THESIS FOR THE DEGREE OF DOCTOR OF PHILOSOPHY

Advancing Pelvis Computational Models for Automotive Safety Assessment

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Cover figure: Finite element human body model, subjected to a frontal impact in a simplified vehicle environment, at the point when the lap belt loses contact with the pelvis and pushes into the abdomen, *i.e.*, submarining.

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To my family

Advancing Pelvis Computational Models for Automotive Safety Assessment
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Abstract

The pelvis is a key load bearer in vehicle safety due to its relatively high load tolerance and shape, which is utilized to control occupant kinematics in accidents by engaging with vehicle restraint systems. However, epidemiological studies have shown that the pelvis is also a highly exposed structure, as pelvic fractures are common outcomes due to interaction with the vehicle interior and restraint systems during a crash. Furthermore, fracture risk is not equally distributed over the population and vulnerable sub-populations have been identified depending on the load scenario. In addition, future autonomous vehicles are expected to allow for a more relaxed occupant posture by reclining the seatback, which increases the risk, in frontal impacts, of the pelvis sliding under the lap belt, *i.e.* submarining. Together, this motivates a deeper understanding of the potential of the pelvis as a load bearing structure, as well as its interaction with the vehicle restraint systems across the entire population, in various crash scenarios.

While vehicle manufacturers try to minimize variability in product development, human individual variability is an intrinsic property that must be considered to capture the vulnerable population and maximize the efficiency of vehicle safety systems. Finite element human body models (FE-HBMs) are the most advanced tool available to use in the virtual design of restraint systems and they provide the opportunity to include both geometrical and material variability through population based models and assessments.

In this thesis, methods enabling inclusion of population variance in FE-HBMs were implemented for the pelvis. Key findings include that sex, age, stature, and Body Mass Index (BMI), only cover a limited part of the population variance in pelvic shape, which is relevant for state-of-the-art FE-HBM development, population based simulation studies, and post-mortem human subject (PMHS) experiments. In addition, pelvic shape was shown to be an influential factor for both pelvis response in side impacts and belt-to-pelvis interaction in frontal impacts, which warrants consideration in future safety assessments.

To conclude, this thesis advances the field of pelvis computational models for automotive safety assessment and enables a population based evaluation for future vehicle safety systems, which can result in more robust systems, reducing the risk of injuries in real-life accidents.

Keywords: Human Body Model; Passive Safety; Pelvis; Population Variance; Reclined; Restraint Systems; SAFER HBM; Submarining; Vehicle Safety

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List of Appended Papers

This thesis is based on the work contained in the following appended papers, referred to by Roman numerals in the text:

Paper I **Brynskog E.**, Iraeus J., Reed M., Davidsson J. (2021).
”Predicting Pelvis Geometry Using a Morphometric Model with Overall Anthropometric Variables”.
In: *Journal of Biomechanics, Volume 126*.
DOI: 10.1016/j.jbiomech.2021.110633

Paper II **Brynskog E.**, Iraeus J., Pipkorn B., Davidsson J. (2022).
”Population Variance in Pelvis Response to Lateral Impacts – A Global Sensitivity Analysis”.
In: *Proceedings of the IRCOBI Conference*, Porto, Portugal.

Paper III **Brynskog E.**, Iraeus J., Pipkorn B., Davidsson J. (2024).
”Simulating Pelvis Kinematics from Belt and Seat Loading in Frontal Car Crash Scenarios: Important Boundary Conditions that Influence the Outcome”.
In: *Annals of Biomedical Engineering, Volume 53*.
DOI: 10.1007/s10439-024-03631-9

Paper IV **Brynskog E.**, Östh J., Larsson K.-J., Iraeus J. (2025).
“Effect of Occupant and Restraint Variability in Reclined Positions on Submarining Probability in Front Car Crash Scenarios”.
Under review at: *Frontiers in Bioengineering and Biotechnology*.

List of Additional Work

Other relevant presentations and publications, not appended to the thesis:

Brynskog E. (2024).
"Scoping Review on Validation Data for Submarining Prediction in Automotive Crashes."
In: *REPORT 2024:02*, Chalmers University of Technology, Gothenburg, Sweden.

Brynskog E. (2024).
"SAFER HBM Model Development for Enhanced Submarining Prediction."
In: *REPORT 2024:03*, Chalmers University of Technology, Gothenburg, Sweden.

Brynskog E., Iraeus J., Davidsson J. (2024).
"Pelvis Response Sensitivity of Human Body Models in Reclined Postures."
In: *Proceedings of the IRCOBI Conference (Short communication)*, Stockholm, Sweden.

Iraeus J., **Brynskog E.**, John J., Östh J., Pipkorn B., Davidsson J. (2024).
"Development of SAFER HBM v11."
In: *REPORT 2024:06*, Chalmers University of Technology, Gothenburg, Sweden.

Abbreviations

ADAS	Advanced Driving Assistance Systems
AIS	Abbreviated Injury Scale
ASIS	Anterior Superior Iliac Spine
ATD	Anthropometric Test Device
AV	Autonomous Vehicle
BMI	Body Mass Index = weight/(length)^2 [kg/m^2]
CT	Computed Tomography
Euro NCAP	European New Car Assessment Programme
EuroSID	European Side Impact Dummy
FE	Finite Element
FE-HBM	Finite Element Human Body Model
FMVSS	Federal Motor Vehicle Safety Standards
GM	Geometric Morphometrics
GPA	Generalized Procrustes Analysis
GSA	Global Sensitivity Analysis
HBM	Human Body Model
IIHS	Insurance Institute for Highway Safety
IRF	Injury Risk Function
LASSO	Least Absolute Shrinkage and Selection Operator
LHS	Latin-Hypercube Sampling
MC	Monte Carlo
M-DRM	Multiplicative Dimensional Reduction Method
MVC	Motor Vehicle Crash
NFR	Number of Fractured Ribs
NASS-CDS	National Automotive Sampling System – Crashworthiness Data System
NHTSA	National Highway Traffic Safety Administration

PC	Principal Component
PCA	Principal Component Analysis
PMHS	Post-Mortem Human Subject
PS	Pubic Symphysis
PSIS	Posterior Superior Iliac Spine
RBF-TPS	Radial Basis Function with Thin-Plate-Splines
SI	Sacroiliac
SPCA	Sparse Principal Component Analysis
SSM	Statistical Shape Model
THOR	Test device for Human Occupant Restraint
UMTRI	University of Michigan Transportation Research Institute
UN/ECE	Economic Commission for Europe of the United Nations
VTC	Virtual Testing Crashworthiness
WorldSID	World Side Impact Dummy

CONTENTS

Part A

Thesis Summary

1 INTRODUCTION

Historically, the efforts made to increase safety in new vehicles have been largely successful, with reduced risk for both serious and fatal injuries. However, the risk of injury is not evenly distributed within the population, nor between different body regions (Forman et al., 2019; Kullgren et al., 2020). Pelvic fractures, specifically, have been identified as the 3rd most common injury of moderate or above severity (Abbreviated Injury Scale (AIS) 2+) in motor vehicle crashes (MVCs) (Weaver et al., 2013), and the dominating AIS2+ lower extremity injury in side impacts (Pipkorn et al., 2020). Furthermore, pelvic fractures are associated with the highest early mortality rate of patients with orthopedic injuries (Tile et al., 2015). In addition, factors such as sex, age, stature, and Body Mass Index (BMI) correlate with pelvic fracture risk in real-life MVC data (Schiff et al., 2008; Sochor et al., 2003; Stein et al., 2006), indicating that vulnerable sub-populations could be identified. Pelvic fracture tolerance from post-mortem human subject (PMHS) experiments is also known to show significant variability (Bouquet et al., 1998; Cesari & Ramet, 1982; Guillemot et al., 1998; Salzar et al., 2009). Together, this shows that pelvic injuries with serious consequences are common in MVCs, and that population variance can have a substantial effect on the injury outcome.

Common tools used in current vehicle safety assessments are the Anthropometric Test Devices (ATDs). These are mechanical systems built to replicate the human response in vehicle crash scenarios. However, due to their mechanical nature, these models are restricted to a limited set of anthropometric measurements and are designed for a specific loading direction. The most advanced tools available in the design of vehicle safety systems via computer simulations are finite element human body models (FE-HBMs). Compared to ATDs, FE-HBMs accurately represent human anatomy, anthropometry, and physical properties to predict human kinematics, kinetics, and internal strains, *i.e.,* a biofidelic response, to omnidirectional external loading. However, while FE-HBMs provide the opportunity to include both geometrical and material variability in the analysis, at a reasonable cost compared to ATDs, state-of-the-art models of today are, like ATDs, typically defined for specific cohorts of the population, *e.g.*, 50th percentile male/female, 5th percentile female, and 95th percentile male (Gayzik et al., 2011; John et al., 2022; Matsuda et al., 2023; Pipkorn et al., 2021). This approach fails to capture the individual variability which could require consideration to protect vulnerable populations and maximize efficiency of vehicle restraint systems. This is particularly true for the complex geometry of the pelvis where interindividual differences sometimes are more pronounced than even the sex differences

(Standring, 2008). Recent studies have highlighted the need for FE-HBM development to focus on assessment of pelvic fractures (Pipkorn et al., 2020), and to specifically include subject-specific factors, such as pelvic shape and position, when analyzing how occupants interact with restraint systems (Richardson et al., 2020, 2024).

Future advanced driving assistance systems (ADAS) have been predicted to reduce the total number of accidents, while simultaneously increasing the relative frequency of intersection crashes (Östling et al., 2019). However, even in the new crash scenarios, frontal impacts will still cover a substantial part of the remaining accidents. An increase in the ratio of intersection crashes might accentuate the risk of pelvic fractures, given its prevalence in side impacts as the pelvis interacts with the vehicle interior (Pipkorn et al., 2020), while frontal impacts might become more prominent given the increased interest in reclined seating, that allows for a more relaxed occupant posture, in future autonomous vehicles (AVs) (Jorlöv et al., 2017; Östling et al., 2017; Östling & Larsson, 2019). Using computer simulations with FE-HBMs, *e.g.,* (Rawska et al., 2019), reclined seating in frontal impacts has been shown to increase the risk of submarining, an event where the occupant slides under the lap belt, which could induce injurious loads to the abdomen and spine. Tests on PMHSs have shown that submarining in reclined postures can be prevented, however, prevention has only been achieved at the expense of high restraining forces causing multiple pelvic, sacrum, lumbar spine, and ribcage fractures, in most of the PMHSs, *e.g.* (Baudrit et al., 2022; Richardson et al., 2020; UMTRI AVOK-Series, 2020-2023).

In summary, the pelvis is a key load bearing structure in MVCs as it interacts with the vehicle interior and the restraint system. Future ADAS and AVs might accentuate the risk of pelvic fractures due to a higher prevalence of side impacts, and efforts to prevent a submarining outcome for reclined occupants in frontal impacts. Furthermore, previous research has indicated that vulnerable occupant sub-populations in terms of pelvic fracture can be defined, highlighting the need to include population variability. The ambition to reduce all injuries in future vehicles emphasizes the need for a deeper understanding of how to best utilize the pelvis as a load bearing structure, and its interaction with the vehicles restraint system, across the entire population of vehicle occupants.

Prior to this thesis, a pelvis finite element model (FE-model) accounting for population shape variance has not been developed. This motivates further research on pelvis anthropometry, development of pelvis FE-models, and the use of these models in FE-HBMs, to address challenges in traffic safety and provide guidance for future research and automotive safety assessments.

1.1 Research Objectives

The main objective of my Ph.D. is to enable pelvis related automotive safety assessments for a population of female and male vehicle occupants, by introducing advanced methods in FE-HBM development and providing guidance for future research. As part of this objective, a generic FE-model capable of morphing to the population variance in pelvic shape has been developed, validated using published PMHS data, and integrated in a state-of-the-art FE-HBM, the SAFER HBM (Iraeus et al., 2024), for use by academia and the vehicle industry.

To fulfil the main objective, four sub-objectives were defined:

1 Describe the pelvic bone population shape variance and develop a statistical model capable of predicting pelvic shape based on anthropometric variables.

2 Develop and validate a generic morphable FE-model of the human pelvis for assessment of both shape and material variations.

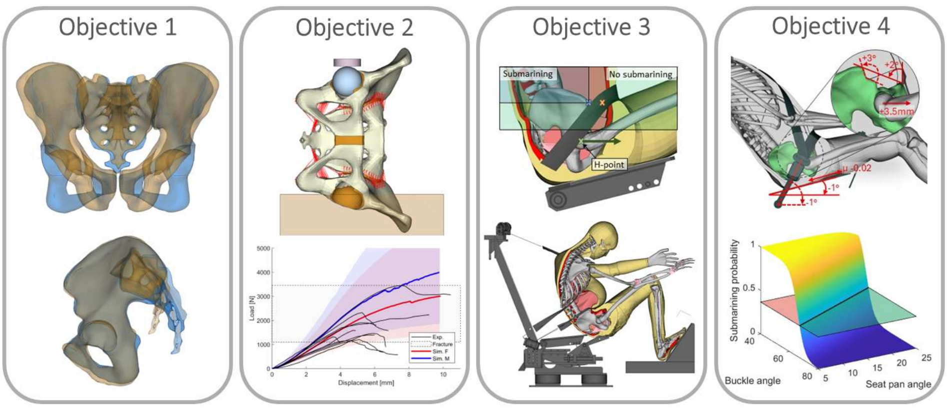
3 Validate a full-body FE-HBM for submarining scenarios, in nominal and reclined seating, and assess the influence of boundary condition variations on pelvis kinematics.

4 Study the pelvis-to-belt interaction in frontal car crash scenarios with a population of reclined occupants subject to variability in restraint design.

1.2 Thesis Overview

To achieve the sub-objectives, four papers were published, see Figure 1 and Summary of Appended Papers.

Paper I addresses Objective 1 by developing a mathematical description of pelvic shape variance. From this, a multivariate linear regression model was generated to answer how much of the pelvic shape variance can be predicted based on sex, age, stature, and BMI. **Paper II** addresses Objective 2 by developing a new pelvis FE-model based on the average geometry found in **Paper I** and integrating it with the mathematical pelvic shape model through mesh morphing techniques. A global sensitivity study, evaluating the effect of shape, material properties, and cortical bone thickness on pelvis response in lateral impacts, was conducted. **Paper III** addresses Objective 3 by presenting the pelvis model integrated with the full-body SAFER HBM. A parameter study, evaluating the effect of varying boundary conditions on pelvis kinematics from frontal car crash scenarios, was conducted. **Paper IV** addresses Objective 4 by running a sensitivity study on a population of reclined occupants in frontal car crash scenarios, including both occupant and restraint variability. A metamodel of submarining outcome based on initial occupant and restraint parameters was developed, to identify important parameters when designing robust submarining countermeasures and a possible vulnerable occupant.



Paper I

Shape Modeling
Linear Regression

Paper II

FE-development
Validation
Lateral Impacts
Sensitivity Analysis

Paper III

Full-body integration
Validation
Frontal Impacts
Parameter Study

Paper IV

Reclined Occupants
Frontal Impacts
Submarining Prediction
Logistic Regression

Figure 1 – Overview of thesis sub-objectives with associated papers and their content.

2 BACKGROUND

2.1 Anatomy and Biomechanics

Pelvis is the Latin word for basin, which can easily be understood by its shape and how it supports the organs of the lower abdomen. The pelvis also acts as a load transferring structure as it connects the upper body, via the sacrum, to the lower body. Its deep, basin-like structure is formed by two innominate bones (also called hip bones or coxal bones), the sacrum and the coccyx, which together are referred to as the bony pelvis, see Figure 2. The bones are connected in a ring-like structure at the anterior pubic symphysis (PS) joint and posteriorly at the two sacroiliac (SI) joints.

Sexually dimorphic measurements have been analyzed by several authors, *e.g.*, (DelPrete, 2019; Luis & Carretero, 1994), and books on anatomy and physiology offer descriptions of the female and male pelvises as; *“strikingly different”* (Marieb & Hoehn, 2010); *“a difference in features of bones are readily apparent”* (Tortora & Derrickson, 2009); and *“the pelvis provides the most marked skeletal differences between male and female”* (Standring, 2008). The differences between females and males are linked to function where the female pelvis has adapted to enable childbirth while the male pelvis, generally, must transfer greater locomotive forces. Hence, a clear distinction between the average male and average female pelvis geometry should be expected. However, since the range of most features overlap between the sexes**,** the inter-individual differences can sometimes be more pronounced than the sex differences (Standring, 2008).

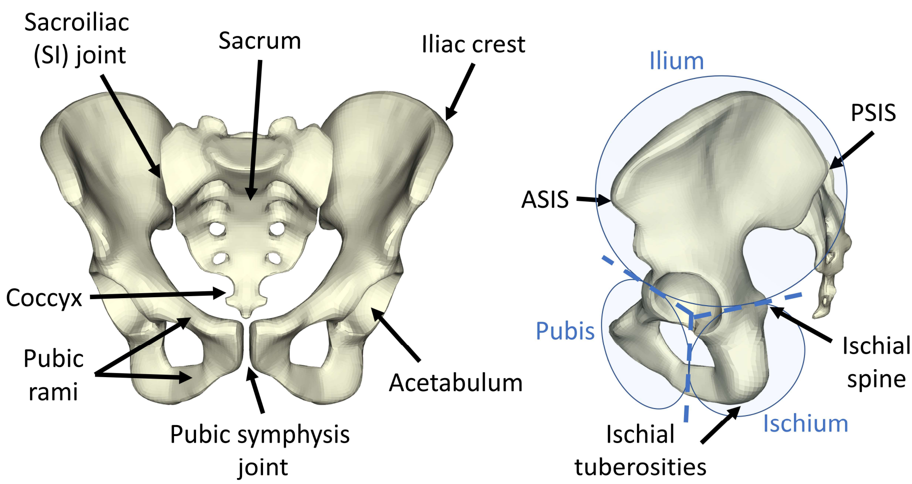


Figure 2 – Anatomical features of the pelvis.

2.1.1 Bones

Each innominate (hip) bone consists of three separate parts: the superior ilium, the inferior-anterior pubis, and the inferior-posterior ischium. The three bones are fused together by the age of 16 at a deep hemispherical socket called the acetabulum (hip socket), giving it the structure of a single bone. The ilium is the largest of the three bones, composed of a body and a large wing-like section called the ala. The thickened superior margin of the ala is called the iliac crest and can easily be felt at the lateral top edge when you rest your hands on your hips. Two distinct landmarks are: the anterior superior iliac spine (ASIS) and posterior superior iliac spine (PSIS). The ischium comprises a superior body that connects to the ilium and the thinner inferior ramus which connects to the pubis anteriorly. Two distinct landmarks are: the ischial spine and the ischial tuberosity. The ischial tuberosity is the strongest part of the innominate bone and when we sit, it holds our entire weight. The pubis is a V-shaped bone comprising a body and two rami, the superior and inferior ramus. The inferior ramus connects to the ischial ramus and the superior ramus connects to the ischium and ilium at the acetabulum. At the junction of the left and right pubic bones in the median plane is the pubic symphysis joint. These rami constitute the weaker regions of the pelvis but represent an important load path for lateral forces. In adults, the sacrum is fused together as one triangular bone by the sacral vertebrae (S1-S5). It articulates superiorly with the L5 vertebra, inferiorly with the coccyx, and laterally with the innominate bones, creating the SI joints. The vertebral canal continues inside the sacrum as the sacral canal, guiding blood vessels and spinal nerves.

Biomechanically, and when classifying injuries, the pelvis is often categorized into the pelvic girdle and the acetabulum. The pelvic girdle is a ring structure consisting of the sacrum and the left and right innominate bones. The ring can be thought of as a posterior arch, consisting of the upper sacral vertebrae and the bone connecting the SI joint with the acetabulum, and an anterior arch, consisting of the pubic bones and their superior rami. However, in total the pelvic ring should be considered as a single anatomical structure.

2.1.2 Joints and Ligaments

The pelvis consists of three joints, the anterior PS joint, the posterior SI joints, and the lateral hip joints. The PS joint connects the articulating pubic bones via a broad flat disc of fibrocartilage, which is covered by ligaments. Functionally it only allows for slight movement between the bones. The SI joint includes a synovial cavity between the articulating surfaces of the sacrum and the two innominate bones. This creates a smooth, low-friction surface that does not bind the two bones together. However, due to the irregularity of the articulating surfaces, and the strength of the surrounding ligaments, the actual movement of the SI joint is small. The hip joint is also a synovial joint which is formed by the articulation of the femoral head and the acetabulum socket, creating a ball-and-socket joint which allows for multiaxial movement.

The stability of the pelvic girdle is achieved by strong ligaments, primarily by the posterior ligaments called the posterior tension band. The major posterior ligaments are interosseous SI, posterior SI, anterior SI, sacrotuberous, sacrospinous, iliolumbar, and lateral lumbosacral. These must work together to ensure that the posterior pelvis is stable and have been described as the most powerful ligaments of the body. The anterior ligaments are called the inguinal and pubis ligaments. The pubis ligaments help stabilize the PS joint in external rotation and anterior shear forces. In addition to the pelvic ring ligaments the articular capsule of the hip joint is covered by the iliofemoral, ischiofemoral and pubofemoral ligaments, providing strength and stability.

2.1.3 Muscles, nerves, and blood vessels

Muscles attached to the pelvic bone can be categorized into three groups: trunk movement, support of the abdominopelvic organs, and thigh movement. For the trunk movement group, the muscles can be categorized from a posterior/anterior view of the body where the posterior muscles attach at the posterior part of the iliac crest or the transverse processes of the sacrum and lumbar spine, while the anterior muscles attach at the anterior part of the iliac crest, the pubic crest, and the PS joint. To support the abdominopelvic organs, and control urination, defecation, and sexual function, several small muscles are utilized. This group of muscles are referred to as the pelvic floor and connects the pubis and ischial spine with the coccyx and sacrum, effectively sealing the inferior opening of the bony pelvis. The thigh movement muscle group can be categorized from a posterior/anterior view of the body. The posterior muscles primarily control trunk/thigh extension and thigh abduction, while the anterior muscles primarily control trunk/thigh flexion and thigh adduction. The thigh muscle group attaches at multiple locations on the anterior, inferior, and posterior sides of the pelvis.

Nerves and blood vessels are at major risk of injury in pelvic trauma. Many of the nerve branches run through the sacral canal passing the sciatic notch, as they continue down to the thighs, making this a critical area. Furthermore, massive hemorrhaging from the blood vessels can occur through dislocation and laceration of bone fragments, representing a major complication in pelvis injuries.

2.2 Injury Mechanisms

Pelvic fractures can cause both direct loading to the soft tissue and visceral damage by bony fragment penetration and laceration. Given the major nerves, blood vessels, and organs inside the pelvic girdle, such injuries are associated both with a high acute mortality rate and a high degree of residual disability (Tile et al., 2015). Every biomechanical structure has several injury mechanisms depending on differences in geometry, tissue properties and the expected loading scenario. As such, fracture injury mechanisms for the pelvis have traditionally been separated for the pelvic girdle and the acetabulum. However, in the context of vehicle safety, it is reasonable to also evaluate the injury mechanism due to interactions with the restraint systems and other hard structures of the interior.

2.2.1 Pelvic girdle fractures

The severity of a pelvic injury often refers to the degree of instability, either from fracture or from dislocation, caused by the trauma. The instability can be described as stable, partially unstable, or completely unstable, and the degree of instability correlates with the energy of the trauma and the patient's physiological status (Tile, 1988). In the case of fracture, the resulting fracture pattern will be governed both by direction and magnitude of the force applied. For general pelvic fractures, three main force patterns can be identified: anteroposterior compression, lateral compression, and vertical shear (Burgess et al., 1990). For each case, different injuries to the pelvic structure can be expected. Due to the ring-like arrangement of the three pelvic bones, a fracture in a single location is highly unusual (Tile, 1988; Tile et al., 2015). Hence, a patient who has been identified as having sustained an anterior pelvic fracture, should also be assumed to have suffered a concomitant posterior fracture or ligament injury. Fracturing both pubic rami will have a significant effect on pelvic stability and its ability to maintain shape (Tile et al., 2015).

2.2.2 Acetabulum fractures

Acetabulum fractures are caused by the interaction between the acetabulum socket and the femoral head. Injuries are usually a result of either a direct impact to the greater trochanter or an impact to the foot or knee, which produces an axial force along the femoral shaft. Since the acetabulum is composed of two columns and two walls that meet at a dome, all fracture types are variations involving these anatomical structures. The type of fracture is highly dependent on both the position and the orientation of the femoral head when impacted (Rupp et al., 2003; Tile et al., 2015).

2.2.3 Injuries from interacting with the vehicle restraint/interior

The typical injury mechanisms describing pelvic fractures can also be defined in the context of pelvis-to-vehicle interaction during a crash. In side impacts, the pelvis is hit by the intruding door structure, or pinched between the door and the center console, causing lateral compression of the pelvis resulting in fractures to the pubic rami, iliac wing, sacrum, and acetabulum (Petit et al., 2015; Tile et al., 2015). In frontal impacts, several different load paths can result in pelvic fractures. First, an axial force through the femoral shaft resulting in fractures to the acetabulum (Rupp et al., 2003; Tile et al., 2015), typically caused by a dashboard interaction where the force is applied through the flexed knee of a seated occupant. Second, a vertical force due to interaction with the seat resulting in sacrum fractures. Finally, an anterior-posterior load applied through the lap belt resulting in fractures to the ala of the ilium, and in high severity cases, disruption to the posterior arch by fracture/dislocation of the SI joint. Such injuries, caused by interaction with the seat and belt system in frontal impacts, have been confirmed by recent PMHS studies (Baudrit et al., 2022; Guettler et al., 2023; Richardson et al., 2020; Uriot et al., 2015a).

The importance of successful belt-to-pelvis interaction in frontal car crash scenarios cannot be understated, since belt-to-pelvis coupling is utilized to control the occupant kinematics. Failure can result in lap belt disengagement from the pelvis, effectively loading the abdomen, *i.e.*, submarining. Instead of resulting in a recorded pelvic injury, the submarining outcome can also lead to an increased risk of laceration and abrasion in the neck area, thoracic injuries mainly to the lower thorax, and injuries to the abdomen and to the knees (Adomeit & Heger, 1975).

2.3 Epidemiology

Pelvic fractures have been associated with the highest early mortality rate of patients with orthopedic injuries, often caused by hemorrhaging (Tile et al., 2015). However, it should be noted that pelvic injuries commonly occur together with other injuries, due to the high impact energy required to fracture the pelvis, and that these co-injuries often are responsible for the mortality associated with pelvic AIS2+ injuries (Weaver et al., 2013). Furthermore, chronic pain and/or long-term disability can be expected regardless of the treatment chosen when compared to the non-injured population, which comes with a substantial cost to society (Tile et al., 2015). A strong emphasis on improved protection for the pelvis can be motivated when considering the acute mortality risk, the complex surgery required, and the long-term effects of severe pelvic fractures.

Pelvic fractures have been identified as the 3rd most common AIS2+ injury in MVCs, based on National Automotive Sampling System – Crashworthiness Data System (NASS-CDS) data from 2000-2011 (Weaver et al., 2013). Furthermore, it has been identified as the dominating AIS2+ lower extremity injury in near side impacts, based on NASS-CDS data from 2000-2015 (Pipkorn et al., 2020). This highlights the exposed conditions for the pelvis in vehicle accidents, especially in near side collisions where the pelvis is close to the impacting object with a relatively small crumpling zone. Additional analysis of the crash data in (Pipkorn et al., 2020), not part of the publication, revealed that pelvic injuries are the most common AIS2+ lower extremity injury, when grouping frequency weighted injuries from all impact directions for both driver and passenger. The pelvis then accounts for 28% of the lower extremity injuries, followed by ankle joint (18%) and knee joint (18%). In addition, pelvis injuries have been associated with sex, age, stature, and BMI, showing that females and low BMI subjects are at higher risk in near side impacts, while taller and heavier male subjects are at higher risk in frontal impacts (Sochor et al., 2003; Stein et al., 2006). This indicates that the definition of a vulnerable occupant for pelvis related injuries varies depending on the crash configuration.

However, both (Weaver et al., 2013) and (Pipkorn et al., 2020) used the AIS98 code to classify injuries. In the crash data from (Pipkorn et al., 2020), 96% of the pelvic injuries were coded as *"Pelvis fracture, with or without dislocation, of any one or combination: acetabulum, ilium, ischium, coccyx, sacrum, pubic ramus"* with an AIS score of 2 (3 if specified to be open/displaced/comminuted). Since 2005, the AIS code for pelvic injuries has changed substantially with separate codes for pelvic ring and acetabulum fracture in which the pelvic ring is now considered a single anatomical structure with overall severity depending on stability (AIS2 – stable, AIS3 – partially unstable, AIS4 – totally unstable, with +1 for each if open and AIS5 if >20% blood loss). Hence, the old pelvic injury AIS coding generally assigned a lower AIS severity and, *e.g.*, in a frontal impact, did not distinguish between acetabulum fractures, possibly caused by axial femur loads, and pelvic ring fractures, possibly caused by belt loading. While this distinction is useful, regardless of AIS version it is still not possible to further classify pelvic ring injuries as, *e.g.*, pubic rami fractures, different versions of sacral fractures, or avulsion fractures of the ilium. Hence, it is difficult to link pelvic fractures from epidemiological data to a specific injury mechanism as described previously.

Given the indirect nature of the injury caused by a failing belt-to-pelvis interaction, as the belt leaves the pelvis and loads the abdomen, retrospective epidemiological studies of this type of event are not readily available. One method to circumvent this problem has been to study hollow-organ abdominal injuries among belted occupants in frontal impacts (Lamielle et al., 2006; Poplin et al., 2015). An association has been identified between abdominal injuries and higher speeds, increased dashboard intrusion, older occupants, and increased weight, together with a higher risk of hollow-organ injuries for belted occupants compared to a higher risk of solid-organ injuries for unbelted occupants. However, based on these results, it is not possible to distinguish between belt misuse, the occupant placing the belt incorrectly above the pelvis, and safety system failure, the belt initially loads the pelvis which then slips under (submarining) during a crash. From volunteer studies it has been shown that belt misplacement in relation to the pelvis is correlated with both older occupants and higher BMI (Reed et al., 2013), which could explain some of the associations identified with regard to abdominal injuries. Furthermore, fractures to the pelvis due to a failing belt-to-pelvis interaction have, to date, not been well described in the epidemiological literature (possibly due to the limitations associated with distinguishing pelvic fractures by the AIS code, or because this type of injury is not common in current vehicles). However, PMHS studies in frontal car crash scenarios have resulted in pelvic fractures including the iliac wings, the complete pelvic ring, and the sacrum (Baudrit et al., 2022; Guettler et al., 2023; Richardson et al., 2020; Uriot et al., 2015a), highlighting the relevance of this loading scenario when designing future reliable safety systems.

2.4 Current Tools for Automotive Safety Assessment of Pelvis

Current requirements on automotive safety relating to occupant pelvic injuries are evaluated based on physical tests with ATDs. Agencies that define the requirements include, *e.g.*, Economic Commission for Europe of the United Nations (UN/ECE), National Highway Traffic Safety Administration (NHTSA) – Federal Motor Vehicle Safety Standards (FMVSS), Insurance Institute for Highway Safety (IIHS), and European New Car Assessment Programme (Euro NCAP). The assessments typically include full frontal, offset frontal, and side impacts with ATDs in different driver/occupant configurations. The ATDs evaluated include the 50th percentile male Hybrid III, THOR, EuroSID-2/2re, and WorldSID, as well as the 5th percentile female Hybrid III and SID-IIs. To evaluate the risk of pelvic injuries, limits on femur compression force in frontal impacts and pubic symphysis force in lateral impacts, have been defined.

In addition, to mitigate the risk of submarining outcomes, rating agencies like IIHS and Euro NCAP have added assessments on force drop at the iliac force transducers in frontal impact scenarios and confirmation via high-speed video analysis. If submarining is detected, a score penalty is assigned to the rating. Legal requirements from UN/ECE and FMVSS have instead added restrictions on seat and belt anchor point designs to achieve pre-defined targets on lap belt angles, intended to result in sufficient belt-to-pelvis interaction to mitigate the risk of a submarining outcome.

While current requirements, to a limited extent, cover acetabulum fractures from loading via the knee (as femur compression) and pelvic ring fractures in pure side impacts (as PS force), it does not protect against the pelvic injuries seen in PMHS tests with reclined occupants. More specifically, there is no limit for iliac spine loading from the belt or sacrum loading from the seat, both potential fracture locations associated with submarining prevention in reclined PMHS tests. Furthermore, one can question the relevance of the ATDs for submarining detection as neither the Hybrid III, nor the THOR dummy, can replicate PMHS submarining outcomes in matched rear seat frontal crash scenarios (Guettler et al., 2023).

Another limitation with ATDs is the inability to evaluate omnidirectional loading scenarios and inclusion of stochastic variations found in human anatomy and real-life accidents. Augmenting the physical evaluation with simulation-based evaluations would allow for a more complete assessment of the vehicle safety performance. Some agencies have started to move in this direction as, *e.g.*, Euro NCAP, included Virtual Testing Crashworthiness (VTC) applied to far-side impacts using the WorldSID dummy for monitoring in 2024, expected to be fully in force by 2026 (Klug et al., 2023). As indicated in the Euro NCAP 2030 roadmap (Van Ratingen & Jacobsen, 2022), virtual testing is also intended to be implemented in frontal protection assessment including variations in impact angle, reclined seating positions, and occupant variability. As soon as viable HBMs with assessment criteria are available, they will replace the WorldSID dummy models. The main objective of this thesis is in line with these efforts.

3 METHODS AND MODELING

3.1 Principal Component Analysis

Historically, anthropometrical studies have focused on analyzing a limited set of measurements that describe the geometry of a body part. A more modern approach is to study a much larger set of anatomical points defined by cartesian coordinates. By shifting the focus to coordinate data, the complete spatial arrangement of the anatomical points, *e.g.*, the shape, is captured, and any distance (or angle) between them is automatically retained. By adopting this approach, a large set of data is usually attained, and statistical analysis methods are warranted. One category of such methods, defined by multivariate statistics on dense sets of corresponding anatomical points, are known as geometric morphometrics (GM) or statistical shape models (SSM) (Slice, 2007).

A common method utilized in SSM studies is called principal component analysis (PCA). This method captures the mean shape and the covariance structure of the data around that mean. As a mathematical method, PCA identifies the orthogonal set of vectors that most efficiently captures the sample variance. This can be used as a dimension-reduction technique by transforming the data into an orthogonal set of loading vectors and only retaining the principal components (PCs) that capture variance up to a predefined threshold, see Figure 3 for a 2D visualization. Once the reduced set of PCs are known, correlation of shape with other factors can be explored to find new associations (Slice, 2007).

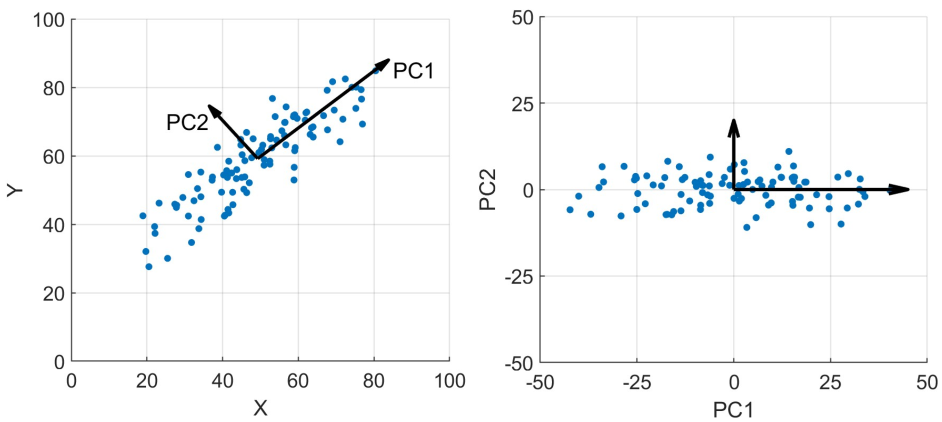


Figure 3 – 2D visualization of PCA transformation, left figure showing the original data with substantial variance in both the X and Y direction, right figure showing the transformed data with majority of variance (≈95%) captured by only PC1.

One drawback with PCA is that PCs are global, which can make shape interpretation difficult, especially for the non-dominant PCs (Zou et al., 2006). The PCs are considered global since they are defined by a variance-maximizing criterion on the entire dataset. This means that local variations in the data might become mixed over several PCs and therefore be difficult to distinguish. An extension to PCA called sparse principal component analysis (SPCA) has been developed to address this problem. SPCA frames the PCA problem as a regression-type optimization problem and utilizes variable selection techniques from multiple linear regression, to penalize the weight of distant data points to zero. This results in sparse loading vectors that describe localized variance in the data (Zou et al., 2006). The general problem formulation can be expressed as:

$$
\begin{gathered}
(\widehat{\mathbf{A}}, \widehat{\mathbf{B}}) = \underset{\mathbf{A},\mathbf{B}}{\arg\min} \sum \|\mathbf{X} - \mathbf{A}\mathbf{B}^{\mathrm{T}}\mathbf{X}\|^2 + \psi(\mathbf{B}) \\
\textit{subject to } \mathbf{A}^{\mathrm{T}}\mathbf{A} = \mathbf{I},
\end{gathered}
\tag{1}
$$

where $\mathbf{B}$ is a sparse weight matrix, $\mathbf{A}$ is an orthonormal matrix and ψ denotes a sparsity inducing regularization such as the least absolute shrinkage and selection operator (LASSO) or the elastic net (Zou et al., 2006). The PCs, $\mathbf{Z}$, are formed as:

$$
\mathbf{Z} = \mathbf{X}\mathbf{B} \tag{2}
$$

and the original data can approximately be recreated using:

$$
\widetilde{\mathbf{X}} = \mathbf{Z}\mathbf{A}^{\mathrm{T}} \tag{3}
$$

To solve this optimization problem a method has been presented by (Erichson et al., 2020), implemented as an R-package "*sparsepca*" (Erichson et al., 2018).

By design, SPCA is not suited for capturing global effects like volume and these should be removed a priori, as noted by (Sjöstrand et al., 2007). This can be achieved by including both translation and scaling when registering the set of data points to a common reference using generalized Procrustes analysis (GPA). The scaling variable includes the volumetric difference between samples and can be studied separately to the shape effects. This approach is further motivated by (DelPrete, 2019), who states that geometric normalization prior to shape analysis is important for understanding how the shape of male and female pelvises differ without the influence of difference in size.

The SPCA method was used to generate the SSM presented in **Paper I**, which became the basis for the pelvis FE-model in **Paper II**, **III**, and **IV**.

3.2 Sensitivity Analysis

To gain confidence in computational biomechanical models, a thorough assessment of model sensitivity is crucial to ensure that not only the average is considered, but also the distribution of possible outcomes (Cook et al., 2014). In addition to providing confidence in the model, sensitivity analysis also enables us to answer questions like; how much does a specific variable contribute to the predicted response, and which variable could result in a predicted response exceeding a predefined threshold. Given the substantial variability in biological systems, Cook et al. encouraged the biomechanical research community to broaden its current scope and make parameter variation a standard component of their analyses.

The literature describes several sensitivity analysis methods (Borgonovo & Plischke, 2016; Saltelli et al., 2008). These can broadly be distinguished as local, where one parameter at a time is varied to compute a constant effect over the entire parameter range, and global, where all parameters are varied simultaneously and evaluated at multiple sample points (Saltelli et al., 2019). However, the local/global terminology should not be confused with the local/global description of shape components used previously in this thesis. While being intuitive and easy to set up, local sensitivity analysis is only valid if the model can be seen as linear with no interactions and only explores a limited range in a multi-dimensional space. Global sensitivity analysis (GSA) on the other hand, is also valid when the model response is both nonlinear and includes interaction effects amongst parameters.

Sensitivity analysis aims to quantify the contribution of each input variable on the random response of a system. A common approach is a variance-based decomposition, where the total variance of the output response is defined by the sum of the contribution of each input variable. Given a function $Y = h(\boldsymbol{X})$, where $\boldsymbol{X} = [X_1, X_2, \dots, X_n]^T$ are n independent random variables, the variance of the output can be decomposed as:

$$V_Y = \sum_{i=1}^{n} V_i + \sum_{i<j} V_{ij} + \cdots \tag{4}$$

where V_i is the variance from variable X_i, V_{ij} is the variance due to the interaction of variables X_i and X_j, and the dots represent higher order (more than two variables) interactions, as presented by (Zhang & Pandey, 2014). Dividing Equation (4) with the total variance V_Y, one obtains the Sobol's sensitivity indices as:

$$1 = \sum_{i=1}^{n} S_i + \sum_{i<j} S_{ij} + \cdots \tag{5}$$

where S_i is referred to as the primary (or first-order) sensitivity index. For a model with no interaction between input variables, $\sum S_i$ equals one since all higher-order terms are zero, for all other cases $\sum S_i < 1$.

The most complete GSA method of a general response function is Monte Carlo (MC) simulations (Zhang & Pandey, 2014). However, this method is time consuming and typically requires tens of thousands of model evaluations, making it unrealistic for computationally demanding models like FE-HBMs. An approximation for variance-based GSA with sensitivity indices called the multiplicative dimensional reduction method (M-DRM) was first presented by (Zhang & Pandey, 2014). The method approximates high-dimensional integrals associated with the variance analysis by a product of one-dimensional functions, and computes the primary sensitivity indices as:

$$S_i \approx \frac{\theta_i / \rho_i^2 - 1}{\left(\prod_{k=1}^{n} \theta_k / \rho_k^2\right) - 1} \tag{6}$$

where ρ_i and θ_i are one-dimensional integrals. As such, these can be computed numerically by Gaussian quadrature as:

$$\begin{aligned} \rho_i &= \int_{X_i} h(X_i, \boldsymbol{C}_{-i}) f_i(X_i) dX_i \approx \sum_{j=1}^{N_{GP}} w_{ij} h(X_i^j, \boldsymbol{C}_{-i}) \\ \theta_i &= \int_{X_i} [h(X_i, \boldsymbol{C}_{-i})]^2 f_i(X_i) dX_i \approx \sum_{j=1}^{N_{GP}} w_{ij} \left[h\left(X_i^j, \boldsymbol{C}_{-i}\right)\right]^2 \end{aligned} \tag{7}$$

where $f_i(X_i)$ is the distribution function for parameter X_i, $\boldsymbol{C}_{-i}$ is the cut point vector with all variables but X_i fixed to their nominal values, $h(X_i^j, \boldsymbol{C}_{-i})$ is the functional evaluation for each input variable, and w_{ij} are Gaussian quadrature weights. Using Gaussian quadrature with N_{GP} Gauss-points and n variables the total number of simulations are at most nN_{GP}, which is several orders of magnitude less than the MC method.

The GSA method M-DRM was used to identify the variance explained by pelvis shape, material properties, and cortical thickness on pelvis response to lateral impacts in **Paper II**.

3.3 Regression

Regression modeling, or regression analysis, is a fundamental concept in statistics and supervised machine learning, where a set of input features are used to predict a continuous output. The goal of a regression analysis can either be a model that predicts an outcome or a deeper understanding of the relationship between input and output, *i.e.*, which input has what effect. As such, regression can be used both as an alternative to variance-based sensitivity methods and as a common metamodeling method, where a computationally demanding model is approximated over the parameter space by a mathematical expression that is much cheaper to evaluate than the original model.

In regression modeling, an outcome (response / dependent variable) is estimated given a known input (predictor(s) / independent variable(s)). In its simplest form this is done to estimate a continuous value from one predictor by linear regression:

$$y(x) = \beta_0 + \beta_1 x \tag{8}$$

where $y(x)$ is the continuous response, β_0 is the intercept, and β_1 is the coefficient (slope) associated with the predictor x. The function is fitted to the observed data as a linear line which minimizes the residuals, *i.e.*, the difference between observed data and the values predicted by the line. The minimization is done by the least squares method which minimizes the sum of the squared residuals, yielding the best-fitting line to model the data based on a linear relationship. In situations where the response is a function of more than one predictor, multivariate linear regression is defined as:

$$y(x) = \beta_0 + \sum_{j=1}^{N} \beta_j x_j \tag{9}$$

where N predictors are fitted by their corresponding coefficients β_j. In its general form, the predictors can be of first order (main effects), higher order (squared, cubed, etc.), and/or interactions (e.g., x_1x_2, $x_1x_2x_3$ etc.).

When the best-fitting function is computed, significance of the estimated coefficients is evaluated by hypothesis testing. For each coefficient β_j, the test checks if it significantly contributes to predicting the response variable by stating a null hypothesis of $\beta_j = 0$, *i.e.*, the predictor has no effect, and an alternative hypothesis of $\beta_j \neq 0$, *i.e.*, the predictor has an effect. From the estimated coefficients, the t-statistic of each is computed and its associated p-value, which indicates the probability of observing the value of β_j, if the null hypothesis were true. A common threshold when identifying significance is 0.05, meaning that if $p < 0.05$ we can say that the coefficient is significant at the 5% level.

In real-life, a perfect linear relationship between the predictors and the outcome variable is rarely observed. Instead, if a linear relationship can be established, there is always some spread around this linear estimate. To evaluate how much of the observed variance that is captured by the regression model, measurements like R-squared (R^2) can be computed:

$$R^2 = 1 - \frac{SS_{res}}{SS_{tot}} = 1 - \frac{\sum_{i=1}^{n}(y_i - \widehat{y_i})^2}{\sum_{i=1}^{n}(y_i - \overline{y_i})^2} \tag{10}$$

where R^2 ranges from 0 to 1 (0 meaning that none of the variance is explained by the model and 1 that all the variance is explained), n is the sample size, SS_{res} is the sum of squared residuals, SS_{tot} is the total sum of squares, y_i is the observed data, $\widehat{y_i}$ is the predicted response, and $\overline{y_i}$ is the observed mean. To estimate the error between the observed values and the regression line, the standard error of the regression can be computed as:

$$SE_{res} = \sqrt{\frac{\sum_{i=1}^{n}(y_i - \widehat{y_i})^2}{n - k - 1}} \tag{11}$$

where k is the number of predictors in the model. If the residuals of the model are normally distributed, which is an assumption that should be confirmed in regression modelling, the standard error of the regression is analogous to standard deviation around the mean, in that it estimates how much predictions typically deviate from the observed values. About 68% of the residuals should fall within $\pm SE_{res}$ and 95% should fall within $\pm 2 \times SE_{res}$. This can be utilized to generate a distribution of possible response values for a given set of predictors.

In cases where the response is not a continuous value, but rather binary (0/1), the response of the linear regression equation can be rewritten with a logit (or log odds) function as:

$$\ln\left(\frac{p(x)}{1 - p(x)}\right) = \beta_0 + \sum_{j=1}^{N} \beta_j x_j \tag{12}$$

where p(x) is interpreted as the probability that the response variable $y_i = 1$ given a known input vector of predictors $\boldsymbol{x}_i = [x_1, x_2, \dots, x_j]$. Solving for $p(x)$, we get the standard logistic regression function as:

$$p(x) = \frac{1}{1 + e^{-(\beta_0 + \sum_{j=1}^{N} \beta_j x_j)}} \tag{13}$$

The output of the logistic regression function is a continuous value that ranges from 0 to 1, indicating the probability of the event (1 = 100% probability).

Regression is not suitable when, *e.g.*, the relationship between predictors and response are highly non-linear, and cannot be modeled by transformation, or when data are insufficient. The first limitation should be controlled when performing the regression analysis. The second limitation is a sampling problem which, depending on the context of the regression, can take different forms. Either the sampling involves collection of real-life data or, in the context of metamodel development, it involves evaluation of a computationally demanding model at specified sample points. As the parameter space grows the multi-dimensional parameter space becomes a hypercube of possible parameter combinations, and a conflict quickly develops between possible sample size and desired parameters.

A common method to address the issue of sampling for computationally demanding models is space-filling sample designs. These methods aim to distribute the sample points as uniformly as possible, ensuring that all regions are covered for the specified sample size. One method applied in space-filling designs is Latin-hypercube sampling (LHS). In LHS, each parameter is divided into N equally probable intervals. From each interval, one sample is randomly generated, ensuring uniform coverage of the space. To further improve the space-filling properties of the design, methods like the *maximin criterion* can be added to optimize the placement of the sample points such that the minimum distance between two points is maximized.

In **Paper I**, multivariate linear regression was utilized to generate the SSM of the pelvis. In **Paper II** and **Paper IV**, the SSM was utilized to predict an average male/female pelvis geometry, while the standard error of the regression was utilized to generate random pelvis shapes from the population, associated with the average predictions. In **Paper IV**, a metamodel of submarining outcome was developed by logistic regression. Tests for significant predictor coefficients were utilized to generate the shape model in **Paper I** and to identify parameters influencing the submarining outcome in **Paper IV**.

3.4 Finite Element Model Development

This chapter summarizes the FE-model development carried out as part of this thesis. The new pelvis FE-model was used in **Paper II** and the updated full-body SAFER HBM, including the new pelvis, in **Paper III**. The updated version became the starting point for developing SAFER HBM v11 (Iraeus et al., 2024), which was used in **Paper IV**. More details on the FE-model development relating to this thesis can be found in **Paper II** and in (Brynskog, 2024a).

The pelvis FE-model developed in this thesis consists of both innominate bones and the sacrum, with separate representations for trabecular and cortical bone. The bone models were connected anteriorly with a model of the PS joint and posteriorly by models of the two SI joints. The former consists of a fibrocartilage disc and the surrounding ligaments. The latter consists of articular cartilage, interosseous ligaments, anterior ligaments, and posterior ligaments. The inferior sacrum and the ischial bone were connected via the sacrospinous and sacrotuberous ligaments. Finally, elements that make up for the articular cartilage were modeled on the lunate surface of the acetabulum. See Figure 4 for a model overview.

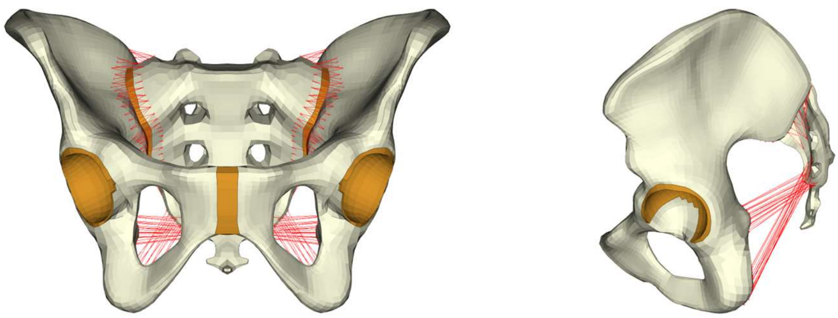


Figure 4 – Frontal (left) and sagittal (right) view of the pelvis FE-model.

One sub-objective of this thesis was to develop and validate a generic morphable FE-model of the human pelvis, capable of running shape variation evaluations. To meet this objective, the model was developed based on the outer surface of an average pelvis geometry described in **Paper I**. The average of the complete sample, including both females and males, was chosen as baseline to minimize the overall morphing distance and mitigate mesh distortion when covering the entire shape space. In addition, mesh quality and structure were highly prioritized to allow for substantial variations in geometry while retaining a numerically stable model. Details on the mesh quality achieved are outlined in **Paper II**.

The pelvic cortical bone thickness is not uniform over the pelvis surface, hence, it is important to include this variability to accurately describe the structural stiffness. However, the data used in **Paper I** do not provide the cortical bone thickness associated with each subject. To include cortical bone thickness in the model, a separate study on cortical thickness distributions of the innominate bones from 10 normal controls (5 females, 5 males) (Harris et al., 2012) was considered. The distribution of nodal thicknesses within each subject was found to be lognormal with a mean of 1.64 mm. Hence, a lognormal distribution was fitted to the nodal thicknesses of each subject, and the subject with the closest distribution to the average was chosen as baseline, see Figure 5 for cortical thickness of the baseline subject. Since the minimum cortical thickness was 0.5 mm, a solid mesh was deemed unfeasible to comply with restrictions on minimum time step length for the SAFER HBM. A quadrilateral shell mesh with a distributed nodal thickness was hence implemented for the cortical bone on the surface of the hexahedral solid elements of the trabecular bone. To have the shell element placed in the midplane of the cortical bone, the outer cortical surface of the average pelvis geometry was offset in the normal direction by half the baseline cortical thickness of each element.

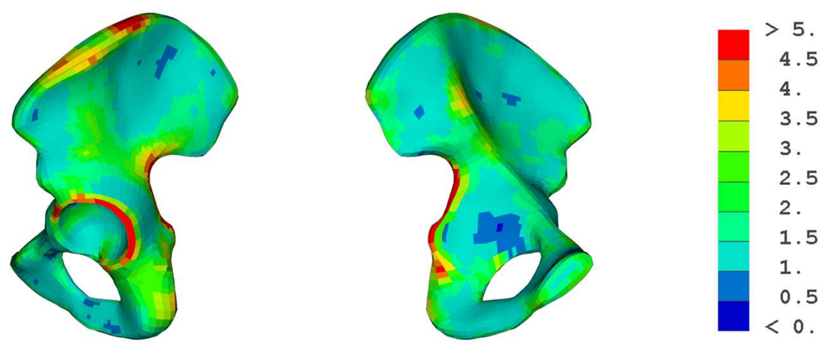


Figure 5 – Cortical bone thickness in the baseline model from a medial (left) and a lateral (right) view.

Material property data for pelvis trabecular and cortical bone are limited in the published literature. For the baseline model, a homogenous Young’s-modulus of 70 MPa (Dalstra et al., 1995) was implemented for the trabecular bone, while an elastic-plastic stress-strain curve from pelvic coupon tests (Kemper et al., 2008), was implemented for the cortical bone. The elastic-plastic curve was generated by reanalyzing the reported coupon results and using a weighted average between groups, see **Paper II**.

Morphing the FE-model based on the geometrical shape variance results from **Paper I** was done by a MATLAB script (Mathworks, Natick, MA, USA), developed in collaboration with the University of Michigan Transportation Research Institute (UMTRI) (Hu et al., 2016). The script utilizes a radial basis function with thin-plate-splines (RBF-TPS) interpolation and sets the outer cortical surface of the average pelvis geometry as source. The target for the morphing was the outer cortical surface of the pelvis geometry as predicted by the SPCA results. The FE-model was then morphed based on the displacement needed to move from source to target in 3D space. For further details see **Paper I** and (Hu et al., 2016).

To integrate the pelvis FE-model with the full-body SAFER HBM, the SAFER HBM v10 (Pipkorn et al., 2021) was used as starting point. To begin, the model was updated with a newly developed lumbar spine (Iraeus et al., 2023) and the pelvis model from **Paper II** morphed to a predicted 50th percentile male (age = 45 years, stature = 1.75 m, weight = 77 kg). The pelvis angle, defined as the angle between a line that connects the superior margin of the PS with the ASIS relative to vertical, was set to a reported average of 45° for a vehicle occupant seated with a 24° seatback angle (Izumiyama et al., 2018). To match the new position of the hip, the legs were translated until the femur head matched the new acetabulum position without changing the femur shape, angle, or length.

With the new skeletal models and position, a new skin geometry was implemented for the hip and thighs based on the HumanShape^{TM} (www.humanshape.org) data (Park et al., 2021). An outer contour of a target male subject was collected in a seated posture, again with the anthropometry of the baseline SAFER HBM. Since the scans for HumanShape^{TM} were generated from subjects seated on a rigid flat surface, and since smoothing occurred over the abdomen-thigh transition in the mesh fitting algorithm, some modifications were made to the predicted surface to produce an initially unloaded buttock as the baseline geometry and to make the abdomen fold in towards the pelvis in the abdomen-thigh transition, see (Brynskog, 2024a).

The resulting skin and bone surfaces were used to build a full hexa mesh of the soft tissues surrounding the hips, thighs, and abdomen, see Figure 6. As for the pelvis model, the target was a high-quality, all hexahedral, mesh to allow for mesh morphing covering the population variability, which was achieved using pre-defined meshing requirements as outlined in (Iraeus et al., 2024). The element formulation used for the solids was a fully integrated 8-point hexahedron with an assumed strain approach to avoid shear locking behavior seen in standard fully integrated elements, intended for elements with poor aspect ratios. While the elements typically have an aspect ratio below 5 in the lap belt area, morphing to a thinner subject or compression from belt loading in simulated crashes may increase the elements aspect ratio and make them sensitive to shear locking at the time of submarining.

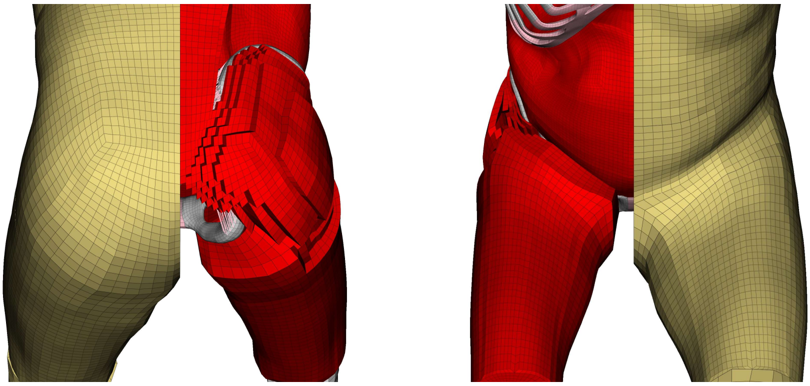


Figure 6 – Muscle (red) and fat (beige) surrounding the pelvis when implemented in the full-body SAFER HBM.

The thigh muscle elements along the femur surface were tied to the femur shaft, while the elements representing the hip muscles were connected to the pelvis model in areas approximately matching their origin/insertion. The pelvis-to-hip muscle connections were achieved by extruding a one element thick solid layer, that shares nodes with the skeleton on one side and are tied to the hip muscle elements using a contact on the other side, see Figure 7.

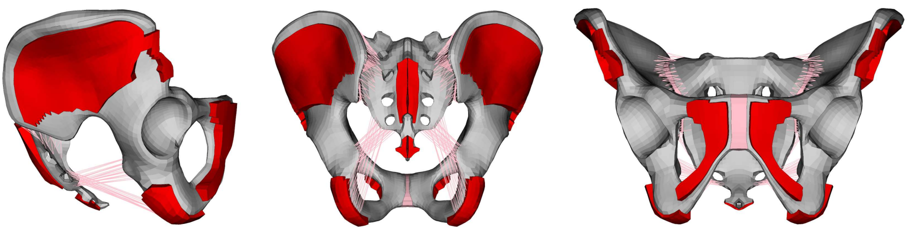


Figure 7 – Pelvis-to-hip muscle connections.

The hip, knee, and ankle joints of the SAFER HBM v10 were found to be stiffer than in the reported literature (Amankwah et al., 2004; Riener & Edrich, 1999). Since the stiffness of these joints will affect the force going through the legs, which can influence the pelvis kinematics in a frontal impact, it was decided to remove the ligaments and tendons of the hip, knee, and ankle joints, and instead use the kinematic joints already implemented for the active version of the SAFER HBM (Östh et al., 2014). The flexion/extension stiffness of each joint was prescribed using passive joint moments and compared with results predicted for volunteers (Riener & Edrich, 1999), see (Brynskog, 2024a).

3.5 Model Validation and Calibration

As part of this thesis, substantial efforts have been made to search the published literature for validation data, and to calibrate/validate both the component level pelvis model and the updated full-body SAFER HBM. For a summary of identified side impact validation data on pelvis response, please see Chapter 3.3.1 in (Peldschus & Wagner, 2021), and for frontal impact validation data on submarining, see (Brynskog, 2024b).

From the identified literature, the following have been evaluated as part of this research (see Figure 8 for a selection of cases):

Calibration

- Tension/compression of the PS joint (Dakin et al., 2001), see **Paper II**
- Displacement/rotation of the SI joint (Miller et al., 1987), see **Paper II**

Validation

- Iliac quasi-static, acetabulum quasi-static, and acetabulum dynamic loading (Guillemot et al., 1998), see **Paper II**
- Gravity settling on rigid seat (Linder-Ganz et al., 2007; Tanaka et al., 2021; X. Wang et al., 2021), see (Brynskog, 2024a)
- Free-back rigid-bar abdominal impacts (Hardy et al., 2001), see (Brynskog, 2024a)
- Whole body lumbar flexion (Uriot et al., 2015b), see (Brynskog, 2024a)
- Stationary test with rotating belt system (Uriot et al., 2006), see **Paper III**
- Sled test in nominal position on rigid seat (Luet et al., 2012), see **Paper III**
- Sled test in nominal position on semi-rigid seat (Uriot et al., 2015a), see (Brynskog, 2024a) and **Paper III**
- Sled test in reclined position on semi-rigid seat (Richardson et al., 2020), see **Paper III**
- Sled test in nominal and reclined positions on semi-rigid seat (UMTRI AVOK-Series, 2020-2023), see **Paper IV**

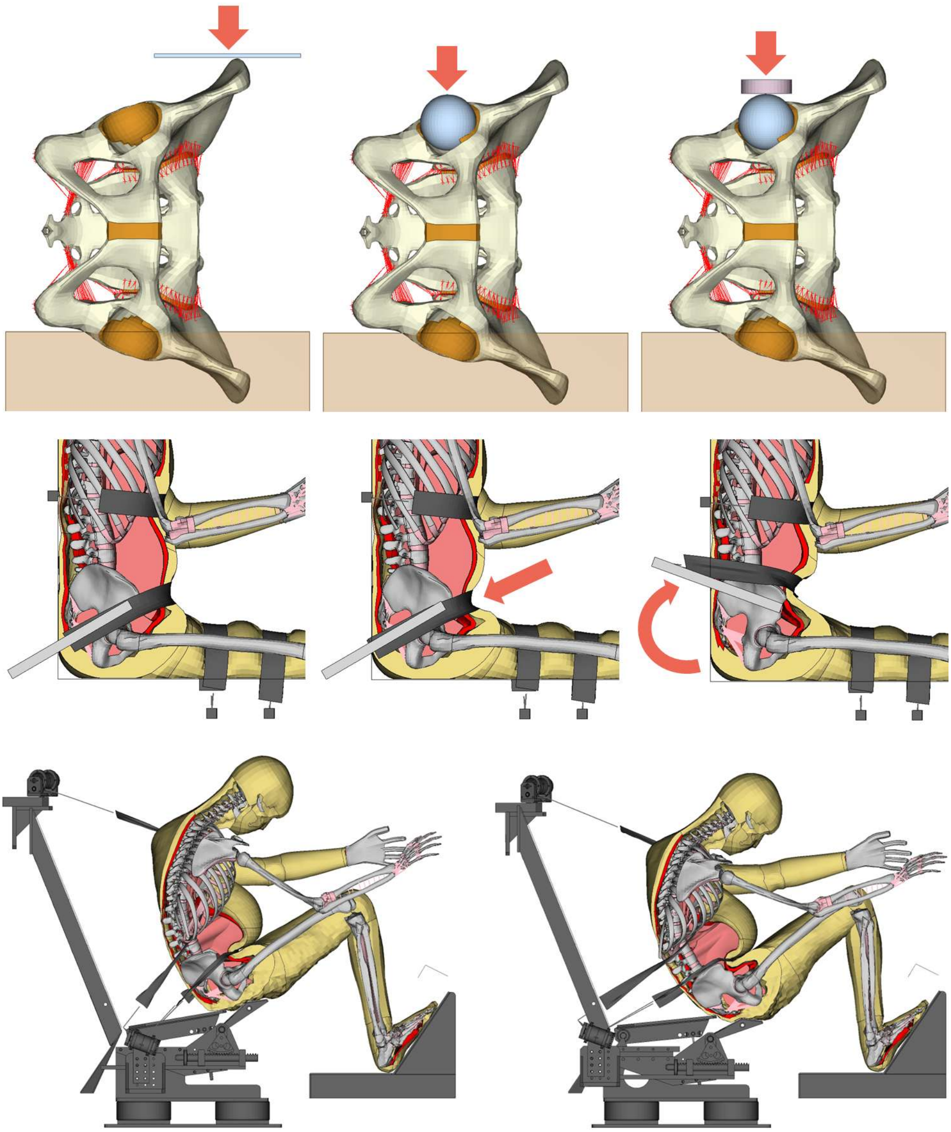


Figure 8 – A selection of validation cases; iliac/acetabulum quasi-static and dynamic loading (top row, **Paper II**), stationary test with rotating belt system (middle, **Paper III**), non-submarining and submarining scenario from sled test in nominal position on semi-rigid seat (bottom, (Brynskog, 2024b)).

4 SUMMARY OF APPENDED PAPERS

This thesis has resulted in the four appended papers summarized below. In **Paper I**, the author of this thesis (the author) contributed by landmarking segmented geometries and morphing a template to each subject, implementing a new method in the form of SPCA, generating and evaluating the multivariate linear regression models, visualizing the results, and writing/revising the article. In **Paper II**, the author contributed by building the new pelvis FE-model, performing calibration/validation simulations, defining the parameter distributions, implementing the GSA method, visualizing the results, and writing/revising the article. In **Paper III**, the author contributed by conceptualizing the study, integrating the new pelvis FE-model in the full-body SAFER HBM, modeling the new hip and thigh soft tissues, identifying simulation scenarios from the literature, preparing and running all simulation scenarios, visualizing the results, and writing/revising the article. In **Paper IV**, the author contributed by conceptualizing the study, developing the method for pelvis sampling, defining the parameter space, preparing and running validation simulations, preparing all simulations for the sensitivity analysis, generating and evaluating the logistic regression metamodel, visualizing the results, and writing/revising the article.

Paper I: Predicting Pelvis Geometry Using a Morphometric Model with Overall Anthropometric Variables

The objective of **Paper I** was to describe the population shape variance of the pelvic bones using SSM, and to develop an associated statistical model by linear regression using overall anthropometry (sex, age, stature, and BMI) as predictors. The study was performed to facilitate the development of FE-HBMs capable of representing the pelvic shape and perform population based evaluation in future automotive safety assessments.

In this study, SPCA was utilized to describe the pelvis population shape variance based on surface segmentations of clinical computed tomography (CT) scans. The sample included 132 subjects (75 females, 57 males), retrospectively obtained from clinical imaging studies at the University of Michigan, Department of Radiology. Figure 9 shows an overview of the methods used.

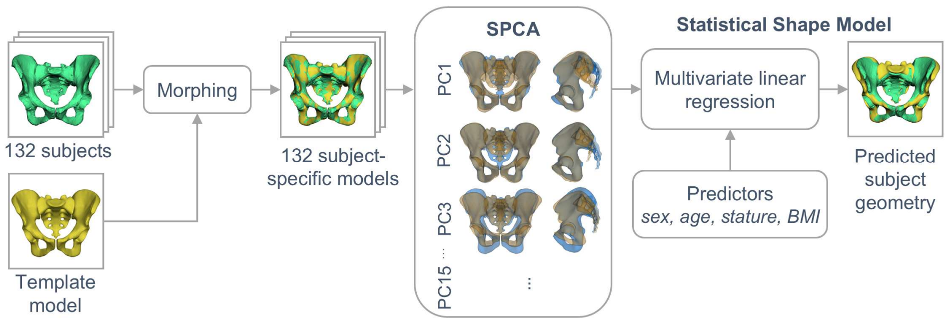


Figure 9 – Overview of methods used to develop the SSM in **Paper I**.

Conclusions from this study include:

- The population variance in pelvis geometry can only partially be explained (29%) by anthropometric variables such as sex, age, stature, and BMI.
- Inferior-anterior regions of the pelvis were primarily captured, while local sacrum features, shape and position of ASIS, and lateral tilt of the iliac wings, were not captured by the regression model. This could have implications for the occupant’s interaction with the vehicle restraint systems.
- Shape features overlap for females/males and, while significant differences can be identified between sexes, substantial inter-individual differences remain even after controlling for sex.

Paper II: Population Variance in Pelvic Response to Lateral Impacts - A Global Sensitivity Analysis

The objective of **Paper II** was to develop, calibrate, and validate a pelvis FE-model as well as identify and quantify the most influential variables on the denuded pelvis response to lateral impacts using GSA. The variables studied were pelvic bone shape, material properties, and cortical thickness.

In this study, a new detailed pelvis FE-model was built from the average geometry in **Paper I**. Using the SSM, a 50th percentile female (50 years, 162 cm, 63 kg) and male (50 years, 175 cm, 77 kg) baseline pelvis geometry were generated. The morphable FE-model was validated against published PMHS experimental results from static and dynamic lateral loads on denuded pelvises (Guillemot et al., 1998). The distribution of possible outcomes was considered, not just the average, by drawing 50 random females/males around each baseline model using variable distributions for shape, material properties, and cortical thickness. To study model sensitivity to variations in input variables, the GSA approximation M-DRM was utilized for the dynamic lateral load case. Figure 10 shows an overview of the analysis.

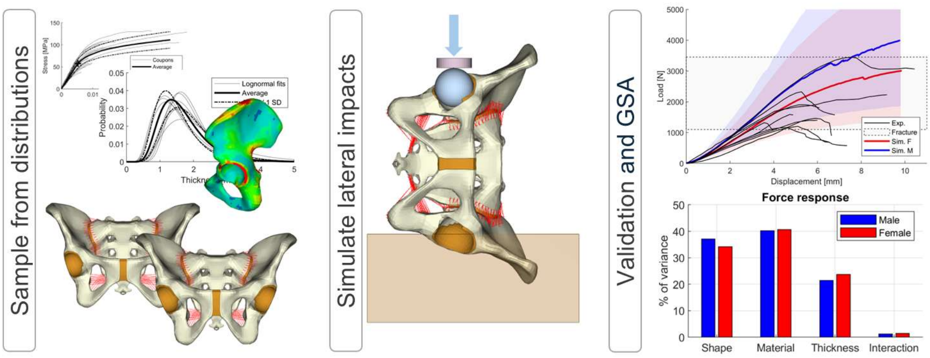


Figure 10 – Overview of the analysis in **Paper II**.

Conclusions from this study include:

- In lateral impacts to the pelvis, shape contributes to the model response variance by the same magnitude as bone material properties, and each of these contributions are approximately twice that of the contribution of the cortical bone thickness.
- To model pelvis response for a general population accurately, variability in both shape and material properties should be considered in the analysis.

Paper III: Simulating Pelvis Kinematics from Belt and Seat Loading in Frontal Car Crash Scenarios: Important Boundary Conditions that Influence the Outcome

The objective of **Paper III** was to evaluate the pelvis response sensitivity to variations in boundary conditions that directly influence the pelvis loads and are deemed important for the submarining outcome. The study aimed to present which boundary conditions that should be prioritized in future experimental and numerical studies, to facilitate a more precise comparison between FE-HBMs and PMHSs.

In this study, the 50th percentile male SAFER HBM from (Brynskog, 2024a), including the new pelvis FE-model from **Paper II**, was evaluated in multiple loading scenarios under varying seat and belt friction, seat stiffness, and belt bending stiffness conditions. The analysis included a stationary scenario with a rotating belt system (Uriot et al., 2006), three upright dynamic scenarios on a rigid seat (Luet et al., 2012), two upright dynamic scenarios on a semi-rigid seat (Uriot et al., 2015a), and a reclined dynamic scenario on a semi-rigid seat (Richardson et al., 2020). The parameter variations were done one-at-a-time for a total of 62 simulations. Figure 11 shows four examples of evaluated loading scenarios, and a pelvis related kinematic response.

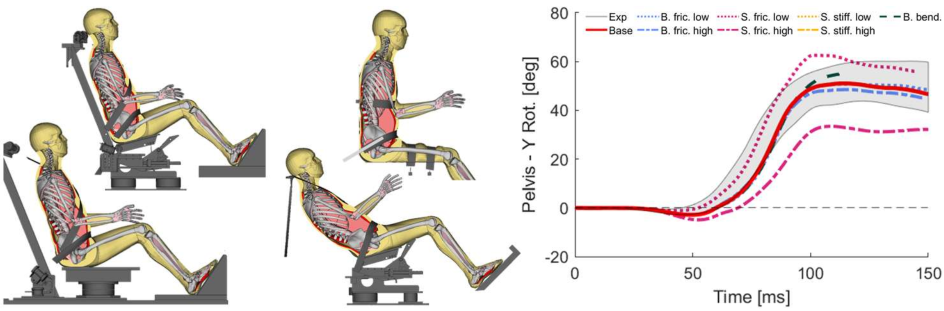


Figure 11 – Example of loading scenarios and kinematic responses from **Paper III**.

Conclusions from this study include:

- To reduce uncertainty in boundary conditions affecting the external pelvis loads, future experiments should evaluate the PMHS to seat friction coefficient and develop new modeling methods to capture belt folding when interacting with soft tissues.

Paper IV: Effect of Occupant and Restraint Variability in Reclined Positions on Submarining Probability in Frontal Car Crash Scenarios

The objective of **Paper IV** was to analyze the influence of both occupant and restraint variability on submarining outcome of reclined occupants in a frontal car crash scenario, using a simplified vehicle setup with a semi-rigid seat. More specifically, the study aimed to answer if intrinsic occupant variance associated with a predicted 50th percentile male is comparable with the restraint design variance on submarining outcome for reclined occupants.

In this study, the newly developed 50th percentile male SAFER HBM v11 (Iraeus et al., 2024) was utilized. The model was validated against PMHS experiments (UMTRI AVOK-Series, 2020-2023), by qualitative comparison to the reported PMHS kinetics and kinematics, before sampling a large number of occupant and restraint variations to generate metamodel training data on submarining outcome, for reclined occupants in frontal car crash scenarios. Significant parameters for submarining outcome prediction were identified and the parameter combinations resulting in a non-submarining zone were presented.

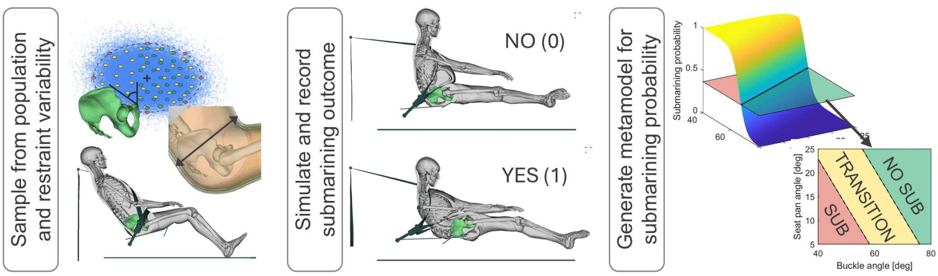


Figure 12 – Overview of the analysis in **Paper IV**.

Conclusions from this study include:

- Residual occupant variability around a predicted 50th percentile male is comparable with restraint design variability on submarining outcome for reclined occupants in frontal car crash scenarios.
- In future vehicle safety ratings, the study implies that variations in submarining outcome can be expected for different FE-HBMs if harmonization of the target occupant anatomy/posture is not established.
- Based on the response in a semi-rigid seat setup, the results indicate that current legal requirements on belt buckle angle might need a shift towards more vertical angles to enable reclined occupant retention.

5 DISCUSSION

With the main objective of enabling pelvis related automotive safety assessments for a population of female and male vehicle occupants, this thesis has focused on four sub-objectives. In short, these objectives include: (1) evaluate and describe the pelvic shape variance, (2) develop a new pelvis FE-model capable of including population variability, (3) validate the model and assess its sensitivity to variations in both occupant and boundary conditions, and (4) study the pelvis-to-belt interaction in frontal car crash scenarios focusing on submarining. The following chapters discuss these objectives and aim at providing guidance for future research in FE-HBM development, PMHS experiments, and population based vehicle safety assessments.

5.1 Population Variance in Pelvic Shape

Previous studies have used PCA to describe global shape variations of the pelvic bone, *e.g.*, (Arand et al., 2018; Audenaert et al., 2019), however, the study presented in **Paper I** is the first to use SPCA to describe localized shape variations. In addition, it is the first study to develop a SSM of the pelvis using only overall anthropometry (sex, age, stature, and BMI) as predictors. This allows for population based predictions of pelvic shape with the parameters typically used to define population cohorts in the context of traffic safety, *e.g.*, a 5th percentile female, 50th percentile male, or a 95th percentile male, although the population variance captured by these predictors was shown to be quite limited (29%). Furthermore, the full SPCA results, in addition to the SSM, allows for a more complete inclusion of population variability, which makes the method flexible for future evaluations and in line with the aim of this thesis.

The SPCA method used resulted in improved interpretability of the pelvic shape variations compared to standard PCA, since the local features were more distinguishable in shape than their global counterparts. As an example, Figure 13 shows two separate PCs, the local features width of ischial tuberosities and lateral tilt of the iliac wing, where the local effect is highlighted by movement in one area while the remaining surfaces mostly remain stationary. This allowed for more precise identification of variations captured by the SSM. It will also enable future studies with the pelvis FE-model, where a specific local feature is varied to evaluate its effect on a predicted response or safety assessment, *e.g.*, to evaluate the effect of population variability in lateral tilt of the iliac wing when interacting with the door panel in a side impact.

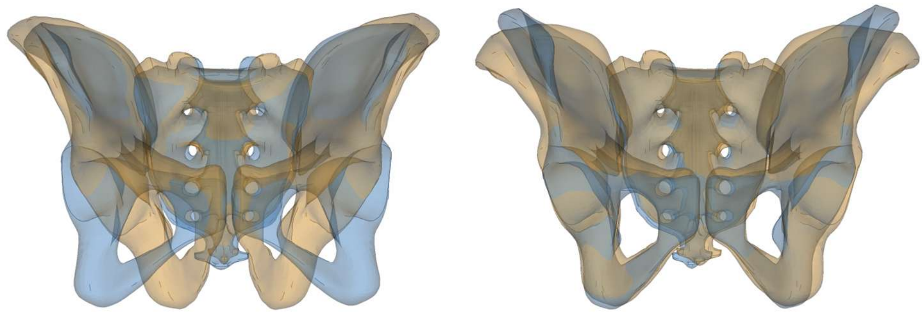


Figure 13 – Local features from SPCA describing width of the ischial tuberosities (left) and lateral tilt of the iliac wings (right). Results represent ±3SD from the studied population in **Paper I**.

Certain shape features were, however, better predicted by the SSM than others. Results showing the prediction error of the model versus the true geometry show that the inferior-anterior regions and the pelvic brim (except the sacral promontory) correlate most with overall anthropometry (median nodal error ≈4-5mm). The results are mostly explained by a strong coupling with sex for these areas (DelPrete, 2019; Luis & Carretero, 1994). On the other hand, the model prediction of the anterior-superior margins of the ilia and the inferior/superior ends of the sacrum were shown to be least correlated with the predictors (median nodal error ≈9-10mm). The shape features found to correlate poorly with overall anthropometry could have important implications for traffic safety analysis. For example, relative distance between the iliac wing and the trochanter can affect impact timing and force transfer from lateral loading, while sacrum length and curvature can affect seat interaction, and ASIS to sacrum relative position could affect lap belt to pelvis engagement. The coupling between sacrum position and lap belt engagement can be realized if the connection with the lumbar spine is taken into consideration. For a given lumbar spine shape, variations in sacrum endplate angle will produce a tilt of the pelvis in the sagittal plane, effectively shifting the ASIS position relative to the lap belt. Such a shift might cause a correctly placed lap belt, loading the iliac spines, to be placed at or above the ASIS instead, consequently increasing the risk of a submarining outcome.

The benefit of describing local features through SPCA is, however, associated with certain disadvantages. In SPCA, the strength of the localization effect is governed by a parameter setting in the SPCA solver. Since there are no discrete boundaries for this localization, it is not possible to define a feature which is the only shape variation associated with a specific PC. A local feature should, hence, be interpreted as the dominating shape variance in a sub-volume of the pelvis geometry, defined by a manual parameter that determines the size of this sub-volume. Visualizing the resulting PCs, it is possible to qualitatively interpret what the dominating shape feature entails. Variations in shape that are too small to generate their own sub-volume will not be captured by a separate PC and could, hence, exist over the entire pelvis surface. In addition, the number of PCs is set a priori and are computed through optimization. Since the variance captured is dependent on the number of PCs specified, the resulting variance captured will be unknown when setting up the problem. Furthermore, since the problem is solved by optimization, the resulting shapes will also depend on the number of PCs specified. In PCA, the complete variance is always captured, and the retained variance can be decided a posteriori by specifying the number of PCs. In **Paper I**, one scale parameter and 15 PCs were specified in the SPCA and found to capture 90% of the total variance, which was considered sufficient for the purpose of the study.

For both PCA and SPCA, the resulting PCs are constructed by a variance-maximizing criterion based on the analyzed coordinate data. However, guarantee that these variations are aligned with the variation relevant to the biological question being addressed is not granted (Slice, 2007). For example, the first PC describing most of the shape variance might not correlate with risk for a specific injury. For this particular injury, a different PC may instead be more relevant despite describing the total shape variance to a lesser extent. In addition, a very localized feature, relevant to a specific injury, could potentially be found in the variance not captured/retained. Hence, with regard to the results presented in this thesis, at most, the evaluations performed can consider the geometrical variations found in approximately 90% of variance captured by the SPCA. However, while this is the theoretical limit, it might not be feasible to include the full shape variance from 16 separate parameters, given limitations in sample size. In **Paper IV**, this was solved by generating a large sample of random pelvises using the full shape model, and then defining the population based on two measurements, believed to be related to a submarining outcome, and drawing a smaller sample from this population to be included in the analysis. Therefore, other shape features were not explicitly controlled, and it is not possible to state the extent of the total variance that was included in the analysis. Since, a priori, it is unknown which feature that contributes to the injury risk, there is a possibility that other relevant shapes were not identified due to the measurements used to define the population.

Even though the SPCA captured 90% of the total shape variance, the developed SSM using multivariate linear regression with overall anthropometry as predictor variables, only captured about 30% of the total variance. This is an important finding in relation to current state-of-the-art FE-HBMs, which are typically modeled starting from CT data of a single subject (Butz et al., 2017; Gayzik et al., 2011; Matsuda et al., 2023). While the chosen subject is considered average by certain evaluations on global measurements, the pelvic shape might vary substantially from that of the true average. Similar conclusions have also been made for other body parts (Yates et al., 2016; Yates & Untaroiu, 2018), where select global measurements did not correlate strongly with local anatomical shape. Furthermore, since state-of-the-art methods to include population variance in traffic safety analysis utilize multivariate linear regression, with overall anthropometry as predictors (Hu et al., 2019), this further highlights the importance of knowing that sex, age, stature, and BMI, are poor predictors of general pelvic shape. While this might be a suitable approach for more one-dimensional structures than the pelvis, such as long bones (Klein et al., 2015), the results from **Paper I** and other studies on more complex geometries such as the ribcage (Wang et al., 2016), indicate that the variance captured shrinks with increasing dimensional variability. This suggests that local measurements, based on anatomical landmarks that require more intrusive measurement techniques, are needed to reliably predict the population variance in structures such as the pelvic bone. Alternatively, the variance not predicted by the statistical models (beyond 30% for the current pelvis model) could be included by random sampling from the standard error of the regression, as was done in **Paper II** and **Paper IV** of this thesis. This allows for the flexibility of both predicting an average occupant for a given cohort of the population, and expanding the analysis by also including the residual variability found around the prediction, *e.g.*, around the average female/male as in **Paper II**. For structures where most of the variance is captured by overall anthropometry, the effect of the residual variability might be irrelevant but for other structures, like the pelvis, this effect could be substantial.

5.2 Model Validation

Model validation was performed on component level pelvis response to lateral loading and full-body response to various frontal loading scenarios. Pelvis related response was the focus of the full-body validations, which was mainly evaluated by different quantitative methods, although other signals relating to both the FE-HBM and the boundary conditions were also evaluated, at least on a qualitative level. For a full presentation of model validations, the reader is referred to the appended papers and (Brynskog, 2024a).

In **Paper II**, no significant difference between simulated and experimental mean stiffness of the denuded pelvis could be identified, for both quasi-static and dynamic loading through the acetabulum in a lateral direction. Furthermore, rejecting that the simulated and experiment results came from the same distribution was not possible. In dynamic loading, the model was successful in capturing the response of the no fracture and only anterior fracture groups, but deviated when compared to the cases with complete fracture, since fracture through element erosion is not included in the model. In the experiments, the fractures mainly occurred at the pubic rami and around the SI joint. However, when quasi-static lateral loading was applied to the iliac wing, a significant difference in mean stiffness was found as the model gave a weaker response than the experiment. Figure 14 shows example figures from **Paper II** highlighting the response to dynamic loading through the acetabulum.

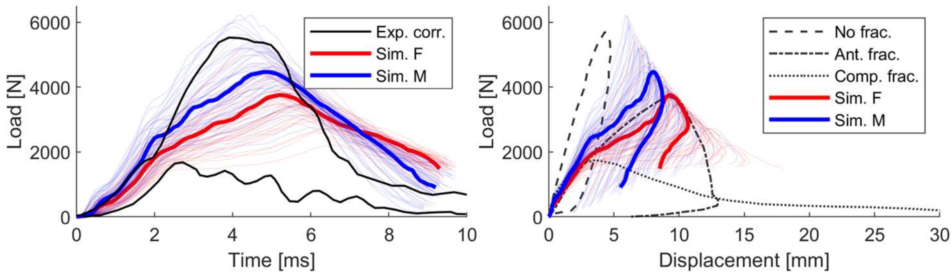


Figure 14 – Impulse response (left) and force-displacement response (right) vs. experiment corridor, from dynamic acetabulum loading to randomly sampled female (red) and male (blue) pelvises. Results from **Paper II**.

In **Paper III**, the pelvis FE-model had been integrated into the full-body SAFER HBM (Brynskog, 2024a) and pelvis forward displacement and forward/backward rotation was evaluated using the CORrelation and Analysis method (CORA v4.1.1) (Gehre et al., 2009), for varying boundary condition settings in three different loading scenarios. The boundary condition variations included belt friction (0.2 – 0.4), seat friction (0.2 – 0.5), seat spring stiffness (±10% from baseline), and belt bending stiffness (by varying element formulations), which were evaluated one-at-a-time. Evaluated as the average CORA score for each scenario, using the best boundary condition settings, the resulting score was between 0.88 and 0.93 (0.0 indicates no correlation and 1.0 a perfect match between curves). However, considering all evaluated boundary condition settings, the lowest average score was 0.43, highlighting the sensitivity for pelvis kinematics to variations in boundary conditions. The updated model was also found to capture the submarining outcome recorded in the experiments. Figure 15 shows example figures of pelvis kinematics from **Paper III**.

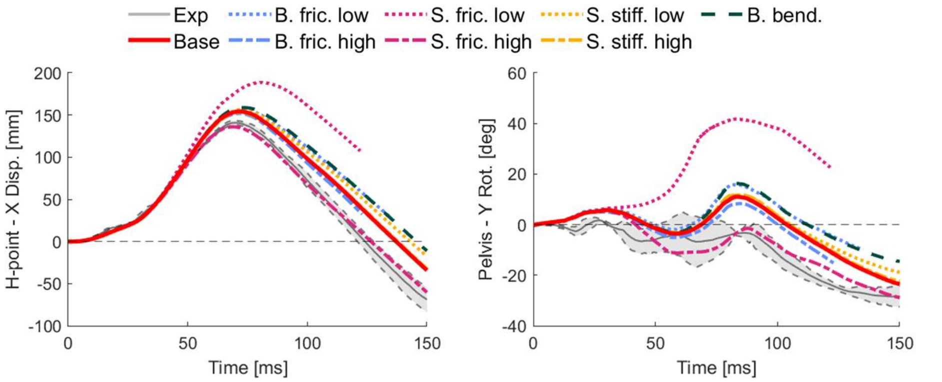


Figure 15 – Pelvis forward displacement (H-point – X Disp.) and forward/backwards rotation (Pelvis – Y Rot.), using various boundary condition settings, compared to PMHS response in a frontal impact with a reclined posture (Richardson et al., 2020). Results from **Paper III**.

In **Paper IV**, the pelvis response was only qualitatively compared to the reported experimental data, since missing data in the tracked signals did not allow for a quantitative evaluation such as the CORA method. Both phase and magnitude of pelvis forward displacement and forward/backward rotation were found to be mostly in the reported envelope. Main deviations from the experiments were identified in the rebound phase of the impact, where the success/failure of the restraint system have mostly been determined. Figure 16 shows example figures from **Paper IV,** highlighting the pelvis response for a reclined occupant in a 50 kph frontal car crash scenario.

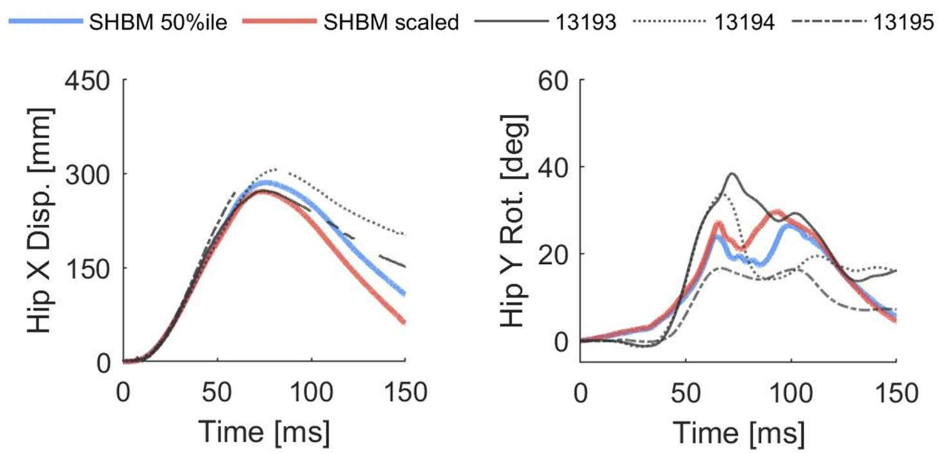


Figure 16 – Pelvis forward displacement (H-point – X Disp.) and forward/backwards rotation (Pelvis – Y Rot.), compared to PMHS response in a 50 kph frontal car crash scenario with a reclined posture (UMTRI AVOK-Series, 2020-2023). Results from **Paper IV**.

Considerable efforts have been spent on model validation in this thesis, which has identified some limitations regarding available data. To generate reliable fracture predictions through future strain-based injury risk functions (IRFs), more research on material properties for the pelvic bone is needed, especially with regard to fracture mechanics in different loading scenarios. To obtain a more complete validation of the component level pelvis response, the existing literature should be complemented with more PMHS experiments evaluating anterior-posterior and vertical loading of the pelvis. Having high quality validation data in these loading directions will be increasingly important in tackling the development of biofidelic models, for use in submarining assessment of reclined occupants. Furthermore, the current literature on reclined occupants has explicitly targeted the response of a non-submarining occupant. While this is important for model validation, it does not allow for validation of the model's sensitivity to accurately predict the submarining threshold. As such, a "validated" model for the non-submarining outcome could miss a scenario in which submarining would have been the result. Future PMHS experiments should, hence, target both submarining and non-submarining outcomes of reclined occupants. In this thesis, the model sensitivity to predicting submarining outcomes was verified against experiments with occupants in a standard driving (upright, not reclined) position. However, due to the lack of validation data, it is unknown how well this translates to the reclined scenario.

Another challenge for submarining validation is the different definitions used to identify and classify submarining in both simulations and PMHS experiments. To begin, one must decide if submarining constitutes the event of belt movement or the effect of belt loading. In the first definition, the belt is initially placed correctly on the pelvis but is then displaced during the crash producing a load to the abdomen. In the second definition, it is irrelevant whether the belt was placed correctly initially, since it is only the loading of the abdomen that is considered. The second definition is much broader and includes both restraint system failure and misuse, which potentially could be beneficial since the outcome is similar irrespective of cause and the aim is to mitigate the resulting injuries. In cases comprising obese subjects, it might be difficult/impossible to achieve an initial belt placement that allows loading the pelvis (Reed et al., 2012), automatically discarding a submarining outcome when applying the first definition. However, a drawback of the second definition is that model validation for submarining might mean very different things depending on initial belt placement. In one case, a complex interaction occurs when the belt load moves from the pelvis to the abdomen, while in the other case, the belt is already placed on the abdomen and the event is defined by a much simpler interaction governed solely by the abdomen stiffness. Furthermore, since the misplaced belt will always load directly onto the abdomen, validating a threshold for submarining is not necessary, instead it is the degree of severity related to the abdominal load that becomes relevant. Once the definition is specified, the difficult task of identifying if submarining did occur in the recorded physical experiments or not comes next, as the pelvis is not visible in video recordings and the belt is often obscured by the protruding abdomen at the time of submarining. Several signals for identifying a submarining outcome have been suggested such as lap belt force drop, belt-to-pelvis angle, pelvis rotation, knee forward displacement, hip-to-torso relative kinematics, abdominal injury pattern, pelvic strain, video analysis, etc. In these circumstances, the most robust evaluation of submarining occurring is when multiple signals are combined, as presented by (Rouhana et al., 1989) in their tests using the Hybrid III dummy and again emphasized by (Trosseille et al., 2018) in their tests using 5th percentile female PMHSs. Using a single measurement signal can sometimes be misleading and could result in inaccurate classification of submarining occurrence, which would harm future model validation efforts. In simulations, the complication of submarining identification seen in physical experiments is not an issue, as long as an agreed upon definition of what constitutes submarining is used, since the exact position of all parts is known throughout the simulation. Throughout this thesis, submarining has been defined as the midline of the lap belt moving superior and posterior to the ASIS in the left/right ASIS aligned sagittal planes, while the hip (H-point) still has a forward velocity relative to the vehicle, as presented in **Paper III**. As such, this defines submarining to only occur when the lap belt is initially placed correctly on the pelvis, and not when placed above the iliac crest directly on the abdomen. This definition was chosen since it is believed to match the spirit of the classic definition presented by (Adomeit & Heger, 1975): *"the lap belt slides over iliac crest with lap belt forces effecting the internal abdominal organs during forward displacement of the lower torso"*. Reaching consensus within the research community on what defines submarining, and how to identify its occurrence in PMHS experiments, would greatly improve validation of future models, making them both more robust and less prone to subjective assessments.

For a summary on available validation data in lateral impact scenarios, please see Chapter 3.3.1 in (Peldschus & Wagner, 2021), and for submarining scenarios, see (Brynskog, 2024b). To date, the literature is mainly limited in terms of females, young, and obese occupants. However, some of the gaps listed above are currently targeted by ongoing research.

5.3 Pelvis Response to Lateral Impacts

The study presented in **Paper II** is the first to include population variance with regard to both material and geometrical shape, when analyzing lateral impacts to the pelvis. In addition, it quantified the relative importance of different input variables on the pelvis response. Other pelvis FE-models for lateral impact evaluations can be found in literature, *e.g.*, (Kikuchi et al., 2006; Konosu, 2003; Kunitomi et al., 2017; Untaroiu et al., 2008), however, this is the first model built based on an average pelvis geometry that includes a SSM enabling parametric evaluations based on the population shape variance.

Fracture tolerance from lateral impacts to the hip is known to show substantial variation in experimental studies using PMHSs (Bouquet et al., 1998; Cesari & Ramet, 1982; Guillemot et al., 1998; Salzar et al., 2009). For example, Cesari and Ramet performed 60 impacts to 22 PMHSs in a seated position, using a rigid spherical impactor centered on the greater trochanter. They found a force tolerance of 10 kN for a 50th percentile male subject and close to 4 kN for a 5th percentile female and concluded that *"the value of the tolerable impact force varies greatly with anthropometry"*. **Paper II** have confirmed the statement by Cesari and Ramet by simulating the substantial variability in pelvis response seen as a result of shape and material variation. Furthermore, it elaborates on the statement by showing that pelvis shape contributes to the model response variance by the same magnitude as pelvic bone material stiffness, and that each of these contributions were approximately twice that of the cortical bone thickness. In the study, shape variability was included as the residual variance associated with a 50th percentile male/female based on the SSM and SPCA results, material variability was included as an elasticity-density relationship for the trabecular bone (Dalstra et al., 1993) using an apparent density distribution from L3/L4 human vertebra samples (Galante et al., 1970) and by cortical bone Young's-modulus from pelvic cortical bone coupon tests (Kemper et al., 2008), while cortical bone thickness was sampled based on the estimated distribution of ten control subjects (Harris et al., 2012).

It should be noted that these findings come from a well-defined impact on a denuded pelvis structure. In real-life MVC lateral impact scenarios, with an intruding side structure hitting the soft tissue of a seated occupant, the variability in boundary conditions is expected to be substantially more pronounced. For this scenario, the contribution from pelvic shape in relation to all other sources of uncertainty remains unexplored and warrants further research on lateral impacts with a full-body FE-HBM in a complete vehicle interior. To visualize the effect of the random distribution in shape, the weakest and stiffest model included in **Paper II** can be seen in Figure 17. The difference in response comes from variation in size, shape, material properties, and cortical bone thickness. While the joint material properties were kept constant, the size and shape variation of, *e.g.*, the PS joint also affected the response.

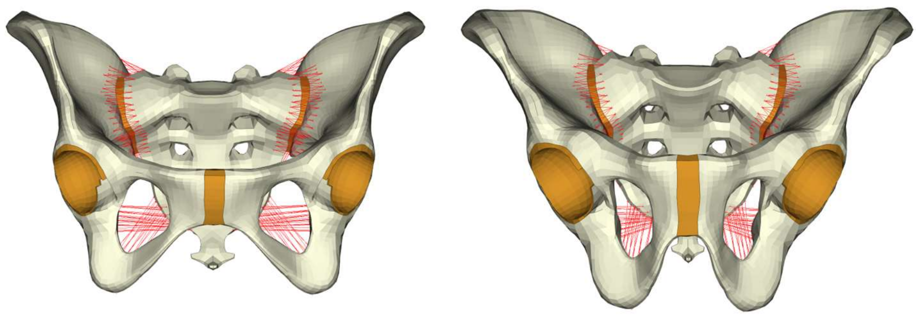


Figure 17 – Weakest (left) and stiffest (right) model from the sensitivity study in **Paper II**. The resulting stiffness was achieved by random sampling of shape, bone material properties, and cortical thickness.

A suitable tissue level criterion for fracture in human cortical bone is strain-based, as demonstrated experimentally (Nalla et al., 2003; Trosseille et al., 2008). This has been utilized in prior studies using both 1st principal strains for the rib cortical bone (Iraeus et al., 2019; Larsson et al., 2021; Pipkorn et al., 2019) and effective plastic strain for the bones of the cervical spine (DeWit & Cronin, 2012). However, a detailed understanding of the fracture mechanics due to lateral compression of the pelvis is lacking in the literature and it remains unclear if the most representative strain-based fracture criterion comes from tension, compression, shear, or a combination of these. Furthermore, the PMHS experiments reconstructed in **Paper II** do not present strain measures against which the model would benefit from being validated. Therefore, it was decided to perform a sensitivity analysis for both peak applied force, and maximum nodal averaged effective plastic strain in the superior pubic rami, since this has been identified as the fracture initiation point in lateral impacts to the pelvis (Petit et al., 2015). The force-based sensitivity can be considered when evaluating the variance in historical PMHS data, where peak force is typically reported, while the strain-based sensitivity can provide guidance for future development of a strain-based criterion.

5.4 On the Importance of Boundary Conditions

While performing a literature review on submarining related validation data (Brynskog, 2024b), it was found that some aspects of the boundary conditions are rarely reported from PMHS experiments. Setting up the simulation scenarios to validate the FE-HBM response, it was noted that some variations in such boundary conditions can have substantial effect on the simulated kinematics. This was first reported in (Brynskog et al., 2024) and later expanded upon in **Paper III**. The boundary conditions included as parameters in **Paper III** include belt/seat friction (range: 0.2 – 0.4 / 0.2 – 0.5), seat stiffness (±10% spring stiffness from baseline), and belt bending stiffness (by varying element formulation). The seat friction was motivated by substantial differences in experimental procedures and PMHS preparation, in which some used Lycra jumpsuits and stated that the fabric to seat interaction was wet, while others used cotton shorts and stated that the interaction was dry. Furthermore, the friction force from the seat interaction loads the pelvis with a relatively long lever arm to its rotation center, which means that even small changes can have a substantial effect on the resulting moment. For comparison, force from the lap belt, the spine, and both femurs were applied by shorter lever arms since they pass closer to the pelvis rotation center and, despite potentially having greater magnitude, can produce smaller moments as a result. Belt friction was included since it was hypothesized that it would be more closely related to submarining and slipping of the lap belt. Seat stiffness was included, since the semi-rigid seats used in most current PMHS tests are built by different institutes, which could cause some variation. Furthermore, production seats in real cars include some degree of variability, which could influence the outcome in a real vehicle setting. Finally, belt bending stiffness was included since it was noted that in several of the simulations, the belt folded in towards the midline, effectively resembling a rope. It is challenging to assess this particular effect in PMHS experiments, since the belt is typically obscured by the protruding abdomen. However, by analyzing the video data it was confirmed that for several of the experiments, the trailing belt section on either side of the PMHS abdomen was folded to some degree, although exact quantification was not possible.

Four different scenarios were evaluated, one stationary with a rotating belt system, in which the belt was first pulled to a target force before the anchor points were rotated to change the belt-to-pelvis angle until the belt lost contact with the pelvis and slipped over onto the abdomen, and three sled tests with various belt and seat configurations. From these scenarios, it was found that seat friction had the greatest influence on pelvis kinematics, while belt folding has the potential to influence submarining timing. Hence, to increase confidence in FE-HBM to PMHS validations, it was recommended that the PMHS to seat friction coefficient is evaluated in future experiments and that new belt modeling methods which accurately capture belt folding, when interacting with soft tissues, are developed. Given the strong influence on predicted kinematics, it is also strongly recommended to include seat friction as a free variable in future sensitivity studies, as was done in **Paper IV**, since it is reasonable to expect seat friction to vary in real vehicles depending on, *e.g.*, seat fabric and the clothes that the occupant wears. While obtaining more precise estimates for the seat friction coefficient in specific setups is crucial for accurate model validation, it is not possible to fully control this particular parameter in real-life scenarios.

Based on the experiment data, it was not possible to confidently establish whether the simulated belt behavior was correct or not. Analyzing the results, it is of the author's opinion that a certain amount of belt folding occurs in most experiments, however, the folding observed in simulations may be excessive. The folding behavior has been noted in other simulation studies, *e.g.* (Richardson et al., 2024), in which it was decided to remove such cases from the analysis, since folding was deemed to affect the lap belt-ASIS load transfer. It is possible that this effect is not seen as much in ATD experiments or with older FE-HBMs, which have stiffer soft tissue properties, while newer FE-HBMs with more biofidelic fat tissue materials create a pocket for the belt when loaded, in which the belt easily collapses towards its midline due to out-of-plane forces. In medical literature, the presence of a symptom referred to as a seatbelt sign, an abrasion within the dermis which leaves a mark where the belt made contact with the occupant, has been associated with high incidence of significant intra-abdominal injuries (Demetriades et al., 2011). Looking at case reports showing seatbelt signs, possible examples of both a non-folded belt, which potentially causes a wider abrasion (Poplin et al., 2015), and a folded belt, potentially causing a narrower abrasion (Demetriades et al., 2011), can be identified. Based on the simulation results of **Paper III**, it was found that a non-folded belt resulted in a submarining outcome more easily than a folded belt. It was hypothesized that this effect was due to the folded belt having greater penetration into the soft tissue, allowing for a stronger coupling with the pelvis below the ASIS. However, while folding potentially could reduce the risk of a submarining outcome, it would also generate greater pressure at a smaller area on the iliac spine, likely increasing the risk of belt induced fractures. In this thesis it was decided to keep the simulations regardless of belt folding behavior, however, it is recommended that this topic receives increased attention in future research.

The semi-rigid seat was introduced to be a simplified version of a real front row production seat (Uriot et al., 2015a), enabling a more realistic boundary condition than a non-deformable rigid plate, while still being easy to share between different institutes and to replicate with computer simulations. In the original source, it is stated that the parameters were determined to reproduce the behavior of a real seat, used in two prior studies with the Hybrid III, however, neither the seat model version nor the validation results are presented in any of the publications. Since its introduction, the semi-rigid seat has been utilized in multiple simulation and PMHS studies (Baudrit et al., 2022; Boyle et al., 2019; Grébonval et al., 2021; Lopez-Valdes et al., 2024; Richardson et al., 2020; Shin et al., 2023; Somasundaram et al., 2022, 2023; UMTRI AVOK-Series, 2020-2023). However, to date and to the best of the author's knowledge, no study has been published evaluating the semi-rigid seats' capability of replicating the interaction between an occupant and a real production front row seat, or how representative the semi-rigid seat is compared to a population of such seats.

To get some insight, hip kinematics, which is partly a result of seat stiffness, can be compared between tests performed using semi-rigid and production seats. First, studying hip kinematics in frontal impact scenarios with PMHSs in a standard driving position on a production seat (Albert et al., 2018; Forman et al., 2006; Shaw et al., 2018), the general response is that the hip moves down into the seat at peak forward excursion. This forward and downward hip movement has been confirmed by automotive industry simulation engineers as typical in frontal impacts with regular production seats. For PMHS experiments with an impact velocity around 50 kph using force limited belt systems, the downward movement is approximately 30-50 mm. Second, studying a similar scenario on a semi-rigid seat (UMTRI AVOK-Series, 2020-2023), instead the hip moved up by approximately 30-50 mm at peak forward excursion. Simulating the front seat configuration from (Uriot et al., 2015a), the hip also moved up by 50 mm, but the PMHS vertical response was not published for comparison. Studying the reclined PMHS experiments on a semi-rigid seat, the hip either moved up (UMTRI AVOK-Series, 2020-2023), or was initially forced down by the lap belt pre-tensioner before moving approximately horizontal until it interacted with the submarining ramp and was pushed back up (Baudrit et al., 2022; Richardson et al., 2020). The above references on reclined occupants target a hypothetical future autonomous vehicle scenario, and so far the stiffness properties of a seat developed for this scenario remain unknown. However, using language that describes the implemented semi-rigid seat as a representative version of today's front row production seats might result in incorrect assessments of future injury risk, should the assumption be flawed. The discrepancies between hip vertical movement in production seats versus the semi-rigid seat, indicate that the current implementation (current spring stiffness) of the semi-rigid seat produces a stiffer vertical response than real production seats, effectively forcing the hip to move towards the lap belt. Seat stiffness was included as a parameter in **Paper III**, spring stiffness varied by ±10%, with the focus of evaluating smaller differences potentially caused in production at different institutes. These variations were not enough to result in substantial differences in FE-HBM kinematics and were potentially too small to capture the difference between semi-rigid and production seats. It is, hence, recommended that future studies evaluate the properties of the semi-rigid seat compared to a population of production seats and present modifications if necessary.

5.5 Evaluating Safety for Reclined Occupants

Several PMHS frontal impact sled studies on reclined occupants have recently been conducted with multiple pelvis, spine, and ribcage fractures as a result, see Table 1. Summarizing the results from sled tests with a delta velocity of approximately 50 kph, 12 of 14 PMHSs received pelvic fractures (including sacrum), 10 received spine fractures, and 11 received multiple rib/sternum fractures (results for four PMHSs by (Lopez-Valdes et al., 2024) are unavailable to date). Common for these tests is that a semi-rigid seat was utilized together with a 3-point belt system and in certain cases, an angled foot support. No other alternative load paths were included. Of these experiments, only one case of partial submarining (one side) (Richardson et al., 2020) has been confirmed, while one or two additional cases have been debated, meaning that submarining in reclined positions have been largely avoided. However, as seen by the reported injuries, that does not necessarily result in a safe outcome for the occupant.

Previous simulation studies have shown increased lumbar spine load associated with the prevention of submarining for reclined occupants (Boyle et al., 2019). Similarly, by additional analysis of the reclined simulations on a semi-rigid seat performed in **Paper IV**, a median peak compressive force of 4.8/3.1 kN and a median peak flexion moment of 153/141 Nm was found for the lumbar spine of the non-submarining/submarining groups, respectively. However, using the lumbar fracture IRF presented in (Iraeus et al., 2023), a median fracture risk of 87/94% (non-submarining/submarining) was found when evaluating a 45-year-old occupant. The seemingly contradictory outcome of greater risk at lower loads for the submarining group is explained by the fact that the peak flexion moment shifted from the lower lumbar spine to the upper lumbar spine when moving from a non-submarining to a submarining outcome, while the peak compressive force remained at the upper spine in both cases. The combined effect of both compression and flexion (at the upper spine), generated a higher risk for the submarining group despite the lower absolute values, showing that the loading scenario is more complex than just peak axial compression or flexion, individually. For both groups, the greatest risk was predicted at L1.

Table 1 – Summary of injuries seen in reclined PMHS testing
*Injury outcome from Lopez-Valdes et al. (2024) are unavailable to date.

Reference	Nr. PMHS	*Fractures*		
		Pelvis/ Sacrum	Spine	Ribcage (multiple)
(Richardson et al., 2020) *Speed: 50kph*	5	4 of 5	3 of 5	3 of 5
(Baudrit et al., 2022) *Speed: 48kph*	3	2 of 3	3 of 3	2 of 3
(Somasundaram et al., 2022) *Speed: 32kph*	3	0 of 3	1 of 3	2 of 3
(Somasundaram et al., 2023) *Speed: 32kph*	3	0 of 3	1 of 3	3 of 3
(Shin et al., 2023) *Speed: 50kph*	3	3 of 3	1 of 3	3 of 3
(Lopez-Valdes et al., 2024)* *Speed: 50kph*	4	N/A	N/A	N/A
(UMTRI AVOK-Series, 2020-2023) *Speed: 32kph*	3	0 of 3	3 of 3	3 of 3
(UMTRI AVOK-Series, 2020-2023) *Speed: 50kph*	3	3 of 3	3 of 3	3 of 3

In addition, the median risk of five or a higher number of fractured ribs (NFR5+) for a 45-year-old was estimated (Larsson et al., 2021) at 83/93% (non-submarining/submarining), indicating that even with the implemented shoulder belt load limit (parameter range: 2.0 – 5.0 kN), the chest will be critically loaded when only protected by a shoulder belt and loaded in a more inferior-superior direction caused by the reclined position. Only including cases where the shoulder belt load limit was in the range 2.0 – 3.0 kN, the median risk of NFR5+ dropped to 9%, while the NFR2+ remained at 87%. Reducing the shoulder load limit further is probably not possible, since the occupant must be restrained to avoid hitting other hard structures in the vehicle interior. Adding restraint forces that distribute the loads, *e.g.* via an airbag, could redirect the occupant's kinetic energy to the lower body, potentially resulting in a reduced injury risk for the chest while increasing the risk of submarining as a consequence.

Comparing the simulated response with the closest matching PMHS experiment, average male in 50 kph recline (UMTRI AVOK-Series, 2020-2023), found that all PMHSs sustained multiple T11-L5 spine fractures and a bilateral flailed chest. In all cases, the most severe spine fractures were located around the T12-L1 transition, indicating that the SAFER HBM accurately predicted the location of greatest lumbar fracture risk. In addition, while none of the three PMHSs had a confirmed submarining outcome, they all had pelvic ring fractures with an incomplete or complete disruption of the posterior arch. Given the substantial pelvis damage, it is unclear how relevant classic submarining definitions are, and it is possible that the belt moved into the abdomen after breaking the pelvis without the event being considered submarining. To protect against both pelvic fracture and submarining when designing new occupant restraint systems, future safety assessments and rating protocols would benefit from implementing a two-stage condition, applying a pass/failure to both fracture prediction and submarining outcome, independently.

Common for all PMHS experiments in reclined scenarios listed in Table 1, is the use of the semi-rigid seat. The reported injuries are, hence, partially a result of the interaction between the PMHS and this particular seat. If the properties of the semi-rigid seat are non-representative of a real front row production seat, as indicated in Chapter 5.4, the identified injuries and PMHS kinematics might not accurately depict the true safety risk for reclined occupants seated in current production seats. For example, if the vertical stiffness of the semi-rigid seat forces the occupant's hip to move up, while a real seat would have allowed for the hip to move down, the semi-rigid seat response will produce higher forces on the pelvis and stronger coupling with the lap belt. This would likely increase the risk of pelvic and lumbar fractures and reduce the risk of submarining, which appears to match the reported results from PMHS experiments. Furthermore, if the results of these experiments are used as the benchmark when exploring future restraint principles without considering this limitation, there is an obvious risk that proposed solutions do not target the actual risks associated with reclined occupants, and that the risk of, *e.g.*, submarining will be overlooked. In **Paper IV,** the semi-rigid seat was used to achieve a close match with the validation scenario and allow for easy parameterization. However, this has limited the findings of **Paper IV** in terms of real-world relevance for an actual vehicle environment including current seat properties. If the properties of the semi-rigid seat, as implemented, are too stiff, a real seat would likely have resulted in more submarining outcomes and the influence of the identified predictors could have been stronger, *i.e.*, further vertical buckle angles than indicated may be required to stop submarining.

Fundamental biomechanical principles for impact trauma protection state that one should aim to restrain strong body parts. The flailed chest, fractured spine, and fractured pelvis in current PMHS experiments indicate that strong body parts might already be loaded to their limit, when evaluated on a semi-rigid seat relying solely on the seat and a 3-point belt. This highlights the apparent challenge in protecting a reclined occupant at this crash severity without alternative load paths. Other principles include early coupling, distributed loading, increased ride down, lower impact speeds, and minimizing relative motion between body parts, which could, in part, be achieved by adding, *e.g.*, knee support or airbags. Such solutions are already well established in current vehicles and might be necessary for efficient protection also in reclined scenarios. Designing safe vehicles for the future will require consideration of both the risk of submarining and risks associated with submarining prevention.

5.6 Implications for PMHS testing and FE-HBM development

As described in the Background chapter of this thesis, fracture of the pelvis can occur in many ways, *e.g.*, fracture at the pubic rami, sacrum or the iliac wing due to lateral compression, iliac wing or pelvic brim due to belt loading, sacrum due to seat interaction, or acetabulum due to femur axial or lateral load. While the current literature includes a decent volume of PMHS studies on full-body lateral impacts, see Chapter 3.3.1 in (Peldschus & Wagner, 2021), information is very limited in terms of fracture properties at material level. Hence, to enhance knowledge about pelvic fracture mechanics, and facilitate the development of IRFs for pelvic fracture prediction based on material data, additional experiments on pelvic bone coupons are motivated. Specifically, it would benefit the development of future models if studies were carried out on: the anisotropy of the pelvic cortical and trabecular bone, osteon orientation over the cortical surface, population distribution of material properties, tuning and validation data for strain-based measurements, and fracture tolerance of pelvic cortical bone in multiple loading directions.

The complexity and variance in injury mechanisms from different loading scenarios means that, to assess pelvic fractures with FE-HBMs, several strain-based IRFs might be necessary. Development of such IRFs based on material testing would be an important contribution to pelvis safety assessment using future FE-HBMs. The injuries which should receive highest priority are pubic fractures in lateral impacts, iliac wing fractures from belt loading, sacral fractures from seat and/or lateral loading, and posterior acetabulum fractures from femur axial load (should knee support continue to be part of the restraint solution in future vehicles). However, the author is of the opinion that a submarining IRF based on a simulation result would be less useful. IRFs with simulation results as predictors are typically created to predict the risk of an event that does not occur in the simulation, *e.g.*, measuring the strain in a rib that never fails and associating that strain with a risk of fracture. Submarining, on the other hand, is an event which current FE-HBMs are explicitly designed to accurately predict. Consequently, a submarining IRF cannot be defined using a simulation result, that on its own classifies the submarining outcome, since this would just create a step function with a threshold value where the risk increases from 0 to 100% for the given FE-HBM occupant. Simultaneously, it cannot be a result measured at the time of submarining, since this time is undefined for the non-submarining scenarios. Instead, it must be a result which is related to the submarining outcome, without explicitly defining that outcome, and which can be evaluated as, *e.g.*, min/max recorded over the entire simulation regardless of outcome. Such a measurement could be, *e.g.*, minimum lap belt angle or belt-to-pelvis relative angle. However, the critical angle is likely dependent on the design of the restraint system and anatomy/position of the occupant, such that any variation could influence the IRF, limiting its usefulness when evaluating new systems. One hypothetical alternative would be to construct a FE-HBM that will not submarine (similar to a rib that does not fracture) and use this model to develop an IRF that predicts the submarining risk for a given simulation output. However, this would require an artificial interaction between the FE-HBM and the belt which goes against the ambition of developing a virtual human-like surrogate, that replicates a biofidelic response to omnidirectional external loading. Another alternative is to run parameter studies with FE-HBMs representing the population at large. From these studies, occupants associated with a high risk of submarining could be identified and used to check for submarining in future development/rating evaluations, see further discussion on population based evaluations in Chapter 5.7.

While the metamodel developed in **Paper IV** could be considered an IRF of submarining outcomes, it uses the initial conditions of the current occupant and restraint configuration as predictors, rather than a result of the simulation. The metamodel predicted risk should, hence, only be considered meaningful if the same scenario is evaluated. The motivation for developing this particular metamodel was that despite the threshold for each predictor being dependent on the load scenario, the significant predictors are expected to be general, such that a similar set of predictors would be identified irrespective of changes in the load scenario. This assumption is reinforced by the fact that the identified predictors have individually been associated with submarining in previous studies, however, their relative importance has not been considered. Knowledge gained about these predictors will remain useful even when designing new systems for which the metamodel was not developed, however, the identified threshold lines for non-submarining/submarining outcomes will most likely change.

To increase confidence in future FE-HBM to PMHS comparisons and avoid *"false"* model validations, obtained by tuning specific simulation settings, it is recommended that future experiments emphasize the interaction between the PMHS and the restraint. **Paper III** highlights that seat friction and belt folding properties influenced pelvis kinematics and submarining timing, which motivates a more rigorous control of these properties in PMHS tests. It was hypothesized that a substantial portion of the variance stemmed from different institutes applying different protocols/guidelines. For instance, dressing the PMHSs in Lycra jumpsuits / cotton shorts and whether the interaction with the seat was wet/dry. Harmonizing PMHS testing protocols regarding, *e.g.*, clothing and preparation, or including test-specific evaluations of these parameters, would facilitate more robust validations of FE-HBMs. Furthermore, since pelvic shape is poorly predicted by overall anthropometry, as presented in **Paper I**, it is recommended that future PMHS experiments targeting the pelvis include detailed descriptions of the resulting pelvic shape.

Running population based FE-studies prior to the PMHS experiments, to identify potential measurements that should be controlled, would be another approach to enhance PMHS experiments for FE-HBM validation. For example, in **Paper IV,** iliac spine hook angle was identified as significant for the submarining prediction and a correlation was identified between the iliac spine hook angle and the hip center (H-Point) to ASIS distance. Having access to this knowledge prior to a PMHS experiment, collection and reporting of these measurements could be included in the test protocol, and subsequently be used to confirm the FE-HBM prediction or indeed as a measure to include more subject specific validations. Having a population based understanding of the expected experimental outcome would also increase the likelihood of the experiment being successful.

Due to pelvic shape having been identified as an important variable for both lateral impact and submarining, as well as being poorly predicted by overall anthropometry, the author is of the opinion that FE-HBMs should target a population shape average, rather than using the shape of an "average" individual. By this approach, certain challenges are involved in terms of assembling a complete body, however, the pros are believed to outweigh the cons. Furthermore, population shape averages would reduce the risk of different FE-HBMs providing different result in injury or kinematic predictions, due to a subject specific shape variation associated with that particular FE-HBM. This would lead to more robust safety assessment across different FE-HBMs when implemented in future rating programs or legal requirements. To emphasize safety for the entire population, future rating protocols could even be specified for the shape average of a vulnerable sub-population, associated with each evaluated loading scenario.

5.7 Implications for Automotive Safety Assessment

Augmenting current physical (real-life) crash tests for automotive safety assessment with simulation based (virtual) evaluations has the potential to open the doors to more realistic assessment of safety. This potential has been recognized by consumer crash safety rating organizations, such as Euro NCAP, who have stated that they will complement their crash testing with FE-HBM evaluations once viable HBMs become available (Van Ratingen & Jacobsen, 2022). Physical tests are often limited by physical and monetary restrictions, while simulations on the other hand, are much less restricted by such limitations and enable the inclusion of real-life variability. This variability can originate from many sources, *e.g.*, crash severity, crash configuration, vehicle design, and occupant posture or anthropometry, all relevant factors to assess true safety performance of a system rather than just the few highly controlled scenarios included in the physical tests. While the relevance of physical tests would continue to be high, as would replicating them with simulations to build confidence in the validity of the simulations, the true potential of simulation based safety is achieved when computers are utilized at their optimum, *i.e.*, when processing large amounts of data, far beyond what any physical experiment could ever include. However, an issue which must be considered is the model's validity range, and the confidence in the simulated result over the whole spectrum of variability that is being evaluated. Since it is not possible to run physical tests of every combination of input parameters, it is important that tests for model validation cover a broad range of possible scenarios, in order to safeguard the FE-HBM and make it robust to variations for which it has not been explicitly validated. In addition, to build confidence in the model predictions, it would also be advisable to not only consider a-priori validation of the model but instead try to verify the model predictions by confirmation in real-life data. For example, if a population based evaluation renders a certain cohort of the population at higher risk, future studies could aim to confirm/reject this finding through epidemiological studies or other experiments.

Expanding the evaluation from a single physical test to a population based safety evaluation using simulations, could develop the assessment from a single pass/failure to an outcome based on probabilities. This opens the discussion for what can be considered *acceptable risk* for a population of occupants when designing robust systems, since it will not be feasible to achieve zero risk of injury for all occupants and all crash scenarios. To keep the possibility of traveling at high speed in a flexible and affordable manner, such as been accomplished through cars, a certain level of risk must be accepted. Population based studies could also be used to identify the *occupant at risk*. This would be an occupant from the complete population with characteristics such that it can be evaluated at risk of a specific injury, when subjected to a given loading scenario. Instead of running a full population based evaluation in every future design iteration, a pass/failure assessment of the occupant at risk could be considered a proxy for meeting the acceptable risk criteria on a population level. This is a useful method for assessing, *e.g.*, submarining risk where, as discussed in Chapter 5.6, IRFs are considered of limited use. However, the occupant at risk would vary with the loading scenario and injury target, as observed by studying epidemiological data in which different cohorts of the population are associated with different injuries. Hence, to achieve complete vehicle safety assessments for the population, a set of occupants would need to be evaluated, similar to what is proposed in (Larsson et al., 2024). This thesis does, however, not aim to answer what would be considered acceptable risk, and more research as well as ethical considerations, are required by the community at large.

A population based approach was utilized in **Paper IV** to simulate the risk of submarining for a reclined occupant in frontal car crash scenarios, using a simplified vehicle setup with the semi-rigid seat, and an occupant at risk was defined based on pelvis angle and iliac spine hook angle. Based on the results, this study indicates that the current legal requirements on buckle angle, if the restraint system only consists of a seat and a modern 3-point belt, might need to be shifted towards more vertical angles to efficiently protect against submarining for reclined occupants. This showcases how the above argumentation around population based safety assessments can be implemented to evaluate both current and future systems.

5.8 Limitations

As with all research, limitations should be considered when reading this thesis.

First, the data used to generate the pelvic shape model all came from a single source. This means that even though the sample is representative of a modern US population based on age, stature, and BMI (Fryar et al., 2016), further generalization based on, *e.g.*, ethnicity, is not possible and cannot be analyzed with the developed FE-model. Furthermore, while the pelvic shape model is built based on a relatively healthy sample of 132 individuals, it is not possible to capture the tails of the population variance with this sample size. Seeing as the occupant at risk is likely defined some distance from the mean of the population, a risk that the accuracy of the shape model degrades somewhat at this point has been noted. In **Paper II** and **Paper IV,** this limitation was considered by capping the shape model at ±2SD. While this means that the population variance predicted by the model is never fully evaluated, it is recommended that future usage of the model also includes similar caps to avoid unrealistic shapes at the tails of the distribution.

Second, the aim of this thesis was focused on the pelvis in automotive safety assessments. As such, some boundaries were required in terms of research scope. This includes simplifying the hip, knee, and ankle joints of the model to kinematic joints with a prescribed stiffness (variation in joint properties were not considered). In addition, only a single spine was considered, meaning that variations in spine alignment, which could be associated with pelvic shape variations, were not included in the analysis. While the distribution in resulting pelvis angle was comparable with the reported distribution from volunteers seated in a standard automotive seat (Izumiyama et al., 2018), indicating that the resulting pelvis position is realistic, the exclusion of spine variance means that the sacrum to lumbar spine coupling might not be fully representative. A correlation score of 0.49 was computed for pelvis angle and lumbar lordosis for a seated male subject using the data from (Izumiyama et al., 2018). What effects a variation in lumbar lordosis could have on the reported results, remains unexplored.

Third, while **Paper I** aimed to build a shape model for the entire adult population, in line with the main objective of this thesis, **Paper II** only included residual variability around a 50th percentile female/male, and **Paper IV** only around the 50th percentile male. This means that a lot of the occupant variance in pelvic shape and surrounding structures remains unexplored and that conclusions from this thesis should be considered more as a starting point than a complete picture. The decision to focus on variability associated with the 50th percentile male was partly due to a lack of validation data for female occupants and partly due to, at the time, limitations in full-body morphing capabilities with the latest version of the SAFER HBM. However, the mathematical description of pelvic shape in **Paper I**, and the generic pelvis FE-model in **Paper II**, include the capability of running similar evaluations for female occupants in the future.

Fourth, with great occupant variance comes great parameter spaces. Even though computers can handle running many more evaluations than physical tests, the complexity of these models presents restrictions on the number of runs that can be performed given available computer resources. As the parameter space grows it becomes impossible to achieve full coverage. In **Paper II**, this was addressed by using an approximative GSA method to greatly reduce the required sample points. In **Paper IV**, this was achieved by space-filling sequential sampling of the parameter space and evaluation of metamodel convergence, to assess if sufficient sample points were included. Both versions are attempts at running the minimum number of samples while still being able to address the current aim. Including variability does not only create large parameter spaces, but it also adds challenges relating to parameter distribution and potential correlations. Knowledge of how a parameter varies within the population, and which distribution it follows, is often lacking and typically requires some assumptions. When population data are pooled from different resources it might not be possible to evaluate if correlations exist. Missing such correlations can lead to extreme combinations of parameters that would not occur in real life.

Fifth, as in most current PMHS studies, much of the work has utilized the semi-rigid seat. In the original source, this seat is labeled as a front seat configuration, however, a comparison against a real production seat was never published. As outlined in this thesis, a discrepancy in PMHS vertical hip kinematics on production seats versus on the semi-rigid seat, can be identified. As such, it remains unclear whether the findings of **Paper III** and **Paper IV** are mainly relevant to understand past PMHS tests, in which a seat stiffer than current production seats appears to have been used, or if the findings are also relevant when extrapolated to real car seats. It is also unknown whether future seats for reclined occupants will have stiffer seat properties than current production seats, making them comparable with the semi-rigid seat.

Finally, all FE-HBM results were generated with the SAFER HBM. This model does, by design, not simulate fracture, since fracture is a very chaotic and subject specific event. Post-fracture, the restraint system has already failed, making it less relevant as a design target for developers of vehicle safety systems. Instead, the risk of fracture is predicted using IRFs. As such, the predicted FE-HBM kinematics post-fracture for a PMHS can be questioned, which may potentially have influenced the simulated submarining outcome. While substantial efforts have been made to validate this model in multiple loading scenarios, a model is still a model approximating reality. It has previously been said that *"all models are wrong, but some models are useful"*, highlighting the need to understand a model's limitations and intended use, which should always be considered before trusting the result of any simulation.

6 FUTURE WORK

To further advance pelvis computational models for automotive safety assessment, future work should initially target material-based fracture prediction of the pelvis, improved control over both experimental and simulation boundary conditions affecting the pelvis, and full-body validation of other than 50th percentile male occupants.

The first target requires additional experiments on pelvic bone material properties in multiple loading scenarios. When such data has been gathered, IRFs for pelvic fracture risk relating to lateral loading, belt loading, sacral compressive loading from the seat, and femur axial loading, should be developed. These IRFs must be evaluated in conjunction with other safety targets, such as submarining outcome, to avoid the risk of protecting against one injury at the expense of another. To successfully develop the IRFs, future experiments aiming to fill this research gap should consider failure in more than just tension and mapping the osteon orientation over the pelvic cortical surface, to include anisotropy in the material models.

The second target could be addressed by harmonizing PMHS experiment testing protocols regarding, *e.g.*, clothing and preparation, or including test-specific evaluations of these parameters. For simulations, new belt modeling methods that accurately capture belt folding when interacting with soft tissues should be prioritized. Regarding the semi-rigid seat, it is strongly recommended that future studies are carried out to validate its response against a population of real production seats, to also include evaluations on the effect of softer seat properties and avoid incorrect assumptions regarding future safety challenges affecting reclined occupants.

The third target should aim at full-body validations in both lateral and frontal impacts, where the frontal impacts should primarily focus on belt-to-pelvis interaction for female and obese occupants. Experimental data which enables validation of both lateral impacts, *e.g.*, (Lebarbé et al., 2016; Petit et al., 2015), and frontal impacts (Somasundaram et al., 2022, 2023) are available in the literature, and should be considered.

7 CONCLUSIONS

With the main objective of enabling pelvis related automotive safety assessments for a population of female and male vehicle occupants, by introducing advanced methods in FE-HBM development and providing guidance for future research, this PhD project has:

- Developed a shape model describing 90% of pelvic bone population shape variance and identified that only approximately 30% of the variance is predicted by sex, age, stature and BMI. This is an important finding considering that state-of-the-art FE-models, population based simulation studies in traffic safety, and PMHS experiments, typically use such predictors to define different population cohorts. There is, hence, an obvious risk that substantial variations in pelvic shape are overlooked by these efforts.
- Developed and validated a generic morphable pelvis FE-model capable of running population based safety assessments including both shape and material variation, and integrated this model in a full-body 50th percentile male FE-HBM (SAFER HBM v11). Validation has been done for lateral loading on component level (denuded pelvis) and for pelvis kinematics and submarining outcome of full-body upright and reclined occupants in frontal car crash scenarios.
- Identified a strong influence on simulated pelvis kinematics and submarining outcome from uncertain boundary conditions in current PMHS experiments. Specifically, this relates to seat friction, simulated belt folding kinematics, and semi-rigid seat stiffness properties. If not considered, this could influence the validity of FE-HBMs and potentially hide the actual risk associated with, *e.g.*, reclined occupants, resulting in incorrect safety prioritizations.
- Using the developed pelvis FE-model, identified that the predicted pelvis response in lateral impacts is equally affected by variability in shape and bone material properties. This suggests that future assessments on lateral loading to the hips should equally consider both these aspects.
- Using the full-body FE-HBM, identified that occupant variability is comparable with restraint design variability on submarining outcome in frontal car crash scenarios comprising reclined occupants, which warrants consideration in future automotive safety assessments.

To conclude, this thesis advances the field of pelvis computational models for automotive safety assessment by implementing methods that enable population based evaluation for development of robust vehicle safety systems.

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Part B

Appended Papers