The Physical Exoskeleton-User Interface: Investigating Modelling and Design of Coupling Surfaces

by

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Author's Declaration

This thesis consists of material all of which I authored or co-authored: see Statement of Contributions included in the thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

Statement of Contributions

Foreword: The contents of this thesis include presentation of conference abstract works expanded for this thesis, pre-publication drafts for full journal publication, and literature and design perspectives on the field of exoskeleton coupling interfaces.

The majority of the contents of this thesis are slated for publication as articles in journals focused on wearable robotics or rehabilitation engineering. Many were presented at conferences which informed the longer form studies, and are also planned to be extended to full publication. This section lists, either the plans for publication, or successful submissions to conference abstracts. Plans to extend conference abstracts into full publication will be noted.

The chapters containing content that has been submitted or is planned to, is listed below. The chapters containing conference abstracts are mentioned specifically:

- Chapter 3: Evaluating the Effects of Customised Versus Generic Surfaces on Lower-Limb Exoskeleton Kinematic Performance: An In Vitro Mannequin Test Bench Evaluation
- Chapter 4: Impact of Customized Coupling Surfaces on the Performance of Lower Limb Exoskeletons: A Pilot Study
- Chapter 7: Validation of Common Models: Perfectly Coupled and Linear Spring-Damper Approaches and a Simple Extension of the Linear Spring Damper
- Chapter 8: A Novel Method for Modelling the Physical Human-Exoskeleton Interface

Abstract

The goal of Lower Limb Rehabilitation Exoskeletons is to provide active and passive support to the user for rehabilitation goals focused on the restoration of function and independence to the user. In circumstances where rehabilitation is not possible, many of these devices can be utilised as fully assistive wearables. The target populations for these devices vary, but often include individuals with varying degrees of mobility impairment. A critical component that defines and governs the relationship between user and device, and often remains ignored in study, is the exoskeleton coupling interface. The interface imparts forces to the user, and as a result, directly influences the safety and performance of the exoskeleton. An optimized balance between user safety and performance must be met. This thesis is focused on and motivated by assistive rehabilitation robotics, with many of the concepts introduced here extending to wearable interfaces.

Injury risks include interaction forces creating joint misalignments that cause undesirable loading and high, prolonged surface pressures. The current tools for evaluation, sensors and models, are adequate at estimating the basic components of user safety but do not readily inform design for newer exoskeleton coupling surfaces. This is a result of a lack of standardisation and transferability in analysis. Lack of baselines, difficult to utilise metrics, relatively simple models, and inadequate methods for evaluation currently limit evaluation and innovation of new exoskeleton coupling interfaces.

The focus of this thesis is to build upon the groundwork laid by prior academic research to identify the drawbacks of existing exoskeleton coupling surfaces, and to improve upon them. This is done by investigating the most commonly approached and evaluated subtopics of exoskeleton coupling design: 1) iterative and sensor based design processes and 2) modelling based evaluation. To improve upon the sensor based processes, a highly customised fully conforming surface inspired by orthoses and prostheses design was developed to introduce new baselines for coupling interface design as a "best case scenario" supported by pressure and joint misalignment related metrics. Modelling of the exoskeleton coupling surface was investigated by first evaluating commonly used models, then developing two novel models with the express intent of addressing initial strapping conditions and estimating pressure conditions prior to manufacturing. The second of these two models, based on elastic foundations, indicated that strapping conditions could be estimated within 10% accuracy of real conditions without a manufactured surface.

Contributions to both of these subtopics made in this thesis were with the goal of developing better standards and transferability of information in mind. With these tools, better comparison of the exoskeleton coupling interface can be accomplished that is informed by established baselines, models, metrics and methods for evaluation.

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I would like to thank those who helped me in the completion of this thesis.

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Dedication

To Naomi and my family and friends. I likely would not have been able to finish this without your constant words of encouragement, and the support that you gave me to lean on you when I needed it most. I love you all, and I know you'll be there to support me through the next one.

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Table of Contents

List of Figures

List of Tables

Chapter 1

Introduction

In Canada alone, roughly 9.6 percent of the population (2.7 million) people over the age of 15 have some form of mobility impairment [1–3]. Of that group, 80,000+ people are affected by spinal cord injury (SCI, Complete or Incomplete), a severe sensory-motor impairment that vastly reduces the individual's capabilities for upright independent mobility. While solutions do exist for some mobility impairments, the majority of tools or strategies are designed to mitigate the symptoms of, and help rehabilitate or recover, mobility capabilities. Instead of completely solving, or removing the source of injury (i.e., damaged spinal circuits), these tools and strategies seek to alleviate impairments and bring independence, confidence and mobility in some fashion back to the individual. Tools such as walkers, wheelchairs, crutches and assistance bars provide users with independent mobility that increases their quality of life. Rehabilitation strategies provided by clinicians are focused on restoring mobility to individuals, and/or providing strategies to improve their independence when mobility is lost completely [4, 5]. Supervised strategies involving assistive devices and rehabilitation regimens are common for individuals with mobility impairments, such as individuals with SCI [7–9]. Rehabilitation goals are tuned and designed by teams of rehabilitation clinicians and caregivers to aid in individual in regaining mobility, but also independence and for improving quality of life. For individuals with chronic mobility impairments without the potential for regained mobility, clinicians may instead focus on alternative strategies tailored for quality of life [10].

With the rise of an aging population, and a decline in individuals trained as clinicians for rehabilitation (e.g., occupational therapy, physical therapy, nursing), there are fewer workers to deliver acute and chronic care rehabilitation [11]. To compensate, new strategies and devices are researched to reduce the demand on clinicians required to provide quality rehabilitation care. Active rehabilitation tools, such as devices which provide active actuation assistance or sensing, are at the forefront of rehabilitation care and engineering [12–15]. These devices are designed to reduce the physical work clinicians are required to perform, and improve the efficiency and quality of care. By relieving clinicians of physical components of therapy, active devices can free up more time for clinicians and patients for additional therapy time and reduce injury risk on either individual [16, 17]. For activities of daily living (ADL), active assistive devices offer a greater opportunity to bring independence to the user by aiding or fully replacing a necessary function, providing increased safety to the individual, and/or reducing energy cost. Recently, the use of lower limb exoskeletons have been proposed as one such device. The intent of these devices are to aid and alleviate many of the struggles and complications of the rehabilitation process, offering gait assistance to individuals with mobility impairments in clinic and ADL settings [18–20, 39].

Lower limb rehabilitation exoskeletons are active and passive wearable devices that augment, or replace, the gait and muscular activity of their user [37, 47, 76, 77, 135]. With a multitude of designs and applications based on the needs of the user, or the problems the device seeks to address, the use cases and design scope of these devices is wide. From exoskeletons designed for a singular joint or section of the leg, to those that span multiple joints and/or limbs, there are a number of design approaches to mitigate targeted mobility impairments. The primary goal of most lower limb exoskeletons is to augment or replace the capabilities of the lower limb. By doing so, the goal is to assist in generating required joint moments for movement, reduce energy consumption, improve independence, reduce pain and risk to the individual, and aid in the rehabilitation process. This process can occur under supervision in clinics, hospitals, and in the home.

While not (yet) widely adopted, the perception of exoskeletons as rehabilitation tools is generally positive [12, 16, 17, 21–27, 33, 39]. The prospective opportunities these devices bring as tools for rehabilitation, autonomy and empowerment to individuals with mobility impairments drives the interest in utilising and implementing these devices in clinic and home settings.

Additionally there is evidence that exoskeletons perform at parity to traditional rehabilitation techniques when observing independent gait rehabilitation in controlled clinical/hospital task spaces, evidenced by the Lokomat system and other exoskeleton systems in training [20, 22, 28, 29, 34]. The proof that these devices are at least as effective or perform better than traditional rehabilitation metrics does exist, and helps justify their use as tools to assist rehabilitation specialists by reducing the manual load tasks common in rehabilitation [18–20, 23–27, 30–32, 35].

There are many challenges in designing and fabricating exoskeletons that hold back their potential as rehabilitative and augmenting wearable devices. The focus of this thesis is on identifying and characterising the exoskeleton coupling interface.

1.1 Research and Challenges in Exoskeleton Design

The conceptual design of lower limb augmenting exoskeletons has been around since the early 20th century, with designs dating back even further [36, 37]. While the concept of human augmentation for able-bodied (and impaired) individuals has been a topic in science fiction and popular media for decades, only recently have designs, components, and fabrication methods that mimic or reach the point of fiction been realizable. Fully assistive and augmenting lower-limb exoskeletons with active actuators for any number of applications, such as the H3 Exoskeleton (Technaid, Spain), Twiin (IIT, Italy) and the Indego (Parker Hannifin, United States), are relatively recent advancements made to the space. These kinds of exoskeletons are fully actuated at multiple joint centres, providing torque and power for complete and partial assistance strategies. Alternative designs exist that target a single joint, provide active or passive (i.e., stored elastic energy) actuation, and/or dampen the potential for injury to the user during a specific task. Regardless, these designs have only been seen in the last 20 years, with the BLEEX (Berkeley, United States) [38], a load bearing able-bodied assistive exoskeleton, laying the initial groundwork early on for full wearable devices. As a new and active field with potential for reducing rehabilitation cost, improving independence for users with mobility impairments and reducing clinician and user-patient load, there has been a resultant increased focus in academic, and industry research to develop them.

Lower limb rehabilitation exoskeleton research has focused on mechanical and actuator design, controller development, and multi-body system modelling, due to their impacts on performance [36, 37, 40–54]. As active rehabilitation tools, they are designed to follow predetermined or calculated trajectories, provide torques, and predict intent to drive or assist gait. Mechanical structures hold actuators together and provide a link between actuators and joints, allowing transfer of forces and torques. Actuators provide the necessary torque and speed relationship to generate desired trajectories, and provide complete or partial assistance. Controllers are designed to determine control signals to the actuators given state information, trajectory, and rehabilitation goals. Simple dynamic or kinematic models, and multi-body system modelling is used to inform exoskeleton and controller design by providing simulation data, including kinematic and kinetic data for desired trajectory goals, and locations for the placement of mechanical structures and actuators. The fields of mechanical and controller design for lower limb exoskeletons are the most popular in academic and commercial research, as they are both key requirements for an exoskeleton to function, and at present have the most evident effect when developed and implemented, on the function and performance of the device as an assistive or augmenting tool [37, 40, 46, 53, 56–62]. These components are fundamental to the performance and success of exoskeletons, and warrant the effort and research into them.

Despite the aforementioned advancements in controllers, actuators and modelling, exoskeletons are still limited in their performance, with issues that are not specific to any one sub-field of research. This includes seeing relatively low performance for energy reduction, and high variability in gait rehabilitation outcomes (compared with traditional rehabilitation) [15, 32, 35, 63–68]. In some cases, performance improvements are met, but only in highly limited testing scenarios (e.g., steady-state gait only). Models and simulation still suffer inaccuracy when it comes to tracking and estimating device effects on users [69–72, 75]. User injury and safety concerns remain a challenge, despite the implementation of physical stops in actuators, or fail safe controllers designed to minimize the risk of accidents or injury [37, 40, 63, 68, 76, 77].

User safety studies focus primarily on two modes of injury: falls and musculoskeletal injuries as a result of over-torquing [68, 76, 78–89]. Other modes of injury still occur, including the potential for pressure injuries forming as a result of interaction forces from physical human-robot interactions (pHRI). Even in highly advanced exoskeletons, joint and segment misalignment as a result of internal relative movements still occur, reducing the effectiveness of the device and raising the potential for musculoskeletal and/or skin injury [76].

The effectiveness, modelling, control and safety of the exoskeleton are critical in the performance of the exoskeleton. Until recently, the physical exoskeleton-user interface coupling the device to the user the individual utilising the device has received little attention or focus [76, 77, 89–92]. While it is technically necessary for there to be a coupling point for an exoskeleton to function, it is often overlooked and underrepresented in literature and industry due to its inherently passive and (on the surface) simplistic appearance. As a result, designs for interfaces are not well defined. Considering the interface directly contacts the user and transmits forces and torques while securing and keeping the individual strapped inside the device, it has the potential to affect many aspects of exoskeleton design [76, 77, 82, 89, 93, 94]. This includes, but is not limited to: how the user experiences forces transmitted to them, the potential for injury ranging from skin to musculoskeletal, and the performance of controllers and simulations.

1.2 The Physical Exoskeleton-User Interface

The exoskeleton coupling interface consists of a series of points and surfaces by which forces and torques are transmitted from the exoskeleton to the user, and vice versa [76, 77, 89]. These surfaces are typically designed as cuff- or orthotic-like structures that provide support through tension and rigid body contact mechanics. A number of design approaches have been applied in the fabrication and manufacturing of coupling interface surfaces. Fully soft cuff-like structures are more popular in soft exoskeletons [89, 95], while combinations of hard and soft surfaces are seen in the likes of fully assistive exoskeletons such as the H3 (Technaid, Spain) or the Indego (Indego, United States). The exact composition of interface surfaces may vary, but in order for the exoskeleton to function it must meet minimum requirements. Each surface must: 1) provide a point of contact coupling to transmit force from device to user, and 2) cannot allow for the complete separation of contact or removal of the device resulting in the user falling out of the device and risking injury. These requirements do not define how to optimize or make the device effective; in fact only considering these two components are likely to result in an ineffective and painful device [89]. The exact composition, position, surface area and other design criterion that define the final appearance and function of the device are topics in exoskeleton design research with the goal of ultimately delivering optimal force-torque interactions while keeping the user as safe as possible. The goal is to design an optimal surface that maximizes user safety while optimizing the transference of force in a smooth and effective way without pain or injury. The functionality, safety and gait assistance performance of the exoskeleton as a whole is dependent on the device's ability to optimize the transference of force, such that [89, 90].

As the point where the user directly interfaces with the device, it is clear that there is potential for the exoskeleton coupling interface to influence the performance of the exoskeleton. The first, and most prevalent influence that coupling surfaces have on the user is safety. The forces transmitted from exoskeleton to user for the assistance of kinematic-kinetic movement are of those at the physical exoskeleton interface, and those experienced as a result of dynamic reaction(s). These types of interactions are necessary and vital to the function of the device as a whole, and without them the device is not assisting, merely following or predicting. As a result of this interaction, forces need to be imparted to the individual in the form of surface pressures and joint load. The goal, ultimately is to find the optimal combination of pHRI, that minimise user injury, and optimize functionality. Problems arise when describing or characterising these interactions cannot be known or performed reliably. If the forces transmitted through the interface are occurring in unpredictable directions and magnitudes, high forces, shear stresses, and joint misalignments can occur and increase injury risk. These unpredictable components are in excess, and while their baseline components are necessary for force transmission to occur properly for the exoskeleton to be assistive, any additional, extraneous components will affect performance and user safety.

Joint misalignment and offset injuries primarily pertain to the resultant reaction forces and moments that occur at an individual's limb segments as a result of the elastic force-position relationship. When joint axes (both robot and user) are aligned, reaction forces are minimised, but still occur. Small misalignments can cause large reaction moments and normal forces to arise in the joint that are not the desired imparted ones. This issue is only more recently addressed (in the last 10 years). However a good understanding of misalignment has been established as a result of robust kinematic analysis framework relating the resultant offset to the forces that may result in injury. Joint misalignment occurs as a result of natural initial offset during initial donning, as well as the "off radius" movements that occur due to kinematic constraint design during use, and any elastic or frictional that may cause sliding [81, 82, 89, 96–100]. Joint offsets occur when elastic components of contact cause the joint centers to misalign similar to joint misalignment, but not as a result of kinematic holonomic constraints. The characterisation of this offset, or slippage, is much more recent and is highly dependent on the interface's initial strapping pressures and elastic mechanics [69, 93]. The resulting effects, however, are similar and are primarily dependent on the coupling interface design. The design, specifically the elastic and strapping pressures, of the coupling interface directly influence the risk of musculoskeletal injury from unwanted offsets and misalignments causing excess load in joints and segments [70, 89].

As a result of pressure at the interface, and the geometry of the coupling interface, skin injury is also a potential risk to the user. The modes of skin injury occurrence vary, typically presenting as some form of capillary collapse and resulting tissue deoxygenation, or damage done to the tissue from superficial friction wear, or deep tissue shear [101, 102, 154–160]. With such a wide variety of potential sources for skin injury, consideration of how the interface applies both normal pressures (magnitude and area), as well as the shear and frictional components interact with the individual is crucial for preventing injury [89, 101, 102]. Currently, most methods of evaluation involve evaluating existing surfaces for both normal and shear pressures, but not alternative methods of reducing interface pressures [89, 93, 103–107]. High pressures are not necessarily the only cause for formation of pressure injuries, and therefore designing away from high pressure is not sufficient. High, but radially uniform pressure in areas with high capillary oxygenation, still allows for tissue oxygenation as a result of flow rate being uniform if applied uniformly and appropriately [108, 109]. These considerations are critical in the design of the interface, as simply minimising the interaction forces entirely will reduce the transferred assistance from device to user. Instead these interfaces must be designed so that the required forces can be transferred safely, uniformly and ideally distributed in a way to minimize excess shear and peak pressures. Without proper design considerations of how the force is transmitted to the user, and how to properly manage forces and misalignment, the individual will be at a heightened risk of coupling-related injuries.

While user safety is a critical component, there are additional influences that the coupling interface has on the performance of the exoskeleton. The second is the influence of the exoskeleton coupling surface on the kinetic, kinematic, simulation and controller performance of the exoskeleton. For kinematic and kinetic performance, if the exoskeleton coupling surface is poorly designed, relative internal movements or unwanted kinematic configurations may occur. These unwanted configurations may be caused by a number of sources, but likely arise from elastic contributions of both the surface and user and the stored energy as a result of the overlapping geometry [70]. For example, two surfaces made of the same materials, with varying surface area will have different total compliant properties, and differing kinematic-kinetic trajectories during use that may be non-linear as a result of complex user-exoskeleton interaction geometry.

Additional errors may arise as a result of user input, or actuator torques exacerbating those compliant interactions. For the exoskeleton as a whole, models of the coupling interface acts as the governing model predicting the relative position between user and device, as well as forces transmitted at each point. If modelled inaccurately or without the required complexity, particularly the compliant interactions, predicted interactions are likely to be inaccurate to real world scenarios, and may limit the effectiveness of simulations and controllers. The highly elastic and non-linear interactions of material indentation make modelling and evaluating challenging for the simple approximation methods that exist for the coupling interface. As the field currently exists, only the simple characterisation of these elastic components is possible, meaning determining the elastic response of these systems (or designing towards a desired one) is nearly impossible prior to fabrication [89, 116].

In similar fields of study for passive wearable devices (e.g., orthotics), the ideal solution is a device which conforms exactly to the anthropometric and ergonomic needs of its user, tuned and refined by clinical specialists [110–115]. To support this practice, study and comparison on the importance of fit with quantifiable metrics in pressure and force distribution, energy consumption and rehabilitation goals have been used to reinforce their use [110–115]. While this design may be optimal for exoskeleton users, a combination of cost restrictions and infeasibility has limited their uptake in academic study [89]. Furthermore, there is no available evidence examining customized fit approaches on active lower-limb exoskeleton performance, and whether the same benefits would be prevalent in fully active systems.

Review and perspective studies on lower limb exoskeletons have addressed the importance of the interface, and its potential influence on the effectiveness of the exoskeletons performance, user safety and satisfaction, but it has not been able to quantify or explain these influences in a generalised and reproducible manner [76, 77, 89]. This lack of standardization has resulted in a wide range of information, perspectives, metrics, sensors, and design approaches on exoskeleton coupling with little quantifiable evidence (e.g., comparative studies). The general perspective and understanding of how exoskeleton coupling interfaces influence safety, control, kinetics, and kinematics is difficult to assess. The understanding of how these coupling interfaces directly influence the user, and how their design can be improved and innovated upon is crucial in further exploring their quantifiable influence and capabilities as components of the exoskeleton system as a whole.

To estimate the impact that exoskeleton coupling surfaces, and the underlying design rationale have on the performance of the exoskeleton is challenged by a lack of available evaluation methods and models. As a result, explaining how new surfaces can be developed to address or mitigate those issues compared to older designs, or how they can be used to improve exoskeleton performance, is also difficult. There is a lack of standardization and understanding of why certain decisions are made in the coupling surface design process. Many processes, designs and models are built upon ease of access for simplicity (to the designer), and metrics which utilise thresholds that are better for preliminary evaluation than to inform design. As a result, justifying new and novel designs is difficult, and comparing or building upon others is nearly impossible. Baselines (or standards) help our understanding of what we should be comparing against for all facets of evaluation. Currently, there are no baseline standardization known to the author, making it difficult to compare candidate designs against each other. While the individual studies contribute to preventing injury, and identifying points of injury, it is difficult to assess and compare new surfaces against them. There are a number of different approaches to designing and evaluating the coupling interface, including using different metrics, methods for evaluation and models for simulation.

Metrics define what is measured and how sensors are implemented into testing. The majority of metrics that define exoskeleton coupling interfaces rely heavily on explicit thresholds based on reported values such as pain pressure threshold, maximum pressure tolerance and pressure discomfort threshold [89, 90, 92, 120]. These tools for evaluation are useful, but are limited as a result of requiring contextual clues regarding the local geometry of the environment and location of the pressures. Utilised somewhat recently, kinematic offsets and joint misalignments are implemented as metrics for evaluating the potential injury as a result of undesired torques and forces [81, 89, 94, 97, 98, 117–119, 121, 135]. While still threshold based, these type of metrics are more desirable in study as they can be transferred easily between exoskeletons.

Methods for evaluation are the conditions by which test procedures are executed primarily focusing on the participants, constraints and tasks. For most exoskeleton testing, the procedure involves the recruitment of either entirely able-bodied individuals, or individuals with mobility impairments. Participant studies are the most common type of exoskeleton evaluation, and are useful in evaluating and determining the potential benefits and drawbacks of exoskeletons in real use cases [53, 68, 76, 77, 89, 122]. These test conditions are useful for evaluating the performance of the exoskeleton's gait augmentation and rehabilitation capabilities, but less so for safety. Without testbench evaluation information (under highly controlled conditions), the knowledge of what components of the coupling interface are actually influencing safety, and what is the user compensation effects, is difficult to parse. Some studies have employed mannequins for coupling interface evaluation prior to implementation which allowed for the identification of "inherent" high pressure locations induced by the design and adjusted accordingly [103]. Without a testbench evaluation with controlled conditions, isolating where sources of injury may occur from is much more difficult, and risky for the individual if testing is required for basic characterisation.

Modelling provides information for simulation and control of the relative positions and interaction forces that occur between user and device without the need for sensors. There are few established analyses that explicitly look at and evaluate the benefits, drawbacks and concessions of the most commonly used models. The baseline information of how accurate these models are and their drawbacks or inaccuracies in controlled scenarios is only now being investigated, with publications considering the effects of initial securing influencing parameter coefficients, indicating there is a potential for high variability based on initial conditions [69, 93]. Studies which put in considerable effort for robust estimation still see error around points with high acceleration indicating unmodeled non-linearities [70].

1.3 Problem Statement and Objectives

This section contains a distilled and focused problem statement on the field of coupling surfaces. The objectives of this thesis are described in this section, as well as the overarching goal and purpose of the thesis.

The field of exoskeleton coupling interfaces is still relatively under-investigated compared to advancements in mechanical linkages, actuators, controllers and multi-body systems. There are a number of different design questions related to exoskeleton coupling that are relatively unaddressed due to the lack of research that may have a major impact on the performance of the exoskeleton. To name a few, unaddressed design questions range from how the interface should be shaped, what materials should be used, where the interface(s) are placed for optimal force transmission, surface area (maximal) considerations, interface modelling in controllers and simulations, acceptable donning and doffing methods, securing stiffness limits, and shear and normal forces limits. The focus of this thesis is not only on addressing some of these design questions, but primarily on developing new tools for their assessment.

Problem statement: There is a need for novel methods to evaluate and model the exoskeleton coupling interface to better understand why and how we can create better surfaces. For the purpose of this thesis, this will be approached through heuristic and iterative design processes, and modelling frameworks.

The focus of this thesis is expanding the tools and means by which exoskeleton coupling interface evaluation occurs. Two "sub-topics" of coupling interface design are the focus of this thesis. The first of these fields, comprising the proceeding three chapters after this introduction, is the iterative and sensor based evaluation of the exoskeleton interface. The second is the modelling of the interface for estimating interaction mechanics, kinematic-dynamic relationships and interface forces. Both of these subfields are used as tools to evaluate exoskeleton interfaces, and in some capacity inform design. What they lack, however, are the tools and baseline methods and metrics themselves to create new surfaces when compared to older surfaces. Currently the focus of these topics is either on direct evaluation of a design for safety (and not comparison between designs), or on the effects of the exoskeleton kinematics and dynamics as a whole.

With many different potential subfields that require investigation, the scope of this thesis is to investigate the iterative and sensor based design processes of coupling surfaces (including pressures, kinematic offsets, kinetic offsets, shears, torques), and the modelling equations of surfaces for simulation and control. These two subtopics of coupling interface design are the most prominent and forward facing approaches to the design and evaluation of coupling interfaces [71, 89, 91, 93, 95, 96, 107, 113, 117, 123–125, 129].

The iterative and sensor based design of interface surfaces is the process by which most commercial, and research, exoskeleton surfaces are evaluated [89, 92, 100, 103, 106, 123, 126–128, 130]. Whether or not interfaces are actually designed or iterated upon through structured methods to develop new or better interfaces is rarely reported, and actual processes to improve or manufacture surfaces is relatively non-existent [89, 103, 123, 183]. The processes highlighted here, are primarily for the characterisation of interfaces, which can be used to inform future design. Most evaluation relies on sensorization to quantify how the exoskeleton interacts with the user through metrics, including pressure, forces, kinematic offset and qualitative survey. The introduction of this information, how the user interacts with the exoskeleton is critically important to identifying where sources of injury might occur and their modes of occurrence [89]. However, utilising them for developing new interfaces is difficult due to a lack of established common baselines (a singular design to reference back to), metrics which can be used to interpret past threshold values and the complications that naturally arise when utilising human participants. Comparing between surfaces utilising the current methods is difficult, but evaluating the performance of one surface alone can be done [89, 105, 106, 124, 125]. If we wish to innovate new coupling surfaces then simpler, easier to use testing conditions need to be developed that address appropriate metrics that can utilise more nuance and inform each other, all under testing conditions that remove as many variables as possible, and provide a reference for best practice standards. If interpretation of results can reasonably eliminate variables to address safety and surface effectiveness by comparing it to a common design or baseline metric, then all new designs can be compared much more easily than what is currently capable.

The modelling of exoskeleton surfaces is used as a tool in simulation and control to describe the governing relationships of force and position between user and the exoskeleton. The most commonly implemented models rely on simple assumptions regarding the governing equations to describe the position and force relationship between exoskeleton and user. Modelling is incredibly important in characterising the coupling interface for identifying the potential effects of the exoskeleton on the user by calculating interaction forces, joint misalignments and followed trajectories. The models implemented are predominantly linear, requiring already fabricated coupling surfaces to approximate their governing coefficient values [70–72, 75, 129, 131–133]. Due to the linear nature of most employed models, the predicted results of simulations trend towards inaccuracies in conditions where large state derivatives are seen [70], an unavoidable consequence. Poor simulation and control models can cause injury for the individual utilising it if the predicted positional relationship and desired trajectories do not match reality. Crucial advancements are being made in acknowledging the initial conditions that may influence the trends of these models, as well as alternative methods for characterising the surfaces, but the trend of error as a result of linearity still exists [70, 75, 129]. Additionally, there is currently no simple method of moving from the desired patterns and behaviours of a simulated model to that of a physical coupling surface without the need of fabrication and similar tools utilised in iterative design.

These two sub-topics, until more recently, have been treated separately due to a lacking ability to evaluate the elastic simulation, and safety components of the exoskeleton coupling interface at the same time [89]. This is partially due to the lack of tools which allow for the evaluation and rational analysis of the coupling interface past basic safety requirements, and the fact that the design of a specific geometric interface for user safety and exoskeleton performance before manufacturing is complete is still difficult to perform [69, 70, 75, 89, 93, 100, 116, 120, 129, 134].

The objectives of this thesis are predicated on the problem statement and the concept of creating better tools and models for rationale as to why design is undertaken. As a result, the thesis is split in two parts. Each section is led by a short literature perspective that lays the groundwork for each section's specific objective. It is then proceeded by an evaluation or design perspective on existing testing, modelling and experiment standards.

The overarching objective of this thesis is:

To evaluate existing coupling interface evaluation, modelling, and simulation approaches to advance methods and establish baselines towards the goals of maximising user safety without sacrificing the performance of the exoskeleton by over designing or removing components entirely. This is performed through the specific objectives.

Part 1: Evaluation and Design Process of Coupling Interfaces:

- Review the currently used and evaluated baselines and conditions for iterative design evaluation
- Develop a new baseline to remove human interference for better preliminary evaluation
- Propose and evaluate a new baseline of surface design comparison that draws on traditional wearable device fields (orthoses, prostheses) so new designs have a common supporting reference
- Define appropriate metrics for evaluation which aid in supporting or making clearer the contributions of coupling interface design than current standards

Part 2: Modelling and Simulation of Coupling Interfaces:

- Investigate the commonly implemented and evaluated coupling interface models
- Evaluate the performance and accuracy of the commonly utilised model under highly controlled testbench scenarios
- Develop new models that more accurately represent the interaction mechanics at the user interface for use in simulation or coupling interface design

1.4 Thesis Organization

The remaining chapters of this thesis is organized as follows:

- Chapter 2: Evaluation and Design of Exoskeleton Coupling Interfaces for Performance and User Safety
- Chapter 3: Evaluating the Effects of Customised Versus Generic Surfaces on Lower-Limb Exoskeleton Kinematic Performance: An In Vitro Mannequin Test Bench Evaluation
- Chapter 4: Impact of Customized Coupling Surfaces on the Performance of Lower Limb Exoskeletons: A Pilot Study
- Chapter 5: Contributions to the Methods, Metrics and Baselines of Exoskeleton Coupling Interface Design
- Chapter 6: A Brief Review on Modelling Approaches for Lower Limb Exoskeletons
- Chapter 7: Validation of Common Models: Perfectly Coupled and Linear Spring-Damper Approaches and a Simple Extension of the Linear Spring Damper
- Chapter 8: Developing a New Algorithm and Model for the Physical Human-Exoskeleton Interface
- Chapter 9: Conclusions and Contributions to the Modelling of Exoskeleton Coupling Interfaces
- Chapter 10: Conclusions and Future Work

Chapter 2

Evaluation and Design of Exoskeleton Coupling Interfaces for Performance and User Safety

This chapter highlights and explores the currently used and evaluated baselines and conditions for iterative design evaluation. Discussed here is also an introduction to proceeding chapters to address the remaining objectives for Part 1. These are: developing new baseline methods to remove human interference, proposing a new baseline surface designs, and developing new design criteria/metrics for comparative and iterative design.

The exoskeleton coupling interface is crucial in the performance and function of the exoskeleton as a whole. For lower limb exoskeletons, coupling interfaces transmit generated forces and torques from the actuators to the user's limbs. Designs for the coupling interface range from a purely rigid body design to a combination of soft and hard elements [89–91, 130]. Regardless of design, the exoskeleton coupling interface must provide support and points to transmit force, shear and torque loads between the user and the device, while keeping them coupled and in the same relative position and orientation. With that support brings risk to the user, including high loads, shear forces and joint misalignments that can cause skin and/or musculoskeletal injury [24,76,77,81,89,90,94]. Pressure injury formation has occurred during exoskeleton use, resulting from excessive non-uniform normal forces, shear forces causing deep tissue injury or relative movement causing skin irritation [24,37, 68,76,77,87,89–92,100,107,118,136]. Musculoskeletal injuries arise when large undesirable torques or forces are applied to the user, damaging joints, segments and musculature. The motivation and goal when designing exoskeleton coupling interfaces is to balance the capability to deliver forces for functional movement, while minimising the risks to the individual.

Most exoskeleton coupling interfaces take the form of a combination of a hard thermoset plastic lined with a soft foam material to provide padding, and fixed with a soft malleable and adjustable "strap" (e.g., velcro, cable tie) to secure the limb in place. Exoskeletons such as the H3 (Technaid, Spain), the Indego (Indego, United States), or the Keeogo (B-Temia, Canada) all employ these interfaces, with one of the H3 Straps shown in Figure 2.1. These interfaces are "semi-rigid", with a combination of hard aluminum surfaces Completely soft designs exist, such as the Suitx range of exoskeletons (Suitx, United States), relying only on soft strapping components to provide resistive forces through tension applied at the surface. Exoskeletons with fully enclosed coupling interfaces are far fewer in number than the alternatives, often appearing in research papers such as the QPRESA (IIT, Italy).

Figure 2.1: Example of the H3 exoskeleton strap: hard-body aluminum surface covered with an EVA foam layer (top), with accompanying hook and loop securing surface (bottom)

These designs serve the function of imparting assistance through stored or active energy, focused usually on supporting the limbs at their attachment points. Oftentimes, they are designed to fit as wide a range of individuals as possible to maximize the potential user base of the device. This tolerable range changes from exoskeleton to exoskeleton. For example, the H3 Exoskeleton claims to have a wide range of usability from 40 to 100 kg and 110 to 210 cm in height. Understanding which of these surfaces provide the most benefit to the user, and perform the best during standard use cases, is not well understood [63, 89, 100]. Furthermore, the process of how to improve upon them and design better interfaces suffers similar issues in a lack of common knowledge base.

Discussion and evaluation of the design of exoskeleton coupling interfaces is relatively under-reported compared to actuator, mechanical structure, controller design and walking function studies. To the author's knowledge, no comprehensive design strategy or guides have been published to draw on when designing new coupling surfaces for exoskeletons. Most studies instead rely instead on heuristic based support or threshold values as indicators for interface safety [89, 103, 107, 123]. Heuristics define common practices, such as interface placement or material composition, that drives design. Current interface designs combine hard supporting interface and securing strap(s) that many exoskeletons possess [37, 47, 89, 90, 130, 135]. Based on physiological studies, threshold values (e.g., pressures, forces) have been defined as maximum tolerances and limits for avoid injury to the user [37, 89–92, 94, 130]. These are good indicators of the baseline safety of the device, but are difficult to compare against other designs. Due to the general lack of comparison between exoskeletons, their performance and safety, evaluating how one design performs relative to another is limited by both the users tested on, the conditions of testing, and a lack of common ideal goals or metrics [63, 89].

Considering commercial designed exoskeleton surfaces are commonly designed with the intent of fitting as many users as possible, many of their designs appear similar both in function and form. While anthropometric design is a well-established goal, there has been little innovation in coupling interface designs.

A recently conducted literature review compiled and evaluated the techniques, methods, and models used to evaluate the physical exoskeleton-human interface [89]. This paper comprehensively covers, and highlights, the means by which researchers evaluate the performance of the coupling interface. While this review is not fully comprehensive on all literature surrounding the design, function and evaluation of coupling interface surfaces it does highlight the most common methods and means of fabrication. Common designs are mentioned, as well as the importance of the analysis framed by user safety and the performance of the exoskeleton as a whole. Some commentary is made on the direction of design, but primarily focuses on the variability in testing and drawbacks in approaches, and the variability and lack of consensus on how individuals should interface with the device [89].

The vast majority of publications rely on "threshold metrics" of pressure, shear or interaction torques to measure performance. These thresholds are used to compare metrics (e.g., maximum or average forces, pressures or torques) against published biomechanical and physiological limits as a means of evaluating physical interactions. The most common approaches are maximum tolerable point load and maximum joint torques, the latter of which is not directly linked to interfaces. Maximum tolerable point load, or pain pressure threshold (PPT), indicates the maximum pressure at which intolerable pain occurs from an applied static load [89]. Tested by algometry, results are highly dependent on location, and a wide range of factors including diet, age, gender, and comorbidity. Considering the subjectivity and inherent variability associated with pain thresholds, allowable values remains unclear. Testing condition were also inconsistent with physiological PPT testing conducted under static conditions, and generalizability to mobility conditions (i.e., gait) is also unclear. Furthermore, exoskeleton interfaces distribute over large surface areas, and the generalizability of point-based tests used in PPT evaluation is unclear [89]. Additionally, point-based tests have been shown to vary under shear, combined loading conditions, and time something that most studies do not account for [89]. The general takeaway from this review is that user variety, task, device greatly influence the interaction between user and device, while the metrics and available methods for analysis make generalisation and evaluation difficult [89].

In some cases, pressure distribution was analysed to identify high pressure locations and explain their phenomena, but was not utilised as a metric for describing improved design or iteration within the same publication [89]. Shear forces, misalignment and relative motion were also analysed, but were far less common as the requirements for measurement and sensorization were much higher. Many of the studies which covered forces, torques, pressures, misalignments and relative motion focused on identifying and characterising these phenomena. Few, if any, studies included in the review proposed and executed means of mitigating these components [89]. However, many of the studies reviewed were focused on identifying threats. Many of the measurements performed on potential error, drawback or injury are one dimensional, limiting analysis [89]. One dimensional analysis are those threshold values not informed or accompanied by additional analysis.

As an inspiration for improving upon or creating "dimensionless" metrics, joint misalignment is a prime example [81, 89, 93, 94, 98, 117–119]. Joint misalignment is quantified by absolute position, and is used to calculate extraneous applied force and torque to joint centers [89, 98]. This value is useful for evaluation of the potential risk for exoskeleton components due to it's relative ease of explaining how to eliminate the source of error. The tools for identifying and mitigating the sources of misalignment only exist for kinematic misalignment however, and not offset generated by elastic components, but the beneficial groundwork can be used as inspiration when developing and investigating appropriate metrics for evaluation. One study, conducted by Langlois et al., lays the groundwork for analysing and identifying the potential sources of elastic offset by comparing a generic exoskeleton to a fully customised orthosis and observing kinematic offset [136].

Issues identified within literature on testing and evaluating the coupling interface can be boiled down to one word: standardisation. Prior studies often focus on singular values and thresholds, from forces and pressures, as a means of evaluating or characterising exoskeleton interfaces, or the utilisation of questionnaires for performance. Implementation of values such as joint misalignment, however, have been used as additional tools for investigation. As mentioned prior, they can be useful as inspiration for dimensionless and transferable evaluation as the sources of potential injury are clear and easily mitigatible as kinematics are simple to use. Kinematic joint misalignment remains relatively simplistic, however, as it does not inform other safety criterion when analyzed alone. These studies lay the groundwork on interface surface effects, particularly safety and performance of the exoskeleton. Considering these characterisation studies investigate distinct exoskeletons, conditions, users and experimental procedures that make transferring knowledge from one study to another is challenging. Likewise, studies investigating improvements or building new sensor systems and devices to assess often do not compare against existing designs [63, 87, 89, 100, 125, 130, 135, 137]. The current issues are both a lack of tools and a lack of transferability between designs, that contribute to a heterogeneous analysis space.

Currently, no clear guides exist to innovate interface coupling design. While many metrics have been studied and evaluated with new designs, few tools to synthesize or interpret that information to truly justify a design choice have been proposed. Considering the inherent complexity of the coupling surface influenced by multiple factors (e.g. gait strategies, mechanical structure, coupling location, users), there is a definite need to establish new methods to implement identified metrics into the design process. Ideally, there would be a clear and distinct path between every type of coupling interface, with each modification or design having an explicit, reasoned need to be implemented.

In the established fields of orthoses and prostheses design and manufacturing, many of the issues listed above are solved by common knowledge in the field. The importance of conformity to body shape is well understood as a value in assessing whether or not a rehabilitative device will be effective at the physical interface [111–115]. In passive rehabilitation devices, the more conforming a surface is to the individual, with proper pressure distribution, the risk to the user is lowered, and efficiency and comfort increase. This understanding is often the “gold standard” in designing prosthetic and orthotic devices, and is used as justification in many socket or surface design studies [111–115]. While it is not proven as of yet for active assistive devices, it would be useful to explore fully conforming surfaces as an alternative to traditional surfaces, and others have hypothesized that the same benefits could be seen for rehabilitation surfaces.

The objective of my work on the iterative and sensor based evaluation and design of coupling surfaces is to establish better means of comparing surfaces between each other, and for evaluating new surface designs. This is performed by concurrently investigating methods of evaluation, including appropriate metrics and baselines, towards developing standards.

Methods of evaluation include conditions and subjects selected for testing. The most common and prevalent method for testing involves studies with human participants. User influence and the complicating factors associated with participant pools, sizes and use cases may lead to a design appearing to be safe or effective in transmission of force. While in reality, among a much larger sample size, designs often demonstrate poor performance [76, 77, 81, 82, 85, 86, 89, 138]. To the best of my knowledge, and authors of recent reviews , no replication studies on the performance of interfaces has been conducted to validate reported safety criterion [47, 85, 87, 89, 92, 125, 130]. While completely removing human testing from the design process is not an option, introducing intermediary steps that allow for characterisation is proposed through the use of mannequins.

Used in testing controller designs, mannequins have yet not been actively used as a means of evaluating the performance of different surfaces to existing designs [89, 103]. Mannequins have many benefits when utilised in controller or exoskeleton mechanics and simulation based testing. While no comprehensive review or rationale of why the mannequins are used, they are utilised primarily as a tool for validation and evaluation under controlled conditions. Typically they are used as user proxies allowing for the removal of unknown, unmeasurable or uncontrollable factors, while increasing safety [99, 139–149]. With a fabricated segment chain for a specific location, all mass and inertial properties can be known for simulation which is much more difficult to guarantee in user testing. User torque is removed by design, with joint properties being easily controlled. Translating dynamics can be removed by securing the base, limiting kinematics and kinetic relationships to just those created by the exoskeleton actuators themselves. These constraints and controls allow for simpler and easier to justify governing equations, while allowing for high repeatability without putting risk on the user [**?**, 58, 106, 139, 142, 150, 151, 153]. While the papers cited here are primarily for control and initial performance evaluation, the same principles can be extended to use in isolating and evaluating physical human-robot interactions [99, 103, 123].

First, evaluating against a mannequin removes many inconsistencies and variability associated with human participants. Second, safety risks associated with human testing are well-mitigated with a mannequin. If a design cannot exhibit a physical safety performance improvement in a rationalised manner on a mannequin, it will likely not show improvement on a user. As an intermediary step in the design process, providing early results prior to major studies involving user participants may allow for replication and validation of results.

The focus on appropriate metrics for evaluation is mainly on expanding and introducing new tools for evaluation that require context based evaluation, to complement threshold based ones. "Context based evaluation" in this thesis refers to the multi-dimensional and focused approach which considers context for analysis in addition to values. Multiple facets, such as force, pressure, surface area, distribution and offset can be considered together to highlight both low risk and high performance design. Limiting average or peak pressure and force values (or any other of the kinetic related variables) are useful approaches to promote safety, but may not inform design for exoskeleton coupling surfaces for performance. Single linear threshold values cannot account for the previously highlighted components which may influence skin injury, which requires extensive knowledge and analysis to investigate and diagnose (e.g., pressure distribution, shear, maximum pressures, friction etc. [155–160]). A high surface pressure compared to a low one may indicate a simple increase in risk, but cannot account for whether or not the low surface pressure is a result of the interface not providing support. This context based evaluation, with supporting appropriate metrics, aims to indicate coupling interface design performance. Relying less on thresholds, also allows for more transferability between users, instead comparing whether one design performs better than another, supported by many metrics, as opposed to one.

While analysis does occur on the pressure distribution side, it is not based on metrics, and relies primarily on visual analysis to highlight trends in differences [89, 103, 123]. This analysis is common in surface evaluation that employs distributed approaches, commenting on location based high pressure occurrences and potential methods to solve them, but often does not evaluate alternatives, or potential optimisation approaches to mitigate pressure. Additionally, the focus is primarily on high point locations and not time based or distribution based [89, 92, 103–106, 123, 125, 161, 162]. Taking inspiration from existing metrics, we look at joint misalignments and offsets for finding new appropriate metrics. This metric, when isolated properly, has easy to explain correlation to the kinematic and kinetic properties of the surface, and it's influence on performance and control. The objective is to identify other metrics which can be used similarly to misalignment as a means of comparing the relative performance of coupling surface design both for safety and to relate it to performance.

The next issue addressed is the lack of baseline comparison tools when designing new surfaces. Other components of exoskeleton design (e.g., controllers, actuators) may report improvements in design by comparing new algorithms or mechanics to a simpler or well reasoned and reported on implementation. Taking inspiration from those fields, as well as the design of custom orthoses and prostheses, a goal was set to establish a set of baseline surfaces which could be easily used and implemented for improvements in the design of interfaces. This was done with the primary intent of reporting on the effects of fully customised coupling interfaces compared against both existing metrics and means, as well as proposed metrics. A secondary motivation was establishing the baseline benefits of a fully customisable surface, and how it can be used as a defacto reference when evaluating new designs. If a fully conforming surface is the currently best performing interface, creating new interfaces can use the metrics to evaluate the performance benefits of a new surface to the best possible result. If each individual is likely to benefit greatly from a customised surface, then utilising it as a reference tool may be more effective when compared to a generic one. Each generic interface may perform relatively different for each individual, which may generate a much wider range of variability in performing an engineering analysis for new custom interfaces.This hypothesis has been approached by one other group, indicating benefits in reducing kinematic offset when utilising an actuated ankle-foot orthosis [136]. Partial inspiration was taken from this protocol and extended to full lower limb exoskeletons, and investigation was extended past kinematic offset.

This first section of the thesis focuses on developing the tools necessary for sound and justifiable design of new exoskeleton interfaces, by investigating new methods, appropriate metrics and baseline surface designs. The next chapter provides a brief perspective on mannequin use in various applications, and describes development and testing using a novel mannequin towards support surface evaluation of lower-limb exoskeleton.

Chapter 3

Evaluating the Effects of Customised Versus Generic Surfaces on Lower-Limb Exoskeleton Kinematic Performance: An In Vitro Mannequin Test Bench Evaluation

This chapter is aimed at the objective of developing a new baseline to remove human interference for improved preliminary evaluation, a new interface design baseline for comparison, and defining appropriate metrics for exoskeleton safety evaluation. This was performed by investigating the design and fabrication of mannequin testbench systems in academic exoskeleton studies, and implementing a functional mannequin to replace user interactions on a secure testbench. The mannequin was then utilised in a preliminary study to simplify the measuring of kinematic offset as a means of evaluating exoskeleton coupling interface performance.

The design of the mannequin was initially formulated as a new method of evaluation in the approach of the iterative and sensor evaluation design of exoskeleton coupling surfaces. With a mannequin, user influence could be removed in the calculation of surface pressures and kinematic offsets/joint misalignment. The main advantage of mannequin use is in isolating and identifying components of exoskeleton use that cannot be measured exactly with human participants (e.g. joint torques, misalignments, forces, weight distributions and inertial matrices). The design of this mannequin was entirely passive, while testing implemented external sensors and vision systems to calculate interaction mechanics.

Joint misalignment occurs when the joint center axes of the exoskeleton and user are not aligned as a result of segment length inequalities, complex joint motion, and/or kinematic trajectory differences. Kinematic segment offsets occur as the result of elastic interaction between user segment and exoskeleton segment that cause leading, or lagging, angular offsets during motion. Both are related, and if one occurs, the other is likely to as well. An able bodied individual is likely to compensate against this offset, and removing user compensation would allow for the full characterisation of a surfaces elastic and natural misaligning tendencies. Both are analysed together in this section as preliminary investigative tools used as means to evaluate the effectiveness of custom interfaces, and to evaluate whether they can be used as effective tools themselves.

3.1 Introduction and Background

Lower-limb rehabilitation exoskeletons (LLREs) are powered or passive devices that provide support and ambulation assistance for individuals with lower-limb mobility impairments. Rigid exoskeletons provide support directly through torques and forces applied by actuators according to trajectories designed through optimization, inspection, or tracking of non-affected gait [36, 37, 40, 46, 47, 49–52, 54, 57, 60, 135]. The potential applications of LLREs are wide-ranging and new designs are constantly being explored in rehabilitative or augmentative capacities in clinical and research settings.

The assistance and control of exoskeletons is regulated through actuators and the forces and torques provided through them. Regardless of design, the exoskeleton coupling interfaces act as the points where forces and torques are transmitted from exoskeleton actuators to the user. Without a coupling interface, the transference of energy and motion from device to user is not possible. While the interface is critical and necessary to the function of the device focus on how the interface is designed, its characteristics and models remain relatively unexplored [89]

The coupling surface brings the user into contact with the exoskeleton by strapping and holding the individual to the exoskeleton through securing and binding forces. The design of these coupling interfaces may vary based on the function and actuation of the exoskeleton, usually some combination of soft and rigid elements together to interact with the user. The more effective the design of the interface, the better these forces and torques are transferred to the individual without sacrificing their comfort and safety. The understanding and investigation of what makes an exoskeleton interface effective, however, is relatively unexplored. Pinpointing what design features and decisions make an interface effective is difficult, and designs which identify and quantify design influence are few and far between [89, 136]. One approach that is employed is taking inspiration and cues from established works in passive rehabilitation devices.

In the development of lower limb rehabilitation devices (e.g. prostheses, orthoses), consideration of the interface surface between the user and device is critical. In particular, certified clinicians (e.g. orthotists, prosthetists) design interfaces to maintain an effective "quality of fit" to minimise risk of skin and musculoskeletal injury while controlling joint motions, allow for long term use, and ensure effective rehabilitation [115]. For Lower-Limb rehabilitation exoskeletons there has been little focus in literature on interface design, the risks they pose to the user, and their influence on the effectiveness of the device in short and/or long term use in the same ways they have been explored in traditional devices [77].

In order to more accurately define the benefits of a coupling interface, the use of a highly controlled mannequin testbench was implemented. The use of mannequins – inanimate proxies for human subjects – has been taken up as an alternative method of benchtop testing that removes the risks to users, without sacrificing components of interaction [64, 139, 148, 149]. In vitro testing where the exoskeleton has no participant inside the device can be performed to analyse basic actuator and mechanical components but is limited in use as it removes the complex interactions that occur between user and device, making it better suited for robotic characterization [64, 121, 139, 148, 149, 164]. Many different examples exist of characterization, but are often relegated under actuator "test-benching". Mannequins solve this problem by adding mass loads and complex interaction mechanics (e.g., limb segment cross-sectional geometry) to the evaluation. This testing can also be conducted with the physical exoskeleton system without the risk of harming participants, avoiding delays in acquiring research ethics board (REB) approval.

Measuring and evaluating the effectiveness and safety of exoskeleton interfaces is focused on the interactions between user and device at the skin and joint interfaces. Kinematic joint and segment misalignment results in unwanted reaction forces occurring in the joints that may cause injury as a direct result of poor interface design. Misalignment and offset may cause any number of injuries, those mostly prominent and evident being joint overloading, with other effects such as excess shear at the interface as a result of offset forces [98]. Joint load could not be measured directly in this study, however it has been proven from simulated static mechanical analysis that reducing angular offset and misalignment directly reduces joint load [94, 98].

The present study aims to evaluate a generic strapping interface against custom surfaces in LLREs in a controlled mannequin testbench environment, to aid in identifying how well custom interfaces perform in reducing musculoskeletal injury risk. From testing and analysis, we can then identify how certain decisions for interface design directly influence safety and performance under controlled conditions. This process is performed through the evaluation of a strongly supported baseline with precedent (e.g., fully conforming surfaces), a testbench environment that simplifies the interactions between user and exoskeleton framed by a novel approach to analysing how interface design directly influences user safety and exoskeleton performance (i.e., kinematic offset and misalignment).

The H3 exoskeleton (Technaid, Spain), was employed as the testing platform for evaluating the influence of custom and generic interfaces on kinematic offset. The H3 has 6 degrees of freedom, bilaterally at the hips, knees, and ankles, limited to the sagittal plane. The mannequin proxy designed for the H3 was created with similar rotary joint constraints to match the device and simplify the relative mechanics. Additionally, the mannequin was fabricated as a passive mannequin system, and as a result, evaluating the design decisions taken to reduce injury will be easier to identify when limiting mechanics to solely the exoskeleton.

The two major segments of the H3 (Shank, Thigh), have two coupling interfaces each. These surfaces consist of a rigid aluminum body layered with a soft EVA foam layer to provide padding. Designed to fit a wide range of individuals, ranging wider than 5th percentile female to 95th percentile male anthropometry, these interfaces were treated as the ideal generic interfaces.

A set of customised surfaces, modelled and designed after the same processes utilised by prosthetists and orthotists, with the aid of an external expert (Orthopaedic Bracing Solutions, Kitchener, Ontario) were fabricated as our custom interfaces. The interfaces themselves were defined to be our customized surfaces, as they were designed to fit a single surface curvature: the mannequin.

The methods of assembling both the testbench with mannequin and the customized straps, and the basic rationale behind the design decisions made for testing are presented first. Additionally the testing, capture and processing methods utilised to gather the kinematic offset data between mannequin and exoskeleton are detailed. The results presented focus on the kinematic angular offset, and the finalised custom interfaces, compared to those of the H3 Exoskeleton's generic interfaces. Lastly, the impact and design rationale are more thoroughly discussed on the testbench conditions and customized interfaces. The main focus of the discussion is placed on analysing the custom interfaces design and their influence on kinematic offset and user safety when compared to generic interfaces. A secondary focus is placed on the benefit of the testbench design in clarifying and simplifying results for analysis.

3.2 Methodology

The H3 exoskeleton (Technaid, Spain) has 6 degrees of freedom, bilaterally at the hips, knees, and ankles. Each major joint center is controlled by a fully assistive rotary actuator. The H3 exoskeleton is accompanied by a proprietary controller app which can command full, or partial, torque(s) to assist with gait. The gait cycle speed can be controlled and are modelled after standard straight-line walking cycles. A mounting frame was assembled to suspend the exoskeleton by the hips to isolate sagittal plane motion without ground reaction forces or translational dynamics.

A simple two joint mannequin was fabricated consisting of thigh, shank and feet segments. These segments were made utilising 3D printed polylactic acid (PLA) segments, designed and fabricated matching the shape and curvature of a 95th percentile male, with height of 183 cm, and mass of 86kg volunteer scanned into CAD software. The mannequin is designed as a simple unilateral, 2 degree of freedom proxy that allows for accurate and consistent collection. The mannequin was split into three segments: thigh, shank and foot, connected by two simple rotary joints. The focus of this test was on evaluating simple elastic interface reactions of a supporting surface. To ensure that the driving elastic factors of the support surface were the driving force-position relationship, the mannequin limb was manufactured as a rigid contact surface, stiffer (PLA) compared to the EVA foam and polyurethane supporting surfaces. Surface curvature was scanned to match the volunteers leg to simulate a custom surface contact interaction without highly elastic or compliant components, characteristic of tissue. Additionally, to support the surface as the driving interaction, the mannequin joints were constrained to rotary motion to ensure interactions associated with joint axis translation as a result of shifting radii did not occur further complicating already nonlinear interactions. The finalised mannequin with nylon surface covers is shown in Figure 3.1.

Figure 3.1: The 2 DoF mannequin leg fabricated of PLA segments with nylon sheath covering. Each joint is a rotary pin joint fabricated and designed to match the H3 exoskeleton joint actuators.

Two sets of interface surfaces were evaluated. The “generic” surface is designed with a curvature intended to fit across a wide range of individuals. The H3 exoskeleton interfaces fit a range of individuals, including 5th percentile females to 95th percentile males. The backing material is a thin aluminum shim (5mm thickness), layered by a piece of EVA foam. The curvature of this surface resembles a series of curved semi-circular surfaces, with a hook and loop securing strap. The hook and loop strap secures the user to the generic surface, consisting of a thin fabric webbing surface with an additional EVA foam pad for pressure distribution. The generic interface is shown in Figure 3.2.

Figure 3.2: H3 Exoskeleton generic interface with aluminum backing and EVA foam compliant surface.

The “custom” surface was designed with a curvature intended to fit a singular individual (i.e., mannequin). The curvature of this interface was designed with the aid of a certified orthotist (Orthopaedic Bracing Solutions, Kitchener), to match the curvature of the mannequin surface. Utilising the CAD model of the mannequin leg, an offset surface was generated with added additional surface area (compared to the generic surface area) in radial and height of surface directions. Additionally, a thicker padding base was added, consisting of EVA and polyurethane foams to increase dampening and compliance. An alternate clamping form was designed to hold the user by creating a conforming clamping interface shell. The general form of this interface takes heavy inspiration from the H3 interface with an emphasis on increased surface area, conformity and increased compliance. The exoskeleton’s final custom surfaces are shown in Figure 3.3.

Figure 3.3: Top down view of the H3 Exoskeleton custom surfaces manufactured from PLA

For each set of coupling interfaces, generic and custom, the devices were fixed to the exoskeleton at the same locations. The mannequin leg was placed into the exoskeleton for both sets of interfaces and secured at the same tensioning force (5 kg). Studies indicate initial pressures influence reactions, and to ensure consistency between tests the same pulling force was used during fixation [69].

The testbench is the same as shown in Figure 3.4. VICON motion capture was used to capture kinematic knee-joint data of the H3 and Mannequin leg. Markers were placed on both the exoskeleton frame and supporting surfaces, as well as the mannequin limbs. The mannequin limb surface is rigid, making marker shifting a non-concern. VICON post processing software was utilised to find the kinematic position and angle data of both exo and mannequin. While initial offsets were a concern during the process of inserting the exoskeleton, extra care was taken to avoid misalignments, and any minor misalignments were removed in post processing to calculate global angular values.

Figure 3.4: H3 Exoskeleton suspended with 2-DoF mannequin on the stationary testbench. The exoskeleton is mounted with the custom surfaces, and VICON markers are placed along the exterior to capture kinematic information.

After securing, the H3 was commanded using a proprietary mobile device app to go through the standard gait cycles. The lowest possible speed (4.5 second gait cycle) was selected. 10 cycles were commanded for both securing interfaces. After collection and post processing, knee angle offset was selected as the joint and metric of interest. For both collections, 10 gait cycles were segmented and overlaid to visualize the variance of the angular offset over all cycles.

Analysis of these results were performed by visual inspection and statistical evaluation. Visual interpretation of the gait cycles was conducted to compare magnitude differentiation in angular offset, and the increase in variance/decrease in predictability and performance over the ten cycle collection. "Quality of Fit" was evaluated as the root mean squared error (RMSE) between the exoskeleton and mannequin knee angles, indicating relative movement between user and device. Increased RMSE of the angular offset indicates an increased separation during the gait cycle, and an increased risk for misalignment related injuries.

3.3 Results

The mannequin performs well in providing clean, repeatable and interpretable data. Swing cycle segmentation using phase and angular data was applied splitting the known periodic swinging segments into 10 separate cycles.

Relative knee angle difference between the generic and custom interfaces for ten steps are shown in Figure 3.5. The 10 gait cycles are segmented and overlaid and labelled. The generic data (left) exhibits higher peak offsets in both positive and negative directions compared to the custom (right) interfaces. Additionally, the offset of the generic interfaces remains consistent, with a constant relative movement and internal shifting that occurs during swing phase.

The custom interfaces exhibited a smaller offset variability, with peaks in offset occurring at similar points during the gait cycles. The peak offsets of the generic interface are reduced compared to the generic interfaces. The generic interfaces experience an offset range of roughly 7 degrees, while the custom interfaces range only 2.

RMSE of the generic strapping ranged from 1.41 to 2.03 degrees across 10 steps, with higher values observed in the later steps indicating an increase in separation. RMSE of the custom strapping ranged between 0.41 to 0.47 degrees with no pattern or increase over early and later steps.

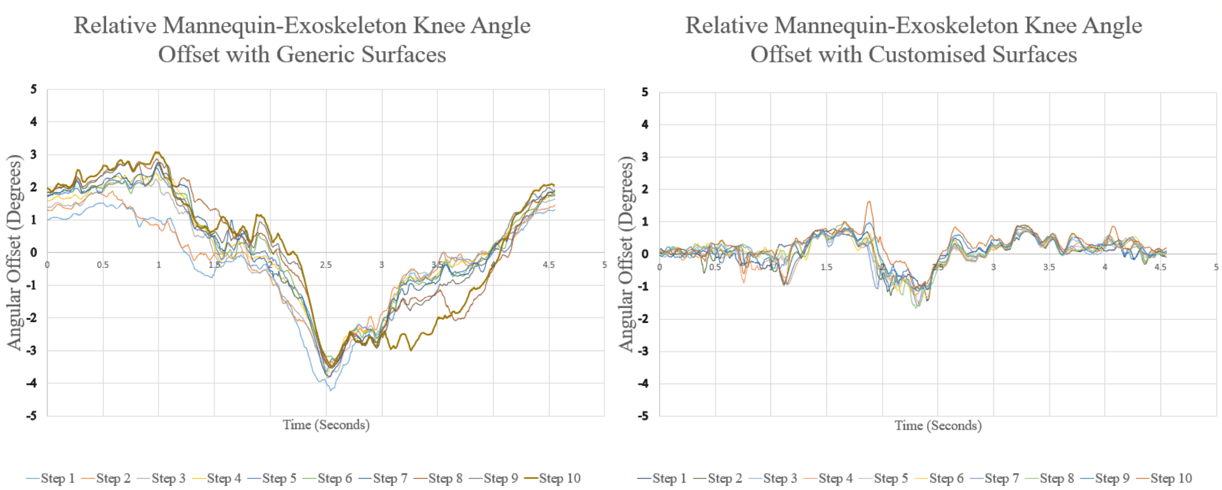

Figure 3.5: Kinematic Data capture from Generic (left), and Custom (right) interface surface data comparing the relative knee-ankle segment angle between mannequin and exoskeleton for 10 gait cycles.

3.4 Discussion

The discussion examines first the influence of a fully custom surface compared to a generic design, then discusses the impact of the testbench conditions and mannequin.

In comparing the custom and generic strapping designs, the knee joint offset results indicate key performance differences. Kinematic results show higher angular offset RMSE within the generic strapping, which increases with each gait cycle. The generic interface has an angular offset range spanning more than 7 degrees and exhibits major peak offsets during forward and backswing. The mannequin and exo only remain relatively closely coupled during the segments of gait where the mannequin is the stance leg, or crossing the centerline from backswing to forward swing.

The customized interface exhibited a lower angular offset RMSE compared to the generic interface. The custom interface has an angular offset range spanning just over 2 degrees, a notable decrease when compared to the generic interface. No major dramatic peak offsets, like the generic interface, were observed; however, the peaks occurred at similar timings in forward and backward swing. The mannequin and exo remain relatively closely coupled during motion, with much of the difference appearing to be non-linear shaking or noise causing separation. When compared to the generic interface, it is unclear at which point(s) the driving inertial component is causing separation (such as the peak back/forward swing). Instead, a major driving factor of separation appears to be the custom interface resisting through frictional and pressure forces applied to the user that are then overcome as a direct result of the more conforming interface.

The gait cycle angular offset of the generic interface, and the relative lack of offset in the custom interface, may be attributable to the design characteristics. The increase in angular offset characteristic of the generic interface is likely caused by the lack of conformity and lower available supporting surface area. With less surface area, forces are concentrated in much smaller areas causing deformation of soft supporting surfaces to increase as well as stored elastic energy. The generic interface's lower surface area is not just in the cylindrical height direction, but also the circumferential perimeter distance. The decrease in in-plane surface area (sagittal) reduces the availability of points of contact for frictional resistance. This ties directly to surface conformity, as even with a reduced surface area compared to the generic interface, the generic interface's surface area is further reduced by it not conforming. What little area there is may not provide a supporting resistive force in the sagittal plane that would prevent shifting from occurring. Due to this, the generic interface only has the hook and loop strapping to effectively resist motion in the sagittal plane. Due to elastic and nonlinear components in this securing method, the hook and loop methods does not resist motion effectively.

The increasing movement over the course of the generic interface tests is likely attributable to inelastic deformation with successive steps. This inelastic deformation is likely to be occurring at the hook and loop straps as force is applied on them during the backswing and forward swing phases of gait, the hook and loop may be slipping.

With these factors in mind, and taking into account the performance of the custom interface, it is easier to extrapolate why it performs better in reducing angular offset. The increased surface area and conformity reduces any one location from experiencing excessive pressure causing elastic deformation. The increased surface area in the circumferential direction also increases the available sagittal plane surface area for frictional resistance. Being a two shell design, as opposed to a single shell with hook and loop as the major resisting surface also removes any of the sliding and nonlinear effects of the securing method from interfering. Additionally the second front shell adds a much larger surface area for pressure distribution compared to the singular small pad used on the generic interface.

The custom interface's conformity and increased surface area redistributes pressure better than the generic interface and offers more points of contact for resistive frictional forces. As a result, relative movement and joint angle offset magnitude is reduced in custom interfaces when compared to generic ones.

The use of the mannequin leg was motivated by the lack of supported or referable work that evaluated the influence of the coupling surface's conformity on relative kinematic offsets for powered wearable devices [136]. Without a basis of how testing should be performed, interpreting the results from initial user trials proved challenging. Particularly, in cases where the user is able bodied, the influence of confounding variables such as input torque, or soft skin mechanics, make interpretation particularly difficult. Isolating what phenomena or physical reaction was a result of user related factors, compared to surface related factors essentially became impossible without an understanding of the baseline benefits and effects that the surfaces themselves provided.

The implementation of a mannequin simplified the issues of user input, and the removal of dynamic components and ground reaction forces further removed complications of dynamic nonlinearities. The mannequin leg, being two degrees of freedom as rotary joints, which matched the H3 exoskeletons actuation methods added to the process of simplification to ensure that the desired interaction for observation being interface conformity and the design's effect on exoskeleton segment offset, would be the only driving one.

The benefit of this testbench is seen through the captured VICON kinematic information. With the mannequin each gait cycle remains tightly consistent through time. Consistent marker placement without elastic reactions ensured that each cycle measured the limbs offset similarly. The mannequin's simple joint structure, and lack of user input ensure that each cycle had simple interactions about the joints and were consistent through the cycles. It is clear that the driving reactions in this experiment were the interface elastic mechanics, and the exoskeleton actuators.

3.5 Conclusion

This experiment was performed with the purpose of evaluating and quantifying the baseline benefits of a customized interface inspired and guided by best practice designs in orthoses and prostheses. A baseline evaluation of the kinematic angular offset of customized and generic coupling surfaces was performed by removing many of the confounding variables commonly seen in exoskeleton testing that make analysis difficult. A mannequin and test bench setup were used to remove user input, tissue elasticity, ground reaction forces and moving body dynamics to isolate the reactions to a simple static swinging gait protocol. With these constraints, the major defining elastic components of interaction and driving factors for kinematic offset should be limited to the coupling interface and the securing straps associated with them. Initial conditions were ensured to be similar through the use of additional external tools, and collection was performed by VICON camera systems to evaluate kinematic angular offset.

These initial findings indicate conforming custom surfaces offer more consistency and contact during use of the exoskeleton when compared to that of generic interfaces. While the trade off of number of suitable users to effectiveness of surface is quite clear, if misalignment and joint injury are a concern during the design and use of an exoskeleton interfaces which conform well to the user with higher surface area should be pursued. Currently, generic interfaces are designed to fit a wide range of users as effectively as possible. Knowing that the conformity and effective surface area of the interface matters in performance indicates that future interfaces would benefit greatly from reducing the number of potential users to improve the effectiveness of the device.

3.6 Steps Taken Post Study

After completing this preliminary evaluation, a new set of surfaces were fabricated with feedback from Al Moore at OBS, with the addition of a torso orthosis so the exoskeleton may more closely match a Reciprocating Gait Orthoses (RGO). Figure 3.6, is an image of Christian Mele working with Al Moore on measuring and fabricating the Torso Orthosis relative to the H3 Exoskeleton. The new surfaces expanded on those presented here and simplified the manufacturing process from fully enclosed interfaces, to open strap surfaces, to be much closer to that of an RGO and the original generic H3 straps.

With these modified surfaces, a real-use pilot study test was performed to further evaluate conforming customised surfaces in active device use, and to introduce the other values for use in comparative surface evaluation in the proceeding chapter. In the study highlighted in this chapter, the only metric evaluated was kinematic offset. While this is a potential source of injury, it is not the only one with major indicators, and prior studies, pointing to skin pressure as a source of injury during exo use [76, 77, 89]. To evaluate skin injury a user pilot study was conducted using the adjusted interfaces to evaluate both numerical and visual effects of custom and generic surfaces on compliant human tissue.

Future plans are in place to evaluate the benefits and designs of mannequins or "phantoms" in exoskeleton testing. User based testing was undertaken for the purpose of this thesis to evaluate soft interface pressure interactions, and as the visual effects of interface pressure would be much more apparent when compared to the mannequin. Mannequin use still remains a great preliminary aid in evaluating interface designs, and utilising all of the metrics collected through this thesis (kinematics and pressures), new interfaces may be designed.

Figure 3.6: Photo of Al Moore - right (OBS) working with Christian Mele - left on fabricating and measuring out a Torso Orthosis for the H3 Exoskeleton.

Chapter 4

Impact of Customized Coupling Surfaces on the Performance of Lower Limb Exoskeletons: A Pilot Study

This section covers the pilot study undertaken during the Fall and Winter of 2022-2023. The objectives approached in this section include investigating new baselines, and defining appropriate metrics. This is accomplished by using common testing methods to evaluate the fully customised surfaces (new baselines), utilising surface pressure and metabolic consumption (appropriate metrics) against generic surfaces.

4.1 Introduction

Exoskeletons are used by individuals with mobility impairments and a range of exoskeletons exist to meet specific needs. The general purpose of lower limb exoskeletons is to provide support through predetermined trajectories or torque profiles to aid in ambulation. In the development and use of lower limb exoskeletons, the physical interfaces which couple the user to the device are often overlooked as critical components [76, 77, 85, 89, 92, 135]. These surfaces, and the methods by which they couple to the user, vary from machine to machine, but they functionally provide the same purpose in the control of exoskeletons. They impart forces and torques through the contact surface to control a desired trajectory making them crucial components in the successful use of lower limb exoskeletons.

For hard-body exoskeletons, the coupling surfaces which join the user to the exoskeleton are also the method by which forces and trajectories are delivered to the user. In order to transmit forces, contact must be maintained for the duration of the path trajectory. If the contact surface area of the system is poorly designed, shear and normal pressures become highly variable, with high dynamic loads adding complexity. The introduction of complex non-linear forces makes trajectory prediction and modelling difficult which reduces the effectiveness of the system in tracking predefined trajectories [69,70,89,93,168]. Furthermore, excessive shear forces and high pressures increase the risk of skin breakdowns.

To counteract this problem, surfaces may be designed with higher surface areas to reduce contact pressure. However, straps with a larger surface area are less likely to fit the broader population which increases the potential of forming localized high-pressure areas known as pinch points. Poor alignment and fit may also lead to musculoskeletal injury during use and may also present challenges in exoskeleton modelling and control [76, 77, 81, 89].

The coupling surfaces used in exoskeletons can be differentiated into two categories: those designed for a range of individuals, termed “generic”, and those designed for a specific individual, termed “custom”. Generic surfaces are motivated by anthropometrics and are used in most exoskeletons as they are intended to fit most users and are inexpensive to fabricate. Conversely, custom orthotic wearables are designed using casts of an individual’s limbs and modified to reduce pressure points. In general, a high amount of contact area is desired to distribute pressure and reduce movement between the user and the surface. However, methods of creating custom wearables are costly in terms of time and labor which may explain why they are rarely pursued in exoskeleton design. Currently, exoskeletons still cost around 100 - 200 thousand dollars and are used by a number of individuals. Creating new, custom interfaces requires that these designs are accommodating to a wide array of individuals, and that their benefits so greatly outweigh the cost of fabrication that it warrants fabrication.

Despite their importance in delivering force, studies that explore and evaluate the sources and magnitudes of surface pressures are underrepresented within existing exoskeleton literature. The severity of secondary health complications from pressure injuries is significant: the user may experience severe pain and setbacks in their rehabilitation [76, 77, 89, 107, 115, 125, 130, 155, 156]. There is a need for further research on the interaction between coupling surfaces and exoskeleton users if these devices are to be more frequently adopted as rehabilitative aids in clinical settings.

The design process of passive rehabilitation devices bases some of its decisions on specific threshold values, but often sees literature focused more on adverse event avoidance, and optimizing the functionality of the device for the user [112, 113, 115]. Wearable rehabilitation devices can pose a significant risk for pressure injuries [76, 77, 89], which can form under high applied normal stresses and are exacerbated by shear stresses [76, 77, 89, 101, 102, 155, 156]. However, orthoses that conform to the body and remove pinch points can reduce shear stresses and frictional forces. Idealized pressure maps are difficult to identify and design exactly, but guiding principles can be used as aids. Designs that permit radially uniform pressures allow for proper blood flow without causing localized pressure injuries, or a reduction in flow resulting in pooling and strain in other locations [108, 109]. Most importantly are pressure maps designed for appropriate distribution and uniformity during use that does not overly promote shear, or pinch at singular locations [89].

The purpose of this pilot study is to examine the potential benefit that interface contact surface customization has on the performance and safety of exoskeletons when compared to generic surfaces. This pilot study evaluates the differences between a set of customized surfaces and generic surfaces using a variety of collected empirical results relating to contact pressure, interpreted from a set of higher end pressure sensors. A single participant completed a series of walking tests with both generic and custom surfaces; the subject spent over an hour walking in each configuration. The exoskeleton used in this study was the H3 Exoskeleton (Technaid, Spain) which provides complete gait assistance in the sagittal plane. The trajectories commanded by the onboard controller are predetermined and will generate required torques to match that trajectory and does not require any effort from the user. An array of force-sensitive resistors at the hip and thigh recorded pressure data. Skin integrity was evaluated by visual inspection of the body following walking trails.

It is hypothesized that customized surfaces will exhibit significant improvements in performance when compared to generic surfaces on an active lower limb exoskeleton, indicated by metrics known to be associated with skin injury formation.

4.2 Methodology

The impact of customized support surface designs on pressure distribution and metabolic cost was examined in this pilot study involving a single participant.

4.2.1 User Information

The individual participant in this pilot study is the lead author who is 183 cm tall and weighs 86 kg. The participant was of good health, had no motor impairments or skin conditions, and had extensive experience with the H3 exoskeleton.

4.2.2 Surface Design and Comparison

The H3 is a six-internal-degree-of-freedom (6DoF) sagittal plane, bilateral lower limb exoskeleton used primarily for research. The H3 uses bilateral DC motors at the hip, knee, and ankle to provide trajectory-based control and contains adjustable lower-limb linkage to accommodate individuals between 110 and 210 cm tall, as noted in the manufacturers manual [169]. The generic surface couplings of the H3 (Figure 4.1)include eight leg straps (4 per leg) and a torso orthosis. The two-piece torso orthosis is made of flexible plastic with a padded foam layer fixed on top, also shown in Figure 4.1. It is designed to sit above the hip centres and can be adjusted to fit the width of individuals with wider hips. Four large hook and loop straps with ethylene vinyl acetate (EVA) foam pads secure the two torso segments.

The eight-leg straps (two shank and two thigh straps per leg) are composed of a circular cross-sectioned aluminum hardbody covered with EVA foam. Each strap is "open" and approximately two-thirds of the circumference has no rigid surface to permit the leg to enter and exit the strap. Hook and loop closures tighten the straps. The straps are narrow and have low surface areas, causing high pressure build-up and strong elastic effects on the skin in that region.

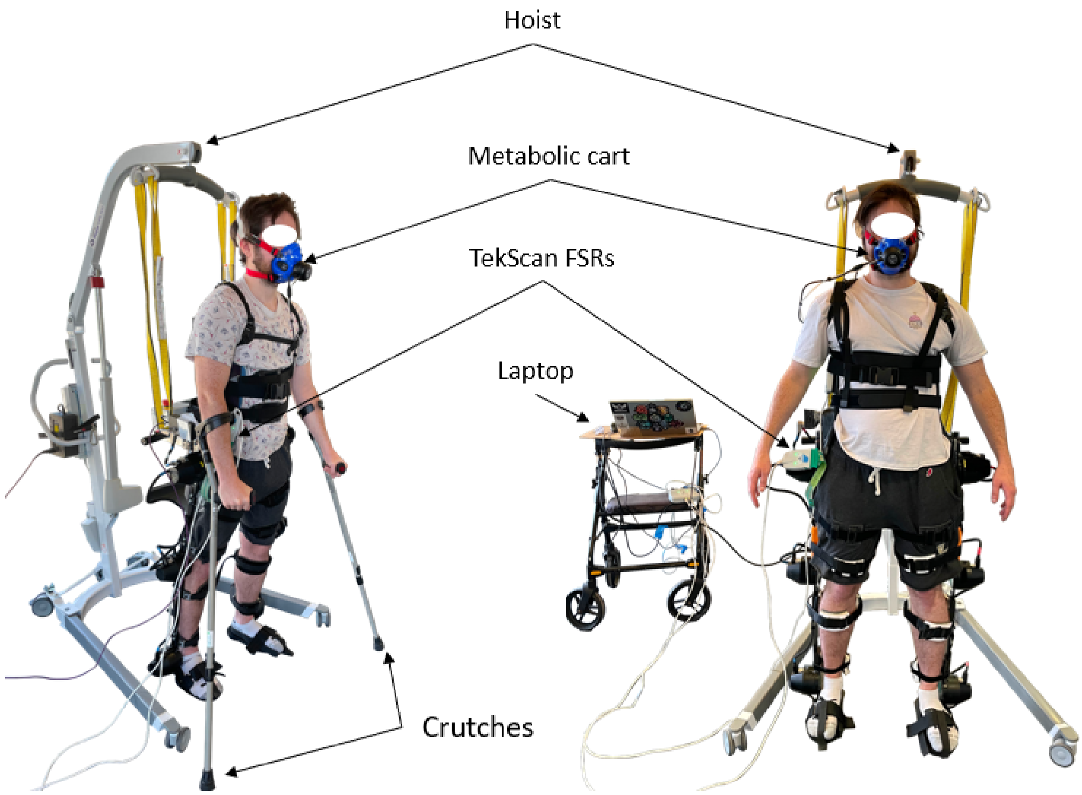

Figure 4.1: Participant in generic and custom straps, pictured with hoist and laptop for data collection. The portable hoist provided fall stability. All wiring for sensor collection was led directly over the hoist to the laptop setup. Crutches are required for use in all tests, second image without crutches for clarity. A metabolic cart system was used for monitoring general breathing and heart rate levels to end testing under high stress, and for potential data analysis.

The shape and form of the 'custom' surfaces were designed to conform to the subject's body (Figure 4.2 right). Positive moulds and 3D scans allowed the orthotist to proactively identify and remove pinch points, and the pressure profiles of all surfaces were designed to be radially symmetric about the centre of the torso and leg.

Figure 4.2: Generic (left) and custom (right) surfaces mounted to the H3 exoskeleton. The two piece generic torso orthosis has a much lower surface area compared to the one-piece custom orthosis. The custom limb surfaces are layered thicker than the generic interfaces, and have a larger circumferential surface area length.

The custom torso surface was fabricated by a certified orthotist at Orthopaedic Bracing Solutions (OBS) in Kitchener, Ontario. Following industry-standard casting and moulding techniques, a single-piece back brace was manufactured out of thermoset plastic. Reducing pinch points was achieved by matching the topography of the subject's waist fold region to increase contact area and snugness around the hip joint centres. The custom torso surface was affixed to the exoskeleton using the same aluminum mounting bracket as the generic 2-piece torso orthosis.

The custom leg straps were designed so that the segment linkages would remain in the same position as the generic system. In consultation with OBS, a technique for leg strap fabrication was developed. A 3D model of the subject's legs was generated in SOLIDWORKS (2022b, Dassault Systems, Massachusetts). Each surface followed the curvature of the subject's leg and was half open (Fig. 4.2, right). The straps were 3D-printed out of acrylonitrile butadiene styrene (ABS) filament and covered with polyurethane foam and EVA. The hard-body surfaces are rigid, with the most compliant components at the bolt-secured location on the exoskeleton, allowing for some effective bending. Anterior closing hook and loop straps and EVA-covered ABS pads were added to secure the leg in place.

4.2.3 Experimental Protocol

In both the generic and custom configurations, the H3 was outfitted with two Tekscan 9811E force sensor arrays (Tekscan, Massachusetts) containing 96 individual force-sensitive resistors (FSRs) each, recorded at 100 Hz. One array was placed on the posterior right upper thigh surface just inferior to the buttocks, and the other was placed on the lateral right hip surface. A Cosmed K5 system was implemented to evaluate safety, and for data collection for future studies evaluating the effects of coupling interfaces.

For each testing session, the following collections occurred:

1. One 1-minute standing to allow for acclimatization
2. Two 5-minute straight walking tests at Speed 1 (4.5 seconds per gait cycle (2 steps))
3. Two 5-minute straight walking tests at Speed 3 (3.8 seconds per gait cycle (2 steps))

Initiating, terminating and controlling walking speed were controlled using the H3 mobile app (Technaid, Spain). For safety, a hydraulic lift was pushed behind the participant by a spotter who also controlled the exoskeleton's speed. Directly beside the participant, a wheeled cart with a laptop collected pressure data. The participant used forearm crutches to help maintain balance, as per manufacturer recommendations. Pictures of the participant's skin were taken immediately after a test session and inspected for visible signs of skin breakdown, such as bruising or redness.

4.2.4 Data Processing

After collection, post-processing was performed on the Tekscan and Cosmed K5 sensor data. While both Tekscan and Cosmed provide proprietary softwares for data processing, custom Matlab (R2021b, MathWorks, Natick, Massachusetts) scripts were used to process and visualise data.

The Tekscan sensor software provides time series data of the 16x6 FSR array (96 individual FSRs) output pressures in millibars and was processed to calculate the average pressure across the coupling surface, the percentage of the Tekscan array that was active as an indicator of surface area, and the centre of pressure (CoP) location on the Tekscan for each time frame.The Tekscan was sampled at 100 Hz, and processed utilising the Tekscan, with each time frame providing a 16x6 grid of pressure values which can be post processed from a CSV file. Post processing was not applied past the Tekscan software, and analysis was performed to extract active surface area, and center of pressure locations. Figure 4.3 highlights the processing for interpreting hip (top) and thigh (bottom) data by rotating collected pressure maps to match the orientation on the surface of the exo.

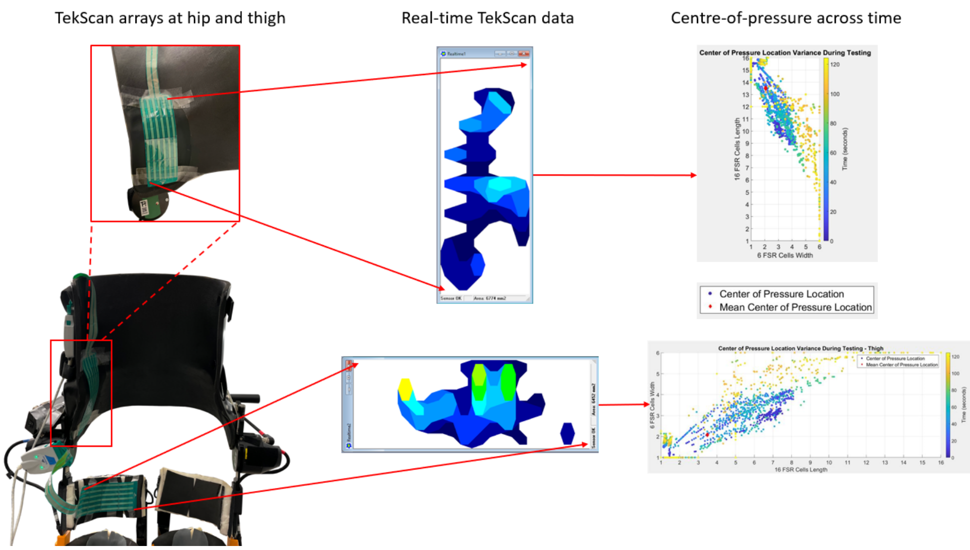

Figure 4.3: Post-processing approaches from FSR arrays. Data was taken from Tekscan proprietary research software, and converted into pressure information. The respective Tekscan mounting locations also inform whether or not the maps should be rotated.The presented information here is center of pressure location throughout the testing trials.

Utilising the Tekscan sensors, 5 values were extrapolated from the pressure map associated with pressure injuries and which can be easily extrapolated and used as values in comparing interface performance. The framework of how these values are used in analysis is critical in understanding why one interface may perform better than another. A short analysis of the value of each is made here:

Mean Active Pressure:

- Describes the mean pressure of any supporting surface. Locations which do not maintain contact with the individual do not count as supporting surfaces and are not accounted for in the mean pressure.
- Mean pressure represents the potential risk for skin injury over the supporting surface area to generate tissue deformation, causing bruising, redness, deoxygenation and ulcerization [155].
- Threshold values can be utilised to indicate risk of injury, however pressure magnitude alone is not enough to describe complex dynamic and distributed interactions.

Pressure Variation Over Surface:

- Variance of pressure over the surface area at each frame/timestep. Larger variance indicates a wider spread of pressures over the surface area. High pressure variance can indicate the existence of pinch points, or regions with large amounts of high pressure support surrounded by immediate low pressure causing tissue deoxygenation in the region.
- Assesses whether or not a gradient is too steep and may cause pinching or flow reduction as a direct result of applied pressure.
- Low variation implies all pressure at any given time-step is evenly distributed over the supporting surface area, and contains peak pressure information (lower peaks mean lower variation).

Mean Percent Active Surface Area:

- The percent of the supporting surface which is actively providing supporting pressure/force. An increased supporting surface area offers more contact area and points for distribution of forces.

- Additionally, increased effective surface area allows for more points to contact the individual to prevent high gradients of pressure application to avoid pinch points forming or vastly reduced flow potential at those locations.
- Depending on the placement of the additional surface area, the interface can provide additional resistive force to prevent relative motion as a result of shear and friction forces.
- Lastly it can reduce stored elastic energy, reducing indentation depth

Variance Active Surface Area:

- The variance of the active surface area for supporting, transmitting forces and preventing limb offset. The variance is calculated across the entire testing time, as opposed to a single frame.
- Large variances indicate that a combination of separation and indentation is occurring that both reduces and increases the effective surface area. This may be caused by a combination of limb volume change during ambulation, and kinematic offset and indentation that occurs as a result of applied and user dynamics.
- A low surface area variance indicates that, whether contact area is initially low or high, then contact will remain relatively stable, and the magnitude of separation will be reduced.
- The implications on user safety as a result of high surface area variability primarily influence musculoskeletal injuries, followed by surface skin injury as a result of offset and chafing/sliding.

Center of Pressure Location Variation:

- Center of pressure location variation indicates the shifting center of pressure across a 3D surface map. The variance is calculated using all time steps.
- A higher center of pressure location variation may cause deep tissue, skin surface and musculoskeletal injury depending on the mode of interaction.
- Center of pressure variation caused by a result of shifting or rotating along the surface with high normal forces may cause deep tissue shearing when contact is maintained.

- When normal pressures are relatively low, the same shifting or rotation along the surface may cause surface level skin abrasion as shear is not maintained and instead sliding occurs.
- Musculoskeletal injury occurs in both of these cases with center of pressure variation allowing for the identification of out of plane musculoskeletal injuries.

Photos were taken post testing of the lower thigh strap location to observe reddening and bruising. The same lighting and conditions post testing (within 5 minutes of finishing tests), were ensured so that observational skin condition evaluation would be consistent.

4.2.5 Results

Experimental results are reported in 2 sections comparing custom and generic strapping: 1) pressure, testing the hypothesis that custom interfaces improve predictability of pressure distribution between, and 2) metabolic cost, testing whether custom interfaces decreases energy cost of movement.

Pressure Data

Tables 4.1 and 4.2 detail the pressure data from the generic and custom straps, respectively, for all 5-minute walk tests.

Table 4.1: Pressure data from generic strap 5-minute walk tests Green highlighted trials are shown in more detail in Figs 3 and 4 Mean values are highlighted in blue

Speed of Test	Test #	Tekscan Location	Mean Pressure (mbar)	Pressure V ariance (mbarˆ2)	Mean % of Tekscan Active	Variance of Tekscan Active	X CoP Variance	Y Variance
1	1	Hip	14.85	2.13	0.27	0.010	0.12	3.21
		Thigh	16.40	20.75	0.04	0.001	7.72	2.69
	2	Hip	14.61	0.98	0.30	0.006	0.06	2.32
		Thigh	15.37	38.65	0.04	0.001	26.04	4.62
	3	Hip	13.55	0.74	0.16	0.003	0.07	1.39
		Thigh	4.61	42.97	0.01	0.000	36.73	4.34
	4	Hip	13.64	0.45	0.16	0.002	0.10	0.76
		Thigh	7.24	36.46	0.01	0.000	31.97	2.86
	Mean	Hip	14.16	1.08	0.22	0.005	0.09	2.77
		Thigh	10.91	34.71	0.02	0.000	25.61	3.63
3	1	Hip	14.85	2.34	0.24	0.008	0.09	2.97
		Thigh	8.84	38.50	0.02	0.000	18.24	2.81
	2	Hip	14.73	1.14	0.27	0.009	0.10	2.83
		Thigh	20.04	135.68	0.03	0.001	28.37	3.89
	3	Hip	13.91	0.51	0.14	0.001	0.07	0.41
		Thigh	5.44	36.92	0.01	0.000	30.59	2.66
	4	Hip	13.75	0.42	0.15	0.002	0.10	0.54
		Thigh	6.69	44.39	0.01	0.000	24.29	2.68
	Mean	Hip	14.31	1.10	0.20	0.005	0.09	2.35
		Thigh	10.25	63.87	0.02	0.000	25.37	3.01

Table 4.2: Pressure data from custom strap 5-minute walk tests. Green highlighted trials are shown in more detail in Figs. 3 and 4. Mean values are highlighted in blue.

Speed of Test	Test #	Tekscan Location	Mean Pressure (mbar)	Pressure Variance (mbar^2)	Mean % of Tekscan Active	Variance of Tekscan Active	X CoP Variance	Y CoP Variance
1	1	Hip	21.54	139.11	0.36	0.012	0.06	0.26
		Thigh	12.11	10.20	0.18	0.013	3.06	0.21
	2	Hip	23.40	199.44	0.46	0.008	0.03	0.34
		Thigh	11.20	4.05	0.19	0.009	0.70	0.07
	3	Hip	21.04	221.27	0.31	0.008	0.13	0.34
		Thigh	11.65	5.45	0.17	0.008	0.65	0.05
	4	Hip	20.19	152.63	0.39	0.007	0.06	0.33
		Thigh	10.75	2.45	0.13	0.007	3.11	0.25
	Mean	Hip	21.54	178.11	0.38	0.009	0.07	0.23
		Thigh	11.43	5.54	0.17	0.009	1.88	0.14
3	1	Hip	21.72	142.49	0.35	0.012	0.07	0.28
		Thigh	10.72	3.37	0.13	0.010	2.46	0.16
	2	Hip	25.25	300.05	0.43	0.010	0.04	0.31
		Thigh	10.52	2.76	0.12	0.007	0.88	0.24
	3	Hip	21.84	232.99	0.36	0.006	0.06	0.34
		Thigh	10.59	1.93	0.14	0.005	1.21	0.11
	4	Hip	19.69	139.94	0.37	0.007	0.07	0.42
		Thigh	10.87	2.37	0.15	0.005	0.70	0.05
	Mean	Hip	22.13	203.87	0.38	0.009	0.06	0.24
		Thigh	10.68	2.61	0.13	0.007	1.31	0.14

Figures 4.4 and 4.5 displays pressure data from selected (green highlighted) tests, chosen because they exhibit the lowest variance in CoP location across their respective speeds for the generic configuration. The equivalent test at that speed with the custom configuration is also provided for visual comparison. The hip test measurements, shown in Figure 4.4, are those at Speed 3, selected based on the generic surface performance and to showcase effects at different speeds. An increased amount of active surface area and pressure is seen, while the CoP locations are much more tightly clustered for the custom configuration.

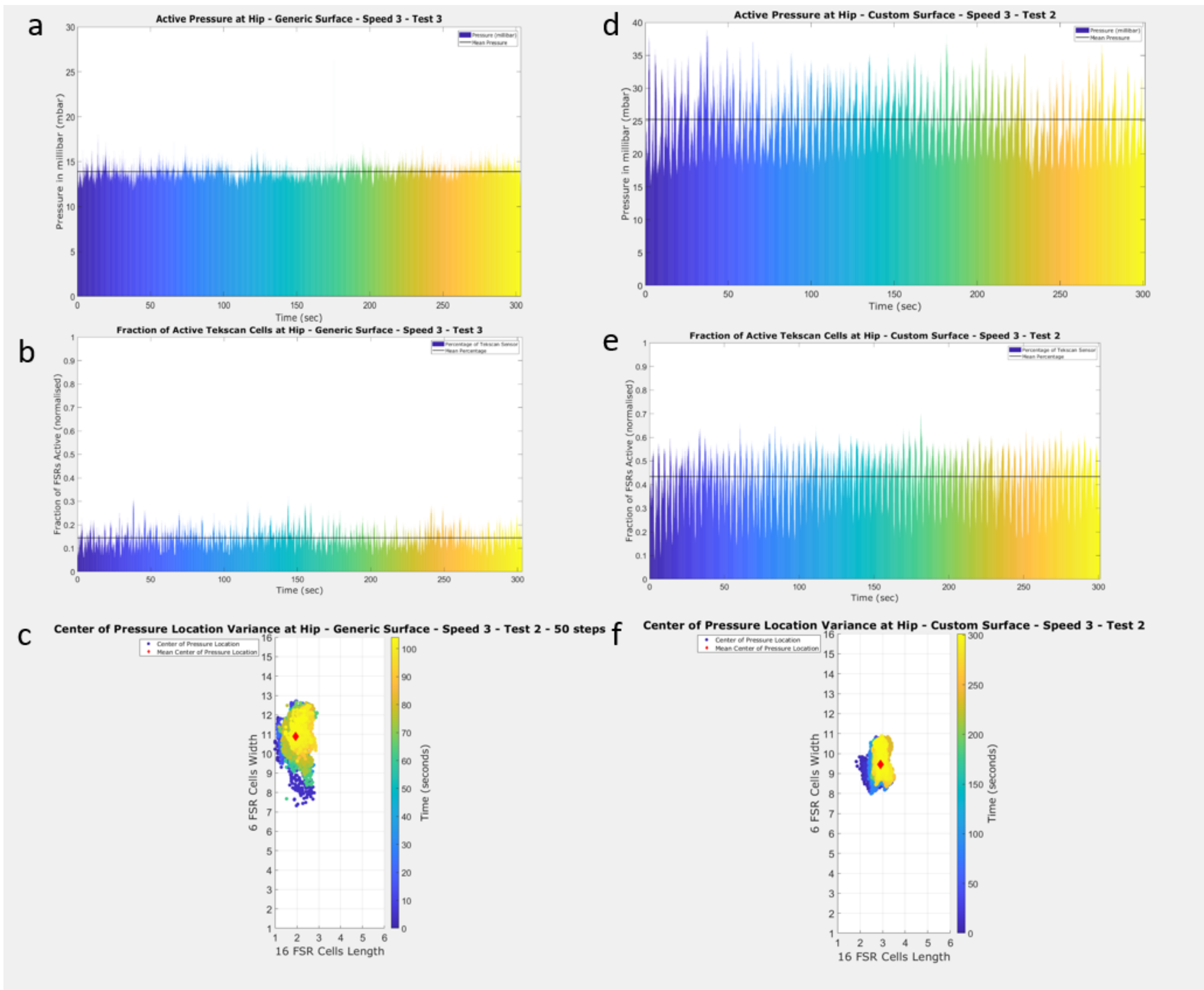

Figure 4.4: Post-processing data from FSR arrays at the hip. (a) Hip Generic surface: pressure during walking test (mbar) (b) Hip Generic surface: percentage of surface area active on surface (c) Hip Generic Surface: Center of Pressure locations during testing (d) Hip Custom Surface: pressure during walking test (mbar) (e) Hip Custom Surface: percentage of active surface area (f) Hip Custom Surface: center of pressure locations during testing

At the thigh, measurements presented in Figure 4.5 are those at Speed 1. Similar to the hip, more active surface area is used and the CoP locations are more clustered for the custom configuration. The custom configuration has a higher average pressure in comparison to the generic configuration, likely a result of the tighter fit as designed by OBS to reduce relative movements.

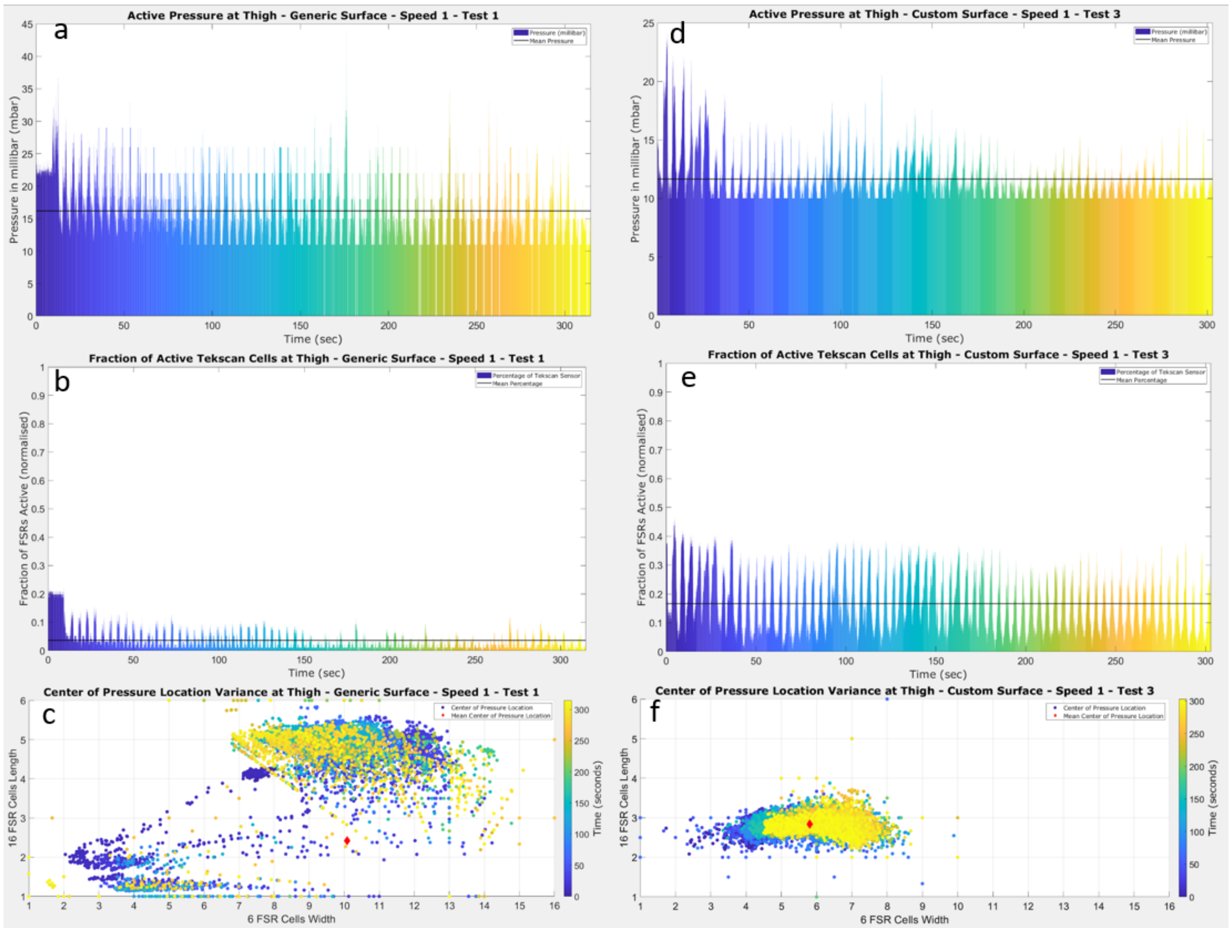

Figure 4.5: Post-processing data from FSR arrays at the hip. (a) Thigh Generic surface: pressure during walking test (mbar) (b) Thigh Generic surface: percentage of surface area active on surface (c) Thigh Generic Surface: Center of Pressure locations during testing (d) Thigh Custom Surface: pressure during walking test (mbar) (e) Thigh Custom Surface: percentage of active surface area (f) Thigh Custom Surface: center of pressure locations during testing

Pinch points exist at the edges of the surface and near the hip joint centers where the pilot study user reported local soreness after use. The user also reported soreness on the posterior lower thigh. The effects of the generic straps can be seen in Figure 4.6. The small surface area of both the supporting aluminum surface and securing straps, and the poorly conforming interface likely contributed to injuries observed. Relative shifting occurred much more often during generic testing, and with a poorly conforming surface rubbing and pinch point shifting was likely to occur much more often. With a reduced surface area in addition to increased shifting this is likely what was occurring as the worst of the reddish-purplish bruising occurred around the sharp edges of the supporting surface and securing straps.

Figure 4.6: Bruising and Injury (red marks) as a result of utilising the generic interfaces at the lower thigh interface.

These types of skin marks were not present after any tests with the custom surfaces, an indicator of the benefit of fully conforming surfaces. The custom interface marks left over from testing can be seen in Figure 4.7. Indentations left over during custom testing can be seen, however the same reddish bruising is not present, instead only a lightening of the skin around interface edges creating the appearance of an outline. Despite having higher interface pressures, the custom interfaces, the conformity of the interface, change in layered material composition and increased surface area have contributed greatly to the indentation impact on the user. As a note, these interfaces are worn for the same duration during testing (around 1 hour).

Figure 4.7: Lighter skin tone as a result of utilising the custom interfaces at the lower thigh interface. Some denting can be made out, however the same characteristic red-purplish color characteristic of skin damage is not seen.

Overall, pressure data indicates that customized surfaces performed better than their generic alternative at distributing pressures and maintaining contact during tests. The improvement of the custom torso orthosis was most substantial in introducing more active supporting surface area, with an average increase of 16.7% at speed 1 and 16.4% increase at speed 3 to 22.7% and 20.8% respectively. The CoP variance for both speed 1 and speed 3 were 5.6 and 4.3 times smaller than that of the generic surfaces.

4.3 Discussion

In this study, the consequences of custom and generic support surface interfaces are investigated. While it is difficult to interpret from solely the results presented, an analysis and assessment of how each metric can be used will inform how pressure data recorded by FSR arrays can be interpreted to evaluate the interfaces. Later in this section, utilising the combined interpretation of recorded metrics, we will show that the performance of the custom surface was better than that of generic interface. The metabolic data is less conclusive, perhaps due to a range of factors that will be discussed further in this section.

4.3.1 Pressure Data

The experimental tests were designed for this study to evaluate common use case scenarios for exoskeletons inspired by walking rehabilitation strategies [64,68,148]. A 5-minute walk test at two different speeds represents longer rehabilitation and exoskeleton use studies. Many studies involving the use of exoskeletons in performance and evaluation review, involve use in straight walking or

Due to the limitations of the Tekscan sensors' capabilities, only two FSR arrays could be used concurrently. The rationale for the placement of each Tekscan sensor was based on common pressure ulcer formation locations for individuals with mobility impairments. Pressure injury formation is common near the ischial tuberosities and hips due to their increased surface height relative to the surrounding topology (i.e. bony prominences), and their constant loading in sitting and lying positions [155], in vulnerable populations. Utilising two common locations consistent for pressure injury formation should also be locations of concern when utilising an assistive device. When utilising an assistive device, the risk for injury formation may exist at many locations. The goal is to optimize pressure at all contact locations so that successful ambulation occurs, designed so that it does not cause injury to the user especially for locations which are common to see injury at.

The equivalent locations on the exoskeleton are the hip and the posterior part of the top thigh strap. The hip actuator naturally delivers torque about the hip contact region and acts as a rigid non-elastic surface. Reaction forces occur here due to the relative location of the hip bone to the surface of the skin. In order for relative motion to not occur as a result of reaction forces causing the torso orthosis to slide away from the user, a significant amount of force is required to provide frictional resistance as an assisting resistive force. Additionally, these hip surfaces provide resistance to side to side motion that accompanies ambulation. The tekscan placement was oriented to capture information along the midline of the torso in the sagittal plane. This placement was made to capture both information of tilting in the frontal plane during ambulation, and static loading pressures at the hip's bony prominence. The posterior surface of the thigh has a large surface area which is well suited to capture information, and is one of two major surfaces crucial to the lifting of the leg. Radial pressures on any of the lower limb segment straps are caused by a static strapping pressure that holds the limb in place. With a much higher surface area, and its significance as a potential location for skin injury, this interface location was selected to capture as much available area as possible.

The metrics selected for analysis draw upon, and are informed by, skin injury mechanics, and surface interaction or energy principles. Traditionally, one dimensional pressure analysis utilises PPT, a difficult to corroborate value that highly values individual thresholds when interaction may be more complex [89]. Additionally, one dimensional pressure and force threshold analysis relies heavily on reaching maximums measured under static, non-dynamic load conditions, bringing into question its usability as an evaluation metric [89]. While it is still a crucial component of injury, there are other context based information we can use for better analysis. While force (and non-equilibrium force) is critical in driving the device, pressure informs the deformation of tissue and bone. For the purpose of this study, only pressure analysis was conducted as force can only be used as a means to evaluate musculoskeletal risk and not tissue risk.

Average pressure was taken as the primary indicator for surface pressure thresholds, a common approach to indicate overall potential for risk as opposed to a binary risk which may arise from one localised pressure or pinch point [89]. Peak pressures can be utilised as a means to evaluate locations where high pressure formation may occur, but the effectiveness of this can be masked by artifacts and noise, and heavily relies on the sensors ability to capture the information reliably. Alternative methods of identifying high risk areas for pinching or localised force utilising alternative surface pressure interpretation may be preferable. Variability in the securing in conditions (e.g. how hard the hook and loop straps are pulled down), may result in a difference in the results. While these values could not be easily controlled between tests, the same securing donning procedure was used, including the same individual securing the participant down with their maximum pull strength during testing. This initial pressure difference may explain some of the differences observed in each of the proceeding metrics.

The first value utilised is pressure variance. The variance calculated here is the variance amongst the tekscan cells, averaged across time. What this analyses is how poor the uniformity of the distribution of pressure is at each frame, and throughout testing. Ideally, a low variance of pressure over the surface indicates that the pressure is relatively uniform across the active supporting surface. Only active supporting surface area is considered here, accounting only for active cells. Ideal conditions would have a low pressure variance, indicating that when cells are active the distribution of the magnitude of pressure is similar. This tool cannot be utilised on its own, but is good at indicating when pinch points may be occuring, as if areas of peak pressure exist, this can be used to indicate it.

Percentage of active Tekscan, is the fraction of active cells at any time frame. An active cell is one that is currently providing a load, as any cell that does not provide assistance is an ineffective component of the supporting surface. For both surfaces, hip and thigh, the Tekscan surface area covers important interaction points. A higher contacting area in those regions indicates more contacting points for proper unloading of pressure. While this metric is not useful entirely alone, it does highlight that a surface is being utilised and is contacting the user. During similar motions, a higher active Tekscan surface area means more points remain in contact when compared against each other. While this might not be enough to indicate adequate distribution of pressure, a higher active area will contribute to avoiding pinch point and highly localised pressures. A high variance of the active Tekscan area may be an indicator of relative motion between the strap and limb or cyclic loading from gait. It may also indicate that large relative movements and offloading are occurring as a result of poor conformity or static holding pressures providing resistive friction. The next metric helps define these conditions.

Center of Pressure location variance is the final metric utilised, and helps tie the preceeding metrics together. A high CoP location variance indicates pressure locations are moving during use. CoP variance may arise from a number of sources, including user influence, contact conditions settling over time, poor contact fit, kinetic influence and soft tissue volume change. CoP variance may indicate no risk, but under conditions where variance is high, and surface pressures or active area remain relatively constant, it may be an indicator that shear and relative movement under contact occurs. High levels of shear and movement under friction can cause deep tissue and friction injuries respectively [155].

Each of these parameters highlights a known mode of pressure injury formation. When analyzed together, they provide insight into the risk of pressure injury formation that is no longer one-dimensional, and instead accounts for performance and safety [89,101,102,107,125,155,156]. The real value of these metrics is when combined. General principles regarding how pressure can be used for analysis is dependent on the subject and test, but this information reinforces it. Average pressure remains a tool to indicate differences between similar condition testing, but should not be used as the ultimate decider in performance.

High pressure variance may indicate under certain conditions that non-uniform offloading is happening during movement, or that pressure may be distributed poorly. Combining this value, with percent active surface area, can indicate whether a change in area is occuring from cyclic offloading. A high CoP variance with low pressure, pressure variance and active surface area may indicate a lot of relative internal movement and shear. A low pressure variance, with a low active Tekscan area and high mean pressure will indicate a pinch point. This analysis, and other combinations of this information can be done without visual inspection. If certain components of the gait cycle are of interest for interaction analysis they can be split apart and performed similarly. For the purpose of this study, the general trends of the system were observed over testing, as opposed to any particular component of the gait cycle.

From a high level perspective utilising these same principles it is possible to analyze the generic and custom surfaces. Starting with the thigh, the generic surfaces exhibited similar mean pressures, a low average active area over time (2% on average), indicating high separation occurences, and a widely variable CoP variance, indicating friction or friction related injuries may occur when compared to the custom interface.

The hip surfaces performed closer relative to each other, and require explanation of benefits for both. The generic interface exhibited lower average pressures, and much lower variance. Center of pressure variance was similarly low when compared to the custom interface, but performed worse in active surface area. The custom interface exhibited higher pressures (likely from the conforming design), but a much higher variance. With the low center of pressure variance, (and a higher variance in percent active Tekscan), this difference may be a result of the allowable offloading that comes from the compliant torso design during gait. The high variance in pressure is likely a result of the non-linear offloading that occurs, as the hip rolls are the last component to unload to help maintain a consistent center of rotation when contacting with the interface.

With this in mind, before diving deeper into the differences, the custom interfaces seemed to on average perform similar (hip) or siginificantly better (thigh) than their generic counterparts.

The CoP of the hip in the generic configuration was located anteriorly to the midfrontal plane, superior to the user's waist fold. The anterior spread of pressures likely reflects the user's tendency to lean forward while using crutches. The custom surface's CoP exhibited less overall variation in the CoP location and the mean CoP was more in line with the midfrontal plane. In both configurations, the CoP shifted anteriorly with time but this shift is less prominent in the custom configuration. The variance of CoP location for both configurations is relatively low, indicating a fit with little shifting during walking.

The major differences between these two designs come from the percentage of surface usage, active pressures and their variances. The generic surface system has a low mean percentage usage of the torso orthosis (14.5 percent) localized about the anterior portion of the torso and waist roll. The custom orthosis has a much larger contact surface area (43.4 percent), localized similarly around the waist and hip region. Both have relatively low variations in active percentage, indicating the generic torso coupling remains in consistent contact despite a relatively smaller contact area. The magnitude of pressure is higher on the customized surface which can be attributed to the tight fit about the waist roll and hip of the orthosis. The custom torso orthosis exhibits a higher variation of the pressure distribution within a frame, which is likely due to the additional surface area which does not provide support and is not accounted for in the variance calculation. This non-supporting surface area are locations within the custom surface sensing area which do not contact with the user as tightly as the generic interface. This variation may also be explained by out-of-sensing area effects, as the vast majority of the generic surface area was captured by the Tekscan sensor, but the custom interface had a much larger surface area, offering more points for support not within the Tekscan area.

With the CoP cluster resting closer to the midfrontal plane, it can be extrapolated that the pressure distribution is more uniformly distributed in the radial direction (or transverse plane). Even with higher mean pressure and variance of that pressure over the surface, the tighter cluster of CoP indicates that the larger active surface area has helped distribute reactionary forces.

The thigh surface collections exhibited the largest differences between generic and custom designs. The generic surface exhibited a wide range of CoP locations with many occurring near the edges of the aluminum support surface. These locations were identified as potential pinch points as the abrupt ends of support surfaces can cause pressure to accumulate. Without a snug fit, the surface will move relative to the limb and fail to provide sufficient resistive forces while causing shear at the limb-strap interface. The custom surface exhibited a tight cluster of CoP locations relative to the generic surface near the bottom-midline of the surface, closer to the lateral edge of the thigh segment. The design of the custom orthosis avoids these pinch points by having a larger width and more gradual curvature at the edges of the strap.

The cyclic nature of gait causes natural movement between the individual wearing the exoskeleton and the device itself due to elastic tissue deformation. In both the generic and customized surfaces, a pattern of increasing and decreasing pressures and contact area are a result of the phases of gait. The observed active support area of the generic surface supports the initial assumption that the pinch points and sharp edges are the major, or only contact points on the limb introducing large amounts of stored elastic energy. During the generic testing session, an average of 3.6 percent of the thigh Tekscan cells were active, with some time frames recording zero contact indicating full separation of the limb segment from the coupling surface. In contrast, the custom coupling surface uses an average of 16.6

percent of the surface area, a significant improvement over the generic surface.

Overall, pressure data indicates that customized surfaces performed better than their generic alternative at distributing pressures and maintaining contact during tests. The improvement of the custom torso orthosis was most substantial in introducing more active supporting surface area, with an average increase of 16.7 percent at speed 1 and 16.4 percent increase at speed 3 to 22.7 and 20.8 percent respectively. The CoP variance for both speed 1 and speed 3 were 5.6 and 4.3 times smaller than that of the generic surfaces.

Whereas static pressure is a poor indicator for pressure ulcer formation in a dynamic interface [89, 155, 156], the high pressure and pressure variance at the hip is likely due to the intentional introduction of the waist roll bump which ensured firm and constant contact around the torso of the subject. It may seem counterintuitive to increase pressure but doing so likely decreased CoP variance.

Across all tests, surface performance during speed 3 tests were worse than those of speed 1 tests, likely a result of increased dynamic interactions. Minimizing CoP location variation by using more snug and conforming designs decreases the potential for translational or rotational effects and resultant shear pressures on the surface of the skin. Across both the thigh and torso straps, walking tests at two different speeds demonstrate that the increased contact of custom straps partially decreases pressure points associated with pressure injuries. Furthermore, low variation in active area demonstrates less relative movement leading to lower shear experienced at the strap-limb interface.

4.4 Conclusion

4.4.1 Limitations

The FSR sensors cover a large surface area, however, the available Tekscan system allows only two FSR arrays to be used concurrently. Future studies should consider a sensor setup used with FSR arrays placed at more than two locations to properly evaluate the forces experienced by the distal straps of the exoskeleton. Evaluation of the pressure maps on additional surfaces may highlight an increased uniformity and performance across all interfaces that reduces the variance across multiple similar purpose interfaces.

The individual involved in this pilot study is able-bodied; the H3 is intended to be used by individuals with mobility impairments. Involving and interviewing individuals from vulnerable populations is recommended to further examine the effects of custom surfaces on assistive robotic devices. The influence of these interfaces may be more dramatic in improvement under conditions with intended use populations.

4.4.2 Future Reccomendations

A future study examining the surface area and pressure relationships of a lower limb exoskeleton across a wide array of individuals whose body types are intended to use the device is recommended. Forces identified in this pilot study may not be repeatable across other studies (a common issue in exoskeleton characterisation efforts [89]), and as such a full-scale investigation should be undertaken to identify which locations experience high pressure in both passive and active exoskeleton use.

Many of the components of design that defined this study for improving the surfaces required the use of heuristics and common knowledge recommended by clinicians. The authors' recommendation is to involve a clinical expert when designing new rehabilitation devices and standardize engineering design practices to reflect the practices seen in clinical orthosis design. The compilation of best practice methods for the manufacturing and fabrication of active and passive wearable assistive devices is crucial in establishing how these devices should be manufactured for effective use. While documentation does exists for design safety already, these safety conditions are minimum safety requirements, and not how certain design decisions impact those conditions.

4.4.3 Conclusion

The design and manufacturing of custom exoskeleton straps is costly and may not be feasible for most exoskeleton users; this study examined the pressure and metabolic consequences. Beneficial and substantial improvements observed in this pilot study indicate that fully customized surfaces reduce the risk of pressure injury by a number of design decisions. By providing a greater surface area to act on and ensuring more consistent contact the pressures distributed on the user - regardless of magnitude are more uniform and avoid sharp gradients which may cause pinch points, unwanted slipping and redness to form on the skin. For these reasons, future exoskeletons designs should integrate custom orthotic fabrication techniques, and take inspiration from increased surface area and conforming curvature in design. Further investigation into relative metabolic cost is required to completely evaluate the impact that custom surfaces have on relative effort.

Chapter 5

Conclusions and Contributions to the Iterative and Sensor Based Evaluation and Design of Exoskeleton Coupling Interfaces

The process for the evaluation and development of new coupling surfaces lacks the transferability for information, required for comparison and sound justification of benefits. As highlighted in these prior chapters, the current process components for evaluation: the methods, appropriate metrics and baseline comparators, are primarily suited for establishing the safety of a device, and not how to improve it. Comparison and replication between studies as a result is essentially difficult from a combination of human factors, limited ability to transfer and interpret recorded data between tests and no common designs that any study relates back to. These issues were addressed in these prior chapters in three distinct ways. Over the course of my thesis, the development of these factors were concurrent, and often informed one another, but analyzing and reporting on them as separate components is critical in establishing where existing studies may be able to implement their own improvements.

The goal for this part of the thesis, “Part 1” was to establish new means of comparing exoskeleton interfaces to inform better designs and innovation for both user safety and performance. I focused primarily on improving the existing and most commonly implemented means of evaluation, and implemented a new set of fully customized surfaces inspired by prostheses and orthoses design to compare against the most commonly used “generic” interfaces which fit a wide range of users.

By focusing on the already existing methods, these new contributions can be more easily implemented into existing frameworks, requiring less effort on the part of the designer to evaluate and design new coupling interfaces. The current means of evaluation are limited primarily to addressing concerns of safety, but not on providing the information required for comparison and innovation. The contributions made in this thesis look to address and mitigate this problem by tackling three of the identified sub-components of the evaluation process to improve the transferability of information.

The first component tackled was the methods of evaluation. Traditional means of evaluation involve the use of an active participant. This brings issues in primarily associated with user input, and the inability to repeat or achieve similar results from studies due to the conditions of testing. Dynamic movement, activities of daily living, the mobility impairment of the users, demographics and sample size are all limiting factors on the transferability and comparison of information between studies. While the utilisation of mannequins is not novel for exoskeleton evaluation as a whole (controllers, actuators), we implemented and investigated the use of mannequins, explicitly for the evaluation of coupling surfaces. The utilisation of mannequin's to validate preliminary data for the performance of a coupling interface is relatively new, and is intended to be used as an intermediary step for comparison [103]. The manufacturing and implementation process for a mannequin limb was demonstrated to be relatively simple and intuitive, and resulted in a much simpler evaluation of the metric (kinematic limb offset) that it was utilised for. With the removal of user input, the influence of each surface, generic and custom, on the limb offset was much clearer and easier to explain. This implementation is not intended to replace human testing, however, but to complement it and act as a preliminary step for transferability and comparison of coupling interface design.

The second component tackled was the appropriate metrics for the evaluation and comparison of exoskeleton coupling interfaces. While no new or novel metrics and sensors for evaluation were tested, the focus instead was on better applying existing metrics that better represented the safety concerns associated with coupling interfaces that have been well addressed in review literature. The most commonly used and analysed metric in evaluation is pressure through the framework of pain pressure threshold obtained through algometry. This metric is useful for evaluating safety, but may not be applicable in dynamic and high surface area use cases, as identified in review [76, 89, 155], and does not describe the modes of error or poor design that may cause injury as a whole. Joint misalignment and joint offset studies were utilised as inspiration to find metrics which do not necessarily require information about the user, and allow for a contextual based evaluation. The implementation of various pressure based metrics (mean, variance, surface area and center of pressure variation), expanded upon typically evaluated pressure information to evaluate the benefit of custom versus generic interfaces.

The last component tackled was baseline evaluation. In order to develop and evaluate how new coupling interfaces function, and whether or not it is worth pursuing said design, a comparison is required that can be drawn on that acts as a testbench baseline or idealized solution or framework to compare against. Currently, no such design really exists, with many of the safety evaluation studies being insular in their investigation of both performance and safety evaluation of coupling interfaces [89]. To address this issue, inspiration was taken from orthoses and prostheses design with the help of a registered orthotist, Al Moore from Orthopaedic Bracing Solutions. With their guidance a set of custom coupling interfaces were developed for both torso and limb segments. The role of conformity in both user safety and device performance is well understood in prosthetic and orthotic design, and was thus applied to exoskeletons. When observing kinematic offset and context based pressure evaluation, both of which are major components in both user safety and device performance (musculoskeletal, skin injury, force transmission), the customised surface performed better than the generic surface. While more work is still required to create a full understanding of the benefits of the customised surface, such as impact on controller, direct manufacturing cost, and an expanded pressure and kinematic evaluation on a larger sample size, this initial groundwork serves two functions.

The contributions made in this section presents an outlook on surface performance that is rooted more strongly in safety for musculo-skeletal and skin injury informed by context based evaluation. It allows for the measurement and comparison of surface performance for both safety and performance where the currently established methods, metrics and non-existent baselines were better suited for surface level evaluation. The contributions made by the researchers whose work came before me on evaluating the safety of exoskeletons laid crucial groundwork and establishing theory for this section to be possible.

Chapter 6

A Brief Review on Modelling Approaches for Lower Limb Exoskeletons

This chapter is aimed at the objective to review the current approaches and models used in simulating and controlling the exoskeleton coupling interface. The review discussed here is also an introduction to proceeding chapters to address the remaining objectives for Section 2 (Investigating commonly implemented coupling models, Develop new models that more accurately represent the interaction mechanics at the user interface).

6.1 Introduction

For the control and rigid body dynamics simulation of exoskeletons, the models which define kinetics and kinematics are crucial to their success. Typically, exoskeleton systems are defined by rigid body dynamics [47,58,71,73,75,152,165,170,202], but can also be modelled as continuous soft systems [69, 71, 103, 145, 151]. Rigid body dynamic models incorporate equations which define the position of each exoskeleton component relative to an origin, and the respective torques and forces required to drive those components through planned or estimated trajectories. Regardless of approach, these models are required to describe the constraints and couplings which define their motion, including the coupling between the exoskeleton and user, the ground and the exoskeleton components itself. Governing constraint equations detail how systems work together, and allow for the calculation of kinetic and kinematic values associated with the components tied to those constraints. A necessary version of these constraint equations, and the focus of this chapter, are the exoskeleton-user coupling equations. If these equations and relationships are poorly modelled, potential musculoskeletal injuries may occur from misalignments or overly torquing joints, skin injuries can form from high shear or normal pressures and models may experience high inaccuracies in tracking user limb segment position, and exoskeleton control values [70, 76, 77, 81, 89, 92, 98, 100, 104, 107, 123, 125, 129].

The modelling of the exoskeleton coupling interactions is critical in understanding the effect of force and torque transmission from the exoskeleton's actuators to the user. If information is known about the torque and position of the exoskeleton system, the exoskeleton coupling equations can be used to calculate and estimate the forces applied and torques applied to the user. These forces can then be extrapolated to estimate the dynamics and kinematics of the user wearing the device, assuming little torque is imparted from the user. For lower limb exoskeletons, calculating the trajectory of the user's leg segments is determined by these equations for simulation, estimation in control, and when only partial sensorization is available.

While "Part 1" detailed the characterisation of the coupling interface utilising sensors and test scenarios, the models presented in this section "Part 2" describe similar reactions through simulation and governing equations. These two facets of the design and evaluation of the exoskeleton coupling interface are the most common means to describe their safety and performance [89]. The modelling and simulation allows for the transferrence of information from characterising a system for one individual to another.

This characterisation and evaluation process is used primarily in simulation and modelling the mechanics of the exoskeleton coupling interface. Both interaction forces and positions can be estimated from these equations allowing for the simulation of interfaces to relate directly to the sensor based evaluation. With proper sensorization, the characterization and evaluation of the same metrics used in section 1 (e.g. pressure, force, joint offset) can be used in the modelling and control of simulated coupling mechanics. This allows for the transfer of information and evaluation of different use cases and controllers for an exoskeleton without the need of a prior fabricated and manufactured coupling interface.

The concept of transferrable simulation is ideal, specifically the ability to evaluate new models or controllers prior to testing, but there are still major drawbacks that limit their accuracy and benefit compared to their theoretical benefit. First, the accuracy of the variables associated with testing are dependent on the data that supplies them, and are variable from person to person, and testing conditions [69, 93]. Due to the high potential for variance in coefficients, the effectiveness of simulated results is highly dependent on matching the scenarios and initial conditions present in the collection phase. Additionally, in order to retrieve the data in this testing phase, an already fabricated interface must exist to test [89, 116]. This requirement puts risks on the user for initial testing, and prevents the design of fabricated with desired or exactly known coupling interface responses. While the capability to transfer information is useful, it still requires user-worn testing and costly fabrication to generate the governing equations necessary for simulation. With more accurate models, and the capability to simulate before fabrication, a more appropriate starting point of evaluation for interaction mechanics and user safety, at the coupling interface can be reached without the barriers to evaluation that currently exist.

As their reliability advances, the potential use cases for physical interaction models opens up past simple kinematic-kinetic simulation. Accurately determining the threshold forces, pressures and shear without the need for an existing physical model is critical for user safety. Doing so also allows for offline rapid prototyping and design, creating new controllers or physical human-robot interfaces that minimise risks through optimal design. Currently, the process of coupling surface design, and modelling that same surface for control, are completely separate processes [89]. As a result, desired properties (e.g., low static pressure, or high coupling surface area) may be designed optimally utilising sensors and heuristics, but can yield poor results in dynamic coupling effects or simulation and control performance. This decoupled design process makes achieving the overarching goal of minimising user risk while maximising exoskeleton performance difficult to achieve.

6.2 Background

Modelling for exoskeleton coupling interactions is primarily for controllers and simulation: assessing kinematic offsets and change in dynamic load during trajectory following. Each coupling surface has a respective constraint equation which represents the linkage between the user's limb and exoskeleton at that point. All those governing equations put together, along with the applied torques, forces and position of the exoskeleton, can define the position of the user's limbs. This information is then propagated forward to the next time step and recalculated based off of the new information, and the prior steps stored information to solve for the new leg position. The governing equations are often formulated as functions of the state space (position, velocity, acceleration), and the resultant forces or energies required to generate motion.

As a result of being used in simulation and control, the most commonly used models are simplistic for computational speed and ease of implementation. The two most commonly used models are the perfectly coupled and linear elastic spring-damper models [58, 89].

The latter of the two being more common in recent literature, and the focus of most state-of-the-art improvements.

Perfectly coupled models assume the user's limb(s) follow the same trajectory path as the exoskeleton without deviating. An exception to other models, the governing equations here are simplistic by "lumping" the inertial components (mass, moment of inertia) directly into the kinetic governing equations that calculate required torque from the exoskeleton [58,60,89,116,165,202,203]. This simplification vastly reduces complexity for the governing equations, requiring fewer equations to solve for the system, with a constantly "known" position of the limb. Instead of requiring multiple equations, a singular matrix driven equation describes the system, such as an Euler-Lagrange governing equation shown in Equation 6.1. M is the mass-inertia matrix, C contains the governing velocity dependent terms, G is gravitational components, T is the applied torque, and J represents the Jacobian for the positions of the applied force loads. While useful in simplifying simulation and control equations, the relationship between user and device exhibits elastic properties [76, 77, 89, 90].

$$\mathbf{M(q)}\ddot{q} + \mathbf{C}(q, \dot{q}) * \dot{q} + \mathbf{G(q)} + \mathbf{J(q)}^T F_{int} = \mathbf{T} \tag{6.1}$$

The linear elastic spring-damper model is the natural progression for adding elastic component consideration to the exoskeleton surface utilising the fundamentals of Hooke's Law. This formulation is widely implemented and is not limited to just exoskeleton coupling surfaces, finding use in a number of robotics and biomechanical applications. It describes the coupling between two points or "nodes" as some combination of spring and damper elements [70, 75, 89, 90, 129, 132, 133, 204, 205]. In the case of linear spring-damper systems, there are only two elements that act as a linear combination of the states. Other applications, such as the Hunt-Crossley model, implement non-linearities to improve accuracy [89, 173]. This model specifically relies on the characterisation of parameters through the use of estimation algorithms, such as least squares approximation which compares simulated results and fits them to measured results. Two spring-dampers are placed at each coupling point along the surface of the limb which takes into account the current and previous time steps of the limb and exoskeleton to calculate the next position. Each spring-damper is activated by displacement from a selected center line, usually the center line between actuators, and determines the force-displacement relationship in the sagittal plane. These springs are typically limited to providing resistive forces to compression. The governing equation and diagram explaining the relationship are shown in Equation 6.2 and Figure 6.1. This governing equation is a constraint added to the previously defined Euler-Lagrange (or Newton-Euler) based equations that couple the inertial equations for the exoskeleton and relate them to the leg it is guiding, separating it into two distinct matrix based equations. Equation 6.2 is the governing force-displacement relationship, where k and b are the estimated spring-damper damper coefficients, and F is the interaction force. This equation is the general formulation and does not include the components to turn this into a piecewise function, such as the simple implementation of the Kronecker delta. Figure 6.1 is a simple representation of the spring-damper models in how springs exhibit elasto-gap models.

$$k(x_{exo} - x_{user}) + b(\dot{x_{exo}} - \dot{x_{user}}) = F_{int} \tag{6.2}$$

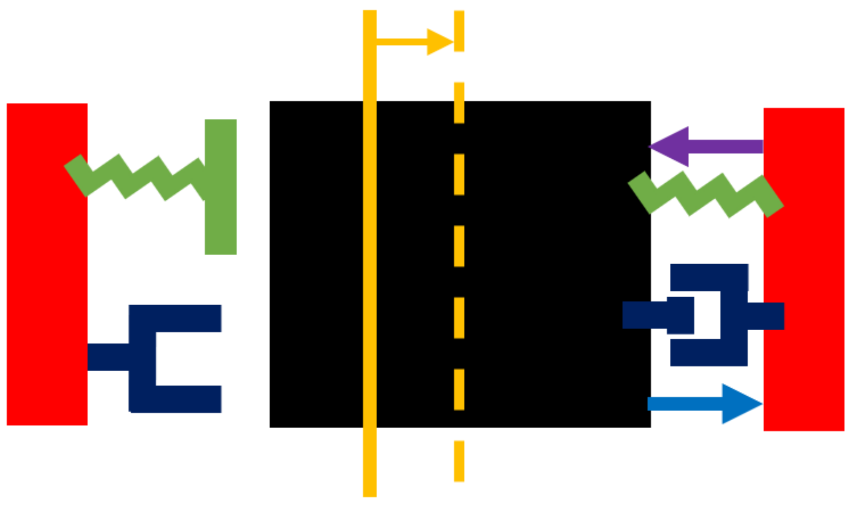

Figure 6.1: Graphic example depicting the spring-damper elastogap nature dependent on the position of the limb. No "sticking" is assumed, so that the material support surfaces only resist in the direction of compression.

These two governing models are commonly used to describe the relative position of the user's limb relative to the exoskeleton during movement for force transmission, and to recalculate the inertial loads required for ambulation. The drawbacks with these models are a direct result of their simplicity, driving the error within the model. Perfectly coupled and linear spring-damper models are simplistic due to the linearisation, or complete removal, of complex interaction mechanics [58, 70, 89, 129, 185]. Instead of representing many contact points coming into contact and separating non-linearly with depth, we utilise simpler models as they can be approximated using linear regression [69, 70, 75, 93, 129].

Models which involve more complex interactions in an attempt to mitigate the error associated with the linearities, such as the Hunt-crossley spring-damper, are more difficult to characterise. Models which involve more complex interactions, such as non-linear state combinations, require non-linear approximation methods as a result. The complexity of these models also effect the ability to calculate simulation. While characterising is difficult, so too is simulating their response due to complex derivatives and non-linearities introduced complicating numerical integration methods, the typical approach to simulation implementation.

State-of-the-art investigations are focused on the potential causes and sources of error arising from the linear spring-damper model, improving those faults or changing how the underlying governing equations of the model function. Only recently has the coupling interface been considered when characterising and modelling the multi-body dynamics present in exoskeleton dynamics. Recent studies have been conducted that acknowledge the influence of initial pressure on linear spring-damper coefficients, which has not (yet) been taken into account as an initial condition when utilising approximation techniques in other studies [69, 89, 93]. Additionally, alternative methods for characterising spring-damper coefficients have been explored to improve accuracy by implementing collocation strategies, assumptions of surface stiffness, force distribution between supporting surfaces, and the governing ODEs to improve accuracy [70, 89, 129]. Considerations of joint misalignment and offset effects is also being undertaken utilising a combination of linear spring-damper models with sensors to improve modelling accuracy and training to prevent injury [81, 89, 93, 94, 98, 100, 104, 106, 107, 117–119, 125, 161]. Lastly, new models are being developed as alternatives to singular linear spring-dampers that involve the discretization of support surfaces into arrays or beds of adjacent supporting spring-dampers to be characterized [89, 174, 177]. A further investigation into the state of the art is present in Chapter 8.

While state-of-the-art research and development has highlighted issues and difficulties with coupling-interaction modelling, it is still relatively limited. The focus of coupling interface studies lay the groundwork associated with the surface-level challenges of linear spring-damper systems, joint misalignment and cuff pressure and their effect on the performance and, in some cases, user safety of the exoskeleton [89]. The studies laying the groundwork in understanding the interaction mechanics between user and device are crucial for design and simulation. Current focus placed on the potential risk and injury to the user by introducing methods to model and evaluate joint misalignment prior to testing, and joint offsets utilising spring-damper models is critical in ensuring user safety and improving controllers. The introduction of newer models and data collected from sensors on the distribution of pressure and the variability that comes from initial conditions contributes strongly to the knowledge that there are challenges with coupling interaction mechanics.

The missing components of these studies, is how to use and implement this information in creating new and better design. Much of the work done characterising the problems has been complete, and only recently has state-of-the-art analysis, design and modelling focused on solving the identified issues of coupling interaction mechanics through modelling: pressures, misalignments, offsets and tracking [69, 89, 98, 174, 177]. Using the directions that modelling has taken as inspiration, and the drawbacks of prior research studies, it is now possible to define the problem statement and objectives of this section of the thesis.

6.3 Problem Statement and Objectives

The overarching goal of "Section 2" is to advance coupling models to allow for more accurate modelling-simulation results. Advancing coupling interaction models requires identifying the primary problems associated with them, and addressing them through focused objectives. Three primary problems were addressed based on literature review, and inspiration taken from state-of-the-art work that broke the mould in modelling interaction mechanics. These primarily focus on improving the force-position relationship models, and the drawbacks arising from an inability to design or characterise an interface's specific elastic and pressure responses without an existing fabricated interface and the safety and cost implications this brings.

The first problem identified was the lack of supporting studies validating the accuracy and performance of both linear spring-damper and perfectly coupled models compared to measured and tested coupling conditions. While understanding baseline offset(s) in estimation as a result of these assumptions is critical in adjusting or compensating for errors, human participant influences and other confounding variables challenges error measurements. Explaining the sources of potential error arising from mechanical interaction(s) that does not involve human input, such as elastic skin properties and volume change in the limbs, is incredibly difficult and likely a contributing factor to the dearth of studies directly addressing robust parameter estimation for exoskeleton coupling interfaces [89].

The second problem is the influence of initial conditions on the performance of perfectly coupled and linear spring-damper coupling assumptions. Most reports of exoskeleton interface spring-damper parameter estimation do not consider the initial conditions associated with characterisation past the initial position of the leg inside the exoskeleton. As a result, context based information such as initial strapping pressures, skin conditions, and material configurations, cannot be integrated into the characterisation process. This leaves position as the sole controlling variable relative to the predetermined center line, instead of the complex reactions that are actually occurring at the user-exoskeleton interface [70, 89, 129].

While studies indicate that initial cuff pressure does influence the spring-damper coefficients, there is currently no acknowledged means of implementing that consideration beyond a proposed scaling term [69, 93].

The third, and final problem, is the lack of published works looking to expand coupling interaction mechanics past single element spring-damper models [89]. Problems that arise from non-constant or poorly conforming contact areas, frictional components, non-linear elastic or dampening components, shear and relative material compositions, cannot be accounted for as adjustable variables or coefficients. As a result of parameter estimation relying on the data collected, and the models it fits, simple models like the linear spring damper cannot account for these complex interactions. New methods for implementing surface geometry are appearing to account for high pressure areas, and out of plane effects (not purely sagittal), but are limited in scope and implementation with focus primarily on simple design adjustments or basic pressure estimation [89, 174, 177]. Additionally, few existing tools comprehensively link the design process stages of exoskeleton coupling surfaces, and the simulation and control stages, with only one concrete example currently available [177]. While complex models utilising finite element methods, or first principle approaches, may exist, these models are not easily implemented for rapid prototyping or implementation in existing models for simulation and control of exoskeleton kinematics and kinetics.

A primary objective for this thesis is to improve the quality of models associated with kinematic and kinetic modelling to extend use past simple control and simulation estimation. Advancing these models for use as a tool for simulation, estimation, and design without the need of fabricating new surface prototypes would elevate the quality of coupling surfaces supported by the heuristic design principals highlighted in the previous chapters. My goals for the scope of this thesis, however, are to lay the groundwork for future studies and explore the potential that new characterising information and preliminary models have on exoskeleton coupling.

The goals for "Part 2" of this thesis are as follows:

- Investigate the commonly implemented and evaluated coupling interface models
- Evaluate the performance and accuracy of the commonly utilised model under highly controlled testbench scenarios
- Develop new models that more accurately represent the interaction mechanics at the user interface for use in simulation or coupling interface design

This chapter accomplished the first goal, and identified the current existing drawbacks. The next chapter showcases the groundwork laid to evaluate the performance and accuracy of the commonly used "perfectly coupled", and linear elastic spring-damper models to identify drawbacks and challenges in those models. This information then informs the proceeding chapters, and results in two new models and framework for analysis of the user-exoskeleton coupling interface.

Chapter 7

Validation of Common Models: Perfectly Coupled and Linear Spring-Damper Approaches and a Simple Extension of the Linear Spring Damper

This chapter is aimed at the objective to evaluate baseline models, and to develop new models that more accurately represent the coupling at the exoskeleton interface. TO evaluate existing baselines a perfectly coupled and elastic model are evaluated and compares them to recorded values to estimate their accuracy. Utilising this information, a preliminary model was developed to evaluate pseudo-static conditions of a spring-damper model loaded with pre-compression to more accurately estimate initial pressures and force profiles. (Investigating commonly implemented coupling models, Develop new models that more accurately represent the interaction mechanics at the user interface).

7.1 Introduction

Lower Limb Exoskeletons (LLEs) are wearable electro-mechanical devices that support the body weight of the user and can provide power at their joints. LLEs have recently become popular in research and rehabilitation settings as treatment aides for those with mobility impairments, e.g., Spinal Cord Injury (SCI) [76]. Often, interactions between end-user and device are mediated through compliant interfacing that responds to internal relative movements. The interaction between the user and interface is critical in the transference of force and controlled trajectories, and involves input from both user and device. The characterisation of the exoskeleton coupling interface is commonly performed through a combination of sensing and modelling to describe the relationship between force and user position relative to the device. With accurate modelling, trajectory planning and optimisation can be to achieve desired rehabilitation and assistive outcomes.

To provide ambulatory assistance, force must be transferred from the exoskeleton to the user through coupling interfaces. These interfaces vary between devices and may contain materials that are rigid, soft, or a combination of the two. To predict and control the trajectory of the leg, the coupling mechanism must be well characterized [89]. Commonly used exoskeleton coupling models are often simplistic or heavily based on assumptions regarding the mechanics of coupling and internal collisions. These mechanics are often represented as perfectly coupled or simple linear spring-damper systems that require experimental tuning or estimation to determine the coefficients.

Perfectly coupled conditions characterise the position relationship between the leg and exoskeleton as perfectly aligned. Forces and torques are calculated and defined as those necessary to drive the system as a connected rigid multi-body system of a lumped limb-exo segment based approach. The distribution of surface or interaction forces and torques is dependent on the solving of simple rigid body conditions under known constraints to determine how forces are imparted at the interface.

Linear spring-damper conditions are more advanced and accurate models when compared to perfectly coupled conditions and represent the interactions between user and device as one-dimensional spring-dashpot constraint equations. These models are most commonly used for their ability to characterize the elastic and viscous interactions during material deformation. The models themselves are one-dimensional, approximating the system utilising simple coefficients which characterise the relationship between deformation and force, allowing for kinematic offset and misalignment to occur as it does when applied force causes soft surfaces to deform. This deformation and offset causes a kinetic and kinematic phase shift, which is calculated using two separate multi-body dynamic system of equations as opposed to one. This system of equations more accurately depicts the exo-user interaction as a push and pull as opposed to a singular lumped robot. These models are primarily utilised in controllers and characterising models for their simplistic formulation, and for its abstract representation of a complex contact problem. The characterisation of elastic spring-damper models requires experimental kinematic or force data collected from a physical experiment, and fit to the governing position-force equation.

The design of these coupling models are simple for integration into modelling and control. Complex surface interactions and soft body mechanics and material deformation are often too complex and difficult to implement into multi-body system equations, and as a result approximation methods become the driving mechanics. The benefit of perfectly coupled and spring-damper models is in their simplicity, ease of implementation and characterisation.Implementing constraints and solving linear equations for coefficients is much simpler when performed online, or from kinematic data. The drawbacks of this simplicity, is the unmodeled components and errors that arise from a highly linearized governing equation. Without the more complex dynamics and material deformation, the models are limited to how closely the system acts to linear conditions.

The magnitude of this inaccuracy has not yet been generalised [89], instead relying primarily on the identification of kinematic-kinetic inaccuracies in modelling and control on a case by case basis. Identifying where inaccuracies arise is difficult due to the limitations of isolating interactions in user based studies and complexities of non-linear interaction. Generalised solutions that identify the magnitude of inaccuracy due to the nature of testing with user participants as the sources of kinematic and kinetic error are hard to identify when both material deformation, and user influence directly impact mechanics.

While it is evident that unmodelled non-linear mechanics are likely to be the cause of inaccuracy in interaction mechanics, only some of the components that would cause non-linearities have been addressed in modelling and control in an attempt to reduce that error. To reduce the magnitude of error associated with non-linear force-position profiles that occur during contact, approaches utilising more complex contact depth models have been used. Non-linear indentation models, such as the Hunt-Crossley contact model, more accurately characterises the force-indentation profile dependent on both depth and indentation velocity have been commonly used in robotics to explain more complex contact, but do not see common use in exoskeleton mechanics

New models have been developed with the desire to expand and build upon existing spring damper models to address issues of non-linearity and initial conditions, but are few and far between. Additionally these models have not seen wide implementation, likely due to the difficulty in characterising coefficients and mechanics when compared to the linear spring-damper models [89, 174].

The purpose of this study is to investigate existing models and to develop new models that address some of the sources of nonlinearity without sacrificing the simplicity of the traditional linear spring-damper model. To investigate existing models, a testbench setup was developed with the intent of removing user interaction through the implementation of a mannequin or "shadow" limb made of a 3D printed PLA and modelled after a volunteer. This testbench also removes ground interaction and translational forces by securing the exoskeleton torso above the ground. The perfectly coupled and compliant linear spring-damper based approaches were compared against data collected on this testbench to evaluate their ability in approximating interaction forces at shank surface-strapping locations.

Utilising the results from the controlled evaluation of a mannequin testbench, an adjusted linear spring-damper model was developed that seeked to address the issues of non-linearity and initial conditions present in the baseline model, without sacrificing its complexity. This new model addresses these issues through the introduction of pre-compression, and allowing for springs to remain in contact past the centerline, a major limiting factor on state space switching. By introducing initial pressures, and one additional state, the system introduces an additional non-linear combination for coefficient solving and initial conditions to be implemented into a relatively simple piecewise model.

The precompression model is evaluated against the recorded testbench data for pseudo-static implementation to assess its value and accuracy compared to traditional rigid and compliant models. This novel model only calculates the required interaction forces to balance gravitational components, and is set to a precompression value known from a recorded sensor to evaluate the performance relative to another model. The method of determining precompression, and coefficients during non-linear states are discussed, however for the context of evaluating the accuracy of the preliminary model spring, damper and precompression values were taken with the aid of sensors and the previously determined spring damper values.

It is hypothesized that the preliminary pseudo-static precompression model will perform adequately, or better, when estimating the interface interaction forces required to drive the mannequin through a predetermined kinematic trajectory.

The precompression model was formulated with the intent of advancing and improving the accuracy of simple one dimensional spring-damper models without vastly over-complicating the characterisation and implementation process. While others, such as the Hunt-Crossley spring-damper model, can accommodate non-linearities, the introduction of pre-compression and initial states vastly changes the problem from individually loaded springs to those in series. Additionally, the model presented in this study may implement those more complex one-dimensional springs if more complexity is required.

In this study we discuss the performance of the new precompression model, its benefits and drawbacks, compared to the traditionally rigid and compliant coupling models with the purpose of improving accuracy and force estimation for exoskeleton coupling interface mechanics. The proposed precompression model investigates the addition of initial conditions and non-linearities through additional states that build upon non-complex spring-damper models to represent complex contact mechanics.

7.2 Methodology

The H3 Exoskeleton (Technaid, Spain) was selected for this study (3 actuated degrees of freedom, DoF, per leg). The H3 is accompanied by a proprietary controller app that provides the control strategy utilised in this experiment. The default control strategies are fully assistive with different selectable speeds for whole gait cycles. Individual steps may also be commanded in place of a continuous gait cycle, where one of the bilateral legs is cycled through a complete step at the prior selected speed. For the purpose of this study, individual steps were commanded at the minimum required speed for evaluation, for simplicity in segmenting and motor noise reduction. With single step commands, residual step dynamics were allowed to dissipate before the next step was taken so the initial steady state at each step was maintained when fitting is applied.

A 3D printed "mannequin" or "phantom" leg, based on CAD modelling of a human leg, was placed within the exoskeleton to prevent tissue compliance and user input torques from impacting data collection. The leg itself is modelled after a 6 feet, 0 inch, 190 pound male near the 95th percent male range. The leg was fabricated with two pin joints: one rotational degree of freedom each in the sagittal plane, simplified to only rotation to match common multibody simulation models for exoskeleton users. The hip joint center is a free end, designed to be easily centered with the hip joint actuator for data collection, and limiting the system to a simple inverted 3-link pendulum. The inertial properties of the system were known from validation and CAD software comparison, and implemented into the simulated models.

The test-bench (Figure 7.1-left) was developed to isolate the exoskeleton single-leg dynamics to those of the supplied actuator torques, removing ground interaction forces and dragging by suspending the exoskeleton by the hips. During data collection, a significant period both prior, and post the singular commanded step cycle was allowed for steady state conditions to be reached and so that cropping could be performed.

For kinetic-kinematic modelling of exoskeleton coupling interaction either position or force information is required to perform dynamic analysis. The purpose of this study is investigating the accuracy of the position-force relationship of common coupling models and developing new ones. To evaluate their effectiveness, both force and position values were collected for use in simulation models. Additionally initial testing conditions were maintained between cycles by resetting both strap tightness and initial position configurations prior to starting an additional cycle to improve the consistency of position and force data.

Tekscan MDL-Medical-Sensor-9811E's were used to map normal interaction forces from the coupling between exoskeleton and strapping interface. 96 force sensitive resistors The front shank surface was selected as the primary collection point, at both of the strap connection locations to evaluate force distribution. This location was selected primarily as a rigid supporting surface, and the Tekscan sensor's ability to capture the majority of the supporting area. Other contact interfaces with higher contact area, such as the thigh, have larger contact areas that are harder to guarantee all pressure data that affects the sagittal plane can be collected. Frictional and torque components cannot be The collected force data was processed and categorized by location for later comparison to simulated data.

Vicon markers were distributed across both mannequin and exoskeleton segments. Kinematic data was collected of the mannequin and exoskeleton, global position and angle were compared to calculate the kinematic offset at each coupling interface. Post processing was performed utilising Vicon proprietary software, and both global position and angular values for segments were output and used for simulation fitting. While the rigidly coupled model only utilises the exoskeleton kinematic data, the linear spring-damper and precompression models require both exoskeleton and mannequin kinematic data.

It has been noted in prior studies that securing forces influence the value of spring-damper coefficients and as such strap tension was kept consistent between tests [**?**]. The strap tension at the shank was controlled using a Newton Scale (initial tension = 4kg), and re-tightened every collection cycle to the same tension value. When fitting for these coefficients in single step simulation

After the mannequin-exo collection was performed, a simulated virtual model was constructed in MapleSim for implementing dynamics. For the precompression model, to evaluate its performance a pseudostatic model was additionally implemented as a preliminary means of evaluating force accuracy without dynamic components. These simulated models are influenced and built upon the most standard form of multibody simulation, with simple pin joints and limited to the sagittal plane, which is common for the

A 3 DoF planar model of the mannequin leg was constructed with no passive joint torques acting on the system assuming that the rotary joints rotated smoothly. Using an existing kinematic model of the H3 Exoskeleton leg [171], a single strapping force for each segment was calculated using both a constrained (Lagrange multiplier) and linear spring-damper approach. The singular interaction force was split into two strap forces (proximal and distal straps) using a least-squares pseudoinverse approach that considered strap position relative to segment center of mass. All dynamics were calculated using Euler-Lagrange formulations, with a constraint equation defining the linear spring damper model as a combination of a spring and damping component, fit utilising a least-squares interior Newton-Raphson approach. The constraint equation representing the linear spring-damper coupling interaction between mannequin and user is equation 7.1.

$$F_{int} = k(x_{exo} - x_{mannequin}) + b(\dot{x}_{exo} - \dot{x}_{mannequin}) \tag{7.1}$$

Utilising the results of the simulation, a novel formulation of the linear spring-damper network was developed. Typically used spring-damper interactions do not consider initial securing force conditions. Spring-damper contact is determined by distance from a pre-set centerline, indicating that only one equivalent spring-damper is in contact at a time. Intuitively and by observation of preliminary results, there is an initial precompression stage that takes place activating both of the springs. This new model is a simple spring-damper coupling equation, motivated by initial coupling forces and their influence on dynamics. This consideration for contact interactions changes the location of contact switching, and allows for both interfaces to maintain contact during motion until an additional set of constraints are met that allows for separation.

Two additional sets of governing equations are added to the linear spring-damper model, and the point of separation. Equation 7.2 represents the equivalent resistance during motion before separation. Equation 7.3 is the point at which separation occurs and is direction dependent. Of note, is before the point of separation these equations are equivalent to two springs in series. The point of separation is dependent on the direction of motion, and is based on the relationship between the indenting spring constant and opposite springs indentation depth. When the indentation depth of the opposite spring is reached contact separation occurs.

$$\frac{\sum F_{bodyforces} - k_1 * x_{static1} + k_2 * x_{static2}}{k_1 + k_2} = \Delta x \tag{7.2}$$

$$\sum F_{bodyforces} = F_{k2} + k1 * x_{static2} \tag{7.3}$$

These state equations result in different trends under response of force. Prior to separation the equivalent spring is much stiffer than that of the single contact spring. This addition may allow for some component of non-linearity to be introduced when switching from two spring to single spring reactions when moving in the same direction. An implementation of this pre-compression model would require information about initial forces along with the typical kinematic data to estimate. For the purpose of this study these initial forces were treated as known, and the spring-damper coefficients were approximated with the aid of those determined in the linear spring-damper model creation.

Rigid, compliant and pre-compressive models were then utilised to estimate the force contact requirements to drive the system during the recorded kinematics. A singular step is simulated and compared against one cycle of recorded Tekscan data.

7.3 Results

Figure 7.1-right details predicted and recorded interaction forces during testing across the process of a single step. Peak test-bench forces occurred at the front of the mannequin shank during the swing-back phase. Maximal kinematic offset occurred near 3 and 3.5 seconds, where the angular offset of the exoskeleton "leads" the mannequin limb before reaching the apex of the backswing. The change in magnitude of the constrained and compliant models greatly exceeded that of the recorded test. The maximum pressure values never exceed that of those measured during testing, likely as a result of initial strapping tension not being considered in the constraint equations. Relative to experimental and compliant-simulated values, the rigid model predicted an earlier occurrence of peak forces. The simulated compliant reaction was fit to the given kinematic data resulting in stiff spring and damper coefficients that accurately track the kinematic position of the exoskeleton, but causes the predicted force to oscillate.

Figure 1 (top-center), illustrates the difference between the simple linear spring-damper and newly proposed precompression model as a simplified graphic. This new model allows for initial coupling to influence mechanics and for both of the effective linear spring-dampers to remain in contact before meeting the separation point conditions. Figure 7.2 (bottom-right) is a similar graph to that of 7.1, instead highlighting the precompression model compared to the linear spring-damper model. When in pseudo-static contact, the precompression model much more accurately matches the magnitude of the recorded Tekscan values. This formulation still results in what appears to be a phase shift that is present in both the precompression and linear spring-damper models. With the inclusion of an initial starting indentation, the mean pressure more closely matches the recorded pressure.

These values represent only the forces required to balance the discrete time steps gravitational component at the angular offset, utilising approximated spring values found based off of these pseudo static conditions.

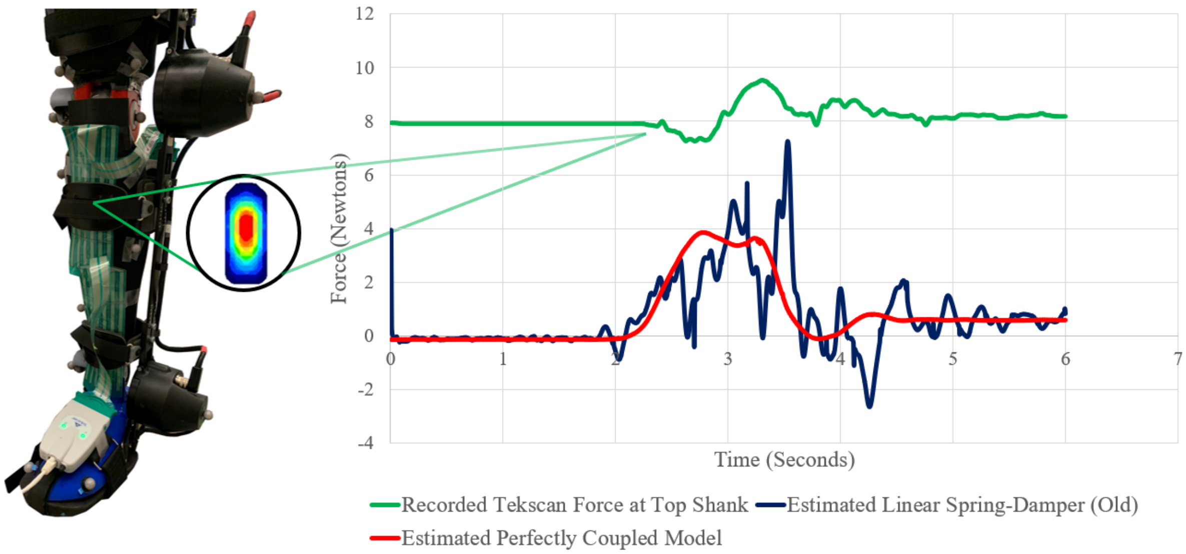

Figure 7.1: Experimental testbench setup with H3-Exo, mannequin leg and Tekscan sensors placed along the shank elevated with floating base support (LEFT) Experimental recorded forces of the Tekscan sensor (Green) from the top shank sensor vs the predicted perfectly coupled rigid constraint (Red), and compliant spring-damper forces (Blue)

7.4 Discussion

Accurate modelling and prediction of surface interaction forces are crucial in risk assessment and design of rehabilitation devices; however, validation can be challenging. The test-bench and mannequin leg utilised in this study help to isolate the contact interactions between exo and "user" without input torques, and complex factors like ground contact reactions and out of plane forces. With a simplified testbench, extensive sensorization was performed to evaluate two commonly utilised interaction models that describe the interaction forces and torques the user experiences. The purpose of this evaluation is to observe their efficacy in not only predicting position, crucial in musculoskeletal injuries, but also forces experienced at the interface itself.

Tekscan MDL sensors and VICON camera systems were utilised to track forces and kinematic position respectively. Utilising position and force data, a simulated 2 dimensional model that represents the exoskeleton-mannequin coupling interaction was created utilising the VICON data and compared against the recorded Tekscan forces. As the most common methods of evaluating interaction forces, evaluating the accuracy of these models, and their drawbacks, is critical in assessing exoskeleton safety and performance. With the results of the preliminary simulation and evaluation, a new framework addresses the drawbacks of traditional coupling models without sacrificing the simplicity of their formulation and implementation.

The perfectly coupled reactions exhibited a phase offset in force interaction caused by the lack of elastic components. As dynamics are tied directly to the leading device in this model, the phase shift in force is easily explained as a direct result of the exoskeleton dynamics occurring ahead of the mannequin limb as it leads it from elastic connections. While rigidly coupled models are commonly used in preliminary evaluation it is evident that they have a phase shift, magnitude change and total magnitude drawbacks in force estimation.

The compliant spring-damper coupling model exhibited much stiffer and oscillatory force interactions likely as a result of the spring-damper least squares approximation methods. Force estimation relies on the spring-damper coefficients estimated purely from kinematics, and as a result of tight fitting is stiff when estimating force. While a phase shift of force interaction is not present, there is a rapid increase in expected/simulated force at points where large separation occurs. This trend in large magnitude increase is not reflected in the measured forces, which when simulated appears to show the system as exerting large changes in force when little change is actually occurring.

In both of these commonly used models, the presence of a static constant securing force is not present. The lack of this estimation, causes a vast underestimation of the maximal forces experienced by the user regardless of interface magnitude change errors. After settling (2-4 seconds) the oscillatory patterns of the compliant model resembled the recorded forces patterns. Experimental steady state forces of the shanks bottom strap differed from the inverse dynamics-derived constrained and compliant forces at the end of the test/simulation likely as a result of strap shifting. While this nonlinearity is likely a result of unmodeled contact interactions the difference is not easily modelable and is not considered in any of the traditionally implemented contact models or in the precompression model. Nonlinearities as a result of shifting caused by friction or nonlinear sliding is difficult to model in one-dimensional systems, and as such remains unmodeled.

The proposed model treats the system as starting initially coupled as a function of the combined spring coefficients. Separation occurs when the compressed surface reaches a displacement equal to the initial compressed length of the opposite surface, allowing for 4 total discrete phases (as opposed to the usual 2). When separation occurs, the system switches and is equivalent to the "Old" linear spring-damper. Damping coefficient influences remain the same regardless as they cannot "push" on the contact surface as it pulls away. Characterising this equation is slightly more complex, as direction is no longer the sole indicator for calculating coefficients. The simulated pseudo-static proposed model followed much closer to the recorded spring-values but could not account for drift likely caused by frictional components.

7.5 Conclusion

The models used often in the characterisation of exo-user interactions lack the ability to describe non-linearities caused by initial coupling forces. The newly proposed model maintains the simplicity of linear spring-dampers, with consideration of multi-contact and single-contact stages to more accurately model force.

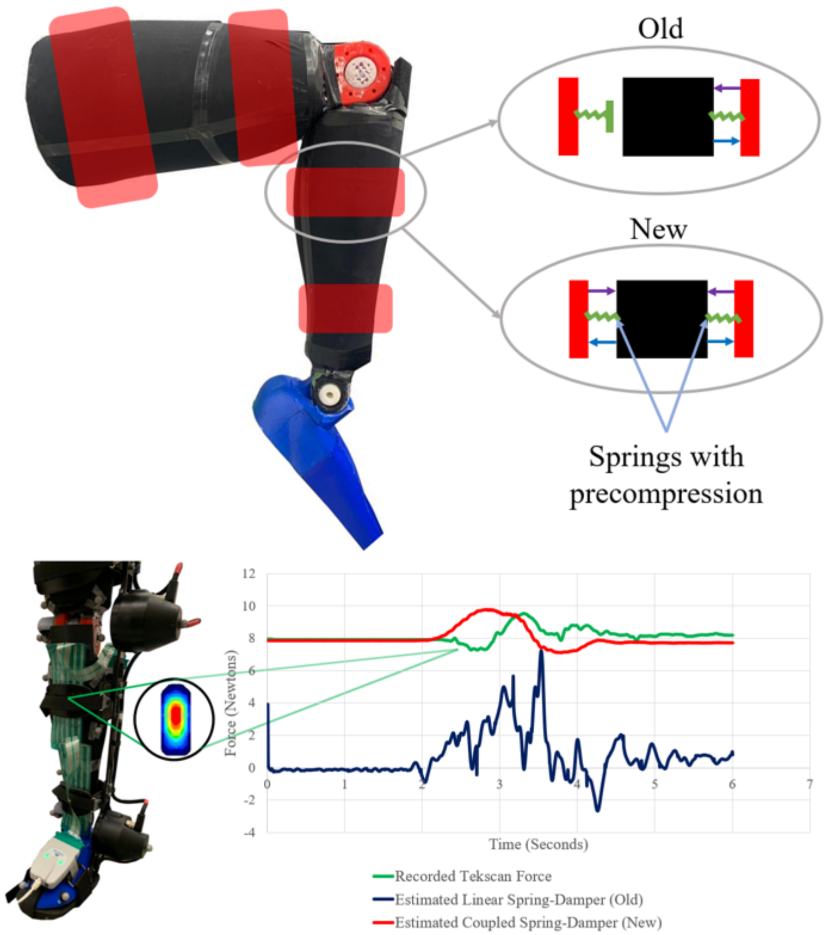

Figure 7.2: Graphic detailing the Proposed Coupled Spring-Damper model with loaded pre-compression against common models (Top). Testbench sensor setup (bottom-left). Recorded Tekscan force (Green) compared against highly stiff linear spring damper (Blue), and the proposed Coupled Spring-Damper (Red), (bottom-right).

Chapter 8

Developing a New Algorithm and Model for the Physical Human-Exoskeleton Interface

This chapter is aimed at developing new models that more accurately represent the coupling at the exoskeleton interface. The paper presented here showcases an elastic foundation model to predict pressures and indentation before fabrication towards developing new models that accurately represent contact, integrated into a "securing in" algorithm that takes into account how donning the exoskeleton affects the initial conditions and pressures.

8.1 Preamble

The prior chapter highlights preliminary work on an alternative model built upon simple linear spring-damper models to improve force based estimation. The benefits of the pre-compression model is the addition of states that represent different levels of stiffness based on contact. The pre-compression model is still limited to two distinct linear springs acting together to provide resistive forces. As a result, the pre-compression model is still limited by the challenges faced by the simple linear spring-damper modelling approach. While initial pressures remove a component of error, primarily offset and series spring dynamics, frictional components and the non-linear contact introduction as depth indentation occurs remain to be addressed .

There are a few options for approaching the problem of missing information when pertaining to the contact mechanics approximation problem. The approximations made for the simplicity of modelling and control are one dimensional approximations, requiring no information of the surrounding geometry. By adding to the considerations of the governing equations and characterising equations or matrices (i.e., stiffness matrices), and how it couples with in- and out-of-plane components will increase both accuracy and complexity. To find more accurate solutions requires adding in more complexity, and may impact their ability to be implemented in simulation and real-time controllers.

Approaches that utilise first principles from contact mechanics or finite element methods are multi-dimensional approaches. The goal is to find an intermediary step that may be applicable in finding a higher dimensional approach, balanced by solving speed requirements for implementation. Figure 8.1 is a flowchart of where the solution presented within this chapter fits in the current framework of contact modelling for exoskeletons, with the red star representing the proposed model. Solutions of the perfectly coupled and elastic spring-damper models were addressed and solved in prior chapters. The potential accuracy of a single element non-linear spring-damper model with coefficients dependent on different conditions, primarily associated with states. Models that seek to improve the accuracy of linear models, such as Hunt-Crossley models are more accurate, but do not add too much in terms of detail for complex contact. Instead, these models allow for non-linear influence of spring depth [173] to effect the damping components. Hunt-Crossley models are commonly implemented in more complex contact cases due to their ability to replicate some non-linearities, but remain limited by their one-dimensional result which prevents it from being characterised without a necessary physical component to estimate from. Hunt Crossley models cannot approximate contact areas, complex initial conditions, pressures or highly non-linear surfaces.

Often times, extra work must be taken to approximate these models and calculate their coefficients. For placement on the graphic below in Figure 8.1 illustrating levels of complexity, single element non-linear models fit behind the model proposed in this chapter.

To act as the penultimate step before approaching the problem from a fundamentals material approach, first principles or finite element method, I propose an elastic foundation inspired model. Complex approaches require significant setup time, computational effort and understanding of there governing equations and relationships to setup. Establishing these for every different system is difficult, therefore a simpler process for early but accurate evaluation of pressure, surfaces and interactions is required.

The model is a component step, or part, of a series of assumptions on the exoskeleton donning process to provide a more accurate initial condition setup to characterise pressure layout and spring-coefficients during compression prior to use. This algorithm is intended

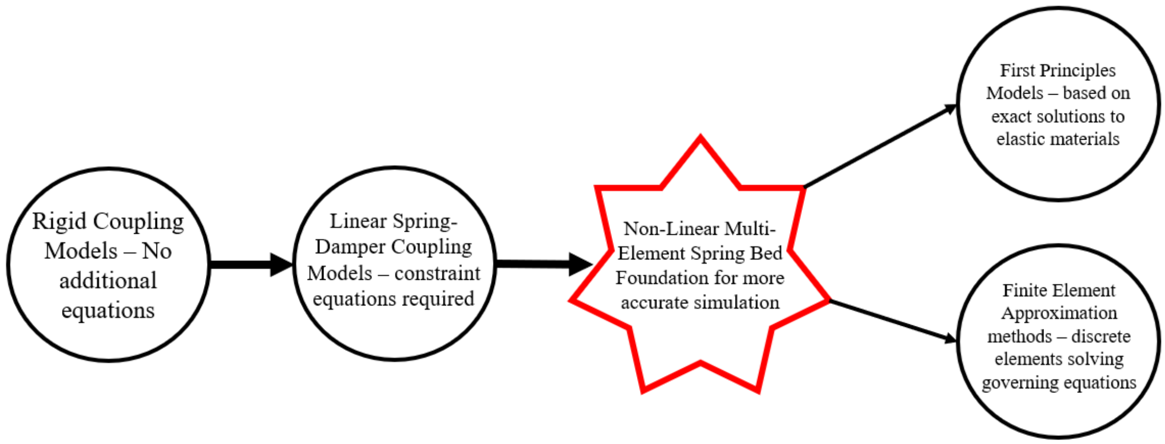

Figure 8.1: Flowchart illustrating the current existing coupling models and where the model presented in this thesis fits. Arrows indicate an increase in complexity and computation time. The proposed non-linear array estimation model fits between linear spring-dampers and first principles approaches.

to be used as a means of characterising the initial conditions present during securing in to find equivalent stiffness values for traditional linear spring-damper systems. Instead of typical stiffness values remaining constant, a depth to spring-damper coefficient curve can be fit that most accurately describes the system. Additionally, this model contributes in-plane forces to be considered representing frictional forces from all contact surfaces.

If time to solve for control is not of concern, this model can be used as a more complex approximation for interaction forces and offset during dynamic use. Otherwise, it is a useful tool for estimating initial conditions and implementing those values into traditional models.

The paper presented in this section was written with the intent to submit to a special review paper on the Physical Human-Robot Interaction mechanics for Exoskeletons. It is currently in the editing and review process.

8.2 A Novel Method for Modelling the Physical Human-Exoskeleton Interface

8.2.1 Introduction

Lower-limb rehabilitation exoskeletons are powered wearable devices that augment or replace the gait of their users. Exoskeletons are used by an array of individuals with mobility impairments, from those who require minimal assistance to those who require complete ambulation control. The primary focus of rigid lower limb exoskeleton design is to achieve specific gait output objectives such as designing trajectories to provide full or partial assistance for ambulation. These objectives are met with a combination of actuator, electrical and controller designs and remain the major focus of most reported exoskeleton research [36, 37, 40, 47–50, 52, 53, 55, 58, 89].

The coupling interface is an under-investigated component of exoskeletons that influences the success of the device. Whether a device provides partial or full assistance, the forces generated by the exoskeleton must be transmitted through contact points and interfaces collectively termed "coupling surfaces".

Research in the field of coupling interfaces for exoskeletons is generally underreported and underrepresented in comparison to innovations in mechanical, electrical, and controller design [89]. However, understanding the interactions that occur from coupling interfaces is no less crucial than exploring new control strategies or actuator designs. With the intent of these devices as rehabilitation tools and gait augmentation devices for mobility-impaired populations vulnerable to injury, minimizing safety risks of the individual is essential, and improving performance critical.

There are two primary considerations in exoskeleton coupling interface design: 1) the influence on the kinematic and dynamic models that directly affect the controller and mechanical design, and 2) the safety of the individual from pressure and musculoskeletal-related injuries directly caused by the coupling surface.

The trajectory of the exoskeleton is influenced by a number of complex factors, including the interaction between the device and the user. Approximations are often made of this interaction to streamline this process, but they often introduce undesirable modelling errors. Many design problems in the control of lower limb exoskeletons may be simplified by having a better understanding of the interfaces between the user and the device and how they interact.

The coupling surfaces also have a large impact on the safety of the device. As the dynamic load of the exoskeleton increases, so too does the risk to the individual from mechanical elastic deformation. Exposure to high magnitude normal and shear forces imparted on the user may cause skin injuries such as bruising, ulceration, skin tears, or tissue necrosis over time [76, 77, 89]. Furthermore, misalignment due to poor coupling may induce musculoskeletal injuries during control from poorly aligned forces [81, 89, 98, 119].

Traditionally implemented models, rigid or compliant, are highly linearised or approximated interactions to abstract singular elements, or ignored entirely. The simplifications made in these models limit their ability to accurately predict interaction dynamics and kinematics, and cannot account for user safety past musculoskeletal misalignment as a result of being limited to force only calculations [89]. These models are also currently incapable of transferring information from simulation (optimized reactions and compliance), into actual interfaces, requiring the fabrication of costly and time consuming interfaces which then need to be tested to approximate coefficients and future simulations. The inability to accurately predict and simulate a coupling interface's interactions with a user limits their designs, and requires costly and potentially risky experimentation to characterise how these surfaces actually interact with the user. This is potentially a driving factor as to why research into coupling interfaces is relatively unexplored, because the development and evaluation of them is so costly.

Developing better models for dynamics, kinematics, and material deformations without sacrificing computational speed is needed to improve control and user safety, and to enable the design of new and improved interfaces [89]. Models which improve the accuracy of kinematic configurations of the coupled exoskeleton-user interface, as well as the safety of the individual, are still relatively uncommon. Most models that characterise interactions focus on simplicity for control, modelling and fitting, and not on how we can transfer ideal simulation results to actual design [89, 94, 98, 174, 176–178]. When developing new models, the focus should be placed on how we may take optimized or desired results in simulation and convert those into manufacturable and realisable interfaces.

This chapter proposes an alternative method to model the interactions between the user and the device through the coupling surface, inspired by contact mechanics and elastic foundations. This novel algorithm calculates the initial configurations that occur during the "securing in" or donning phase of exoskeleton use [89, 174]. The process step can be seen in Figure 8.2, where much. Implementing more accurate models for simulation or control requires knowledge of equilibrium conditions and forces that occur on the surface before dynamic loads are introduced, as indicated by studies investigating the effects of initial conditions on effective spring and damper coefficients [69, 93].

By modelling the interface as a 3D object lined with an elastic foundation of springs, complex contacts, initial securing conditions and non-linear interactions can be accounted for. The algorithm works by first locating initial contacts, applying securing forces and determining the equivalent spring-damper values along the path of indentation. These spring-damper values are then reported, along with effective pressure distributions and kinematic configurations which is output as the initial static coupling conditions. The value of this algorithm comes from its ability to be used in conjunction with already existing models. While not yet implemented in this algorithm: by providing optimized coefficients or parameterized equations for effective spring and damper values, this algorithm may be used to aid in fitting and optimizing for those values through tuning surface parameters.

The main focus of this design document, however, is on the algorithm and framework for analysis, which can be adjusted to achieve the desired level of complexity and granularity for output in various research settings. Although there are alternative combinations of surface interfacing available, the building blocks presented in this study should facilitate ease of transferability. Inputting simulated interface surface curvatures and user parameters will output effective spring-damper values at initial contact (from neutral axis to depth of contact), and initial joint configurations, which can be used to simulate elastic interaction prior to fabrication. By introducing this preliminary model, we aim to advance the modeling and design process of coupling interfaces for both control and user safety, and to promote innovation in interface design.

8.2.2 Background and State of the Art

Background

The history, benefits, and drawbacks of existing exoskeleton models will be presented in this section, followed by the motivation for the novel model presented in the "Methodology" section of this paper. This history will cover both the most commonly used and most recent models. Few models exist that describe the coupling interactions between the user and the device. Prior models are either too simple and introduce large amounts of error, or too complex and computationally expensive for use in controllers.

The most basic model that describes the coupling between the user and the exoskeleton is the "perfectly coupled" assumption. This assumption treats the coupling between the user and the device as rigid and non-deformable. The user's limbs are assumed to move synchronously with the device, requiring no alterations to the complex rigid body dynamics that also describe the serial robot. The simplicity of this model lies in the fact that only a single set of dynamic equations describe the motion of the limbs and exoskeleton since

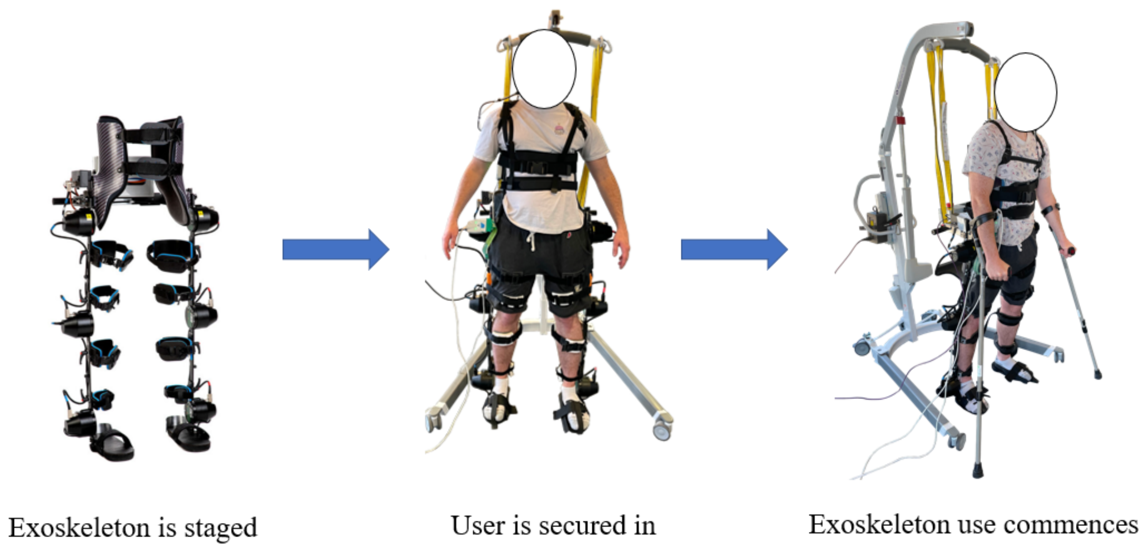

Figure 8.2: Exoskeleton use flowchart. The securing in phase is controlled by clinicians and aides, and often ignored in modelling.

they are assumed to move in union. These dynamic equations are common for rigid-body dynamic systems with kinematic chains that are often expressed as Euler-Lagrange formulations with known or estimated inertia property matrices [47, 58, 70, 89, 129, 135].

Perfectly coupled assumptions are made with the intent of simplifying dynamic equations and complex controllers at the cost of accuracy. Without proper estimations of the location of the limb, the controller has to overcompensate for misalignment caused by elastic contact. The primary benefit of simplistic models are their ease in implementation into controllers as they allow for faster calculation speeds, at the drawback of accuracy.

Commonly used in many controller and mechanical design studies, the next model describes the contact surface as a single point that is both viscous and linearly elastic, forming a linear spring-damper system [56, 58, 60, 70, 89, 129, 132, 133, 179]. This model provides an approximation of the elastic deformation and relative movement of the user's limb, making it possible to estimate the forces present at the surface by abstracting the problem to a virtual spring. The model can be expanded to include two spring-damper contacts that offer non-linearity with alternating "on-off" states from a center-line, which are activated as elastogap springs. However, the mechanics of this coupling model are more complex than the perfectly coupled assumption and therefore require more computational resources when implemented into a controller. To solve the two series of equations, free body diagrams and directional switching during coefficient optimization must be implemented to find these coefficients.

It is important to note this method is still an abstraction of the actual contact mechanics, so spring and damper coefficients must be approximated using techniques such as curve fitting and least squares. Least squares fitting techniques rely on matching a desired fit equation to a sample set of data and solving for coefficients. The approximated coefficients for dynamic interactions become better with more sample points. Drawbacks related to overfitting occur when data is collected only for a single individual. Coefficients found from a single individual's data is personalized to the individual and not easily generalizable reducing the effectiveness of the characterizing equations between users. Additionally initial conditions that define the securing and strapping calculations differ from test to test, and person to person, making the process of generalizing these values increasingly difficult for a wide range of potential users [69, 70, 93, 129]. The process of characterising these model coefficients also requires a physical exoskeleton to test. Determining the coefficients before fabricating or designing the device for specific or intended values is currently not possible using online values, limiting the potential designs and functions of the coupling interface [116].

State of the Art

The rigid body and linear spring-damper models are the most commonly used approaches to exoskeleton coupling in the modelling of exoskeleton interactions. Recent state-of-the-art advancements have highlighted drawbacks of these models and recommend improvements the modelling of exoskeleton coupling surfaces. The improvement or characterization of interface pressures is divided in two overlapping categories based on the focus of the study: control or accurate model prediction. For the purpose of this study, we evaluated studies and models that focused on the framework of modelling the interface without the intent of control, as many of the advancements made for existing controllers primarily focus on the algorithm, and not the model [89].

Recent advancements in modelling contact interactions focus on incorporating the effects of initial conditions, calibration, and joint misalignment [69,70,81,89,93,98,129]. For elastic spring-damper models, initial conditions such as forces associated with the coupling surfaces are often not considered during modelling. During the "securing in" phase of the exoskeleton strapping problem, initial forces are introduced into the system when the primary user is strapped in by a clinician. These forces have traditionally been disregarded due to the complexity of characterizing the component. Evaluating the influence of initial cuff pressures on the finalized parameters reveals that the characterized spring stiffness does change based on the initial pressure [69, 71, 93]. These studies indicate that while damping components are not affected, initial compressions alter the calculated stiffness of the spring-damper system and will change from experiment to experiment depending on the user and initial pressure. As a result, better predictive modelling of resultant pressures or consistent strapping pressures are needed.

Joint misalignments have more recently being investigated as a critical component of interaction mechanics, with a particular focus on controlling them through mechanical-kinematic design and coupling models. The investigation of joint misalignment primarily relies on kinematic chain modelling and holonomic constraints. However, components of coupling models in kinematic offset prediction are also incorporated. Studies quantifying kinematic offsets often utilize force sensors at the interface to directly measure interactions at the interface and characterize joint forces resulting from misalignment [89, 103–106, 118, 124, 125, 137, 161, 180–182]. Joint offset simulations relying on traditional spring-damper models to calculate interaction forces and predicted offsets are often inaccurate. Although they can predict position well, due to the coefficients being approximated using position, the calculated force and torque relationships are stiff and difficult to evaluate. These difficulties occur especially around inflection points, and highly non-linear changes in position and velocity resulting in inaccuracies in tracking force and position from highly linear models [70, 89, 129]. Linear spring-dampers are able to semi-accurately predict forces, when applied under conditions similar to those they were tested on. However, non-linearities present challenges in tracking position and angles [70, 129].

Lastly, new models are being developed and researched that seek to further the framework and approach for contact modelling, innovating and improving on existing ones. Advances in the characterization of linear spring-dampers offer alternatives to current solutions by implementing more rigorous optimization techniques to avoid the stiffness present in other models [89, 174, 176–178]. Despite the improvements made on these alternative techniques, stiffness is still present around inflection points causing innacuracies in tracking. Newly developed models expand on the theory of linear spring-dampers by extending singular springs into arrays or "beds" of springs [174, 177]. This approach sees a more accurate network of pressures across discrete cells, but treats each individual cell as a distinct pressure region. This method of approach is more accurate at the trade-off of computation time as a series of constraint equations must be solved simultaneously across multiple surfaces. As shown in studies by Yousaf, and Varghese [174, 177], these models can be used for design and for approximating interface contact mechanics prior to fabrication. These models are similar in formulation, and allow for simple governing material equations to approximate the indentation depth. For complex modelling of robotic interactions, approaches implementing Winkler-type foundations and integrated overlapping curves serve as much more complex solutions that more accurately model interaction in both static and dynamic interactions, without the complexity of finite element analysis [200, 206]. These previous two models are the inspiration, and driving framework used for the model presented in this chapter.

In the development and innovation of characterising models, the implementation and use of them in controllers cannot be overlooked. Improvements provided by controllers are primarily seen in the implementation and interpretation of models rather than in the development of new models outright [89]. Often, these innovative models rely on sensors and predictive linear spring-damper models to generate appropriate control signals and trajectory goals synthesised from available information, as opposed to implementing more complex governing equations. The combination of measured and predicted interaction forces is primarily used to inform controller signals through iterative design or adjustment, and not the models themselves [58,89,132,133]. In some instances, controllers and iterative design are used in conjunction to improve interaction forces through a sensor-only approach to inform design or guidelines of use [89,100,103–106,123,125,162,183]. The design process for controllers, and their efficiency, could likely be improved with an improved coupling model, where in the past more computational power, or sensor integration was used.

The focus of this study is to establish a framework for analysis to approximate the initial conditions of an array of springs associated with the ”securing-in” phase of exoskeleton use, inspired by elastic foundations. This includes determining the initial equivalent spring coefficients, contact configurations, and quasi-static reactions associated with this step of the exoskeleton donning process, for approximating initial condition interactions that often go overlooked. As noted prior, the initial conditions associated with strapping intensity (initial tension strength), influence the conditions of the effective linear spring-damper coefficients. Incorporating this knowledge, along with the more recent developments made in elastic foundation coupling models, a new method of generating accurate initial states that can predict an array of, or equivalent, spring-damper(s) has been developed to more accurately predict elastic physical interactions between user and exoskeleton.

We propose a novel surface interaction algorithm that considers surface geometry, material properties, and applied forces based on approximations of contact mechanics principles for a simple approximation of initial strap-in conditions of exoskeleton-user contact. By detailing the process for securing-in and calculating equivalent spring values, even simple linear spring-damper algorithms can be improved with more accurate initial estimates. As such, this document details only the initial algorithm required for implementation, and the performance of the elastic foundation approximation relative to common approximations and known closed-form solutions. This initial model showcases and highlights the methods by which an approximation of an exoskeleton-human contact model could be made, and its relative accuracy to known simulated solutions to provide a baseline of accuracy as an approximation method.

8.2.3 Methodology

This section will detail a step-by-step analysis of the general concepts required to implement the static strapping algorithm, followed by recommendations for dynamic implementation. The goal of this algorithm is to establish the kinematic and static conditions of "strapping in" in a simple manner to implement that does not require extensive knowledge. Each subsection will detail a major step of the algorithm with accompanying logic. An overview of the complete securing algorithm process can be seen in Figure 8.9.

This version of the algorithm was is inspired by the H3 Exoskeleton (Technaid, Spain), seen in Figure 8.3, and is intended to be used with similar exoskeletons. These exoskeletons have securing interfaces that are both hard (aluminum hardbody) and soft (EVA foam covering), with an hook-and-loop (i.e., Velcro) strap that provides a compression force to hold the limb to the surface of the exoskeleton. The algorithm itself was made to be adaptable in order to accommodate other interfaces that differ from the H3 and allow alternative loading conditions and use cases.

Parameter Initialization

To accurately model exoskeleton interactions, the values for certain properties such as surface geometry, securing locations, segment lengths, surface materials, and material properties must be precisely defined. Additional "shifting variables" associated with each strapping surface that represent changing material properties values must also be accounted for. This information can be represented in multidimensional matrices that can be accessed for solving governing equations, relative to a global origin, facilitating transferability.

General surface cross-sectional information, as well as leg cross-sectional information, is stored in 3D point cloud arrays (see Figure 8.4), to allow for simple iterative searching of collision or overlap. When initialising the individual springs and "beams" that represent our simulated support surface, we treat each as being perpendicular and parallel to the underlying support surface, respectively. This assumption represents a soft material glued down over a supporting surface, and affects how indentation is split. A volumetric cell approach would likely result in a more accurate result by averaging the indentation profile

Figure 8.3: The H3 exoskeleton and one of the coupling interfaces. The coupling interfaces are "generic" by design, intended to fit a wide range of individuals and as such are not perfectly circular, nor half of the circumference of a circle to allow for ease of use.

of the cell overlapping with adjacent cells, but would require more time and computational resources for implementation.

For the purpose of the strapping-in algorithm, the ankle is considered the origin and point that rotation occurs about. All limb cross-sections are assumed to be circular and are approximated from limb segment circumferences.

Material properties such as shear-spring values are stored as global variables in a matrix that are accessed based on the surface of interest. In this study, only a single EVA foam covering is used, however, if materials are layered, equivalent spring matrices can

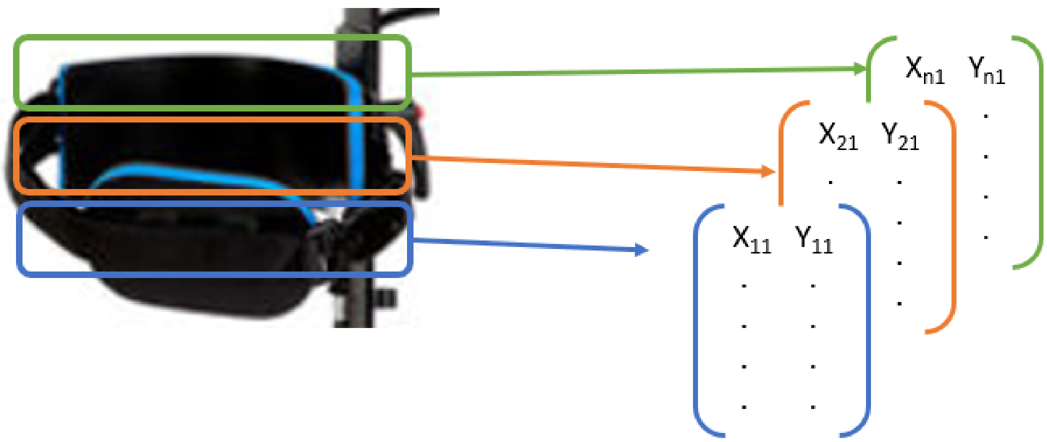

Figure 8.4: Visualization of layer slices for point cloud indentation analysis. Each slice is treated as a discrete independent section for simple analysis.

be calculated similarly to how springs in series are calculated. To save computation time, the limb is treated as a rigid non-deformable structure similar to the hard body of the aluminum surface. Evaluation of a soft limb would be more complex and require shape change estimation at each time step from pressure profiles.

A hierarchical definition of strapping can also be set at this step. For a fully coupled system, the order in which securing occurs will influence the final configuration. In this study, we present the logic for coupling on the same segment. Interfaces further from the segment of interest provide resistive forces and, if possible, allow for rotation to occur. Constraint equations or limit values for segment rotation can also be defined here.

Initial Contact and Localization

The information from the Parameter Initialization step is used as input for this step. The 3D surface and position information is used to initialize a leg in 3D space and find the no-force contact points. An iterative search method finds a set of contact points (an arc or single point) for fastening the limb segment into its initial condition before forces are applied from the soft securing strap.

To simplify our model, we assume that once the origin is set, all modes of translation are either sliding or rotation, but not at the same time. Rotating the limb into position requires altering the cross section within the transverse plane based on rotations made in the frontal and sagittal planes. These rotations would alter the assumed circular cross-sections into ellipses, adding computational complexity and requiring forces to match the new geometry. Instead, we will "rotate" the leg into place while assuming the cross-section remains unchanged. This assumption introduces error into the cross-sectional area estimation, which is already simplified by the initial circular assumption. Any translation required for force balance on the surfaces will have a resultant translation within the segment itself, introducing forces at other locations. If this introduces rotational moments (or the force cannot be adequately balanced), sliding can occur. Each segment is also treated as an individual contact problem, considering the allowable translation and relative rotation that can occur between segments about the knee. This adds complexity and computation time and is difficult to estimate accurately using an unknown contact, double inverted pendulum formulation. Therefore, this aspect will be ignored.

With the protocol for segment adjustment and movement established, determining the initial no-force contact points is possible. No-force contact points are defined as a singular point, or set of points (an arc) that is shared between two surfaces that require no applied force to achieve adjacency [184, 185]. The importance of establishing the no-force contact point is to find where on the supporting surface intersection occurs during indentation. This step represents the "stepping in" or assisted placement component of the donning process that occurs before strap forces are applied. Before fully strapping the exoskeleton user into the device, their limbs should be placed into appropriate positions that are snug within the strapping surfaces' encompassing area. Forces introduced in this phase, such as stochastic pushing and pulling, as well as gravity, can be accounted for but alter the final steady state position. This step in the process allows us to assume a quasi steady-state estimation, with only translation occurring as a result of reaching static force application.

Finding this initial contact arc can be accomplished using a four-point A* search algorithm that searches for an arbitrary origin point that is the prescribed "farthest point" within the open surface area [186]. A strict cost-estimation is assigned to cells that would cause a collision with the surface by using absolute distance to any occupied cell. The algorithm finds the shortest path towards the origin, with a combined weight to the point closest to the origin that maximizes overlapping area between the closed area of the coupling surface and the limb cross section is selected as the initial starting point. The search stops when the "shortest path" left is a high cost. Alternative strategies can also highly value maximized surface area to reduce initial pressures. Figure 8.5 visualizes a potential path trajectory taken by the limb during the search, starting at a point defined by the extreme limits of the supporting surface point cloud.

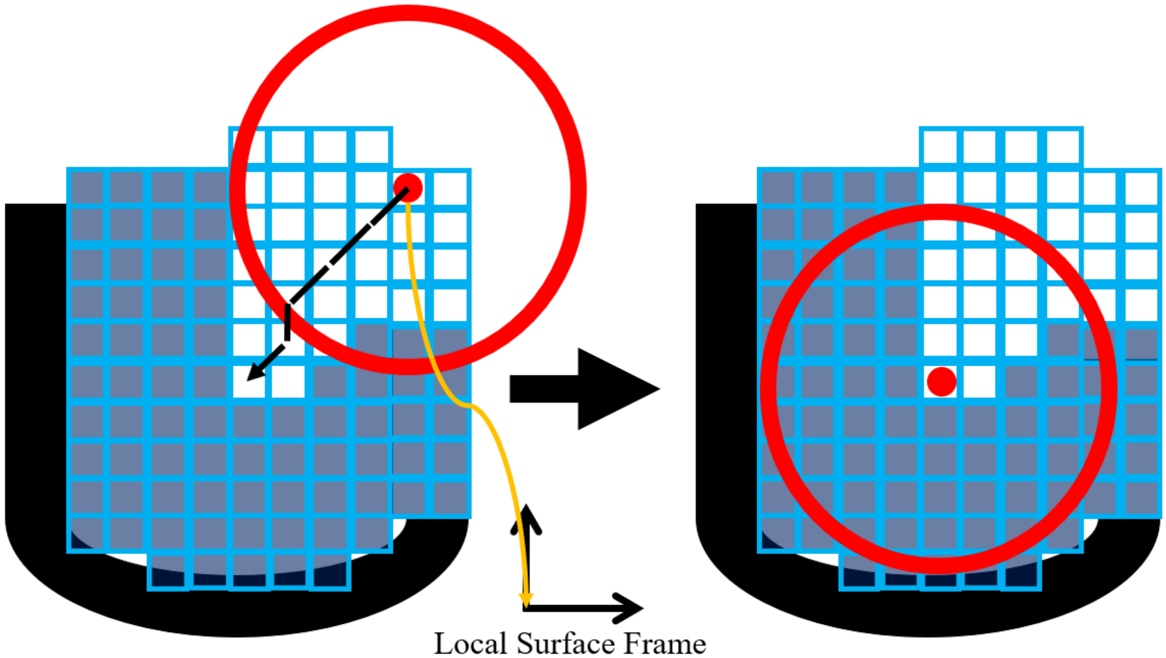

Figure 8.5: A* search algorithm finding the point furthest "within" the coupling interface avoiding points where collision would be met by comparing overlapping cells. Occupancy grids are generated based off an assumed constant, and circular cross section.

Soft Securing Straps

This section uses information from parameter initialization and initial contact locations to apply forces that occur from the rope-like straps to secure the limb segment to the supporting surface. In real-use scenarios, a securing strap attached to a D-ring on one side of the supporting surface is looped over the limb segment, through another D-ring on the opposite side, and pulled taut from an applied tension force, which creates reaction forces that hold the segment in place.

The strap anchoring locations in 3D space and the final location of the no-force contact limb segment are used to generate the initial conditions for the applied strapping forces. Assuming that the strap remains taut, a search is conducted to find the point where a tangential line can be drawn from the anchor to the limb segment. A semi-circular arc is then drawn to represent the strap laying across the limb and is considered the location where force can be applied from tension. The forces applied along this arc represent capstan forces, a force amplification phenomena that occurs from a tensioned rope exerting normal forces and friction on a surface which alters its curvature [175].

After initializing the strap's contact arc, force is applied normally to the contact surface based on a simple capstan force balance, seen in equation 8.1. Normal capstan forces are exerted by the limb segment surface in reaction to the curvature change of the tensioned rope. The change in curvature exerts forces normal to the surface that can be calculated at discrete intervals. These discrete intervals can then be converted into the principal coordinate systems, and forces can be found for balancing in two primary directions.

$$dN = 2Tsin(d\Phi/2) \tag{8.1}$$

Friction is ignored as it contributes a relatively small component to the normal force at that interval. Frictional forces at the surface generate torsion components on the limb that must be resisted by the supporting surface and joints. Each small angle change is summed and mapped to the global or local reference frame as seen in Figure 8.6. In application, implementing sensors or using estimated pulling forces as tension from load cells will inform the applied tension when securing down and preloading the sources with compression.

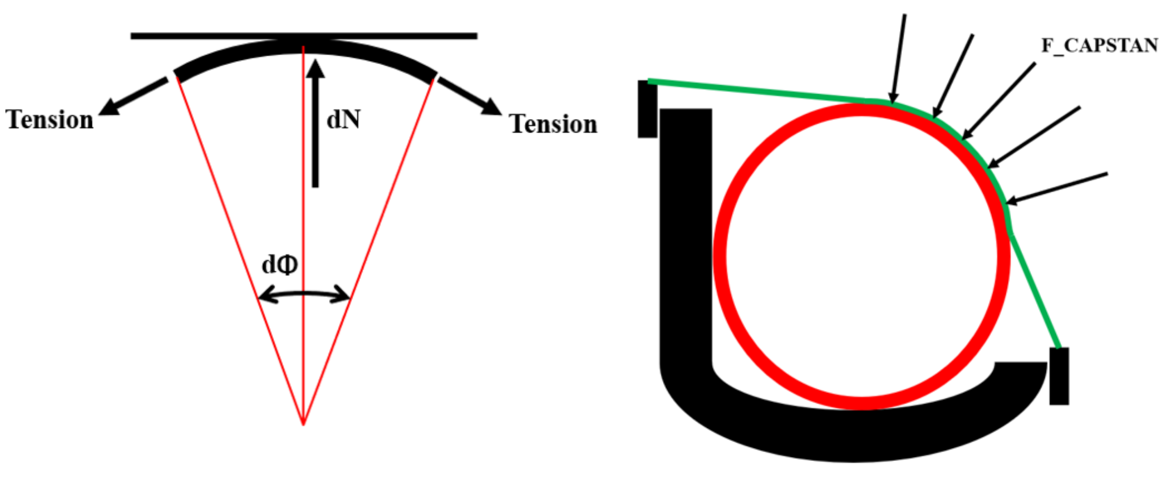

Figure 8.6: Individual capstan element (left) and entire capstan force arc (right). The individual capstan elements are calculated within their own reference frame and converted into the global frame for balancing.

In this algorithm, a few assumptions are made to simplify the calculation for proof of concept. If frictional forces are completely ignored, the total applied force on the supporting surface is overestimated. Additionally, the assumption that these components are purely planar is not accurate. To justify this, however, most existing models assumed that linear springs only acted perpendicular to the supporting attached surface. Implementing out of plane components (shear and friction in the sagittal plane), will increase complexity and accuracy.

Support Surface Reaction and Deformation

This section considers information from all three previous subsections to localize the final position of the limb segment for a given applied contact force. After decomposing the applied capstan forces into the local support surface axes, we can approximate the resultant deformation and pressure distribution that occurs in the coupling surface. The resistive force is a combination of frictional and elastic forces provided by the most compliant components of the supporting surface. The rigid component is assumed to not deform, allowing for a constant and known application of resistive force within the sagittal plane. As a result, the only components assumed to deform are the soft components; for the H3's surfaces this is the EVA foam layer.

Deformation that occurs between two non-conforming surfaces will result in a non-linear contact area increase with a linear increase in penetration depth. The penetration depth is defined as the overlap depth between surfaces from the original, non-deformed cross section. To accommodate this non-linear interaction, an approach that expands and builds upon approximations made within contact theory can be applied. Each surface is divided into a series of cell positions in a 3-dimensional matrix, with the third dimension representing each discrete slice of the surface. The support surface balance accounts for each of these slices in the interaction between limb segment and support surface.

A one-dimensional Winkler foundation is an approximation of a support surface as a bed of linear springs. Each spring represents a discrete area of the supporting surface allowing for an approximate calculation of the force-depth relationship in that area. Hooke's law is used to calculate the spring coefficients. The solution for this relationship is based on Equation 8.2, where g(z) is a selected stress relationship function between vertical and lateral components, and E(z) is the Young's modulus as a function of depth [184, 185, 187, 188, 192, 193, 195–197]. This value is considered constant for this evaluation and many materials, however it can be easily altered as a function of characteristic depth if needed. H is the characteristic depth; for sufficiently large indentations it is the size and thickness of the support surface, however, it can also be treated as the width of the contact arc [185, 187, 195, 197]. In the case of most exoskeleton supporting surfaces however this value should be the thickness of the material. When compressing a small layer of EVA foam, we assume a negligible lateral relationship and use a constant Young's modulus with depth of indentation, as the material is relatively thin.

$$k_s = \frac{1}{\int_0^H \frac{g(z)}{E(z)} dz} \tag{8.2}$$

This relationship does not fully explain the coupled curvature and shear relationship. When a surface is compressed in a local region, a cell "pulls" down on the adjacent cells, which resist this change in curvature, increasing the required pressure at each location. In a Pasternak foundation, this relationship is approximated using a beam layer in pure shear. The force required to shear this beam is a result of the second derivative of the indentation. Moment forces are not taken into account in this support shear beam, or in the Winkler foundation beneath it [185, 187, 199]. The shear layer's shear modulus is normalized by the characteristic depth. Finding exact reasoning for the normalizing factor is difficult, however, the most common approach uses the characteristic depth of indentation [185, 187]. Equation 8.3 shows the pressure relationship that generates the force-depth relationship.

$$\mathbf{q(x)} = k_s \mathbf{w(x)} - G_s \frac{d^2\mathbf{w(x)}}{dx^2} \tag{8.3}$$

A visualization of the Winkler and Pasternak foundations and the resultant curvatures underneath and outside of the indentation surfaces is shown in Figure 8.7. K represents the approximated equivalent spring, while G represents the shear layer. The springs are equivalent between the Pasternak and Winkler, the only difference is the shear layer.

In order to further simplify the problem, these beams only act within their respective planar slice, reducing the partial differential equation to an ordinary differential equation. Height is limited to a function of the slice as opposed to a 3D problem. The supporting surface is simplified into a series of stacked discrete Pasternak foundation problems, each approximating the local contact at that point. The width of these discrete slices can be modified to trade computation time for calculation accuracy. The pressure distribution is solved as a discrete function which results in an overestimation of pressure at the separation point.

Different approaches can be used to calculate the point of separation. For high accuracy, the Euler-Bernoulli beam theory can be employed, which involves solving a fourth-order

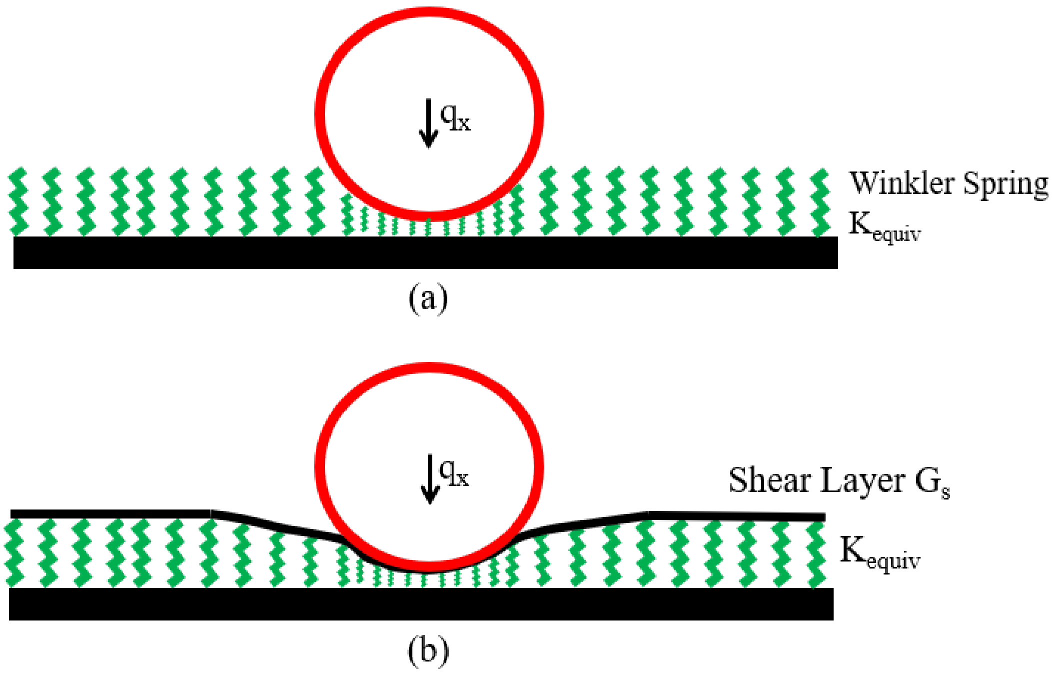

Figure 8.7: (a) Winkler Foundation (b) Pasternak Foundation

ODE numerically. However, this method is challenging to apply to non-linear geometries like coupling surface interfaces and requires boundary and initial conditions. The objective of this function is to find the curvature that minimizes the force offset between the applied force and material resistance, as shown in Equation 8.4. If the computational time is not a concern, using a finite difference approach to solve the governing ODE is recommended for determining the separation point. In this approach, D represents the width of the layer slice, and q(x) is only active under the contact points of the indenter, requiring the iteration of the switching point alongside the w(x) function.

$$\mathbf{q(x)}D = (EI)\frac{d^4w(x)}{dx^4} + k_sD\mathbf{w(x)} - G_cD\frac{d^2\mathbf{w(x)}}{dx^2} \tag{8.4}$$

Another option is to use an estimated curvature to finding the point at which separation; this is much less accurate but is also faster. Using an arcsin-based curvature that relies on the ratio between half the characteristic depth and distance from center of pressure – as seen in equation 8.5 – can solve for an approximate curvature past the point of separation. Arcsin curvature is commonly found as a major defining component outside of the contact area for radially symmetric problems [184, 185]. Finding the point of separation requires iteration along the curve up to the edge of the Winkler contact area, selecting the point with the least steep change in curvature relative to the indenting surface. This approximation is not intended to be accurate, but instead to be relatively fast and simple to implement that mimics the ideal exact solutions for simpler contact conditions [184, 185].

$$\mathbf{w(x)} = arcsin(\frac{a}{r}) \tag{8.5}$$

Frictional forces are treated as coulomb forces and do not cause slippage underneath the contact arc from shear. Frictional components which act outside the "slice" or plane of interest are not considered. For initial strapping conditions, it is assumed that friction does not need to be overcome for the leg section to shift, but it is a contributing factor in the reducing depth of indentation to maintain the quasi-static steady state assumptions. The force is calculated by using the assigned area given to each point cloud. For simplicity, each point cloud location is assigned the same area based on resolution, and the pressures are summed over the indented and active area.

The presented algorithms only showcase flat support surfaces in imagery, however, for complex curvature and geometry, the discrete methods remain the same. To calculate indentation depth requires information about the underlying supporting surface. To calculate the resulting indentation, the overlapping point cloud location searches for the shortest path, perpendicular to the surface through a nearest-neighbour search. The force required for indentation can then be calculated, including forces perpendicular to indentation and frictional resistance forces. Shear forces are applied normally to the original surface curvature. This information is tracked by rotation matrices that step through each discrete cell along the surface. A visualization of this local cell decomposition into its component forces is shown in Figure 8.8.

Surface Coupling Effects

The capstan and support surface deformation is estimated in steps 3 and 4 for a single interface. However, both the thigh and shank segments have multiple support surfaces. These steps ignore the coupled moment arm effects of the other additional surface within the same segment. When movement due to applied strap forces occurs, contact also occurs due to the accompanying surface of the segment. Since all motion is treated as rotation, any movement that occurs within one segment causes motion in the other segment related by a polar coordinate relationship. The new intersecting cross-sectional areas are then balanced by following the same procedure as outlined in step 4.

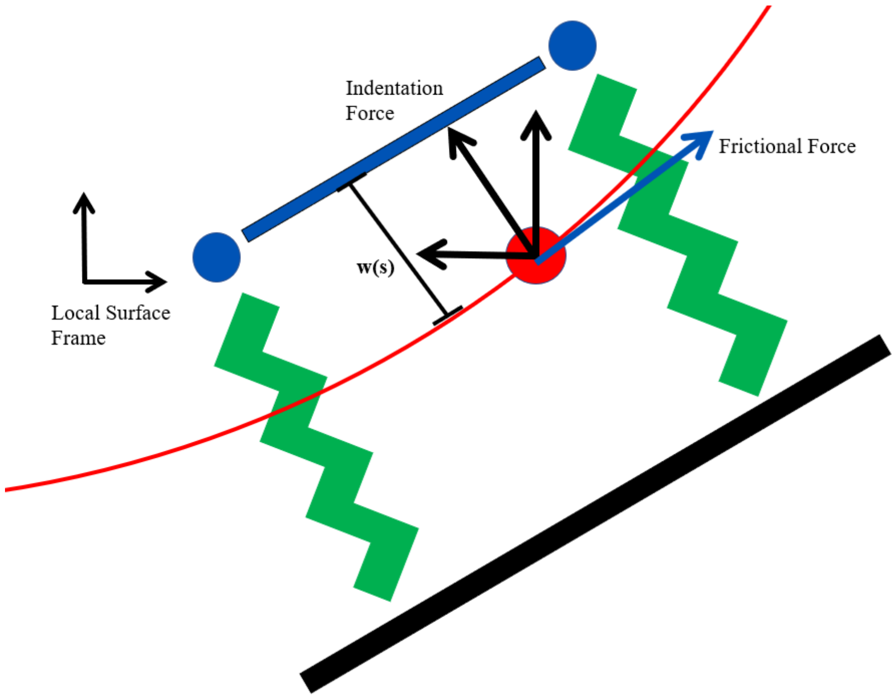

Figure 8.8: Point cloud representation of the contact matrix where indentation depths normal to surface are calculated and broken into component directional forces.

The Exterior Loop

The exterior loop of the algorithm that encompasses steps 3, 4 and 5 for the continual increase of applied tension. In dynamic scenarios, this step represents time steps and the change in applied force. In the initial securing algorithm, this step is purely for calculating the force-position relationship that occurs from a discrete increase in applied tension in a pseudo-static (time-independent) loop.

In order for the system to avoid numerical stiffness and remain numerically stable while solving the force balance, a small step applied tension is implemented. The tension applied through the rope is used as the control input in the global loop. This tension generates a capstan force, deformation, and pressure distribution. The numerical approach for solving the indentation depth given an applied force is the interior loop as it requires a number of iterations using simple gradient descent algorithms with a selected, small, hyper-parameter to reach a local minima. The capstan equations are solved once at the beginning of each step based on the final position of the previously applied tension step to save computational time by remaining constant.

Once the local solution has been found, the algorithm proceeds to the next step in the applied tension governed by a selected step size. At the last iteration step, the capstan force is calculated twice to double check the validity of the local solution by using the same tension applied through the strap. Once the final position of the limb is found, a pressure map and the relative position of the limb in the exoskeleton is output. Generating the equivalent spring-damper coefficients by summing the parallel spring coefficients is also possible. The finalized algorithm flowchart can be seen in Figure 8.9.

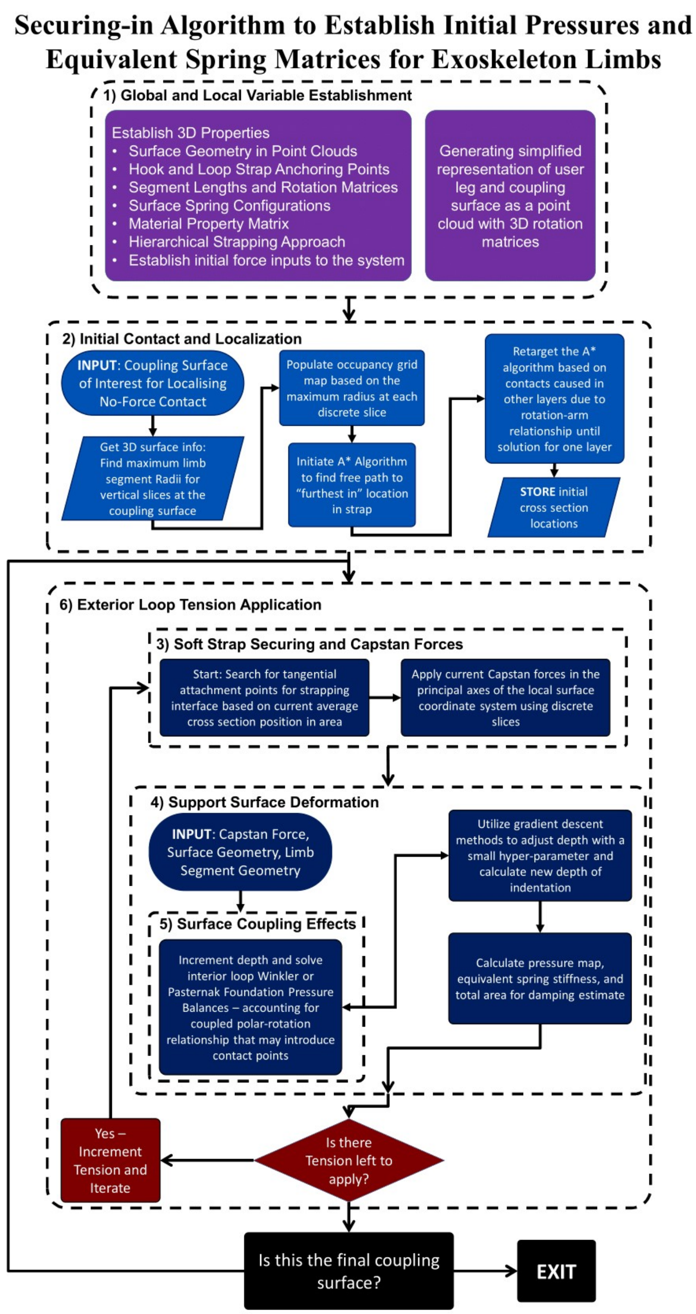

Figure 8.9: Flowchart representation of the logic to solve the total securing algorithm

Validating the Performance of the Elastic Foundation

To demonstrate the effectiveness of the proposed algorithm in matching known closed form solutions, a typical "back of napkin" materials approach [185], and ideal contact mechanics solution are calculated [184, 185]. This algorithm's performance will rely on its ability to replicate exact solutions, as indentation depth and kinematics relate directly to the force-pressure relationships. To evaluate the performance of our algorithm, we compared exact solutions for a sphere indenting a medium from Hertzian stress theory [184, 185] against a point cloud sphere using our pressure search algorithm.

The fast approximation and exact solutions seen in Figure 8.10.The approximated solution assumes that the maximum indentation represents the strain and applies Hooke's law using Young's Modulus.Equation 8.6 represents the governing equation that, given the average indentation depth and surface area, can be used to replicate the response of commonly employed spring-damper systems. In this equation, H represents the characteristic depth. The resultant force is calculated by converting the uniform pressure into force using an approximate equation for the indented surface area.

$$\sigma = \frac{Ed}{2a} \tag{8.6}$$

The exact Hertzian contact theory solution has known well-defined solutions for indentation depth, contact pressure, and required force derived from the theory of elasticity [184, 185]. The governing equations relate indentation depth and pressure, where the pressure distribution is detailed in Equation 8.7 and the maximum pressure in Equation 8.8.

$$p(x) = p_0(1 - (r/a)^2)^{\frac{1}{2}} \tag{8.7}$$

$$p_0 = \frac{2E}{\pi(1 - v^2)}(d/R)^{\frac{1}{2}} \tag{8.8}$$

With knowledge of the algorithm performance against ideal solutions, future studies can identify sources of error that might arise from the assumptions made in the kinematics. Without showcasing the simulated performance of this model against established solutions for similar problems, identifying where error is occurring in the algorithm becomes difficult. It is not possible to determine whether recorded errors are due to pressure offsets or kinematic offsets in the model without extensive sensorization.

By isolating the pressure-indent error, improvements can be made to either the Pasternak foundation or kinematic loading steps in an attempt to improve the algorithm. This allows for the design process to remain decoupled.

This section highlights the performance of the securing algorithm's Winkler and Pasternak foundations using an approximated materials approach against a known closed-form exact solution. The material properties were simulated to represent an EVA foam layer with a relatively low poisson ratio (0.05). For testing, a sphere with a radius of 100 mm was used due to its radial cross section and steep curvature, which introduced potential errors at the bounded edges. The sphere indented each surface by the 1 mm. This was performed to showcase the baseline performance of the algorithm prior to real world testing. Real world testing conditions may vary from session to session, and thus indicate a different in simulation vs measured result that does not accurately reflect the baseline performance. For this reason, a closed form solution was compared against to ensure more accurate adjustments could be made in the future.

8.2.4 Results

In simulation, three metrics were calculated for comparison: indentation separation point, pressure profile, and force required for indentation. The results for the approximated and exact Hertzian solutions, as well as the Winkler and Pasternak foundations are seen in Figures 8.10 and 8.11, respectively.

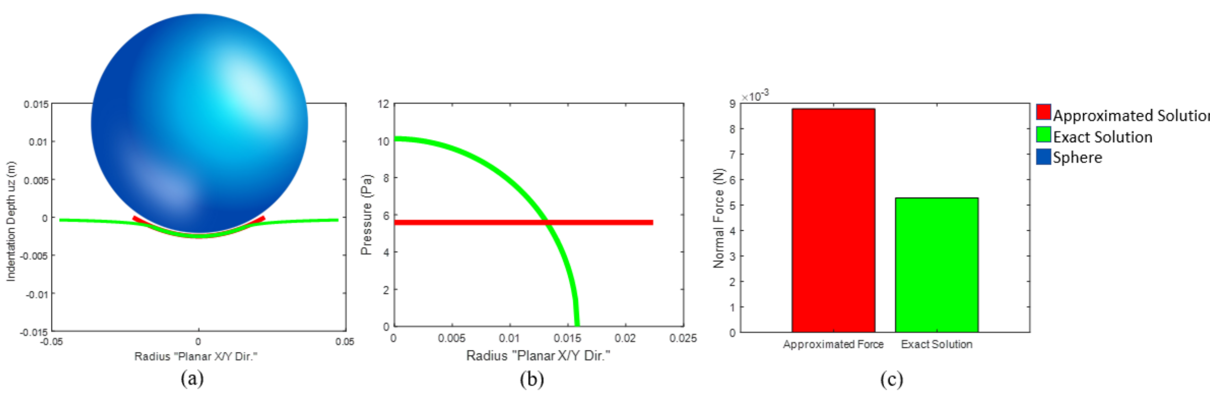

Figure 8.10: Performance of the approximated solution (red) with the exact Hertz solution (green) (a) Indentation profile (b) Pressure distribution from CoP (c) Force required for Indentation

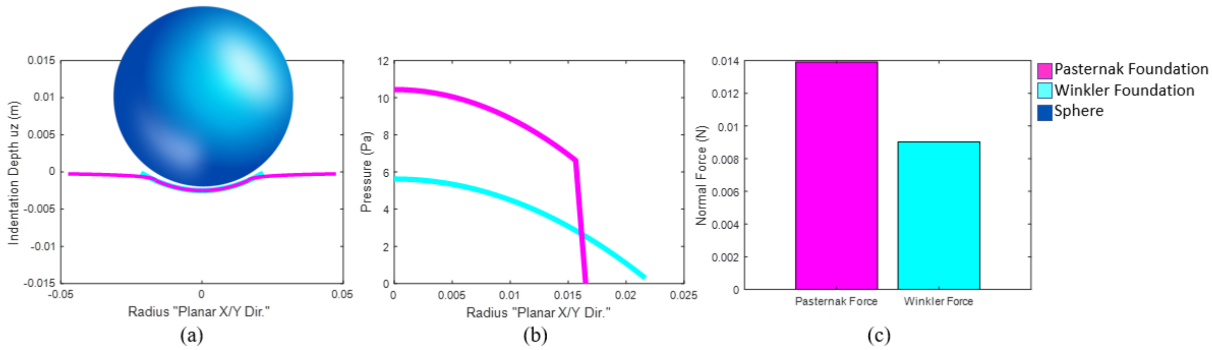

Figure 8.11: Performance of the Winkler Foundation (cyan) with the Pasternak Foundation (magenta) (a) Indentation profile (b) Pressure distribution from CoP (c) Force required for Indentation

Table 8.1 details the results of the above simulations. The pressure profiles for the Pasternak and Winkler foundations appear "flatter" than the Hertzian solution but are more accurate than the approximated model. The Pasternak and Hertzian models exhibited similar maximum and average pressures and similar separation points. However, the force required for indentation is much higher. The Pasternak model exhibits a 10% increase in average pressure in comparison to the Hertizan solution and a similar maximum pressure. The force required for indentation is 160% of the approximated and Winkler solutions. The Winkler foundation performs poorly in terms of maximum pressure and average pressure. It's maximum pressure was 56% of the Exact solution's maximum pressure and its average pressure was 37% of the Exact solution's. Furthermore, the required force was nearly double that of the Hertzian solution due to the large contact arc.

Table 8.1: Performance of the various pressure calculation methods

Model Name	Separation Point (Half Characteristic Depth) in meters	Maximum Pressure - Pa	Average Pressure - Pa	Force Required - N
Approximated ("back of napkin")	0.0224	5.5902	5.5902	0.0088
Hertzian Exact Solution	0.0158	10.0911	7.8378	0.0053
Winkler Foundation	0.0222	5.6254	2.8743	0.0090
Pasternak Foundation	0.0160	10.44	8.6757	0.0139

8.2.5 Discussion

The performance and applications of this algorithm will be discussed in this section. Potential alterations to both the level of complexity of the algorithm and composition of the surface are highlighted to aid in applying this algorithm to other exoskeleton coupling surfaces.

Algorithm Performance

This design paper showcases the performance of a simulated elastic surfaceunder indentation by a rigid sphere. We calculated the pressure distributions, curvature, and force requirements for a known indentation depth under four separate conditions to illustrate the pressure-indentation relationship that can be expected when employing our algorithm.

Relative to the exact Hertzian solution, the Pasternak solution performed the best. The Pasternak foundation approximated the maximum pressures and average pressures accurately, but could not approximate the force well. If characterizing pressure distributions and average pressure is the main objective of simulation, then the Pasternak foundation should be used. However, it will underestimate the indentation depth given the same forces as the Hertzian approach.

The discrepancy between the force approximation and pressure relationship in the Pasternak foundation may be attributed to the coefficients of the spring and shear values. Some formulations incorporate a "decay" feature for the shear and Young's modulus springs, which reduces their influence as distance from the center of pressure increases [187, 195]. By incorporating a similar relationship in our model, we may achieve a more accurate approximation similar to the Hertzian distribution. Selecting an appropriate relationship can be challenging as there is no general consensus on the capabilities of each model. Most models use relational coupling or decaying terms to achieve accurate approximation. For implementation, as long as the cross-sectional areas remain circular, including nested circular cross sections, the performance should be straightforward to approximate. By applying a validated relationship for a specific material property, such as the one presented here and scaled using the Pasternak-Hertzian force relationship, we can obtain the correct depth and pressure relationship for the Hertzian solution.

Computationally, the time it takes to find the point of separation is the rate-limiting step. Since most of the solutions require iterative methodsm, the computational time is dependent on how stiff the relationship is, and how far away the initial estimate is. Establishing a faster method of finding the exact de-coupling location would improve the model.

One major benefit of these discrete approximation methods is how customizable they are to highly complex surface compositions. Solutions for non-radially symmetric problems are difficult to create in contact mechanics. By allowing the model to find solutions on complex surfaces that other models cannot, optimal surface designs can be generated given template structures or bounded ranges to remain relatively accurate.

To validate the pressure-indentation relationship of our algorithm, we validated it under controlled conditions without the need for sensors. However, testing the position assumptions made in our algorithm against real-world conditions presents challenges without a baseline reference for comparison. Although finite element analysis (FEA) is a more accurate procedure, it requires assumptions to be made about loading conditions and may struggle with contact points. Real-world testing, however, runs into complications associated with the kinematic assumptions. Therefore, we present the performance of 3 relative approximations to a known solution.

Applications and Implementation Challenges

There are a many applications that this simple coupling model brings to the design and control of exoskeleton surfaces. Namely, this model will improve surface design prototyping, control, and optimization algorithms motivated by recent studies that acknowledge the effect of initial cuff and surface pressure on performance.

This paper does not cover dynamic estimation, which is a major component of exoskeleton use is control and modelling during gait. For a model to be effective in control and optimization, it must be simple to implement and fast when converging to a solution. The Pasternak foundation, while effective at estimating pressures compared to simpler methods, requires more computational effort and setup time. This complication can be circumvented using similar characterizing methods used in characterizing linear spring-damper models. By fitting a depth-variable spring and damper coefficient to establish a characterizing equation, as seen in a Hunt-Crossley contact model, more accurate curve fitting can be achieved than with traditional linear approaches [173]. These initial simulated coefficients can then be used in typical linear spring-damper formulations, providing the added benefit of known pre-compression for a more accurate estimation of the force-depth relationship, without the added computation time.

The adjustability of this algorithm provides an advantage in designing new exoskeleton coupling surfaces. Compared to simpler models (such as linear spring-damper) and FEA software, this novel model allows for more complex implementation with less computational effort. Traditionally, exoskeleton interface design relies on heuristics and experience-based design principles, which do not promote innovation. However, modern engineering design processes are iterative and incorporate the use of CAD and FEA software to design and evaluate the performance of new coupling surfaces. Surprisingly, few studies have focused on computer-aided and analytical design of support surfaces without relying heavily on heuristics and user feedback surveys. By utilizing this algorithm during the iterative design process, designers can easily make changes to the algorithm itself to suit their needs. They can adjust the cross-sectional curvature to better reflect actual use, increase the complexity of the no-force contact search algorithm, and add more complex coupling equations in the Pasternak foundation. This allows for a wide array of customizability that can accommodate the design of new surfaces. With a simple algorithm, designers can implement a guided iterative design that requires less effort, thereby facilitating the analytically supported design of new coupling surfaces.

The algorithms used for controlling and optimizing exoskeleton kinematics and dynamics are currently rudimentary. For effective control and optimization, the governing equations and constraints must accurately model the coupling mechanics. Poorly modeled equations can result in less effective trajectories and jittery movement, particularly for online controllers. Offline optimization uses desired goals and equations to create new trajectories, but poor coupling equations can lead to significant differences between the desired and actual results. Constraint equations govern coupling by determining the offset between segments under input force. Replacing these equations with a generalized algorithm or minimizing the force-depth relationship presented in this paper would be a significant but worthwhile adjustment. While implementation would require some computation time and information about the lower limb exoskeleton, this new approach would provide a more accurate computation that decouples the system, similar to current linear spring-dampers. This approach offers a middle ground between accuracy and simplicity for exoskeleton designers.

The simplicity of a coupling model that only requires a least squares approximation to generate spring and damper coefficients is a significant advantage. However, the processes for establishing and utilizing the algorithm presented in this paper require a lot more information, particularly the material characteristics of the supporting surface, as well as its geometrical composition. This may limit ease of access for implementation but can allow for a more accurate response if all knowledge regarding the coupling surface and the applied tension through the securing straps is known. To accommodate scenarios in which all information cannot be known, such as cross sectional geometry, or tension applied through the strap, approximations or least squares estimation can be used. Online estimation, using adaptive linear quadratic regulators or a Kalman filter approach, can search for missing components given information of the surface and kinematic configurations.

8.2.6 Conclusion and Future Steps

The algorithm and framework presented here propose a strategy for identifying the initial conditions of exoskeleton interaction. The accuracy of the algorithm relies on the level of detail captured in the point clouds, as well as the effectiveness of the proposed Pasternak foundation in calculating the force-depth relationship.

Traditional approaches to modelling the physical exoskeleton-human interface for control or evaluation, such as the perfectly coupled model or linear spring-damper model, are limited in their ability to describe complex nonlinear problems. In this paper, we propose a model and framework inspired by spring-bed networks that come into contact as the depth of contact is increased, allowing for proper characterization of the contact problem as nonlinear. The proposed model can be used to characterize initial pressure during the exoskeleton donning process, as well as for calculating forces and pressures during typical exoskeleton movement. The implementation process and framework presented in this paper is intended to be used primarily for the initial strapping in sequence during use which until recently was not considered a crucial component in influencing the system dynamics. This algorithm also offers a faster and easier to interpret and implement surface evaluation compared to FEA, which may struggle with the complexity of the strapping in problem.

Preliminary results from this study indicate that the Pasternak simulation performs relatively well in finding separation points and estimating maximum pressures, however, the force-indentation relationship will be under-estimated compared to an ideal solution. While closed-form solutions, such as those found in radially symmetric contact mechanics, or numerical methods, such as boundary surface evaluation, will result in more accurate results, this step is intended for use as an intermediate step between linear spring-dampers and exact solutions. The strength of the Pasternak approach is its transferability to any geometry as the approach is discrete and analyzes the surface using an element-by-element approach. A component of Winkler and Pasternak foundations not explored in this paper, is the ability to approximate the spring and shear coefficients given experimental data. The next step would be to implement this feature to improve the efficacy of the algorithm.

Future work is required to further generalize the presented framework of analysis to accommodate other exoskeleton interface designs, including entirely soft, or entirely rigid interfaces. In this preliminary version, all cross sections are assumed to be circular which future versions should address, however, the additional point-cloud rotations required for this feature would increase computational time.

Finally, in a future study, we aim to evaluate the proposed algorithm and the governing surface interface equations using both static and dynamic simulations, and compare them against recorded kinematic and pressure data. The performance of this interface pressure model will be compared against a linear spring-damper formulation and a FEA approach, in terms of both time and accuracy, to provide a comprehensive evaluation.

8.3 Future Work and Exploration

The model presented herein is far from perfect. Preliminary results from a real test which utilised the Winkler foundation (presented previously), and one of the coupling segments of the H3 Exoskeleton yielded results within 10 percent of the measured values in both magnitude and percent area. This result indicated that, while not perfect, this algorithm can be used to determine the pressure layout, force distribution and general performance of a coupling surface during initial donning. These results were presented as the preliminary version of the securing algorithm at the conference NACOB 2022, and are shown in Figures 8.12 and 8.13.

Further improving and expanding on this model has a few potential directions for development and exploration. Improving the accuracy of the model requires increasing the complexity of the governing equations, or improving the assumptions made during setup. Increasing the complexity of the problem generally requires increasing the order of the defining equations and the considerations of coupling effects. For example, moving from a 2 dimensional slice that considers shear only in plane, can be shifted to a complex plate which represents considerations in three dimensions for the shear component. Further increasing accuracy involves approaching the problem from a first principles perspective [200], or from a finite element perspective utilising boundary constraint conditions. These methods, and assumptions would aid in solving more accurate pressure distributions and maximum forces required for indentation. Improvements can also be made to the approximations in the Winkler and Pasternak foundations, by altering the effective spring values utilising any number of assumptions, such as Vesic's beam on solid approximations [197]. The simplest assumptions were used in this study to show a proof of concept implementation of such a strategy, and lay the groundwork for future exploration.

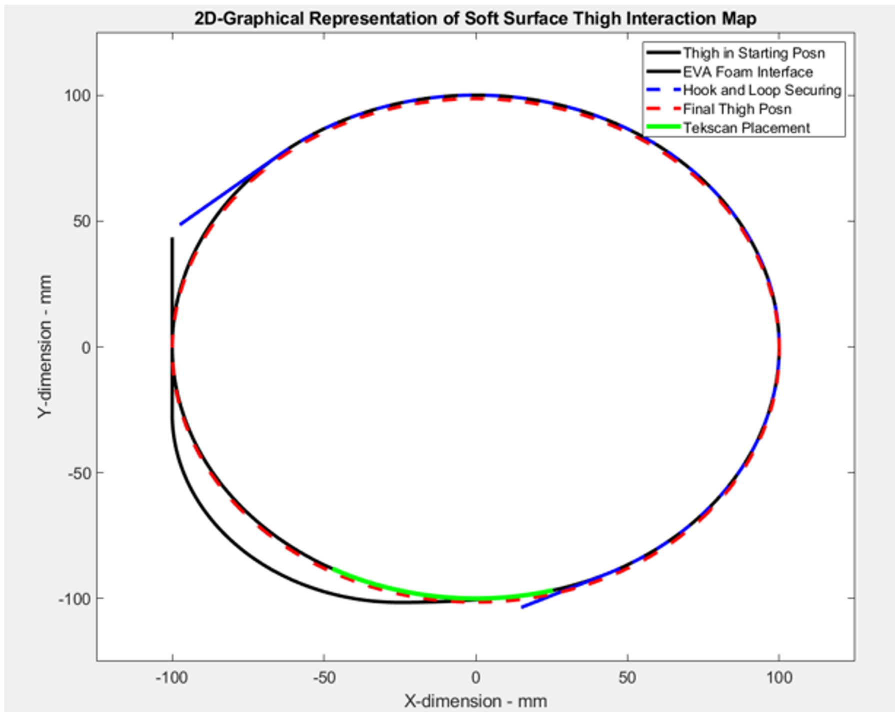

Figure 8.12: Matlab (2021a) Representation of the H3 exoskeleton surface cross section, VELCRO securing strap and user limb

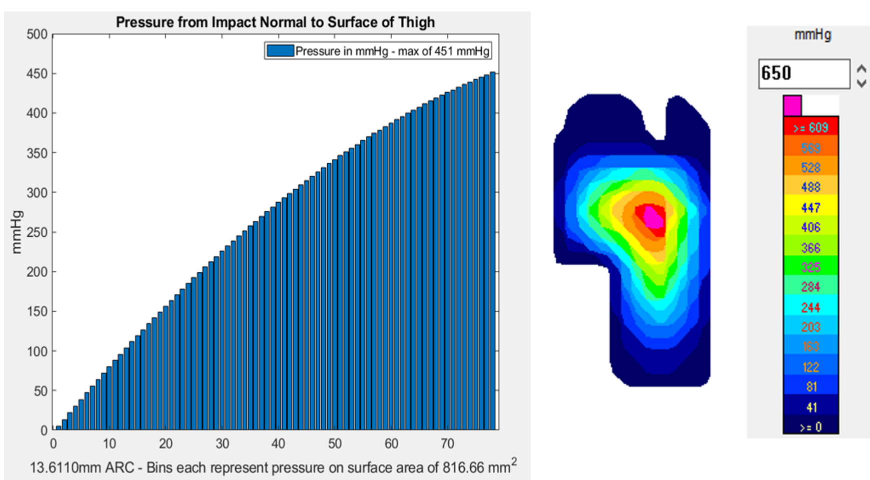

Figure 8.13: Total Pressure across the active area, compared to the recorded Tekscan values. Pressure drop-off cannot be calculated with the Winkler model.

Chapter 9

Conclusions and Contributions to the Modelling of Exoskeleton Coupling Interfaces

The most common approach for the modelling of the exoskeleton coupling interface simplifies and approximates the interactions that occur between user and device for ease of implementation and simplicity in characterisation. The models utilised for describing the coupling between user and device are utilised primarily for kinetic simulation of the relative position between user and device for the purpose of estimating the device effects on the user, and how to control those interactions [58, 69, 70, 89, 129, 179, 202]. Utilisation of these models for predicting risk to the user by means of joint misalignment does exist [94, 98], but the primary applications often focus on kinematic-kinetic estimation for simulation and control [89]. These models are represented primarily by perfectly coupled, or linear spring-damper models which greatly simplify interactions to one-dimensional constraint equations.

The drawbacks and limitations of the existing models are clear. While they are useful in their simplicity for implementation, inaccuracies present in the model tend to poorly represent the force and torque components required to drive the user. Because of these inaccuracies, their reliability in properly characterising the influence of the coupling interface diminishes both for exoskeleton performance (i.e., control, force transference, gait), but also in user-safety (i.e., joint misalignments, limb offsets, skin injury), is brought into question [89]. The limitations of the one-dimensional approximation is especially prevalent for the highly complex contact surface areas, and its inability to predict interface pressures [89, 100, 174, 177].

The overarching goal in approaching the modelling of coupling interfaces was to investigate and review existing models, identify their drawbacks, and develop new models to address limitations. The models investigated throughout this thesis were created with the intent of identifying and better characterization of existing surfaces to evaluate their relative contributions in user safety, and exoskeleton performance. These new models could then be used to more accurately define and characterise interface surfaces without the simplicity of a heavily linearised model.

A secondary objective identified was creating models that allowed for the development of coupling interfaces with desired compliant properties, surface pressures and kinematic responses without the requirement of fabrication, a limitation of current models [116]. While not fully explored within the scope of this thesis, Chapter 8 lays groundwork and proof that models such as those proposed can be achieved and realised for use in better coupling interface design. These goals were accomplished through three major sections, all of which are posed for expansion and future work.

Taking inspiration from recent exploratory studies on the effect of initial strapping on the force-position relationship [69, 93], and the findings in our characterising study that initial pressures were not considered in linear spring-damper models, a novel initial pressure model was developed. It was noted through review that no prior study implementing elastic spring-damper systems considered the initial compression and its potential effect on the coefficients calculated and estimated in their study [70, 129].

Simplistic in nature, the proposed model is considered the "next step" in design that considers initial conditions other than position of the limb. The model is implemented for pseudostatic investigation of the influence of pre-compression forces on the reported forces when compared to a simple linear spring damper. This model essentially turns a singular elasto-gap model into a "springs in series" problem with an equivalent spring-damper coefficient(s). This addition of the spring in series changes the equivalent spring coefficient when both surfaces are in contact, making the position-force curve stiffer. Damping is not affected by this model, as it is assumed that it is unidirectional similar to other models. This initial proof of concept study indicates that initial pressure contributes to the forces experienced at the surface, and suggests that exploring other initial conditions or unmodelled forces may be an alternative route for improvement. Introducing and allowing for an additional spring to be in contact, the proposed model still requires an existing coupling interface to exist. This pre-compression model explored, however, performs much better than that of the traditional linear spring-damper models, indicating that further exploration into initial conditions, and complex contact conditions is warranted to solve the drawbacks of existing models.

The next step, and final contribution of this thesis, was the creation of a coupling model improving upon the established initial force condition approaches to allow for the addition of complex contact arcs and frictional forces, towards addressing complex geometry issues. As the discrete element elastic foundation model, this model was created with the intent of expanding on linear models without adding too much complexity. The concept of an elastic supporting bed of springs for exoskeleton coupling interfaces has only been more recently explored, but is a popular method of explaining and describing soft material elements in other robotics applications [189, 191]. Partially responsible for the direction pursued is the work of Yves Gonthier, who developed a similar model for continuous problems which operate much closer to the idealized solution, and acted as a starting direction for a much simpler model [200, 206].

The strapping in algorithm localises the user's limb under no-input conditions to find the geometry initial condition forces that statically secure the leg in place. The intended use of this algorithm is primarily on the evaluation of and development of ideal pressure maps on coupling surfaces during donning for user safety. By providing the ability to calculate equivalent spring values based on indentation depth, which cannot be done with least squares approximation as the relationship is not a linear combination of state variable influence, the proposed algorithm can be used to evaluate preliminary pressure maps before fabrication. At the same time, depth-variable spring-damper coefficients which can be input into the already existing linear spring-damper and pre-compression models can be created. For the scope of this thesis, only simulated results were compared for the more complex Pasternak models. However, an early formulation of the algorithm and elastic foundation utilising the Winkler model. Both Paternak and Winkler models demonstrated good accuracy in predicting pressure maps and trends relative to simulated and recorded values, with limitations associated with initial conditions. These promising preliminary results are the motivation for future work, and indicate that even simple elastic foundation models may be an appropriate method for developing coupling interfaces for both performance, and safety prior to fabrication and supported by empirical data, as opposed to the traditionally "best practice" focused approaches.

The benefit and novel contribution of this algorithm is its versatility and the ease of implementation to existing models. While simple linear equation driven models that can be integrated into Euler-Lagrange simulation are beneficial due to their relative speed of computation, they cannot be used for prior simulated design. By limiting the use of this algorithm to initial conditions, the proposed approach allows for both characterisation of user safety without sacrificing the simplicity of existing models. The initial conditions - or spring beds in contact, can then be converted into equivalent spring-damper curves that depend on indentation depth that integrate seamlessly into existing models.

While the process of designing, simulating, fabricating and validating a sound and effective coupling interface for either accurate inertial values or desired pressure maps is possible utilising this approach, it is outside of the scope of the current thesis. The model was intended and validated against static strapping conditions. While validation of the equivalent spring values and pressure maps during dynamic use is critical for implementation. In this thesis, the strapping model was developed to establish better baseline estimations.

Chapter 10

Conclusions and Future Work

The coupling interface is a critical, yet underexamined, aspect in the function of the exoskeleton. As a direct result of it being the transmission point for the forces and torques designed to aid the user, it is a contributing factor to user safety, and exoskeleton performance. The focus of this thesis is on the iterative design and sensor based evaluation processes, as well as the models which describe the performance of the coupling interface in simulation and control. The choice was made to focus on these subtopics of exoskeleton coupling interface design as they are the forward facing components of how the exoskeleton interface is designed, evaluated and implemented for user safety and performance. The overall motivation of this thesis was to advance tools for designing coupling interfaces that optimize safety of the individual, without sacrificing exoskeleton performance.

The conclusions drawn from this thesis pertain to the overarching objective and motivation that drove this thesis: how can we create new interfaces optimized for safety, without sacrificing the performance of the device itself?

The iterative and sensor based evaluation of exoskeleton coupling interfaces investigated and developed more appropriate means, metrics, and baselines for the evaluation of already existing coupling interfaces. These tools provide information and methods by which we can evaluate and compare existing coupling interface designs, both amongst each other and against well understood baselines. By evaluating for user safety (i.e., joint misalignment, skin injury) as well as exoskeleton performance (i.e., kinematic offset, supporting forces and energy), we can apply context-based evaluation utilising captured metrics to inform whether a design works better than another.

A baseline set of surfaces for reference as a "best case" scenario for interface evaluation was also proposed, seeking to target issues associated with rationally comparing different exoskeleton surfaces between each other. By advocating for, and showcasing the performance benefits of, a fully customised surface inspired by well-established prostheses and orthoses design, we offered the essential "best case" scenario that others could reference. With the understanding that a fully customised surface will always perform relatively well as an assistive device, it can be used as a reference tool, and design guide for new interfaces. These tools can be used as a means of iterative design and evaluation. If a pressure map highlights points where undesirable pressure is observed, or an interface has poor elastic behavior (e.g., too stiff, too compliant) these context-based evaluation tools can be used to inform safe and effective design. The changes in design can then be evaluated utilising quantitative metrics compared to qualitative observation.

The modelling of the exoskeleton interface investigated and developed new models for coupling to address the shortcomings and drawbacks of existing ones. This work focused on a lack of initial condition considerations resulting in error in modelling, as well as shifting the focus of existing models from one dimensional elastic spring-dampers to multi-dimensional geometry based elastic foundations.

The initial pre-compression model provided support to the hypothesis that initial conditions greatly influenced performance and prediction of the linear spring-damper models. Tackling these initial conditions, we show that even simple integration and consideration into typical free-body diagram approaches will yield more accurate results.

A larger paradigm shift was then taken in changing the coupling model from a "fitted" model which represented a trend as opposed to physical reactions to that of the elastic foundation. The Winkler and Pasternak foundation both exhibited trends of relative accuracy (10%) that could be used to estimate pressure maps for user safety. The complexity in these models limit their implementation in simulations and model, as a result of their computational load in both governing equations and point clouds. To address this issue, and the non-linearities in the system, without trading the efficiency and ease of implementation, a strapping algorithm was introduced that approximated the initial conditions and strapping configurations offline that would output effective strapping conditions. This initial "strapping-in" model's outputs could be used as depth variable spring-damper equations calculated and dependent on the initial applied forces, and distance from the center line, which are easy to implement into commonly used one dimensional spring-damper models. This would aid in both initial condition estimation and accurate coefficient approximation.

A discrete elastic foundation allows for the design and testing of surface coupling conditions that can account for the metrics established in Part 1, while still estimating the same kinematic offsets. More effort is required for setting up the problem space and algorithm for solving (i.e., point clouds, governing curvature equations), but the benefit of not needing to fabricate a surface to establish initial design outweighs the computational effort required. This addition also theoretically allows for the evaluation of surface pressure maps if sensors for surface pressure are not available. As long as one source of data is available then the resultant state-force relationship can be found.

10.1 Future Work

There is still work that needs to be done in the space of exoskeleton coupling interfaces. The content of this thesis scratches the surface in investigating the influence of the coupling surface, established by countless researchers approaching a difficult question. There are many confounding variables that add complexity and confusion to the process of analysing and evaluating the exoskeleton interface. The purpose of this thesis, and my work over the past two and a half years, has been on exploring why it is so difficult to evaluate and model surfaces. It is a multi-faceted and complex issue, that while certain components have been addressed, there is still a lot of work to be done. The future work section will be presented in three sub-sections: 1) the iterative and sensor based design and evaluation section, 2) the modelling and simulation section, and 3) collaborative work between the two.

10.1.1 Evaluation and Design of Exoskeleton Coupling Interfaces for Performance and User Safety Future Work

An important point to acknowledge, and one that is covered to some extent in prior chapters is the limitation of the scope of this (Master's) thesis. Developing these tools to be used as a means of comparison between individual users was deemed too complex as a starting point. Instead I utilised the tools developed as a means of comparing coupling surfaces for one individual, with the intent that others would repeat similar tests. Additionally, there are many components of what make a coupling design "good" involving psychosocial and perception-based factors. For the scope of this thesis, only physical metrics were evaluated with the intent of minimising the components of user risk: skin and musculoskeletal injury. As this problem is a highly multi-dimensional one, only the physical design of the interface was considered, and not the location or number of support surfaces.

Future work is needed that expands upon the findings that fully conforming surfaces are the "best" baseline surface for evaluation. This is arguably the most important component of this thesis. Clinicians and rehabilitation engineers may not be able to simulate full surfaces, designing them utilising point clouds, or governing curves, and instead rely on heuristics or design principles to create a better surface. Creating a baseline of information for how individuals of different shape, size, gender etc. react to fully conforming surfaces, how they benefit from it compared to surfaces like those designed for the H3 Exoskeleton would be a useful database similar to anthropometrics tables generated for designers. It would likely encourage innovation and workarounds to design towards fully conforming surfaces that are common in orthoses and prostheses designs, without making the sacrifice of fitting fewer individuals.

The appropriate metrics investigated in this thesis, primarily the average surface area utilised and center of pressure location variance, were the first step in creating context driven analysis of surfaces. Finding more metrics which can be utilised as a means of evaluating the coupling interface, or creating new sensors to make it easier to do so, is imperative. Tools such as simple, non-vision based joint and limb segment offset estimation sensors would be a major aid, not just in characterisation but also in control. The more tools we can use to characterise when a surface might not be an adequate or desirable design the better.

Lastly, mannequins, and the development of mannequins that mimic human properties as tools in exoskeleton evaluation should be undertaken, prior to human trials. For fields such as orthopaedic implant design and injury biomechanics, investigation is performed utilising cadavers and the most accurate human segments as possible. Replicating this approach, either by directly utilising cadavers to investigate the potential for injury, or creating lifelike mannequins would facilitate early exoskeleton investigation, particularly for user safety. While it is considered a method of evaluation, it does help to create "stable" bases of information collection as well. Seeing more studies utilise a mannequin for real use control algorithms and coupling surface evaluation would make reproduction of results easier.

10.1.2 Modelling and Simulation Future Work

More recent studies which expand upon the work of prior upper limb exoskeleton work characterizing the effects of strapping tension on the spring-damper coefficients at the interface point are helpful in explaining where inaccuracies come from in models, and transferability of information between simulations. Replicating similar studies for lower limb exoskeletons, and expanding the scope of the study to include information on user body type and height, and then comparing those collected values against recorded force sensor values would be useful for future studies. Currently, researchers are aware of the concessions made to implement and utilise these linearized coupling interfaces, but giving a bounding range of inaccuracy under controlled mannequin testing conditions would act as a helpful reference.

Improvement and expansion of both the precompression and strapping spring bed models should be further expanded. Explaining the step by step process of how to characterize the pre-compression model, and fully comparing it against multiple sets of recorded kinematic data, would be ideal for validating its usability. The particular difficulty of utilising this model is proving how easily it can be incorporated into controllers or simulation; hence, highlighting this potential application would be ideal.

For the strapping and securing in algorithm, further exploration and expansion of how it can be built upon, introducing new searching methods, random noise, and more accurate elastic foundation calculation would make it a useful tool for initial strapping condition calculation. Currently, more depth of examination needs to be conducted to further evaluate the accuracy to fit current exoskeleton kinematic-kinetic models by creating more accurate spring-damper coefficients that are non-linear based on indentation depth. By doing so, it may be transplanted directly into existing algorithms, with the only change being the need for an initial calculation, and does not significantly alter the speed of the algorithm. For simulation purposes, exploring the introduction of first principles investigation or finite element analysis for the surface contact would result in more accurate implementation of the elastic bed, and allow for design to implement more refined optimisation algorithms for desired elastic or pressure based interactions. Deriving or investigating the use of more generic elastic foundation coefficients, more reliant and effective than the independent pillar based approach currently taken.

10.1.3 Future Collaborative Work between these two topics

While it would have been a desirable outcome, the scope of this thesis did not leave room for investigating utilising these newly developed and investigated tools. Utilising computer generated models to create surfaces with idealized elastic and pressure properties motivated by skin and musculoskeletal injury, and then characterising them using a mannequin test-bench, the newly developed metrics and fully customised surfaces to compare performance against would be an ideal demonstration of application.

As mentioned in this thesis, there is currently no reliable way of generating a known idealised interface pressure map and desired elastic properties before fabrication. While kinematic trajectories can be estimated if there are no elastic components [116] doing so under alternative conditions or governing equations is not possible. Estimating elastic components (either position-force offset or coefficients themselves), forces imparted onto the interface (not the joint), and the simulated compliant trajectory of the limb cannot be known without a fabricated coupling interface [116]. This severely limits the capabilities of design and innovation by placing a cost barrier to entry for fabrication, and rewards the simplest means of characterisation if they are the most commonly utilised and implemented models. These two subtopics are inherently tied, with shared components in how we can analyze and evaluate interfaces. One relies on existing interfaces, while the other utilises simulated ones.

Connecting the two together would allow for a “direct stream” of design and evaluation process that is justified in its approaches as compared to the current standard for developing new interfaces. By creating better models for simulation informed by materials, a surface can be generated that optimizes for a desired pressure map goal and kinematic offsets during trajectory. Those ideal components can be informed by the baseline design for a customised surface, where iteration for improvement stops once parity is met. Metrics and constraints for user safety simulated would be informed by the iterative and sensor based evaluation process using real surfaces, primarily relying on those comparative metrics, while kinematic misalignment and offsets take inspiration for constraints and goals from both modelling and sensor evaluation. After an initial design is created in simulation, informed for both safety and performance, the device can then be fabricated and evaluated for performance and safety in real use case scenarios. If iteration is still required, this new information can be used and then improved upon for each required iteration or goal after inputting new equivalent material values and readjusting.

The benefit is clear: requiring very little input in terms of getting started with accurate models for both safety and performance is a step up from the current process. This process is not uncommon in other fields, including actuator design in exoskeleton use. By reaching parity in the engineering process and allowing for simulated or more accurate calculated results that address both safety and performance concerns, many problems associated with the interface may be addressed.

The hope is that by tying these two methods of evaluation together, new interface designs may be created when a designer is not faced with the cost and time associated with manufacturing. While the most beneficial design may be the ones that currently exist, without more exploration into other designs and interfaces, we cannot know for certain. Making the barrier to entry much lower will encourage innovation, and drive design towards better, and more effective, coupling interfaces.

10.2 Closing Remark

The work presented in this thesis is intended to be an examination into the exoskeleton coupling interface that addresses the history, current issues and novel contributions that could be added to the research space. By addressing primarily the two most forward facing sub-topics of exoskeleton coupling interface design, this thesis offers alternative methods, novel approaches and new models to use during said process to make them more accurate. Until recently, the field of exoskeleton coupling interfaces has been relatively ignored, in no small part due to the complexity of approaching the problem and parsing complex interactions. I hope that the contributions made in this thesis inspire others and help pave the way for better means of exploring these interfaces.

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