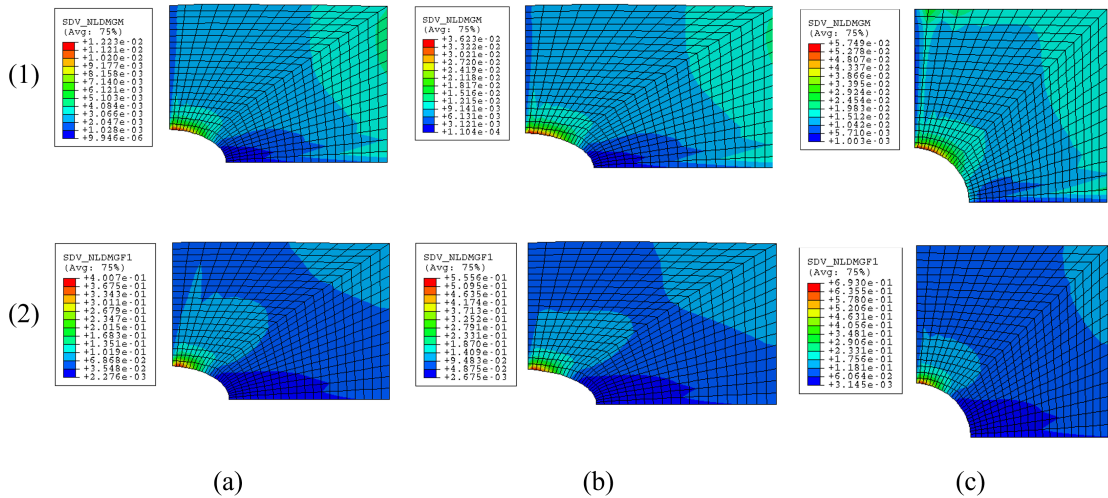


Figure 8.5: Second simulation with initial matrix damage recovery: extracellular matrix (1) and fiber (2) damage contour plot in the deformed configuration at displacements of (a) 14 mm, (b) 19 mm and (c) 10 mm, achieved during the two cycles of the uniaxial test and the one cycle of the biaxial test, respectively.

The predictions of the model indicate that the damage caused is characterized by muscle fiber rupture and extracellular matrix disruption. Similar to muscle injuries, the muscle fibers experience an initial injury while the matrix remains intact until a higher amount of trauma is encountered throughout the loading cycle (Oliveira et al., 2016b).

Besides, it is also possible to see that the model and UMAT are working correctly since, in the second simulations, the damage values are being initialized with the values resulting from the first, and there is an evolution of the damage variables throughout the simulations.

The damage values of the fibers and matrix as a function of the normalized path at the end of each simulation were also obtained. Such results are presented in Figure 8.6. Consequently, in the second simulation, in which the values of the state variables for the matrix and fibers are initialized with the values resulting from the first simulation, the damage at the end of the simulation is much higher than in the first, both for the matrix and the fibers. In the second simulation, in which the matrix damage is initialized to 0, the final damage to the matrix is similar to the damage to the matrix in the first simulation, but the damage to the fibers is much higher, as would be expected.

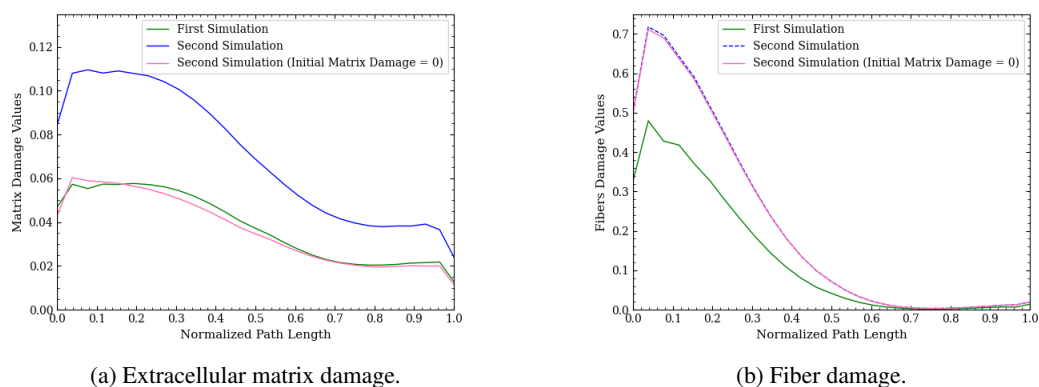


Figure 8.6: Damage values measured on the defined path, at the end of each simulation.

• Stress-strain evolution:

In order to obtain the stress in function of the strain, the node in the top right-hand corner of the plate was chosen, as can be seen in Figure 7.3. These results were obtained throughout each simulation, i.e., for the uniaxial and biaxial tests for each of the three simulations. It is necessary to highlight that the displacement data were extracted in magnitude, and the stress values in maximum principal stresses, directly from Abaqus® software.

As an example, the plot obtained for the second simulation is shown here. Thus, the stress-stretch evolution can be seen in Figure 8.7.

Therefore, these results show that the model is able to capture the typical stress-strain behavior observed in fibrous soft tissues and seems to confirm the soundness of the proposed formulation.

Consequently, it was possible to proceed to the childbirth simulations in the anatomical model.

8.2 Vaginal Birth Simulations

The main objective of this study was to develop computer simulations of vaginal childbirth and to investigate the impact of the damage on subsequent births. Thus, the childbirth simulations were conducted according to the description in Section 7.2.

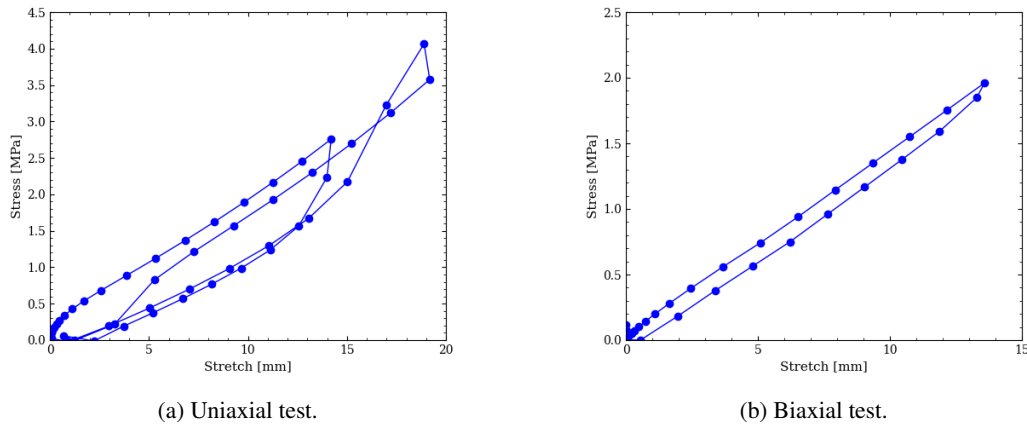


Figure 8.7: Stress-stretch evolution for the second simulation, throughout the uni and biaxial tests.

To analyze the results, the first simulation of vaginal delivery will be called the first simulation, the second simulation in which there is a 50% recovery of fibers damage will be referred to as second simulation (50% recovery), and the second simulation in which there is no recovery of fibers damage, i.e., the state variables are inserted with the values extracted from the first simulation, will be termed second simulation (no recovery). Some of the relevant results are discussed below.

- **Stretch:**

The maximum stretch ratio is influenced by various factors, including the relative sizes of the fetal head and maternal pelvis and the degree of fetal head molding. Hence, the maximum stretch ratio is predicted to fluctuate substantially among individuals ([Ashton-Miller and DeLancey, 2009](#)).

As such, Figure 8.8 shows the stretch along the vertical displacement of the fetal head for the PFM for the first simulation. It is worth remembering that the stretch was measured on the designated path defined earlier (Figure 7.7) and that the initial length of the path was 126.6 mm. As a result, in the present study, the maximum stretch obtained for the PFM was 2.18, occurring at approximately a vertical descent of the fetal head of 65.0 mm, which corresponds to the moment when the fetus begins the head extension.

Analyzing Figure 8.8, it can be seen that the stretch values are similar for the current study and for the work carried out by [Moura et al. \(2024b\)](#), who obtained a maximum stretch value of 2.15, at a vertical displacement of 74.72 mm. This slight discrepancy in values can be explained by the fact that the degrees of freedom of the fetus's head were left free, so the movements of the head may have slightly influenced the stretch values. In addition, although the geometry of the PFM and the fetal head is the same for both investigations, a damage model was implemented in the present one, which may have caused the fetal head to fit slightly differently in the muscles. Such a phenomenon may explain the variations in the values.

In this work, the stretch observed was notably higher compared to prior studies simulating childbirth. For instance, [Parente et al. \(2008\)](#) reported a stretch of approximately 1.63, while [Silva](#)

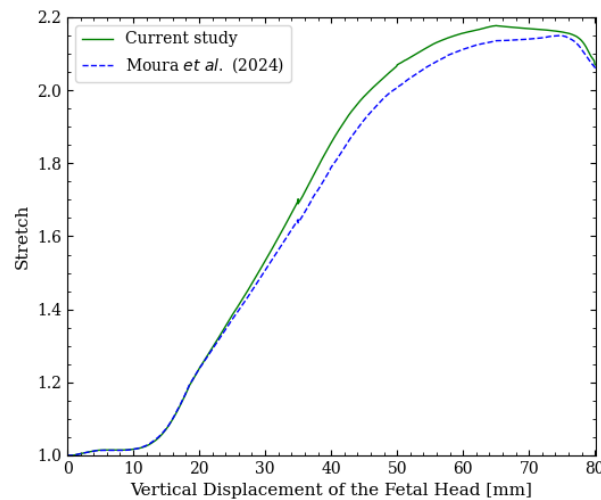


Figure 8.8: Stretch values obtained for the PFM on the defined path, during the vertical displacement of the fetal head.

[et al. \(2015\)](#) found a value of 1.53. However, this variance is likely due to the different geometrical models used, as in the current work, smaller dimensions were employed to model the levator hiatus, providing a more realistic representation of the constrained space available for fetal descent. In conclusion, the findings of this study align with previously published muscle stretch ratio ranges, which typically fall between 1.53 and 3.26 ([Lien et al., 2004](#); [Parente et al., 2008](#)).

Since the geometry of the fetal head and PFM is the same for the three simulations, the stretch values were equal for the first simulation, the second simulation without recovery, and the second simulation with recovery.

- **Maximum principal stresses:**

The maximum principal stresses exerted by the fetal head on the PFM were measured along the normalized path at the point of maximum muscle stretch. This maximum stretch occurred at the end of step 8 in all three simulations, corresponding to a descent of 65 mm. Moreover, the normalized path scale ranges from 0 to 1, with 0 representing one extremity, 0.5 representing the midpoint, and 1 representing the opposite extremity. The resulting plot is shown in [Figure 8.9](#).

In agreement with the literature ([Silva et al., 2015](#); [Parente et al., 2009b](#); [Oliveira et al., 2016a](#)), there are peaks in the stress values at the extremities of the path and also near the midpoint. As anticipated, the highest stress values in this study were found at the insertion points of the rectal area of the LAM in the coccyx, as well as in the symphysis pubis (right and left insertion points). The insertion points in the symphysis pubis, represented by the initial (0) and final (1) normalized lengths in [Figure 7.8](#), are expected to exhibit similar stress values. However, slight differences in these two values were observed across the three simulations, which may be attributed to an asymmetric fetal head engagement with the PFM. This asymmetry may arise because only the fetal vertical movement and head extension are controlled by the reference point in the fetal head,

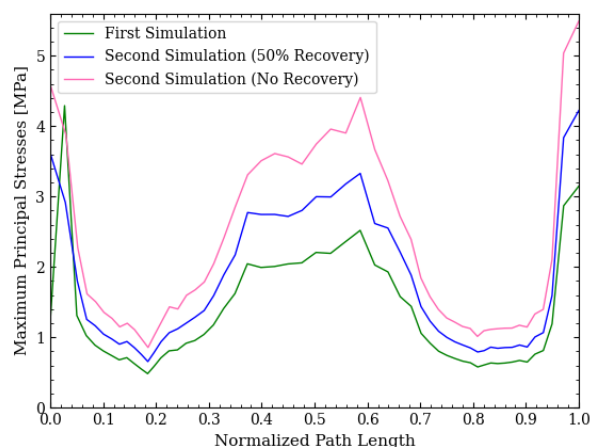


Figure 8.9: Maximum principal stresses in MPa calculated along the normalized path, at the instant of maximum stretch.

leaving other degrees of freedom unrestricted. The middle peak of the maximum principal stress curve corresponds to the posterior (rectal) area of the LAM, approximately at 0.5 of the normalized length along the path, as noted in [Silva et al. \(2015\)](#).

Analyzing the results, the highest stresses recorded for the PFM for the first simulation, the second simulation with 50% recovery, and the second simulation without recovery were 4.29 MPa, 4.22 MPa, and 5.49 MPa, respectively. Considering the entire muscle, the maximum principal stress values were measured at the insertion points of the rectal area of the LAM in the symphysis pubis for all three simulations. The stress values ranged from 1.23 to 4.59 MPa on the right side of the muscle and between 3.14 and 5.49 MPa on the left side.

Comparing the different simulations, these results are not in line with what was expected since it was predicted that the second simulations would show lower maximum principal stresses because the PFM would already be damaged. So, it is anticipated that the maximum principal stresses decrease as accumulated damage increases, reducing the ability of the pelvic floor to withstand the descent of the fetal head. Furthermore, when the perforated plate model was evaluated, it was found that in the second simulations the stresses are lower, as expected. Therefore, the maximum principal stress values obtained for the vaginal delivery simulations are the opposite of what was expected. These phenomena can be explained by the fact that with damaged PFM, the load distribution during the descent of the fetal head might change. The damaged regions might transfer more load to the remaining intact or less damaged areas, leading to higher localized stresses.

To further study the impact of the stresses, the distribution of the maximum principal stresses on the PFM was analyzed in the instant of maximum muscle stretch. The results are shown in [Figure 8.10](#).

Once again, it is possible to conclude that the highest values are reached at the insertion points of the rectal area of the LAM in the coccyx and in the symphysis pubis. By comparing the stress

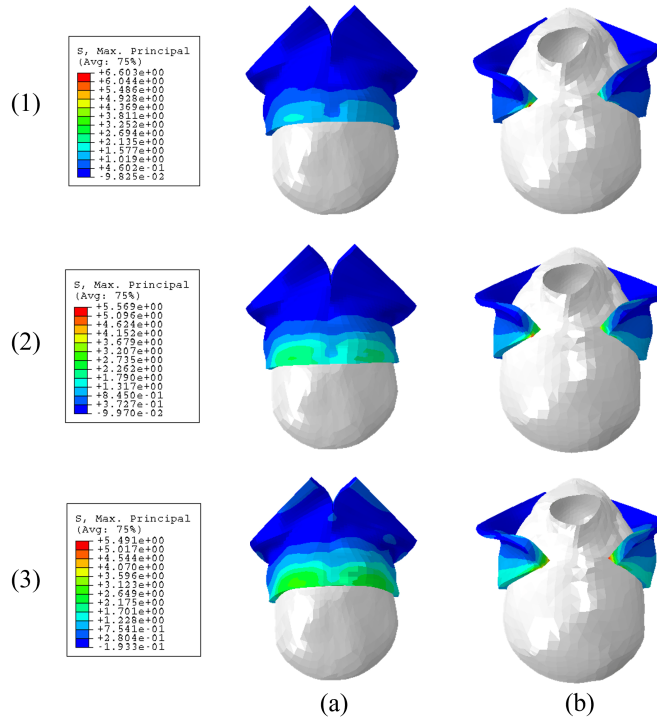


Figure 8.10: Distribution of the maximum principal stresses in MPa on the PFM for the (1) first simulation, (2) second simulation with 50% fibers damage recovery, and (3) second simulation without fibers damage recovery, at the instant of maximum stretch. (a) corresponds to the posterior view, and (b) to the anterior view.

distribution within the muscle for the different simulations, it becomes evident that the impact extends to a larger area of the pelvic floor, its most inferior region, for the simulations of the second births. Specifically, the second simulation without fibers damage recovery shows a greater area with higher stresses values.

- **Damage:**

Similar to what was studied for the maximum principal stresses, the evolution of the (nonlocal) damage to the matrix and fibers was also assessed for the moment of greatest stretch of the muscles as a function of the normalized path. As a result, the plots in Figure 8.11 were obtained. Furthermore, it becomes essential to mention that the damage values were extracted for the path elements at the centroid level.

As can be seen in Figure 8.11a, the behavior of the matrix damage is relatively similar to that of the maximum principal stresses, meaning there are damage peaks at the extremities of and in the middle of the path, which correspond to the insertion points of the rectal area of the LAM in the coccyx, and as in the symphysis pubis (right and left insertion points). For the three simulations carried out, the matrix damage values end up being similar since, due to problems in stabilizing the numerical simulation, the state variables corresponding to the matrix damage had to be initialized to zero in the second simulations. Only in this way were the simulations able to evolve. Therefore, as the matrix damage started from zero in all three simulations, the values are similar in all three

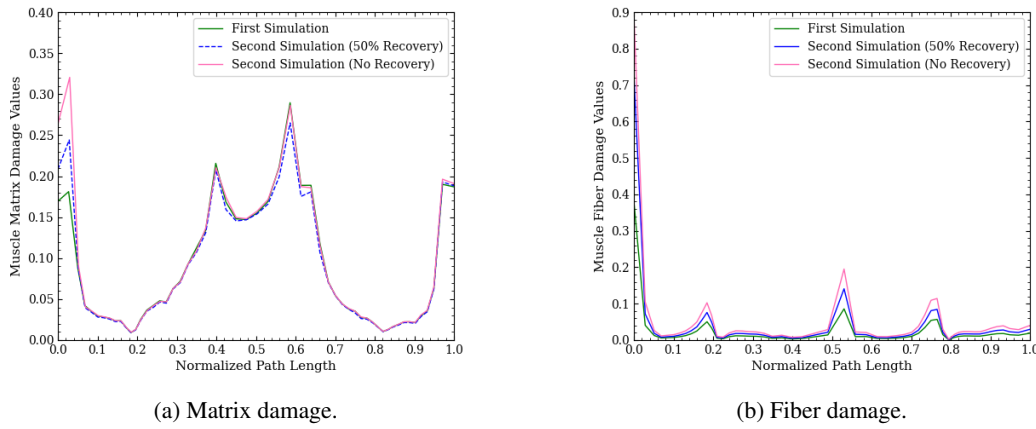


Figure 8.11: Damage values for the matrix and muscle fibers, measure on the normalized path, at the moment of maximum stretch.

of them. However, as can be seen more clearly on the right side of the muscle, the damage values are the highest for the second simulation without recovery and the lowest for the first simulation. This phenomenon could and should be due to the damage imposed by the fibers in the second simulations. However, higher values appear in some situations in the first simulation.

As far as the damage to the muscle fibers is concerned (Figure 8.11b), contrary to the simulation on the plate, the muscle fibers damage is not that different between simulations, although it is more pronounced in the fixed zone. Thus, there is a very prominent peak in the damage values on the right side of the path defined in the muscles, reaching values of 0.38 for the first simulation, 0.73 for the second simulation with 50% recovery, and 0.89 for the second simulation without recovery. This point corresponds to the region near the right lateral origin of the pubovisceral muscle. This area has been consistently proven as the most affected region of the LAM during childbirth in previous studies (Vila Pouca et al., 2022b; Jing et al., 2012; Vila Pouca et al., 2018). In addition, a peak also occurred around the 0.5 value of the path, which corresponds to the insertion point of the rectal area of the LAM in the coccyx, where there is also an accumulation of fiber damage. When comparing the three simulations, once again, the second simulation without recovery shows the highest values of fiber damage, while the first simulation shows the lowest values. This is as expected since, in the second simulations, the values of the state variables corresponding to fiber damage are initialized with non-zero values (values/half values resulting from the first simulation). So, the damage values are accumulated throughout each simulation, resulting in lower values for the first simulation, intermediate values for the second simulation in which the muscles recover, and higher values for the second simulation in which the muscles do not recover.

Consequently, it was also possible to obtain the distribution of matrix and fibers damage in the PFM at the moment of maximum stretch. This analysis can be done by interpreting Figure 8.12. In this way, it is possible to observe a more visual representation of the damage values. Before analyzing the images, it is essential to mention that, in Figure 8.12, muscle fibers damage values

greater than 1 can be seen. These values relate to how Abaqus® interpolates the data, yet these damage values are always less than or equal to 1 at the centroid level.

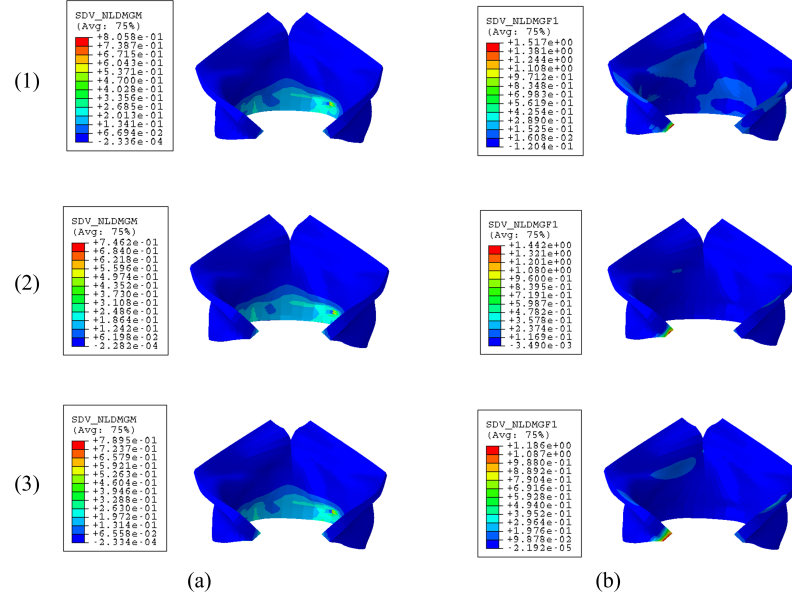


Figure 8.12: Distribution of the (a) matrix and (b) muscle fibers damage values on the PFM for the (1) first simulation, (2) second simulation with 50% fibers damage recovery, and (3) second simulation without fibers damage recovery, at the instant of maximum stretch.

As previously concluded, the region near the right lateral origin of the pubovisceral muscle shows higher values of accumulated damage in the fibers for the three simulations. However, the matrix damage seems to be greater around the posterior area of the LAM in the coccyx. Both these values are lower for the first simulation, intermediate for the second simulation with recovery, and higher for the second simulation without recovery.

In Figure 8.13, it is depicted a quantitative view of the distribution of muscle fiber damage for the course of the first simulation. The boxplot includes the muscle fiber damage values predicted for every element of the pelvic floor muscles (centroid level), grouping the values by the categories defined in Section 7.2.5 (critical, high, noteworthy, moderate, and low).

Throughout the simulation, it is only from step 7 onwards that there are elements with moderate damage; until then, they all had low damage. As can be seen, by the end of the first simulation, only low, moderate, and noteworthy damages are present, with the element with the highest damage having a value of $\widehat{D}_{f1} = 0.449$.

It is also possible to compare the damage of the matrix and the accumulated damage of the fibers at the end of the first simulation. Therefore, Figure 8.14 shows the distribution of nonlocal damage to the matrix and muscle fibers at the end of the last step of the first vaginal birth simulation. Thus, as far as matrix damage is concerned, the elements only have low and moderate damage, with the highest value obtained being 0.22, while there is one element in which fiber damage is considered noteworthy, as mentioned earlier. Besides, it is plausible to conclude that

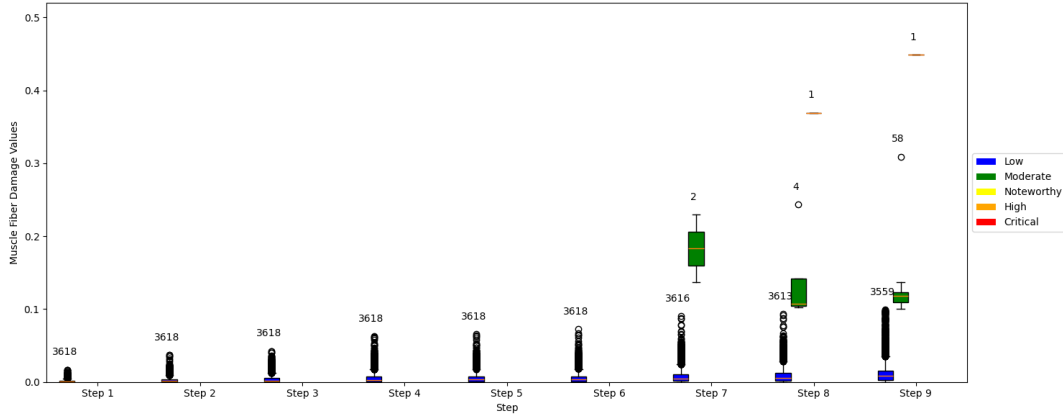


Figure 8.13: Boxplot illustrating the muscle fiber damage values for the PFM elements, throughout the course of the first simulation, for each category (critical, high, noteworthy, moderate, and low).

the number of elements with low damage is higher for the damage accumulated in the fibers. In comparison, the number of moderate damage elements is greater for the matrix.

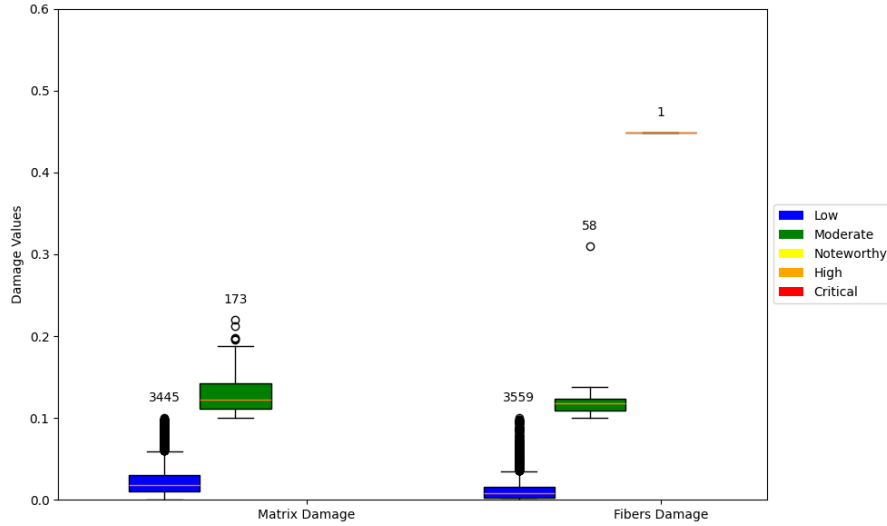


Figure 8.14: Boxplot illustrating the damage values for the matrix and muscle fibers for the PFM elements, at the end of the first simulation, for each category (critical, high, noteworthy, moderate, and low).

The boxplot in Figure 8.15 allows one to directly compare the muscle fiber damage at the end of each of the three simulations performed.

While the first simulation only has one noteworthy damage element ($\widehat{D}_{f1} = 0.449$), the second simulation with recovery presents 2 elements with high damage ($\widehat{D}_{f1} = 0.552$ and $\widehat{D}_{f1} = 0.755$), and the second simulation with no recovery shows one high damage element ($\widehat{D}_{f1} = 0.745$) and one critical damage element ($\widehat{D}_{f1} = 0.903$). In addition, the number of elements with low damage decreases from the first simulation to the second with recovery and, consequently, to the second without recovery, as elements from other categories appear.

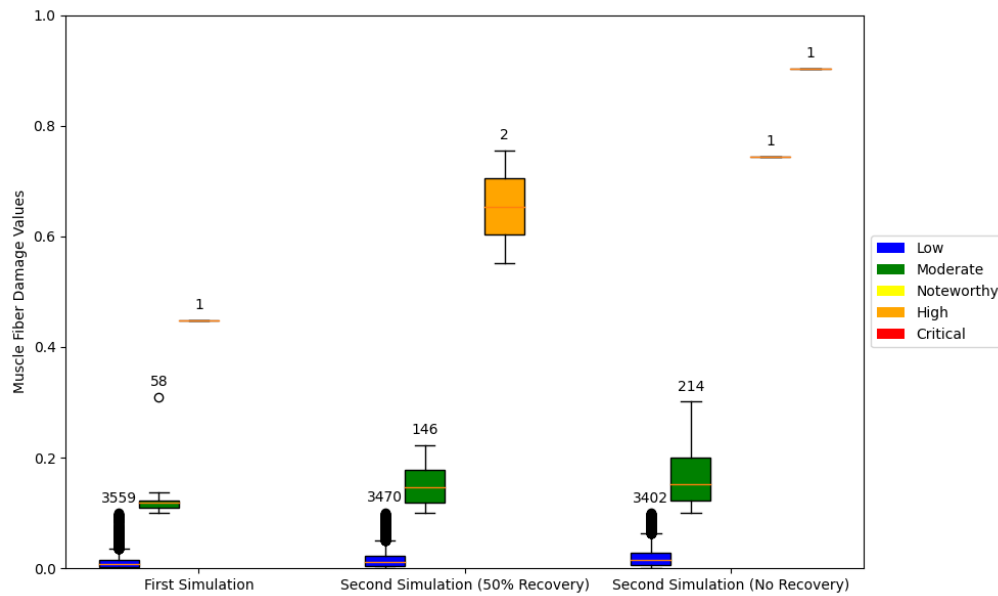


Figure 8.15: Boxplot illustrating the muscle fibers damage values for the PFM elements, for each category (critical, high, noteworthy, moderate, and low), at the end of each simulation.

In conclusion, the second simulation, in which there is no recovery, has more critical, high, and moderate areas than the other two simulations, indicating that it is the case that is predicted to cause the highest amount of damage accumulation, which may result in a partial or complete structural failure. Moreover, it is followed by the second simulation with 50% muscle fiber recovery. This may indicate that the closer in time the childbirths, and/or the shorter the recovery of the woman's PFM, the greater the likelihood of injuries.

It was also decided to study the evolution of the muscle fibers damage at the critical location, throughout the simulations. For that, the fibers damage at the end of each of the nine steps of each simulation were plotted. It is worth mentioning that the critical element is the point where the maximum muscle fibers damage was recorded, and this was the same for the three simulations. That being said, element 729 is highlighted in Figure 8.16.

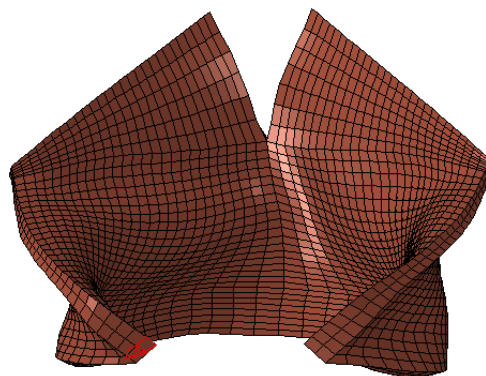


Figure 8.16: Location of the critical element on the PFM, highlighted in red.

Therefore, Figure 8.17a shows the fatigue damage evolution at the critical element for the nine steps of each vaginal birth simulation. It is quite noticeable that the damage values are greater for the second simulation with no recovery, whereas the smallest values are obtained for the first simulation. Although, the first simulation showed a greater damage evolution from step 6 until the end of the simulation, despite the smaller values achieved. These findings can be explained by the fact that there was no prior accumulated damage in the muscle fibers, so the tissue has more resistance and accumulates more damage.

For easier comparison, Figure 8.17b shows the critical element maximum value reached at the end of each simulation.

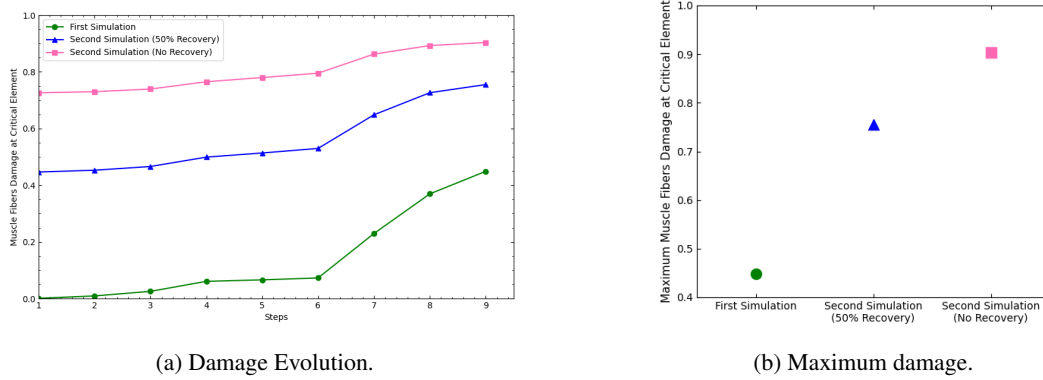


Figure 8.17: Fiber damage at the critical element for the 3 performed simulations.

From all three simulations, only the second simulation without recovery brought the tissue to a critical threshold, with $\widehat{D}_{f1} = 0.903$. This event occurred at the same site where literature reports the highest damage and frequent tears (Vila Pouca et al., 2022b; Jing et al., 2012; Vila Pouca et al., 2018; Dietz et al., 2007; Margulies et al., 2007; Miller et al., 2015; Krofta et al., 2009). Consequently, the critical risk location remained consistent across the simulations, situated near the right lateral origin of the pubovisceral muscle.

In summary, the results showed an increase in the number of damaged elements in the second simulations, as expected.

8.2.1 Simulation of a Second Vaginal Birth with a Smaller Fetal Head

As previously described in Section 7.2.4.4, it was decided to evaluate how a smaller fetal head would influence the stretch, stress, and damage results in a second delivery. Therefore, a 5th percentile fetus head was used in the second simulations of this stage in order to compare with the simulations previously carried out with a 50th percentile fetus head. So, a second simulation was performed assuming a recovery of 50% of the accumulated damage in the fibers, and another second simulation where fiber damage recovery was not considered was also executed. It is essential to mention that the geometry of the PFM remained the same.

Accordingly, in this Section the simulations carried out with a 50th percentile fetal head (previously analyzed) will be commonly referred to as P50, while the simulations performed with a 5th percentile fetal head will have the prefix P5.

- **Stretch:**

Similar to the previous simulations, the PFM stretch values, measured on the path, were also obtained for the simulations with the smallest head. Previous studies have shown a direct correlation between tissue stretch ratios and the size of the fetal head. Increasing the diameter of the fetal head by 9% has been demonstrated to correspondingly increase the stretch of the medial pubovisceral region by the same percentage ([Ashton-Miller and DeLancey, 2009](#)).

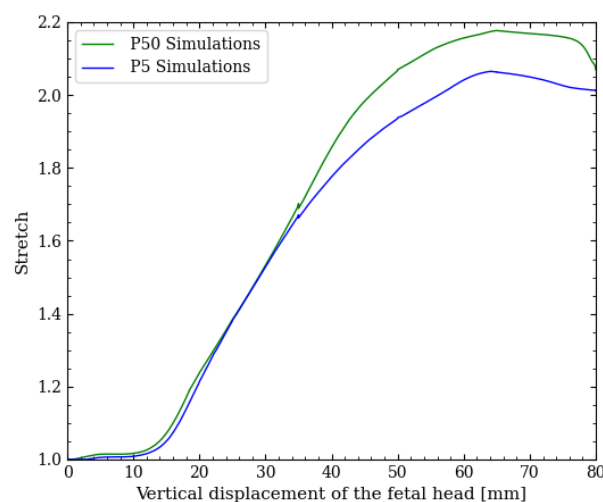


Figure 8.18: Stretch values obtained for the PFM on the defined path, during the vertical displacement of the fetal head, for the simulations with a 50th percentile fetal head and 5th percentile fetal head.

Therefore, a comparison between the PFM stretch resulting from an interaction with a larger head and a smaller one is depicted in Figure 8.18. Analyzing the results, it is possible to infer that the difference between the curves becomes evident from the 35 mm vertical displacement of the fetal head. As Section 8.2 mentioned, the maximum stretch for the P50 simulations was 2.18. Indeed, for the simulations with a 5th percentile fetal head, the maximum stretch value of the muscles was lower, assuming the value of 2.07, for a descent of 64.2 mm. Thus, the prediction that PFM stretch would be higher in a simulation with a 50th percentile fetal head compared to a simulation with a 5th percentile fetal head was confirmed. Furthermore, this maximum stretch was verified for a slightly lower vertical displacement for the smaller head.

It is also relevant to note that the stretch values were similar for both simulations with the same fetal head because its geometry and degree of flexion were consistent in the two simulations.

Hence, the results corroborate the premise that the larger the fetal head, the greater the stretching of the muscles that interact with it.

- **Maximum principal stresses:**

In general, regarding the maximum principal stresses, they are expected to decrease with the amount of accumulated damage since this decreases the ability of the pelvic floor to resist the descent of the fetal head. For this reason, it is anticipated that the simulations of a first delivery will show higher values of principal stresses than the simulations with accumulated damage to the muscle fibers. It is also presumed that in simulations with 50% muscle fiber recovery, the principal stresses will be higher than in simulations without muscle fiber recovery.

The literature describes that a larger fetal head diameter, with constant PFM dimensions, results in higher stresses in the PFM (Parente et al., 2009b). Therefore, it is presupposed that, for the same recovery and damage accumulation conditions, the maximum principal stresses experienced in the simulations with a 50th percentile fetal head will be higher than those with a 5th percentile fetal head.

Therefore, the results of the maximum principal stresses as a function of the normalized path, for the maximum stretch moment, were obtained, as previously. Consequently, the simulation of a first delivery with a 50th percentile fetal head was compared to the two simulations of a second delivery with a 5th percentile fetal head and accumulated damage. Figure 8.19 shows the curves for the three simulations mentioned.

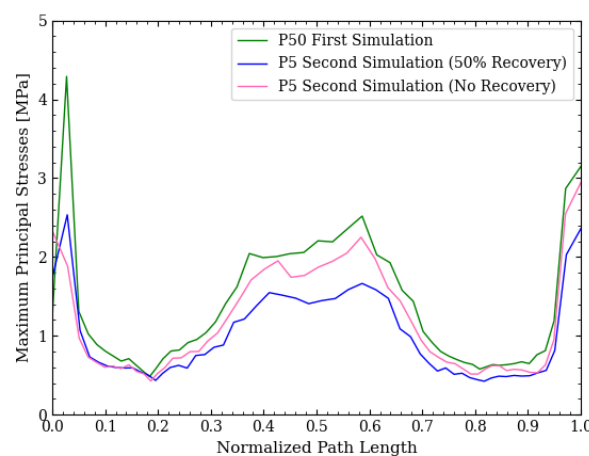


Figure 8.19: Maximum principal stresses in MPa measured along the normalized path, at the instant of maximum stretch, for the first simulation with a 50th percentile fetal head, and second simulations with 5th percentile fetal head.

As expected for the first simulation, the stress values are generally higher than in the second simulations. This is due not only to the accumulation of damage in the second simulations, but also to the decrease in fetal head diameters, which translates into lower head tension in the PFM. Furthermore, in the second simulation in which there is 50% recovery of the fibers damage, it was assumed that the stresses would be higher than in the simulation in which there is no recovery of the fibers damage. However, this phenomenon only occurs in a region near the right end of the path, i.e., at the right insertion point of the rectal area of the LAM in the symphysis pubis.

Upon analysis, the highest stresses values were recorded for the PFM for the first simulation with 50th percentile head, the second simulation with 50% recovery and 5th percentile head, and the second simulation without recovery with 5th percentile head were 4.29 MPa, 2.53 MPa, and 2.94 MPa, respectively. The stress values at the insertion locations of the LAM in the symphysis pubis ranged from 1.23 to 2.33 MPa on the right side of the muscle and from 2.36 to 3.14 MPa on the left side.

Still along the same lines, it became appropriate to compare the simulations with the same accumulation of damage but different fetal heads. In other words, the simulations with 50% recovery of the muscle fibers for the two heads were compared (Figure 8.20a), and the same procedure was carried out for the simulations without recovery of muscle fibers damage (Figure 8.20b).

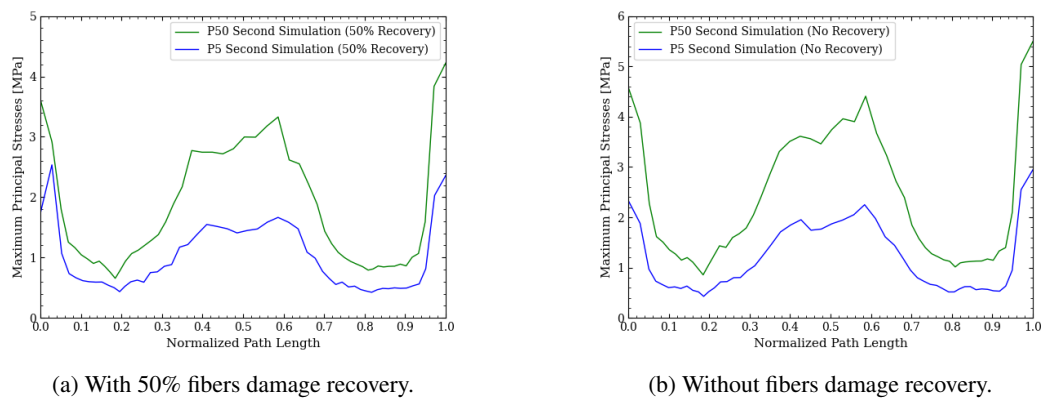


Figure 8.20: Maximum principal stresses in MPa measured along the normalized path, at the instant of maximum stretch, for the second simulations with a 50th percentile fetal head, and with 5th percentile fetal head.

The simulations with the 50th percentile fetal head show higher principal stress values, both in the case where there is 50% recovery of the muscle fibers and in the case where there is no recovery. These results are in line with expectations. Therefore, for the simulations that consider the percentage of recovery, the highest stresses recorded for the PFM for the second simulation with 50th percentile fetal head, and second simulation with 5th percentile fetal head were 4.22 MPa, and 2.53 MPa, respectively. In turn, for the second simulations without recovery, the values obtained using a 50th percentile fetal head and 5th percentile fetal head were 5.49 and 2.94 MPa, in that order.

In general, the plots of the maximum principal stresses of all the simulations evolve in a similar way. More specifically, they show maximum values next to 0.0 and 1.0 of the path (near the pubic symphysis), decreasing as they move away from the extremities and increasing again from 0.2 and 0.8, with a peak in the area near 0.6. In other words, the graphs show some symmetry characteristics (in relation to a vertical axis $x = 0.5$, which corresponds to the location of the coccyx). Thus, it is possible to conclude that, once again, the highest stress values in this study were found at the insertion points of the rectal area of the LAM in the coccyx, as well as in the

symphysis pubis, also for the simulations with a smaller head.

In conclusion, the larger the diameter of the fetal head, the higher the stress values obtained in the PFM, which can lead to more complications during delivery. Thus, larger diameters may correspond to a higher risk of PFM injuries during normal childbirth, leading to a greater predisposition to the development of PFD.

- **Damage:**

Lastly, it was decided to evaluate the muscle fibers damage at the end of each simulation, displaying the data in a boxplot. Therefore, through Figure 8.21, it is possible to directly compare the damage from the first simulation with a 50th percentile head, second simulations with a 50th percentile head, and for the second simulations with a 5th percentile head. It is worth noting that the numbers within the graph area, positioned above each box, represent the number of elements in each damage category.

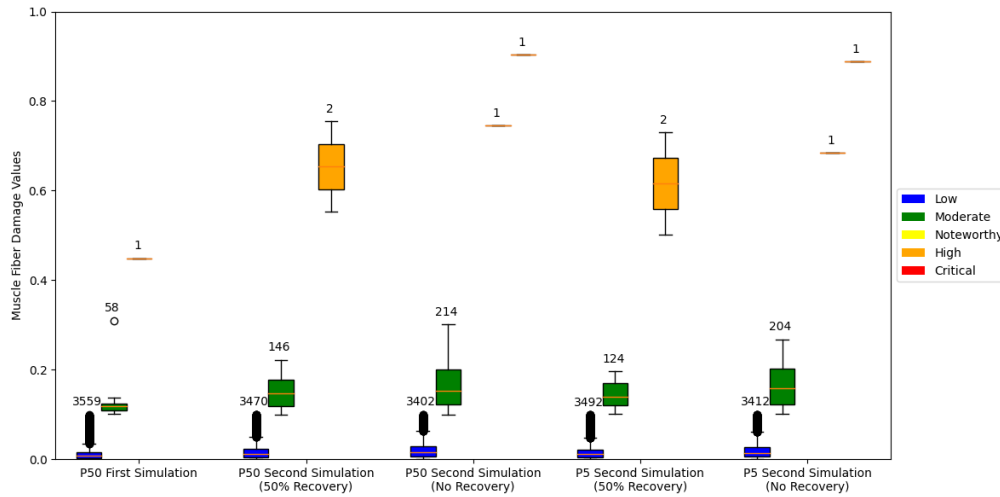


Figure 8.21: Boxplot illustrating the muscle fibers damage values for the PFM elements, for each category (critical, high, noteworthy, moderate, and low), at the end of each simulation, for all the simulations with a 50th percentile fetal head, and the second simulations with 5th percentile fetal head.

As already established, the first simulation with the 50th percentile fetal head only has one noteworthy damage element ($\widehat{D}_{f1} = 0.449$).

When comparing the second simulations performed with the different head sizes, it is visible that for the simulations in which there is 50% fibers damage recovery, both simulations show two elements with high damage, by the end. However, for the smaller head, the values are lower ($\widehat{D}_{f1} = 0.501$ and $\widehat{D}_{f1} = 0.730$), as opposed to the values for the larger head ($\widehat{D}_{f1} = 0.552$ and $\widehat{D}_{f1} = 0.755$). In addition, there are more elements with low damage for the 5th percentile fetus head.

In turn, considering the simulations in which there is no recovery of fibers damage, the simulations for both heads show one high and one critical element. Once again, the values of each

damage category are higher for the simulation using the 50th percentile fetal head. Thus, these values are $\widehat{D}_{f1} = 0.684$ and $\widehat{D}_{f1} = 0.888$ for the smaller head, and $\widehat{D}_{f1} = 0.745$ and $\widehat{D}_{f1} = 0.903$ for the larger head. The number of elements with low damage is also higher for the second simulation with the 5th percentile fetal head.

In conclusion, the second simulations, in which there is no recovery, has more critical, high, and moderate areas than the other three simulations, indicating that it is the case that is predicted to cause the highest amount of damage accumulation. Specifically, the damage values become even more notable when the simulation is performed with a bigger head.

In the literature, it has been stated that the size of the fetal head is directly related to the imposed strain, so larger fetal heads have the potential to accumulate damage at a faster rate ([Vila Pouca et al., 2022b](#)). The results obtained here align with this statement because, in the simulations with the larger head, there are more elements with moderate damage values. In comparison, in the simulations with the smaller head, there are more elements with low damage. Furthermore, for higher damages, such as high and critical, the values obtained are more elevated for the simulations with the 50th percentile fetal head.

Chapter 9

Conclusions and Future Work

9.1 Final Remarks

Vaginal delivery is a natural process that brings about major physiological changes in the mother, making it a highly complex issue with numerous variables affecting the outcomes. It is also acknowledged as a serious risk factor for the occurrence of PFD, emphasizing the need for a deeper understanding of specific mechanisms. However, experimental research in this area is limited due to clinical, technical, and ethical constraints. As a result, computational models have become valuable tools for gaining mechanical insights into childbirth. Studying the impact of vaginal delivery on the PFM can lead to the development of preventive and therapeutic strategies to minimize common injuries. This approach could enhance the understanding of how vaginal childbirth affects the pelvic floor, ultimately improving the ability to define and implement effective treatments and conduct targeted clinical research.

The main objective of the current work is to develop computer simulations of the vaginal childbirth process, and to investigate the impact of the accumulated damage on the female pelvic floor, for a second birth. This premise arises from the need to address the lack of information regarding subsequent births, as most biomechanical studies tend to consider only nulliparous women to simulate the childbirth process.

To achieve this goal, a simplified finite element model of the second stage of labor during vaginal delivery was employed to simulate childbirth with the fetus in a vertex presentation and OA position. This model included three-dimensional geometric representations of the PFM and the fetal head, along with a UMAT subroutine to more accurately reproduce PFM behavior. The constitutive material model used was a quasi-incompressible, transversely isotropic visco-hyperelastic-damage model. Three birth simulations were conducted: a first simulation, a second simulation with 50% recovery of accumulated muscle fibers damage, and a second simulation with no muscle fibers damage recovery. It is important to note that for the second births, full recovery of the matrix damage was assumed. Initially, the second simulations were carried out with a 50th percentile fetal head, and, after, simulations with a 5th percentile fetal head were also performed to assess the impact of head size on the results. Although attempts were made to conduct simulations with a 95th

percentile fetal head for more comprehensive results, convergence issues due to the complexity of the numerical simulations did not allow this. The simulations were performed using Abaqus® software (version 2022) with an implicit solver (Abaqus®/Standard).

Considering the results for the model with the 50th percentile fetal head, the maximum stretch obtained for the PFM, in the present study, was 2.18, at a vertical descent of the fetal head of 65.0 mm. This value falls within previously published muscle stretch ratio ranges. The measurements for the maximum principal stresses are higher at the insertion points of the rectal area of the LAM in the symphysis pubis for the first and second birth (with and without muscle fibers damage recovery) simulations, as expected. However, the simulations for the second births show greater maximum principal stresses values, for the moment of maximum stretch, which is the opposite of what is anticipated, since the muscles would already be damaged.

The accumulated damage in the matrix shows peaks at the insertion points of the rectal area of the LAM in the coccyx and the symphysis pubis (right and left insertion points). Its values are similar for the three simulations carried out since the matrix damage is always initialized at 0. However, on the right side of the pubovisceral muscle, the damage values are the highest for the second simulation without recovery, and the lowest for the first simulation because of the damage imposed by the fibers. The region near the right lateral origin of the pubovisceral muscle has been consistently proven as the most affected region of the LAM during childbirth in previous studies. Accordingly, in the current investigation, the highest values for the accumulated damage in the muscle fibers were verified in that zone, with the more worthy of attention damage being obtained for the second simulation without fibers damage recovery, the intermediate value for the second simulation with 50% fibers damage recovery, and the lowest value for the first simulation. By comparing the muscle fibers damage values at the end of each simulation, it is possible to conclude that the first simulation only has one noteworthy damage element, the second simulation with recovery presents two elements with high damage, and the second simulation with no recovery shows one high damage element and one critical damage element. So, the second simulation, in which there is no recovery, has more critical, high, and moderate areas than the other two simulations, indicating that it is the case that is predicted to cause the highest amount of damage accumulation, which may result in a partial or complete structural failure. This suggests that the closer the births are in time, and/or the shorter the recovery of the PFM of the woman, the greater the likelihood of injuries. Moreover, the first simulation showed the biggest damage evolution throughout the simulation despite the smaller values achieved. Since there was no prior accumulated damage, the tissue had more resistance and accumulated more damage.

Regarding the second part of the hypothesis, where the simulations were carried out with the smaller head, the stretch achieved is lower than that of the simulations with the larger fetal head, corroborating the literature data. The maximum principal stresses and damage values are also smaller for the simulations with the 5th percentile fetal head since it has been stated that the size of the fetal head is directly related to the imposed strain, so smaller fetal heads accumulate damage at a slower rate.

In conclusion, the critical risk location remains consistent across the simulations, situated near

the right lateral origin of the pubovisceral muscle; the number of damaged elements increases in the second simulations, being higher for lower recovery rates, which is in line with the literature since it indicates that subsequent childbirths are correlated with a higher risk of developing PFD; and a smaller head results in less muscle stretch, stresses, and accumulated damage.

Given that this study focuses on a novel topic, it has both strengths and limitations. One of the key strengths is the use of a well-characterized material behavior for the PFM, which incorporates viscohyperelasticity, muscle fiber orientations, and the capacity to accumulate damage under repetitive submaximal loading - making it one of the first studies to do so ([Vila Pouca et al., 2022b](#)). Additionally, although simplified, the geometric models were derived from human data. Besides, the biomechanical model consistently identified the weakest region of the pelvic floor during labor, aligning with the regions where tears are commonly reported in the literature.

The problem studied here is highly complex, and any methodology used will inevitably have limitations and be subject to criticism. Therefore, it is essential to interpret the findings critically ([Parente et al., 2008](#)). Towards the end of the pregnancy, the pelvic floor undergoes many alterations to facilitate birth, potentially reducing the maximal stretches. However, these modifications were not addressed in this study. Furthermore, numerous variables, namely variations in maternal pelvic shape, fetal head shape, the degree of molding during delivery, symphyseal diastasis, types of episiotomies, and presenting orientation, can affect the maximum muscle stretch ratios and, consequently the final results ([Parente et al., 2008](#)). None of these factors were taken into account in this study. Also, the material parameters utilized to define pelvic floor tissue behavior were derived from experimental data on non-pregnant ovine pelvic floor muscles. This approach was chosen due to the lack of human experimental data on pelvic floor low-cycle damage accumulation. Consequently, the results should be interpreted cautiously as there might be an overestimation of stress levels ([Vila Pouca et al., 2022b](#)). Additionally, the perineal body was not included in the finite element model despite evidence suggesting its protective role, which could potentially reduce damage accumulation in the LAM. In all the simulations, the fetus was considered a rigid structure to facilitate convergence, and a more accurate characterization of the fetus would influence the stress and strain values measured in the LAM ([Silva et al., 2015](#)). In addition, it was assumed that the muscles only respond passively due to the effects of epidural anesthesia. Without anesthesia, the active response of the muscles must be considered ([Oliveira et al., 2016a](#)). Finally, minor limitations include the lack of differentiation between muscle fibers and collagenous fibers in bearing the load through the ground substance of the pelvic floor muscle material ([Vila Pouca et al., 2022b](#)).

Despite these limitations, this study complements previous work by considering a novel parameter and represents a step forward in achieving more accurate results.

In the future, an improved and more realistic version of this model could become very useful for healthcare professionals. The results of this study are very relevant for clinical practice, specifically in the treatment and prevention of PFD related to childbirth. Understanding the biomechanical impact of successive deliveries on pelvic floor musculature might influence clinical decisions on the timing and style of delivery, especially for women with a history of pelvic floor injuries.

With this knowledge, healthcare providers can advise patients about the potential risks associated with multiple vaginal deliveries and in designing personalized birth plans that minimize pelvic floor damage. Thus, developing precise numerical models of the female pelvic cavity will provide tools to help clinicians plan the optimum labor approach and implement it in a more controlled and reliable way, reducing delivery complications. Additionally, the findings can be applied to establish targeted rehabilitation protocols to enhance muscle recovery postpartum, hence minimizing the chance of PFD. Ultimately, these insights contribute to improving maternal health outcomes by promoting informed clinical practices and preventative strategies.

Finally, it is essential to mention how important the research on reproductive health is, namely the biomechanics of birthing, concerning women's rights. Women have the right to accessible, informed, and respectful maternal care that not only values their well-being throughout childbirth but also considers and accepts the various physiological experiences connected with it. Ensuring women's rights entails promoting fair healthcare practices, empowering women with information about their bodies, and creating a supportive environment that values and supports informed decisions around birth. Essentially, acknowledging the biomechanical consequences of vaginal birth acts as a driving force for the progress of women's healthcare.

9.2 Future Work

Though the findings of this work are encouraging, more simulations and optimization will be required. Some changes should be conducted in order for this study to become more elaborate and realistic, making it useful for health professionals, particularly for understanding the differences between a first and second vaginal birth.

The constitutive parameters used in this model were the same as those employed by [Vila Pouca et al. \(2022b\)](#) in their investigation, albeit with different geometry. Thus, testing and adapting these parameters would be advantageous. Furthermore, since these constitutive parameters were derived from non-pregnant ovine pelvic floor muscles, using data from damage accumulation in human pelvic floor tissue in pregnant individuals would likely yield more accurate results.

Moreover, in the future, it would be beneficial to start the model with a pelvic floor geometry measured from a pregnant woman, including representations of both the genital and levator hiatuses. The influence of different pelvis types is another valuable modification to consider, as the normalization of a gynecoid shape model is rooted in white-centric female anatomy. Also, due to numerical complications, simulating the second birth processes with a 95th percentile fetal head was not possible. However, studying how a larger fetal head would impact damage accumulation, particularly in a second birth, would be advantageous.

As mentioned previously, this study treated the fetal head as a rigid structure. However, it would be helpful to characterize the mechanical properties of the fetus and incorporate them into the simulation. This hypothesis would enable the investigation of phenomena such as fetal head molding.

Studying the damage accumulation in various fetal positions, such as OP, could provide valuable insights into how the positioning of the fetus influences resulting damage, either for the first or second birth.

Considering that epidural effects are assumed to induce only a passive muscle response, it would be valuable to also incorporate the active response responsible for contractions, as suggested by [Parente et al. \(2010b\)](#).

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Appendix A

A.1 Paper submitted to CMBBE 2024: The 19th International Symposium on Computer Methods in Biomechanics and Biomedical Engineering

EXPLORING THE BIOMECHANICAL CONSEQUENCES OF SUBSEQUENT CHILDBIRTH ON THE FEMALE PELVIC FLOOR

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1. Introduction

Throughout their lives, more than one-third of women will face pelvic floor dysfunction (PFD), a phenomenon extensively investigated in recent years, particularly regarding its correlation with childbirth and pregnancy [1]. Modelling the pelvic cavity enables the thorough examination of PFD, facilitating the analysis of the stresses and strains exerted on these structures during events such as childbirth [2]. Among the array of software available for such simulations, Abaqus stands out as one of the most utilized. Its scripting capabilities allow for efficient parameterization of models, enabling simulations under various conditions with speed and practicality.

This paper aims to delve into the biomechanical repercussions of initial childbirth on subsequent occurrences of pelvic floor lesions in females.

2. Materials and Methods

In Abaqus, an anatomically accurate model of the foetal head and pelvic floor muscles (PFM) was meticulously crafted using medical imaging data as a foundation. These models were subsequently parameterized to facilitate the generation of foetal heads across various percentiles and PFM with distinct levator hiatus dimensions.

The PFM were characterized using a viscohyperelastic transversely isotropic constitutive model incorporating continuum damage considerations [3]. Conversely, the foetal head was modelled using an elastic constitutive model with high stiffness parameters to emulate rigid body behaviour.

A childbirth simulation was conducted with the foetus in the vertex presentation and occipito-anterior position. Regarding boundary conditions, the nodes on the extremities of the PFM were fixed.

3. Results

Beyond variances in the percentages of damage observed after each simulation, there were also slight disparities in the values of maximum principal tension and reaction forces, both exhibiting a decrease.

These diminished tension levels and reaction forces observed in the subsequent simulation may suggest a shorter duration of labour during a second vaginal delivery, a phenomenon commonly observed in multiparous women. Typically, multiparous women experience a shorter second stage of labour compared to nulliparous women.

4. Discussion and Conclusions

PFD represent a significant public health concern and have garnered extensive attention, especially concerning the repercussions of initial vaginal childbirth. Computational modelling allows a better understanding of the influence of obstetric factors on the risk of PFM injury.

5. References

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