



**Cape Peninsula
University of Technology**

Design, Build and Test an Assistive Lower Limb Exoskeleton

by

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ABSTRACT

In 2017, the global elderly population was approximately 962 million, and this figure is projected to reach 2.1 billion by 2050, according to the United Nations. Africa is anticipated to experience an increase of 225.8 million elderly people during the same period. As of 2019, South Africa's elderly population stood at 5.3 million, and this number is expected to continue growing, according to Statistics South Africa.

In recent years, exoskeletons have emerged as a significant technological advancement to assist individuals with limb disabilities. However, many existing LLEs tend to be heavy and cumbersome, thereby increasing the user's metabolic cost.

The objective of this study is to design an adjustable, lightweight LLE that enhances the mobility of elderly users by reducing their metabolic expenditure. This LLE is specifically designed for movement in the sagittal plane; therefore, it is assumed that users will rely on crutches for additional lateral support and balance, as the exoskeleton is limited to forward motion.

The development of this walking robot was achieved using CAD software to model the LLE. During the modelling process, anthropometric data were used to shape the design of the LLE in relation to its intended user. The LLE was specifically designed to accommodate users ranging from 1.505 m to 1.740 m tall, with a maximum weight of 80 kg. After completing the modelling phase, the LLE was imported into Ansys for finite element analysis (FEA), which was conducted segment by segment. Upon confirming the structural integrity of the LLE frame, the walking robot was imported into MATLAB. An adaptation of the walking algorithm developed by Castro and Kim in MATLAB was modified to align with the LLE's design requirements for this research. During this phase, both kinematic and dynamic modelling were established, along with the Linear Inverted Pendulum Model, which was essential for the LLE's walking functionality.

The weight of the LLE frame was measured at 4.942 kg; however, with the actuators included, the total weight rose to 13.338 kg, excluding the 16 kg of the two batteries. FEA results for the LLE frame demonstrated its ability to endure the applied loads without compromising its structural integrity. The yield strength for each segment was maintained below those of aluminium 6061-T6 and 6082-T6 for the frame, and below that of low-carbon steel for the bolts and nuts. Furthermore, the buckling load for each segment of the LLE remained below the critical buckling load during the stance phase, whereas the maximum bending stress experienced by these segments stayed below the specified permissible bending stress during the swing phase.

In a MATLAB simulation, the LLE successfully walked 5.63 metres in 10 seconds, demonstrating its responsiveness to the walking algorithm. The scaled-down prototype of the LLE effectively mirrored the walking gait cycle in practice, with results from both the experiment and the simulations closely aligning.

The contribution of this research is centred on the development of the LLE design to assist the elderly population by improving their walking capabilities, which are affected by aging.

Keywords: Exoskeleton, assistive, lower limbs, zero moment point, centre of pressure, centre of mass, trajectory, dynamics, kinematics, balance, and stability.

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DEDICATION

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ABBREVIATIONS AND DEFINITIONS

ALLE	Assistive lower limb exoskeleton
BLEEX	Berkeley lower extremity exoskeleton
CAD	Computer design-aid
COM	Centre of mass
CPUT	Cape Peninsula University of Technology
DOF	Degree of freedom
D-H	Denavit-Hartenberg
EOM	Equation of motion
FBD	Free body diagram
FOS	Factor of safety
HAL	Hybrid assistive limb
LLE	Lower limb exoskeleton
IPM	Inverted pendulum mode
LIPM	Linear inverted pendulum mode
ROM	Room of rang
SEA	Serie elastic actuator
ZMP	Zero moment point

Some definitions

- An actuator is a device that converts energy into motion.
- Paraplegia is the paralysis of both lower limbs caused by spinal disease.
- Paraparesis is a partial paralysis of the lower limbs.
- The robotic exoskeleton is a wearable electromechanical device that increases physical ability.

CHAPTER 1: INTRODUCTION

1.1 Problem statement

Mobility is a crucial ability in everyday life, but it can deteriorate with age. During this phase of life, elderly individuals often require support with mobility. This assistance can be provided through various assistive technologies, including crutches, wheelchairs, canes, and walkers (Sehgal et al., 2021). Unfortunately, none of these devices resembles a lower limb exoskeleton (LLE) that mimics human movement and capability.

Over the last decade, engineers and scientists have dedicated their efforts to enhance the usability of this technology. Although substantial progress has been achieved, it continues to encounter certain limitations and challenges. One significant concern is the weight and size of LLEs, which can pose difficulties for the elderly population due to their diminished muscle strength (Fosch-Villaronga & Özcan, 2020; Hamza et al., 2020).

The identified limitations of the LLEs need to be thoroughly addressed, as they have a profound effect on both the metabolic cost experienced by the user and their overall comfort experience during use. These challenges not only hinder the user's experience but also influence the efficiency and effectiveness of the walking robot itself (Zhang et al., 2021; Goo et al., 2020).

The weight and size of LLEs are directly determined by the selected materials and the choice of components, such as motors, harmonic drives, and batteries, which are selected during the exoskeleton design process (Rodríguez-Fernández et al., 2021; Goo et al., 2020).

1.2 Background

In 2017, the United Nations (UN) reported that the elderly population was 962 million, and this demographic is expected to reach 2.1 billion by 2050. In Africa, this community will increase from 68.7 million to 225.8 million, which is 7.1% to 10.9% of the global population, respectively (Department of Economic and Social Affairs, 2019). In 2019, the elderly community in South Africa accounted for 9% of the South African total population, or approximately 5.3 million individuals, and this classification refers specifically to those aged 60 and older (Statistics South Africa, 2019). Every year, 3 million elderly people are admitted to hospitals due to falling injuries (Centers for Disease Control and Prevention, 2017). In addition, an annual estimation of 250 thousand to 500 thousand people are victims of spinal cord injury worldwide, where 44% of them have completely lost the function of their lower limbs; this condition is known as paraplegia (Desai & Patel, 2017).

According to a 2016 report, approximately 2,248 million individuals, or 4% of the South African (SA) population, had some form of mobility impairment. Among these, 1,016 million people, or

1.8%, were severely affected, and 174 thousand people, or 0.3%, had lost the ability to walk. Because of these lower-limb issues, assistive devices such as wheelchairs, walking sticks, frames, or crutches have become essential for their mobility. The report further highlighted that 184 thousand people, or 0.4% of the SA population, relied on wheelchairs, while 697.445 thousand people, or 1.5%, used other mobility aids (Statistics South Africa, 2016).

These statistics demonstrate a clear and pressing need for assistive devices worldwide and in South Africa, linked to the number of people with mobility impairments. Moreover, the report projected a significant increase in demand for such devices, particularly among the elderly population and paraparesis victims, which is expected to rise dramatically by 2050. In fact, by then, the elderly population alone is projected to reach 2.1 billion, as mentioned above.

The increasing demand for assistive devices indicates increased usage, making the introduction of advanced assistive technology potentially profitable for suppliers. Chen et al. (2016) state that Over the past sixty years, LLE technology has attracted significant interest from scientists, engineers, academic institutions, and industry, where the primary goal within this field has been to develop human-like LLEs, as these devices provide a more efficient alternative to traditional mobility aids such as wheelchairs, walking sticks, frames, and crutches.

In recent years, exoskeletons have emerged as a significant technological advancement to aid individuals with limb disabilities. As a result, there has been a growing interest in developing these devices due to their potential profitability. According to projections, the industry was expected to be worth \$1.8 billion, equivalent to R32.4 billion, by 2025 (Golabchi et al., 2022). Despite the expected significant market demand for this product, LLEs still face engineering limitations related to weight, adjustability, and ergonomic design (Cumplido-Trasmonte et al., 2023). Chen et al. (2020) observe that a principal factor affecting the weight of the walking robot is its mechanical structure, which is directly related to the chosen material.

The LLE technology is categorised into three primary types: augmentation LLE, rehabilitation LLE, and assistive LLE (Chen et al., 2016). An augmentation LLE is designed to enhance the physical capabilities of healthy individuals, allowing them to perform tasks with greater strength or endurance (Zoss et al., 2005), while a rehabilitation LLE focuses on aiding individuals recovering from injuries or surgeries by supporting movement and improving mobility during the healing process (Alcobendas-Maestro et al., 2012). In contrast, this research is primarily centred on an ALLE, which aims to provide essential support to the elderly population (Ghaddar & Mohammad, 2019). This category is specifically tailored to help seniors maintain independence in their daily activities by offering stability, reducing the risk of falls, and improving overall mobility.

1.2.1 Conclusion

The background proves the relationship among the mobility needs of elderly individuals, clinical significance, and engineering constraints. For example, the excessive weight of a walking robot directly increases the metabolic cost for its user. After reviewing 36 studies on the mobility of healthy people and those with muscle defects, Gavrilă Laic et al. (2024) observed that the use of an LLE device was satisfactory for healthy ageing individuals. In these trials, LLE users demonstrated improvements in walking speed and step length.

1.3 Research questions

The main goal of this research is to develop a lightweight LLE. To achieve this effectively, it was essential to define specific research questions that guide the investigation. These questions provided a framework for examining various aspects of exoskeleton design, including material selection, mechanical efficiency, user comfort, and adaptability across users. By addressing these key questions, the aim is to create a functional, efficient exoskeleton that improves mobility and supports individuals with lower-limb impairments. The following six questions served as guidelines for this research:

- What is the most effective method for designing a lightweight LLE?
- Can an LLE, weighing around 30 kg, replicate the human gait cycle for users who are 1.5 to 1.73 meters tall and weigh up to 81 kg?
- How does the design of LLE have the potential to reduce the metabolic cost for users?
- If the applied load on the segment of the sagittal-plane LLE fluctuates by $\pm 10\%$, will it preserve its structural integrity after multiple FEA simulations?
- If the torque and speed of the hip, knee, and ankle joints of the walking robot change by $\pm 10\%$ in torque and ± 0.100 m/s in velocity, will the robot's stability be maintained during the gait simulation?
- How does an assist-as-needed control system enhance user interaction with the proposed ALLE in this research?

1.4 Research Objectives

The research objectives outlined below provide a framework for addressing the problem identified in this study. These objectives are categorised into mechanical objectives, whose aim is to support the achievement of the simulation objectives, to be accomplished using ANSYS and MATLAB. In addition, there is a mechatronic aim to consider, along with two objectives that focus on the manufacturing and experimentation of the ALLE. The objectives are listed below:

- Develop a kinematic model of the sagittal-plane LLE that replicates the flexion and extension movements of the hip, knee, and ankle joints.
- Develop an inverse dynamic model of the sagittal-plane LLE that can generate sufficient torque to move the leg of this walking robot.

- Develop a 3D CAD model of the LLE, encompassing the frame, actuators, and battery. The robot's total weight must not exceed 35 kg. Furthermore, the robot should be capable of supporting a payload of 81 kg without surpassing the elasticity limits of the robot segment under the applied load.
- Develop a size-adjustable LLE with discretely adjustable segments, whose adjustment limits allow the LLE to accommodate users whose height ranges from 1.5 to 173 m.
- Conduct a segment-centred FEA of the thigh, shank, foot, and waist to assess the structural integrity of the LLE under its own weight and the user's weight.
- Perform the walking gait simulation of the LLE to verify whether the walking robot has sufficient torque to cover 4 meters in 10 seconds.
- Design of a suitable control system,
- Construct a functional prototype of the LLE for experimental purposes.

1.5 Budget

The materials and components needed to complete the prototype of the LLE for the experiment were available in the Department of Mechanical Engineering at Cape Peninsula University of Technology (CPUT); therefore, no budget was needed for this research, especially for the manufacturing of the LLE.

1.6 Significance of the research

Through this research, the CPUT aims to contribute to the development of LLE technology by designing an affordable, lightweight exoskeleton that enhances mobility for individuals with movement impairments. By achieving this goal, the university intends to make this assistive device accessible to a broader audience, including those who may not have previously had the means to get such technology. The intention is to improve the quality of life for many individuals by providing greater independence and mobility through advanced engineering and design solutions.

Furthermore, the researcher intends to develop a compact, portable, and lightweight LLE device capable of generating sufficient torque to assist elderly individuals in walking by supporting their weak lower limb muscles.

In African culture, elderly individuals are esteemed as pillars of their communities, offering guidance and wisdom that influence social cohesion and decision-making. As they age, however, many face physical limitations that hinder their mobility, restricting their ability to participate actively in community life and share their insights (Radar Africa, 2024). If this research on LLE is successful, it could revolutionise the lives of elderly individuals, granting them greater mobility and independence. By enabling them to move freely throughout their communities, they would be able to contribute their knowledge and wisdom more effectively, enriching the lives of those around them and fostering a more inclusive environment that values contributions from all ages.

Injuries and fatalities from falls in the elderly population have profound effects on families and communities (Lord et al., 1993). This research aims to reduce morbidity and mortality from falls among older adults, thereby alleviating the economic burden families and communities face.

The consequence of falling can have a negative psychological impact on its victims by imposing a fear of experiencing the same accident again. Therefore, elderly individuals develop a sense of dependency for their mobility (Kalula et al., 2016). The LLE will improve elderly individuals' confidence in walking, helping them overcome their mobility-related fears.

This research aims to design and manufacture an LLE to help reduce the metabolic costs associated with daily activities for people with lower-limb muscle dysfunctions, such as the elderly.

1.7 Delineation of the research

Research on LLE is known to be complex and diverse. Thus, it is crucial to establish clear boundaries to promote a focused and streamlined workflow. For this reason, the following delineations were established:

- The LLE is only allowed to walk in a sagittal plane.
- The scope of this study is restricted to walking and does not encompass other locomotor activities, such as running, turning, sitting, or ascending and descending stairs.
- Only the ALLE will be the centre of this research.
- Human joints naturally experience movements in various directions, including yaw, roll, and pitch. However, for this research, only the pitch movements of the hip, knee, and ankle are considered, as these are responsible for the forward motion of LLE.
- While several actuation systems exist, this research will concentrate solely on using an electric motor to power the joint of the LLE.
- The frame of the LLE will be constructed from rigid materials, in contrast to some LLE designs using soft materials.

1.8 Expected outcomes.

By achieving the research objectives and addressing the research questions, this study will lead to the following outcomes:

- An adjustable 3D model of LLE,
- A simulation tool that will allow quick and low-cost virtual testing to push the upgrades of the system,
- A technology demonstrator for future studies,
- The student will contribute to a conference presentation,
- The student will obtain a Master of Engineering in Mechanical Engineering.

1.9 Ethical principles

An institutional approval concerning the ethical principles of this research has been granted to the researcher. However, at this stage of the investigation, involving human subjects is not deemed necessary, as the primary objective is to confirm whether the scaled-down LLE model can effectively replicate the LEE's walking simulation. The safety of the research will be prioritised by ensuring that the LLE prototype is set up to prevent hazards associated with mechanical or electrical components.

1.10 Chapter outline

The chapter outline provides a detailed structure of this thesis, offering insights into each chapter, from the introduction to the recommendation.

Chapter 1

This introductory chapter outlines the research problem statement and examines the background to emphasise the study's importance. It formulates research questions based on the identified problem statement and background. The chapter also details the research objectives. Furthermore, it discusses the significance of the research and defines its scope and limitations. Finally, it presents the anticipated outcomes of the study.

Chapter 2

In this chapter, the focus is on conducting a literature review of existing LLEs, with particular attention to ALLEs, analysing their features, capabilities, and limitations. This chapter provides an overview of the information needed to design the ALLE. The literature review is segmented into three primary sections:

First, a summary of the current state-of-the-art related to ALLEs is provided. Subsequently, a gap analysis concerning ALLEs was performed, and finally, the mechanical component of the ALLEs, specifically the frame, was examined in detail due to its critical role in the design of these walking robots.

The second section discusses mathematical modelling, with a particular emphasis on the robot's kinematics and dynamics. The aim is to ensure the stability and balance of the LLE, which is achieved by considering critical points such as the zero-moment point (ZMP), the centre of pressure (COP), and the centre of mass (COM); the selection of materials is also addressed.

The final section of Chapter 2 focuses on the mechatronics of the LLE. This section presents an analysis of a suitable control system for the ALLE, followed by a design process. Furthermore, it examines the critical functional aspects of the exoskeleton, including sensors, actuators, walking controllers, and path generation.

Chapter 3

This research segment is dedicated to methodology and serves as a guide for conducting this study in an organised and structured manner. This chapter provides essential information and techniques for performing a precise and systematic investigation. It elaborates on several key aspects, including modelling a 3D LLE, FEA of LLE, and MATLAB gait-cycle simulation, all of which are fundamental to the research methodology of this thesis. By adhering to the guidelines presented in this chapter, the research can be conducted methodically and efficiently, thereby minimising errors and maximising accuracy.

Chapter 4

This section outlines the results of the 3D LLE modelling, along with the FEA and MATLAB gait-cycle simulations. The necessary data will be collected through these steps, serving as essential tools for assessing whether the FEA and walking algorithm simulations are satisfactory.

Chapter 5

This chapter focuses on the detailed process of constructing a scaled-down model of the LLE. The necessary steps to complete this task are discussed, including material selection and modifications needed for the model. Once the scaled-down LLE model was completed, the experiment was conducted to observe the LLE's behaviour and compare it with the results obtained in MATLAB. The data collected during these experiments served as a foundation for drawing conclusions.

Chapter 6

This section offers the conclusion derived from the results examined in Chapter 4 and responds to the research questions and objectives outlined in Chapter 1. Additionally, it will address the limitations and challenges encountered during the research process.

Chapter 7

This section outlines the key recommendations derived from the results. Addressing these factors will provide insights and serve as tools for future studies.

1.11 Conclusion

This chapter focused on delineating the essential elements required to complete this research. Throughout this process, the problem statement, research questions, and research objectives were clearly established. Additionally, the outcomes and scope of this research were defined to minimise complexity.

CHAPTER 2: LITERATURE REVIEW

2.1 Introduction

This chapter presents a detailed overview of the mechanical structure and components of ALLEs, along with an in-depth discussion of the electronic hardware required for the walking robot to function correctly. The chapter provides necessary information on the forward and inverse kinematics, forward and inverse dynamics, and trajectory generation of the ALLE. Furthermore, this section discusses the actuation system, material selection, sensors, control system, and power supply.

2.2 Overview

Since the 1960s, experts in robotics have been dedicated to developing humanoid robots to enhance individuals' physical abilities. Extensive research has been conducted on the fundamental principles of LLEs and bipedal robots towards achieving this objective. A notable application of this technology was successfully initiated by the United States Army, which used an LLE to improve the mobility and efficiency of a soldier carrying heavy loads, while reducing physical fatigue (Li et al., 2014).

Power exoskeletons are wearable robotic devices that incorporate a control system with a computer board to operate their actuation systems, which may include hydraulics, pneumatics, or motors, to help restore mobility. These devices are also referred to as robotic exoskeletons (Miller et al., 2016). While LLEs are robotic exoskeleton devices, they focus only on the movement of human lower limbs. It is essential to know that there are three types of LLEs: assistive LLEs, augmentative LLEs, and rehabilitation LLEs (Trombin, 2022).

This research focuses on ALLEs. Therefore, it is essential to define this specific type of exoskeleton. Trombin (2022) states that the primary purpose of the ALLE is to assist individuals with lower limb disabilities, including elderly people, in their daily activities that they are unable to perform due to limited mobility.

2.3 Selected existing LLEs

In this session, an exposition on selected existing ALLEs is provided for a thorough understanding of the functionality and composition of these walking robots.

In recent years, significant advancements have been made in the development of LLEs, which represent an innovative leap in mobility assistance technology. Researchers, universities, and companies are actively developing walking exoskeletons that closely replicate human movement while enhancing physical capabilities. Several LLE models are currently available, including the Hybrid Assistive Limb (HAL), ReWalk, Mindwalker, Atalante, and Vanderbilt LLE. Each of these LLEs is discussed briefly in the following paragraphs.

HAL is a remarkable technological advancement created by Cyberdyne Inc. This device serves two primary functions. Firstly, it is designed to support individuals with mobility impairments. Secondly, it can augment physical strength as a power-augmentative tool (Ghaddar & Mohammad, 2019). The exoskeleton has direct current (DC) motor actuators that aid the hip and knee joints. In addition, the system is integrated with sensors that enable real-time monitoring of the environment, with data transmitted to the robot's control system (Kapsalyamov et al., 2019). For illustrative purposes, Figure 2.1 visually represents the front, side, and back views of a HAL model.



Figure 2.1: Hybrid Assistive Limb (Cyberdyne, 2014)

ReWalk is a mobility aid specifically designed to help individuals with spinal cord injuries (SCI) regain independence and mobility. It aims to ease the burden on families and communities. The system consists of a rigid pelvic frame, pads, and a waist belt that work together to provide support for the lower torso, effectively securing the user to the exoskeleton (Esquenazi et al., 2012a). Its hips and knees are powered by electric motors, controlled by a control system in the backpack. Crutches are utilised during movement to maintain the user's balance and safety. A wireless controller is used to program the robot for various transition movements, including sitting, standing, normal walking, and stair climbing (Nacy et al., 2016). The ReWalk LLE is presented in Figure 2.2 for illustrative purposes.



Figure 2.2: ReWalk (Lifeward, 2015)

MindWalker (MW) is an LLE designed to facilitate both roll and pitch movements at the hip joint while allowing only pitch movement at the knee joint. Both joints are powered by series elastic actuation (SEA). MW features an array of sensors, including torque and force sensors, angular sensors for precise joint rotation, obstacle detection sensors, and inertial measurement units. The actuators and sensors are interconnected through an EtherCAT network, ensuring communication and coordination (Heike et al., 2013). As an adjustable LLE, the MW is designed to fit 90% of the adult population in Europe. To determine the adjustability of this LLE, data from the DINED database were used to identify the 5th and 95th percentiles (Wang et al., 2014). Figure 2.3 depicts the MindWalker LLE.

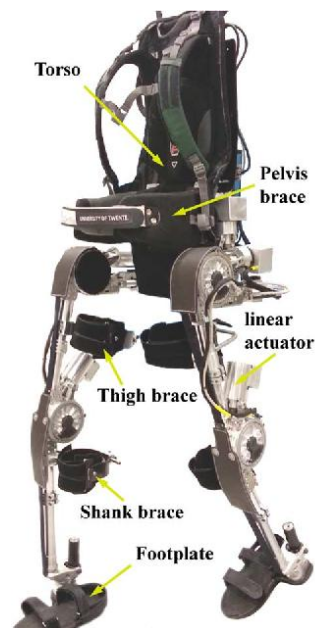


Figure 2.3: MindWalker LLE (Wang et al., 2014)

Atalante LLE is a product of Wandercraft, a French company. The frame of this LLE is constructed from a rigid metallic material. The unique feature of this LLE is to enable a stable walk without the need for crutches (Gurriet et al., 2018). In Figure 2.4, an Atalante LLE model is depicted.



Figure 2.4: Atalante LLE (Wandercraft, 2023)

Each leg of the Atalante LLE features six degrees of freedom (DOFs), comprising three at the hip, one at the knee, and two at the ankle. The hip joint facilitates a variety of movements, including flexion and extension, abduction and adduction, and lateral and medial rotation. The knee permits only flexion and extension. Meanwhile, the ankle allows for dorsiflexion, plantarflexion, pronation, and supination. These degrees of freedom enable the LLE to provide individuals with paralysis with a comfortable walking experience, eliminating the need for crutches (Kerdran et al., 2021).

In Figure 2.5, the Vanderbilt LLE is depicted; this walking robot is limited to operating in the sagittal plane, where the hip and knee joints are powered to generate enough torque to enable the user and the LLE to move forwards (Farris et al., 2012). The hip and knee joint actuators are powered by lithium polymer batteries packaged on each thigh of the LLE. Instead of including an actuator at the ankle joint to control the foot, this LLE uses ankle-foot orthoses (AFOs) to enhance stability at the ankle joint by transferring weight to the ground (Murray et al., 2015).



Figure 2.5: Vanderbilt LLE (Murray et al., 2015)

The existing LLEs outlined above reveal both similarities and differences in their designs. Notably, Atalante from Wandercraft distinguishes itself by not requiring crutches for stability, unlike the other models mentioned, which all utilise crutches to provide essential support during operation. Furthermore, these exoskeletons differ in their degrees of freedom (DOFs), which impacts their functionality and user experience. Some LLEs are designed with 4 DOFs, offering basic movement capabilities, while others feature 6 DOFs or even 12 DOFs, enabling more complex and nuanced movements.

The diversity in design and functionality means that each LLE offers distinct advantages in terms of mobility and user experience, ultimately enhancing users' ability to navigate their environment more effectively.

2.4 Gap analysis

While progress has been made in the field of ALLE technology, several issues remain, including the weight of the LLE, the weight of actuators, metabolic cost, battery weight and life cycle, the use of crutches, and market-related accessibility concerns due to pricing. These matters are thoroughly analysed in subsequent sections, supported by examples that highlight the importance of addressing these challenges.

Chen et al. (2016) state that the heaviness and bulkiness of the LLE are the result of mechanical structures and other additional components located on it. Proper selection of these components will result in a lighter LLE, as the portability of the walking wearable robot is

directly affected by these factors; some users struggle to move around with this novel technology due to its bulkiness and weight.

A typical example of the bulkiness and weight of these walking robots is the Atalante, a crutch-less LLE with six degrees of freedom per leg, capable of self-supporting and self-balancing. This robot has a total weight of 75 kg due to its design decisions (Paredes & Hereid, 2020; Duburcq et al., 2021). The biomechanics of normal human gait are affected and altered by certain mechanical aspects of the existing walking wearable robot. Users of these robots experience higher metabolic costs and discomfort. The LLE design should be modular, allowing adjustments to suit a wide range of users and improving comfort for its user (Chen et al., 2016).

The necessity to utilise crutches may be regarded as a limitation for the majority of existing LLEs, owing to the restricted DOFs available on these LLEs. Lau & Mombaur (2024) indicate that crutches provide lateral stability and safety during walking; however, their use alters the natural gait of both the robot and the user. This is particularly problematic for elderly individuals, as their limited arm strength is required to hold crutches, making it difficult for them.

Schiele & van der Helm (2009) asserts that the user's comfort, functionality, and safety may be compromised due to misalignment between the user and the exoskeleton, particularly when the device is rigid. Most current LLEs are limited by the capabilities of available actuators, including factors like power, weight, and size (Kapsalyamov et al., 2019).

Researchers require high-torque, lightweight, and compact motors to advance lightweight robots. Furthermore, users find the continuous noise generated by these motors during operation uncomfortable (Chen et al., 2016).

The exorbitant price of these assistive devices renders them inaccessible to middle-class individuals, thereby limiting their availability to a small segment of the population. Mertz (2012) states that in 2012, the cost of LLEs ranged from US\$100000 to US\$130000, equivalent to R897270 to R1.166 million at the relevant exchange rate; this high cost was considered an obstacle to middle-class people accessing this technology. Murtagh (2015) provides an example where the cost of a ReWalk LLE was R1.145 million for personal purchase, and R1.359 million for institutional purchase. These values are base prices that include an annual service fee over a 5-year lifespan.

2.4.1 Strengths, weaknesses, and transferability of the Selected LLEs

Contrary to most existing LLEs, Atalante is equipped with a self-balancing feature, allowing the user to move their hand freely without crutches during ambulation (Lejeune et al., 2025). In contrast to the mentioned strength, the weight of this LLE is considered a weakness, as it may cause fatigue for its user over an extended period (Kerdran et al., 2021). This technology can

be adapted for use by elderly individuals, as it inherently eliminates the necessity for crutches through its self-balancing feature (Lejeune et al., 2025).

Esquenazi et al. (2012) notes that ReWalk has effectively demonstrated its capability to support users by allowing them to stand, walk, and ascend stairs. Nevertheless, the walking robot still relies on crutches for lateral stability, requiring users to use their hands to support their upper body weight. This dependence on crutches presents challenges for transferring this LLE to elderly populations, as their potentially weaker arms and shoulders may struggle to maintain this activity Over prolonged durations.

2.5 Human Biomechanics and Degree of Freedom (DOF)

The same method used to analyse the movements and mechanisms of the lower limbs in humans should be applied to study the mobility of LLEs. This method involves creating a rigid sequence of links that mimics human anatomy. Each inflexible link in this sequence corresponds to a specific part of the human body in terms of length. Additionally, the connecting joints between these segments possess attributes like those of human joints, including range of motion (ROM) and DOF (Innab, 2019). Figure 2.6 illustrates the kinematic model of the human, showing the joint angles of each link.

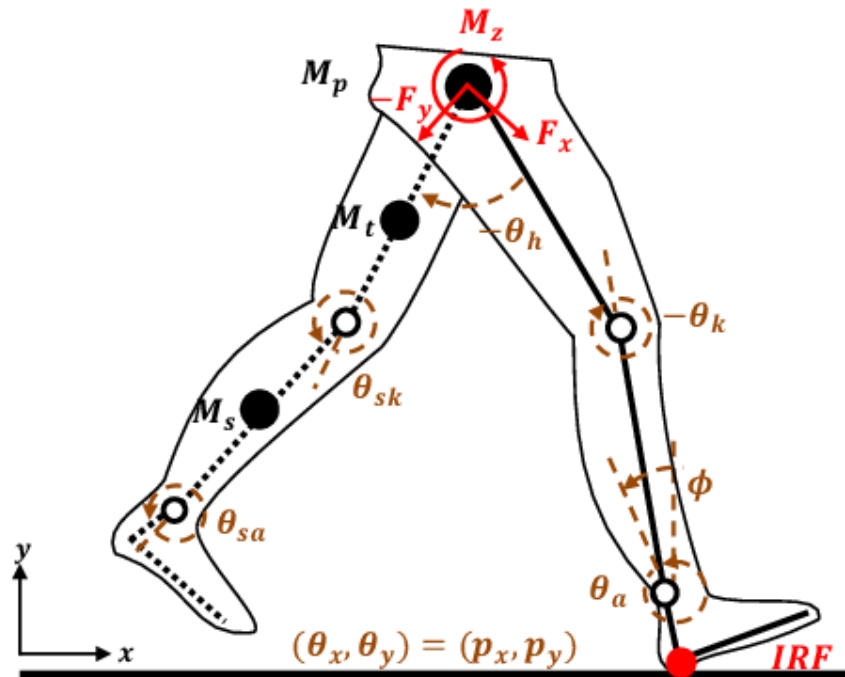


Figure 2.6: Kinematic model of the human body (Lv et al., 2020)

The standard anatomical planes are used to define the ROM and DOF of the lower limb in humans. These planes are arranged perpendicular to each other, as illustrated in Figure 2.7. The lower limb consists of six DOFs: three at the hip, one at the knee, and two at the ankle (Lerner, 2001). The three perpendicular planes that divide the human body are the coronal

plane (also known as the frontal plane), the horizontal plane (also called the transverse plane), and the sagittal plane (Innab, 2019).

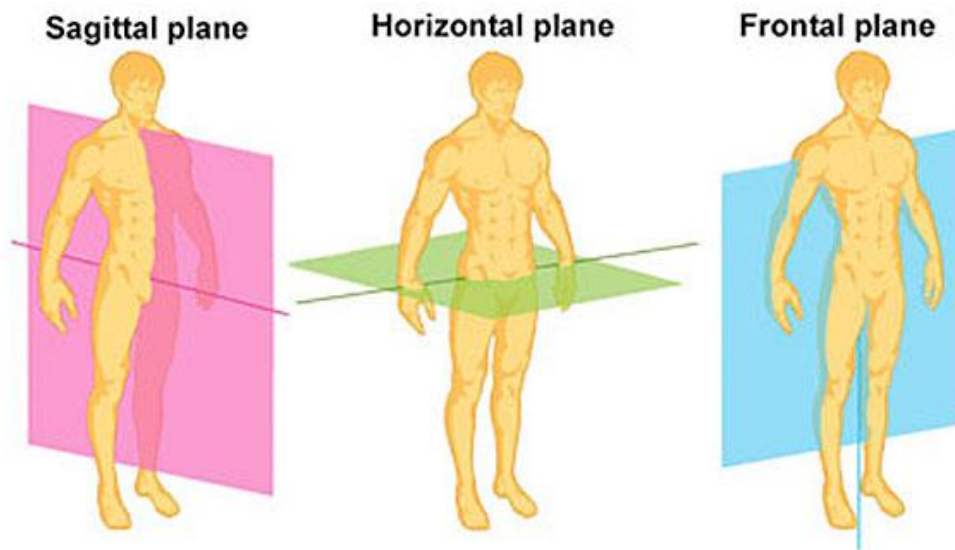


Figure 2.7: Human anatomy plane (Walden, 2019)

The human body has 3 main anatomical planes, each providing unique DOF that enable precise movements. A prime example of this is the extension of the hip joint in the sagittal plane, which propels the body forward during the landing phase. As the heel contacts the ground, the knee flexes to absorb the impact, known as the swing phase of the knee joint, enabling the foot to clear the ground. Additionally, the ankle joint's dorsiflexion and plantarflexion in the sagittal plane contribute significantly to effortless walking by transitioning the foot smoothly from the heel to the toe during the stance phase (Önen et al., 2017). Figure 2.8 illustrates the human gait cycle, showcasing the various transitions the leg undergoes to complete the walking motion in the sagittal plane.

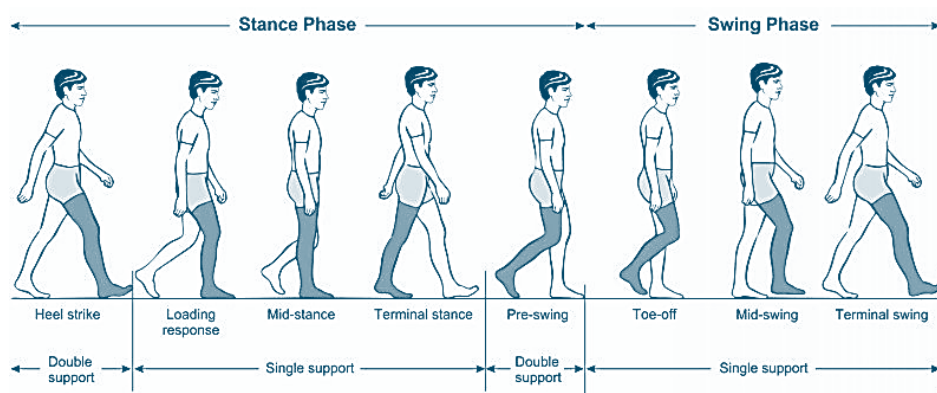


Figure 2.8: human gait walking cycle (Pirker & Katzenschlager, 2017)

In addition to the sagittal plane, the coronal plane allows for hip abduction and adduction, which help stabilise the body by maintaining the centre of gravity within the footprint. Similarly, ankle

joint abduction and adduction aid in adapting the foot to the ground. Finally, the transverse plane allows the human body to change the direction of movement by coordinating the lateral and medial rotation of the hip and ankle joints (Önen et al., 2017).

Ensuring the safety of the LLE is of critical importance and demands attention to detail regarding the range of motion of each joint in the lower limb. To guarantee safety, it is highly recommended that the analysis be based on the physiological range of motion of each joint of the human body. Moreover, to mitigate potential safety hazards, it is advised that the design range of motion of the exoskeleton be set lower than that of each human joint. This approach will ensure that the exoskeleton operates within safe limits for human use (Qiu et al., 2023). Table 2.1 provides the DOFs of each joint and their corresponding ROMs.

Table 2.1: Degrees of freedom of each joint of the lower limb (Zheng et al., 2023)

Number	Joint	Degree of freedom	Range of motion
1	Hip joint	Flexion/Extension	$-120^{\circ} \sim 65^{\circ}$
		Abduction/adduction	$-30^{\circ} \sim 40^{\circ}$
		External/Internal (Rotation)	$-60^{\circ} \sim 25^{\circ}$
2	Knee joint	Flexion/Extension	$0^{\circ} \sim 120^{\circ}$
3	Ankle joint	Flexion/Extension	$-20^{\circ} \sim 30^{\circ}$
		Abduction/adduction	$-15^{\circ} \sim 15^{\circ}$
		External/Internal (Rotation)	$-15^{\circ} \sim 15^{\circ}$

Figure 2.9 provides a graphical representation of the DOFs, enhancing the understanding of the motion patterns and behaviours described in the table.

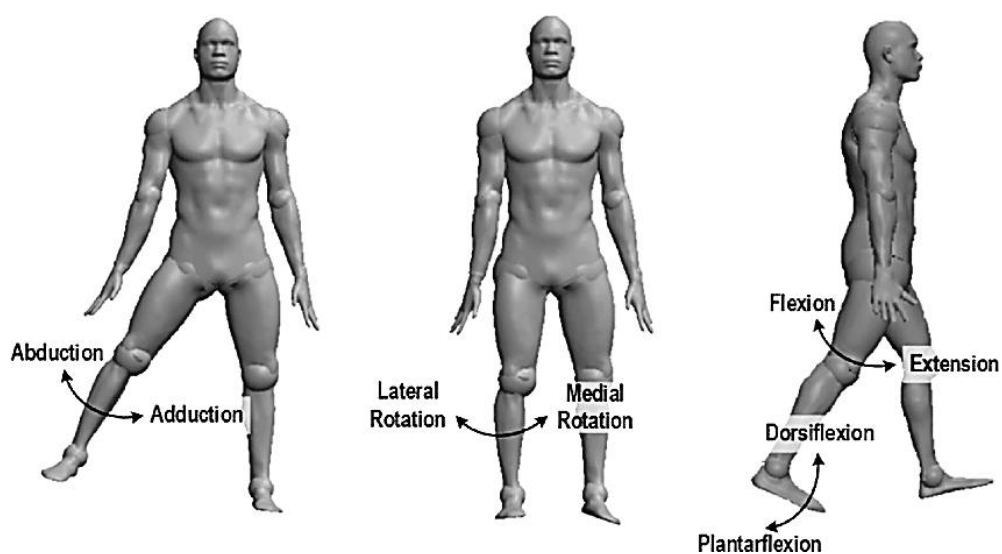


Figure 2.9: Human DOFs in standard anatomical planes (Önen et al., 2017)

2.6 Anthropometry

Exoskeletons are designed to closely resemble the shape and movements of their users. Therefore, it is crucial to analyse anthropometric data, including measurements of height, weight, volume, and centre of gravity for each body segment, along with the range of motion of the hands and feet. This data collection is essential for understanding the human body and designing exoskeletons that fit and function well. The anthropometric measurements help in designing an exoskeleton that meets the physical requirements of its user by determining their exact size, shape, and dimensions (Rosyada et al., 2019). The standard properties of the human body shown in Figure 2.10 relate to the ratio of each segment based on anthropometric data.

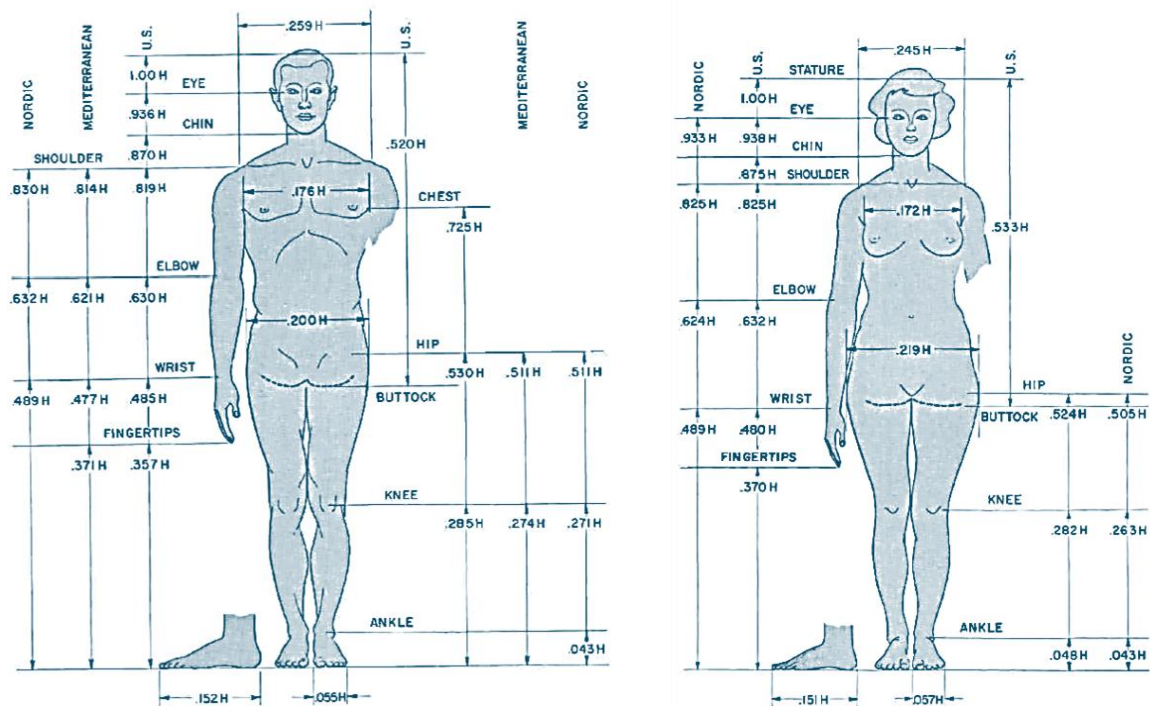


Figure 2.10: Standard properties of the human body (Contini, 1972)

2.7 Theory

This section examines the relevant theories necessary to complete this research effectively. It offers an in-depth analysis of the core philosophies of LLEs, organised into mechanical and mechatronic.

2.7.1 Mechanical considerations

Understanding the interaction between the kinematics and dynamics of an LLE helps elucidate its correlation with lower-limb movement. This can be divided into two sections, which will be discussed in subheadings 2.7.1.1 and 2.7.1.2. Furthermore, the LLE balance is influenced by several factors, such as the zero-moment point and centre of pressure, centre of mass, and the linear inverted pendulum mode, which are discussed in subsections 2.7.1.3 to 2.7.1.5, in

that order. The material selection and the stress analysis are discussed in subsections 2.7.1.6 to 2.7.1.7.

2.7.1.1 Forward and Inverse Kinematics

In mechanical engineering, kinematics is the study of motion that occurs without external influences. Understanding the relationship between forward and inverse kinematics is important when designing an LLE.

Kinematics plays a crucial role in defining the intricate relationships between a robot's joints and its end-effector, as this governs the robot's motion and positioning capabilities. By understanding a robot's kinematics, engineers can generate precise end-effector trajectories, ensuring smooth, accurate movement. Moreover, a thorough understanding of kinematics is essential to guarantee the efficiency and controllability of the robot's actuators, making it immediately relevant in practice. Figure 2.11 illustrates the relationship between forward and inverse kinematics (Minchala et al., 2017).

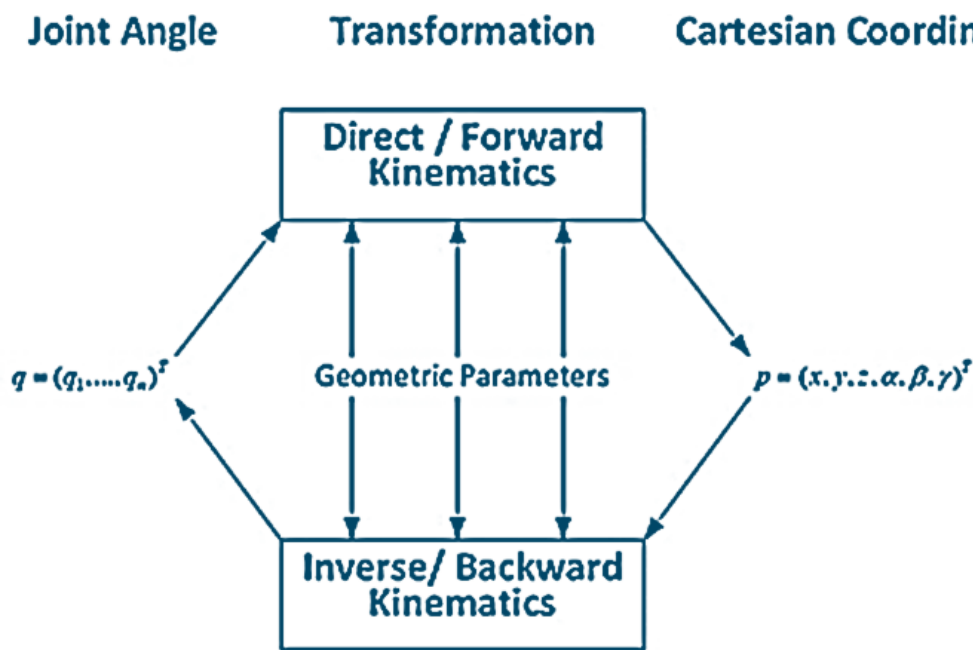


Figure 2.11: Kinematics of a mechanism (Iris Dynamics Ltd., 2023)

a. Forward Kinematics

Forward kinematics is the process of determining the end-effector's position and orientation from joint angles. This can be illustrated by Equations 2.1 to 2.3 and Figure 2.12, where the length of each link is represented by L_1, L_2 , and L_3 , while the orientation of the end-effector frame is represented by ϕ as a function of joint angles θ_1, θ_2 and θ_3 . The Cartesian position is represented by x and y (Lynch & Park, 2017).

$$x = L_1 \cos \theta_1 + L_2 \cos(\theta_1 + \theta_2) + L_3 \cos(\theta_1 + \theta_2 + \theta_3) \quad 2.1$$

$$y = L_1 \sin \theta_1 + L_2 \sin(\theta_1 + \theta_2) + L_3 \sin(\theta_1 + \theta_2 + \theta_3) \quad 2.2$$

$$\phi = \theta_1 + \theta_2 + \theta_3 \quad 2.3$$

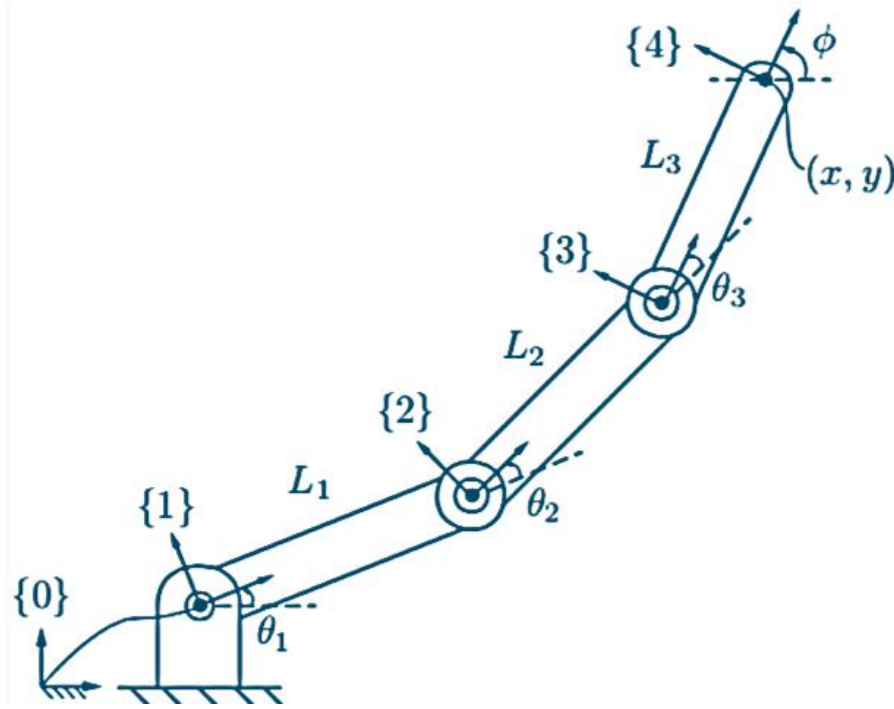


Figure 2.12: Forward kinematics of a 3R open chain (Lynch & Park, 2017)

There are various methods available for calculating the forward kinematics, including the Denavit-Hartenberg (D-H) and successive screw displacements methods. (Kruthika et al., 2016).

One of the initial steps in modelling a mechanical system is to define its kinematic model. This is particularly crucial when dealing with movable mechanisms, such as LLEs and humanoid robots. To determine the motion of a local frame of the LLE relative to the world frame, the robot's kinematic model must be established.

There are several methods for modelling the robot's kinematics, with the D-H parameters being commonly used due to their simplicity and efficiency in producing results. After defining the D-H parameters, the transformation matrices for the sequentially related local frames can be determined. This approach is used to describe both forward kinematics and inverse kinematics. On one hand, when the joint angles are known, the end-effector's position can be determined (forward kinematics); On the other hand, when the end-effector's position is known, the angles of the joints can be calculated using inverse kinematics (Sørensen et al., 2020; Sebe, 2016).

The D-H parameters define the configuration of a robot's joints and links. These parameters comprise pairs that describe the translational and rotational movements, establishing the relationship between the system's frames. Specifically, one parameter rotates about the z-axis, while another rotates about the x-axis. Additionally, one parameter translates to the z-axis, and the other translates to the x-axis. There are two forms of D-H parameters: the classic D-H and the modified D-H, which differ based on the position of the coordinate system. Equations for both classic and modified D-H parameters are provided below, with Figure 2.13 illustrating only the Classic D-H parameter description (Sørensen et al., 2020).

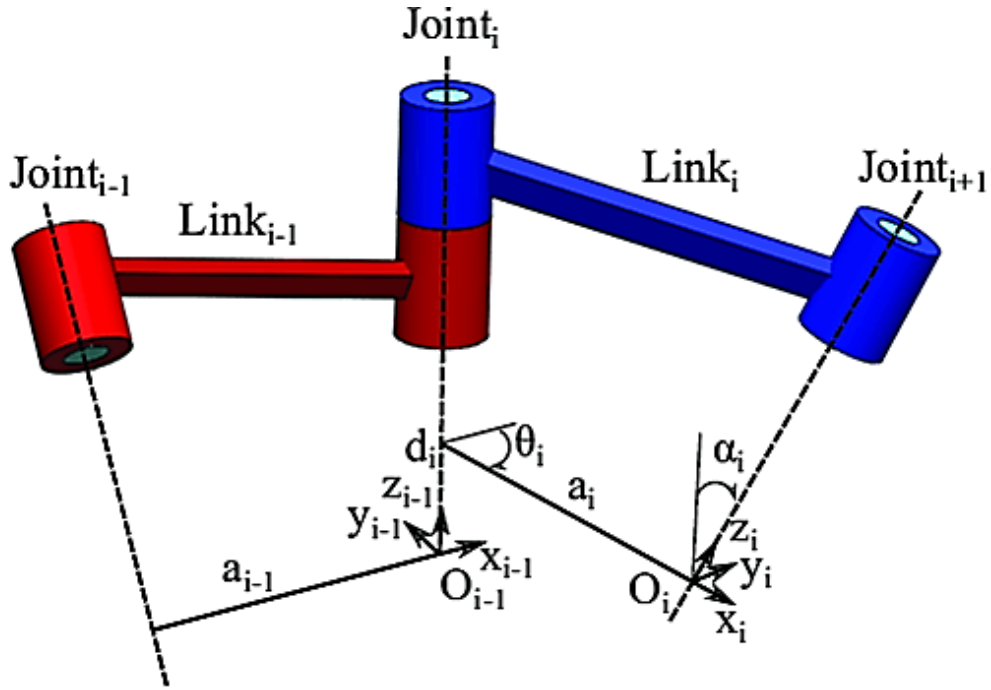


Figure 2.13: Representation of D-H parameters (Cursi et al., 2022)

Equations 2.4 and 2.5 are the transformation matrices obtained from the DH parameters, where the former is the classic transformation matrix and the latter represents the modified one.

$${}^{i-1}T_{ic} = \begin{bmatrix} \cos(\theta_i) & -\sin(\theta_i) \cos(\alpha_i) & \sin(\theta_i) \sin(\alpha_i) & a_i \cos(\theta_i) \\ \sin(\theta_i) & \cos(\theta_i) \cos(\alpha_i) & -\cos(\theta_i) \sin(\alpha_i) & a_i \sin(\theta_i) \\ 0 & \sin(\alpha_i) & \cos(\alpha_i) & d_i \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad 2.4$$

$${}^{i-1}T_{im} = \begin{bmatrix} \cos(\theta_i) & -\sin(\theta_i) & 0 & a_{i-1} \\ \sin(\theta_i) \cos(\alpha_{i-1}) & \cos(\theta_i) \cos(\alpha_{i-1}) & -\sin(\alpha_{i-1}) & -d_i \sin(\alpha_{i-1}) \\ \sin(\theta_i) \sin(\alpha_{i-1}) & \cos(\theta_i) \sin(\alpha_{i-1}) & -\cos(\alpha_{i-1}) & d_i \cos(\alpha_{i-1}) \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad 2.5$$

b. Inverse Kinematics

It is possible to calculate the joint angles using the inverse kinematics method once the end-effector's precise orientation and position are known. As illustrated in Figure 2.14, the joint angles 1 and 2 for links 1 and 2 represent as q_1 , q_2 , l_1 , and l_2 , these variables are used in determining the joint angle of each segment (Lam, 2002). Equations 2.6 and 2.7 provide the means for defining the joint angles.

$$q_1 = \frac{-(l_2 \sin(q_2)x + (l_1 + l_2 \cos(q_2))y)}{(l_2 \sin(q_2))y + (l_1 + l_2 \cos(q_2))x} \quad 2.6$$

$$q_2 = \frac{\cos(x^2 + y^2 - l_1^2 - l_2^2)}{2l_1l_2} \quad 2.7$$

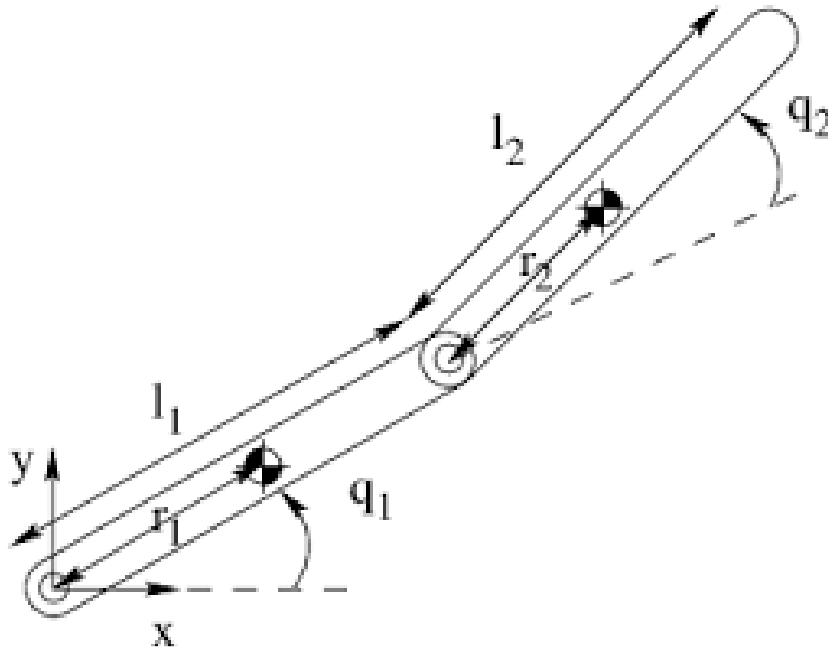


Figure 2.14: Inverse Kinematics of two links (Farooq mir & Anbarasi, 2024)

2.7.1.2 Forward and inverse dynamics

The equation of motion (EOM), also known as the dynamics model, is a critical tool for calculating the forces and torques applied to the different joints of the LLE mechanism. To determine this, there are two effective methods: the Lagrange and Newton-Euler formulations. Equation 2.8 is used in both formulations, where the joint variables are represented by a vector q , the torque by a vector u , the inertia matrix by M , the centrifugal and Coriolis term by C , and the gravity vector by g (Høifødt, 2011); According to Lynch & Park (2017), $h(\theta, \dot{\theta})$ represents the force vector that combines the effects of centripetal, Coriolis, gravity, and friction forces. The value of $h(\theta, \dot{\theta})$ depends on both θ and $\dot{\theta}$.

$$M(q)\ddot{q} + C(q, \dot{q})\dot{q} + g(q) = u \equiv \tau = M(\theta)\ddot{\theta} + h(\theta, \dot{\theta}) \quad 2.8$$

The forward dynamics deal with determining the acceleration of the robot when the state of its position (θ), velocity ($\dot{\theta}$), and the acting joint forces and torques are known; Equation 2.9 illustrates this (Lynch & Park, 2017).

$$\ddot{\theta} = M^{-1}(\theta)(\tau - h(\theta, \dot{\theta})) \quad 2.9$$

Inverse dynamics, in contrast to direct dynamics, is a method for determining the forces and torques acting on a mechanism from its state and desired acceleration. The theory of motion, as outlined in equation 2.8 (Lynch & Park, 2017), is used to calculate these values.

The Lagrange equation is used to investigate the inverse dynamics of the robot, helping to determine the torques and forces acting on a joint based on its motion. Equations 2.10 and 2.11 illustrate the Lagrange formulation, where the Lagrangian symbol is $\mathcal{L}(\theta, \dot{\theta})$, ΔE is the change in energy, $\mathcal{K}(\theta, \dot{\theta})$ is the kinetic energy, $\mathcal{P}(\theta)$ is the potential energy. In addition, θ_i is the angular displacement of joint i , $\dot{\theta}_i$ is the angular velocity of the same joint, and τ_i is the torque applied to joint i (Flynn, 2016).

$$\mathcal{L}(\theta, \dot{\theta}) = \mathcal{K}(\theta, \dot{\theta}) - \mathcal{P}(\theta) = \Delta E \quad 2.10$$

$$\tau = \frac{d}{dt} \frac{\partial \mathcal{L}}{\partial \dot{\theta}_i} - \frac{\partial \mathcal{L}}{\partial \theta_i}, \quad i = 1, \dots, n \quad 2.11$$

2.7.1.3 Zero moment point (ZMP) and Centre of pressure (COP)

Vukobratovic et al. (2001) introduced the concept of a ZMP, which refers to the point under the supporting foot at which all the reaction forces are concentrated. This concept is named after the effect produced around the defined point, where the sum of all moments of acting forces equals zero. In robotics, walking robots leverage the ZMP to replicate the dynamic walking of the human lower limb. The ZMP concentrates all reaction forces at a single point under the foot's contact with the ground, resulting in a zero net moment of gravity and inertial forces about the horizontal axis (Pamungkas et al., 2019). The ZMP location is important for ensuring stability and balance as the LLE moves. When the humanoid robot's feet remain within a specific boundary region on the ground, it prevents the robot from falling quickly. This defined region is referred to as the support polygon, and the ZMP determines both its size and location

(Yu et al., 2014). The polygon formed by the feet and the ground contact is presented in Figure 2.15.

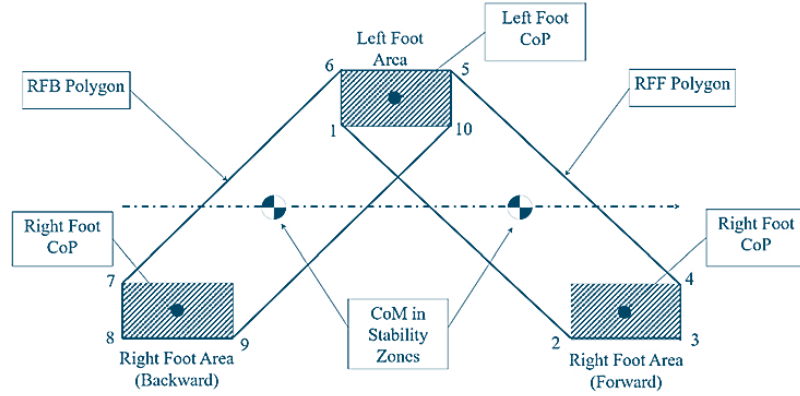


Figure 2.15: Support polygon (Ganesan & Gu, 2021)

Van Dalen (2012) presented a list of humanoid robots that utilise ZMP technology to maintain balance while standing and walking. These robots include Nao from Aldebaran Robotics, HUBO from KAIST, HRP-3 from AIST, and ASIMO from Honda. The mentioned walking robots use this method to achieve quasi-humanlike dynamic control.

ZMP criteria are necessary to determine the stability of biped robots during their motion; this is doable by using Equations 2.12 and 2.13, where $(x_{zmp}, y_{zmp}, 0)$ are the coordinates of the ZMP and $(x_i, y_i \text{ and } z_i)$ are the positions of the COM of the link i on the absolute Cartesian coordinate system. In addition, N is the number of links, I_{ix} and I_{iy} are the inertial components, $(\ddot{\Omega}_{ix} \text{ and } \ddot{\Omega}_{iy})$ are the absolute angular velocities around the x-axis and the y-axis at the centre of gravity of the link i , g is the gravitational acceleration, and m_i is the mass of the link i (Yu et al., 2014).

$$x_{zmp} = \frac{\sum_{i=1}^N m_i (\ddot{Z}_i + g) x_i - \sum_{i=1}^N m_i \ddot{x}_i z_i - \sum_{i=1}^N I_{iy} \ddot{\Omega}_{iy}}{\sum_{i=1}^N m_i (\ddot{Z}_i + g)} \quad 2.12$$

$$y_{zmp} = \frac{\sum_{i=1}^N m_i (\ddot{Z}_i + g) y_i - \sum_{i=1}^N m_i \ddot{y}_i z_i - \sum_{i=1}^N I_{ix} \ddot{\Omega}_{ix}}{\sum_{i=1}^N m_i (\ddot{Z}_i + g)} \quad 2.13$$

The COP is a concept used in biomechanics and other branches of engineering. To differentiate it from other similar concepts, Vukobratovic and his colleagues introduced the concept of the ZMP (Menga & Ghirardi, 2019; Chevallereau et al., 2008).

In certain conditions, the COP is similarly defined as that point under the foot where all the moments are zero as explained in the case of the ZMP. To ensure the stability of the walking robot, its COP must coincide with its ZMP (Li et al., 2015). LLEs are subjected to reaction forces from the ground as well as action forces from inertia, gravity, and extension. The terms

COP and ZMP describe the reaction and action forces, respectively (Sardain & Bessonnet, 2004).

As long as both the ZMP and the COP are within the support polygon, the two concepts are considered equal. However, if the ZMP is outside of the polygon area, it indicates a potential loss of stability, which can be a critical issue in robotics and biomechanics (Chevallereau et al., 2008). Since the LLE robot has only a few contact points, its COP can be calculated using equation 2.14. In this equation, the coordinates of the COP are denoted as x_{COP} and y_{COP} , the coordinates of the contact point i are x_i and y_i , and $F_{n,i}$ represents the component of the reaction force from the ground acting on the contact point i (Kim et al., 2013).

$$(x_{COP}, y_{COP}) = \left(\frac{\sum F_{n,i} x_i}{\sum F_{n,i}}, \frac{\sum F_{n,i} y_i}{\sum F_{n,i}} \right) \quad 2.14$$

2.7.1.4 Centre of mass (COM)

The COM of an LLE is a crucial factor in the dynamics of walking robots, as it defines the robot's total mass position. The COM is that specific point where the combined linear momentum of the LLE is well-defined, and where the resultant force of gravity acts (Orin et al., 2013).

During walking, the COM variable controls the robot's walking motion. It is important to note that this walking motion must adhere to the ZMP's balance constraints. These balance constraints help maintain the robot's stability and equilibrium during locomotion. Therefore, the COM variable must operate within the limitations set by the ZMP to ensure the robot's stability and balance (Carpentier et al., 2015).

The COM of a humanoid robot is calculated by finding the distance-weighted average location of each mass particle. This is achievable by employing equation 2.15, where Pm_i represents the location of a mass particle i , and M_i denotes its mass (Lam, 2002). The COM of the humanoid robot is estimated to be located around the waist, specifically in the lower torso. This estimation considers the distribution of mass within the robot's body and its overall geometry. It is important to note that the COM's position can shift as the robot moves and changes posture or gait. This dynamic nature of the COM makes it a critical consideration for understanding the robot's stability and balance during various motions and activities (Kljuno & Williams, 2010).

When a humanoid robot is walking forward in the sagittal plane, its COM tends to shift in the same direction until both feet contact the ground. This phenomenon occurs as the robot propels itself forward, and the gravitational forces acting on its COM cause it to lean in the direction of its movement until stability is achieved upon landing both feet (Sun & Roos, 2018).

$$COM = \frac{\sum P m_i M_i}{\sum M_i}$$

2. 15

2.7.1.5 Linear Inverted Pendulum Mode (LIPM)

In the planning of gait patterns, the hip joint movement is represented by the peak of the inverted pendulum model when the body's COM is at the hip position. This method forms the basis for all gait pattern planning. To ensure that the exoskeleton can replicate the walking motion of a human, a specific path must be defined in which the position of each link is known (Ren et al., 2020).

The LIPM is a useful tool for simplifying the complex dynamic model of the LLE. When the exoskeleton's motion is restricted to translation, determining its COM becomes easier. Refer to Figure 2.16 for more clarity (Yang et al., 2018; Kajita et al., 2001).

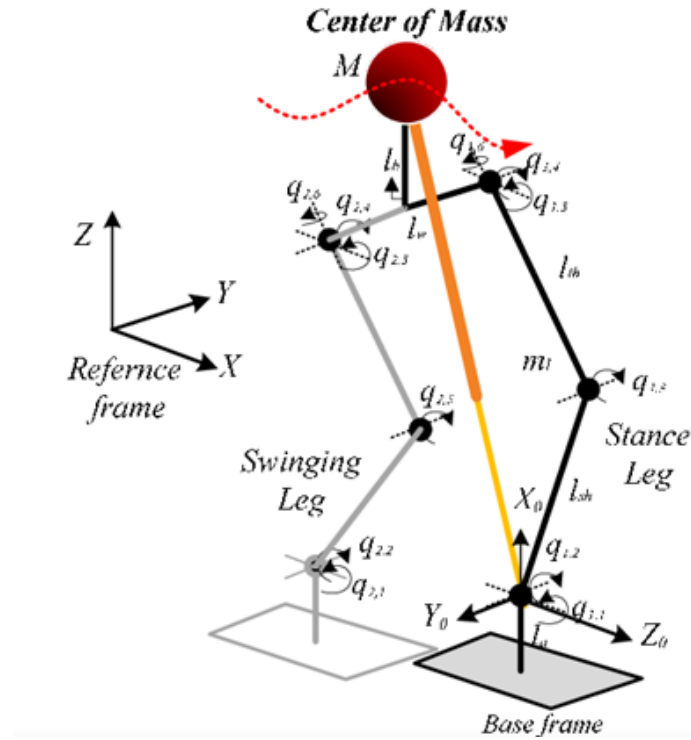


Figure 2.16: 3D Bipedal Model and Inverted Pendulum Model (Shin & Kim, 2010)

Yilmaz (2010) States that a quartic polynomial-based walking pattern generator can be used to determine the omnidirectional gait of an LLE. By leveraging the ZMP and LIPMs, this step model guarantees walking stability. This approach combines the benefits of the least-squares method and periodicity, resulting in a reduced range of fluctuations in the ZMP trajectory across different footprints. The quartic polynomial serves as the foundation for the LLE algorithm, with the desired ZMP defined accordingly.

Ensuring the stability of an LLE under both dynamic and static conditions is decisive, and the ZMP and LIPM criteria are essential. To achieve a smooth, steady gait in the exoskeleton, it is important to adopt a continuous, natural ZMP reference trajectory. When the exoskeleton is supported on one leg, the ZMP reference trajectory moves forward beneath the support foot. Defining the COM of the LLE using the predefined ZMP reference trajectory can be achieved through the Fourier series approximation (Qiao et al., 2024).

When designing a robot like the Raibert hopper, it is important to consider both the swing and stance phases. The swing phase occurs when the robot's foot is not in contact with the ground, whereas the stance phase occurs when it is. When modelling an LLE or a multi-legged robot, one approach is to treat the LLE during the mid-swing phase, while the grounded leg is fixed. In this scenario, the LLE design follows the LPIM principle. Figure 2.17 demonstrates how the intricate structure of the walking robot can be simplified to a linear inverted pendulum (Castro, 2019).

Kajita et al. (2001) developed the provided dynamics equation below based on the principle of the LIPM, where $\ddot{x}$ and $\ddot{y}$ represents the COM, and $(x - p_x)$ and $(y - p_y)$ stands for the ZMP. Equations 2.16 and 2.17 are used to calculate the COM, which is related to the ZMP.

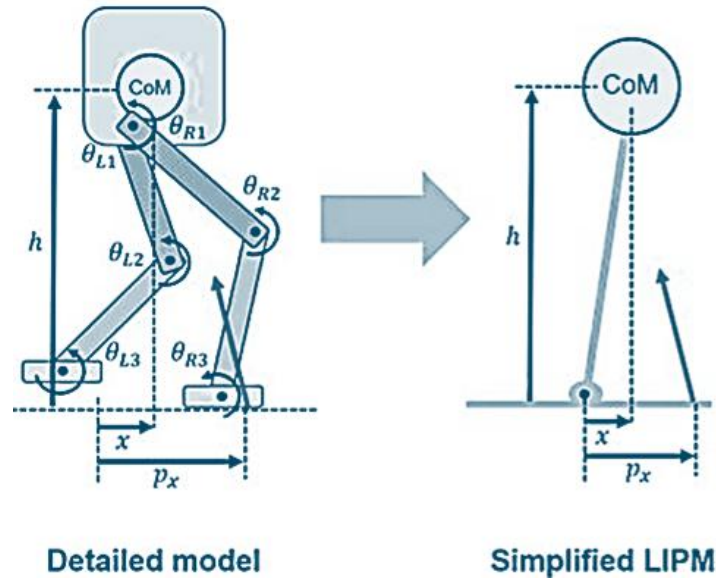


Figure 2.17: Detailed model to simplified LIPM (Castro, 2019)

$$\ddot{x} = \frac{g}{h}(x - p_x) \quad 2.16$$

$$\ddot{y} = \frac{g}{h}(y - p_y) \quad 2.17$$

2.7.1.6 Material

The selection of the frame material is a critical factor in the design of the LLE, as it enables the exoskeleton to withstand the forces and torques produced by each joint's actuator without distortion. It is important to choose a lightweight material to minimise power consumption when operating actuators. Moreover, the material must be strong enough to endure the forces and torques produced without compromising its structural stability (Raut et al., 2021).

Chittar and Barve (2022) state that alloys and metals are the preferred materials for the frames of most LLEs available on the market. This choice is driven by their inherent rigidity and strength, which are essential for the optimal performance of the exoskeletons. At present, the most utilised metals and metal alloys for constructing exoskeletons are as follows:

- Manganese alloy with grades Az91D, AM60, and AM50
- Titanium alloys with grades Ti-6Al-7Nb and Ti-6Al-4V
- Stainless steel with grades 304SS and 316LSS
- Aluminium alloys with grades AL7068, AL7075, AL7475-T7351, AL2024, and AL6061

The production of LLE frames is not limited to metallic materials and alloys. It also includes composite materials and polymers. Kevlar, carbon fibre, and fibreglass are popular choices for exoskeletons and other technical applications due to their high strength-to-density ratios (Chittar & Barve, 2022). The cost of an LLE depends on several factors, including the material used in its structure. The chosen material must be lightweight yet capable of supporting the user's weight and other attached apparatuses. Additionally, the material should possess high strength and toughness with low density (Abdullah et al., 2023).

2.7.1.7 Stress analysis of LLEs

Stress and displacement analyses of static structures are conducted when components, parts, or assemblies are subjected to specific loads. This analysis assumes that damping and inertia effects are minimal and that the load and boundary conditions will remain constant for a certain period. Such static structural analysis is applicable in a variety of software, including SolidWorks, Autodesk Inventor, ANSYS, Abaqus solver, Samcef, and Catia (Özgün, 2021).

Atmaja et al. (2017) Conducted an FEA of an LLE using an ABAQUS solver, where An Aluminium alloy was employed for this analysis. The stress analysis aimed to determine the Von Mises stress. To ensure safety, a limited load was applied to the exoskeleton during testing to guarantee comfort and safety. ANSYS software was used by Siva Suriya et al. (2008) to assess the strength and stiffness of an LLE.

2.7.2 Mechatronics

This section concentrates on mechatronics components employed in the design of LLEs. Subsection 2.7.2.1 delineates several existing actuators, while subsection 2.7.2.2 addresses sensors. Subsection 2.7.2.3 is dedicated to control systems, whereby the Walking controller and gait cycle are examined under subsection a, and trajectory generation is discussed under subsection b.

2.7.2.1 Actuators

It is essential to note that an LLE can operate independently as a mechanical device, without requiring an external power source. However, the addition of an external power source, such as an electric motor, pneumatic pump or hydraulic pump, activates the mechanism. Furthermore, a quasi-passive system exists, where the robot functions as both a passive and an active system, as described by (Nacy et al., 2016).

An actuator is a device that converts input energy into motion to perform work for a mechanism. Hydraulic, pneumatic, and electrical actuators are preferred alternatives for existing LLEs. Although electric actuators are more efficient than hydraulic and pneumatic ones, they are still heavier (Ghaddar & Mohammad, 2019).

To ensure that the selected actuators are suitable for their intended purpose and can effectively support the user's movements, it is crucial to conduct a thorough evaluation of the torque and power requirements needed for both the exoskeleton and the user's lower limb. This evaluation process serves to ascertain the suitability of the selected actuators, as it delineates the capability of the actuator to fulfil the intended task (Bortole, 2013).

The Series Elastic Actuator (SEA) is a novel actuation system that has attracted the attention of engineers due to its numerous benefits. These advantages include low output impedance, reliable amplified force, improved safety, and high back-driveability. Additionally, the SEA can store energy in its spring, thereby reducing power consumption. These qualities make SEA an attractive option for a range of engineering applications. The conducted study highlights SEA's potential as an energy-efficient actuation system (Wang et al., 2014).

When choosing actuators for LLEs, it is important to be cautious, as their selection can affect performance. Electric actuators are a preferred choice because they offer several advantages, such as higher bandwidth, simpler joint control, and better construction when compared to pneumatic and hydraulic actuators (Westervelt et al., 2007).

A successful LLE is a torque-based system that requires more torque than speed to function effectively. To achieve this, harmonic drives are used to reduce the speed of the electric DC motor while increasing its torque. Selecting a flat electric motor and a flat strain wave gear, or a harmonic drive, is important for creating a compact, lightweight walking robot. Harmonic

drives are more advantageous than traditional gear systems because they offer high torque, no backlash, high gear ratios, high resolution, excellent repeatability, and coaxial input and output shafts. These features make the selected harmonic drives highly suitable for repositioning inertial loads (Bortole, 2013).

2.7.2.2 Sensors

Sensors are a fundamental component of robotics, as they provide real-time information about the system's condition to its controller. A control system typically comprises three interconnected layers: perception, decision, and execution. In the exoskeleton controller, the perception layer relies on multiple sensor data, including pressure, biological signal, and angle sensors.

These sensors work seamlessly together to accurately determine the system's condition. The pressure sensor regulates the robot's gait and the user's movements, the mass variation, and the interaction of forces between the user's lower limb and the exoskeleton. The angle sensor comprises encoders, inertial measurement units, a gyroscope, angular acceleration sensors, and an inclinometer (Li et al., 2015).

To effectively control the LLE, it must be highly responsive to its environment. This responsiveness is achieved through sensors. There are two main categories of sensors: interactive devices and navigation devices. Interactive sensors enable precise control of the LLE, while navigation sensors guide its movements (Zegloul et al., 2016).

The number of sensors used in an LLE can vary depending on the design. It is important to note that, for safety reasons, most LLEs have multiple sensors. However, it is necessary to be cautious when selecting these sensors because that decision directly affects how quickly the robot can respond to its environment (Li et al., 2015). There are three main types of data related to human biomechanics and motion. These are kinetic, bioelectric, and kinematic data. Kinetic data concerns the forces and torques acting on joints and links. Bioelectric data is gathered during human motion and includes signals such as electroencephalography (EEG), electromyography (EMG), electrocardiography (ECG), and electrooculography (EOG). Finally, kinematic data is based on body posture and joint angles produced during motion (Chen et al., 2016).

An LLE comprises various types of sensors, including pressure and angle sensors, each serving a specific purpose. Angle sensors, including encoder inclinometers, angular acceleration sensors, and gyroscopes, are mounted at the exoskeleton joints to monitor the expected angle. These sensors determine the angle at which the user and the exoskeleton move together. On the other hand, pressure sensors measure the combined weight of the user and the exoskeleton on the ground and send the data to the control system. The controller then uses data from both sensor types to compute the ZMP position (Li et al., 2015). The

Lokomat LLE is equipped with force and position sensors that work seamlessly to ensure accurate and efficient support to the user. The force sensors assist in regulating the interaction between the exoskeleton and the user, whereas the position sensor accurately measures joint angles, thereby enabling the exoskeleton to deliver the requisite assistance (De Rossi et al., 2010).

Bortole (2013) incorporated a strain gauge into each link of an LLE. These gauges were added as force sensors to quantify the torque generated by the interaction between the user's lower limbs and the exoskeleton. To connect these sensors, a full Wheatstone bridge configuration was utilised, which is an electrical circuit widely employed for measuring unknown electrical resistance. Figure 2.18 illustrates a strain gauge and POT operating as a force sensor.

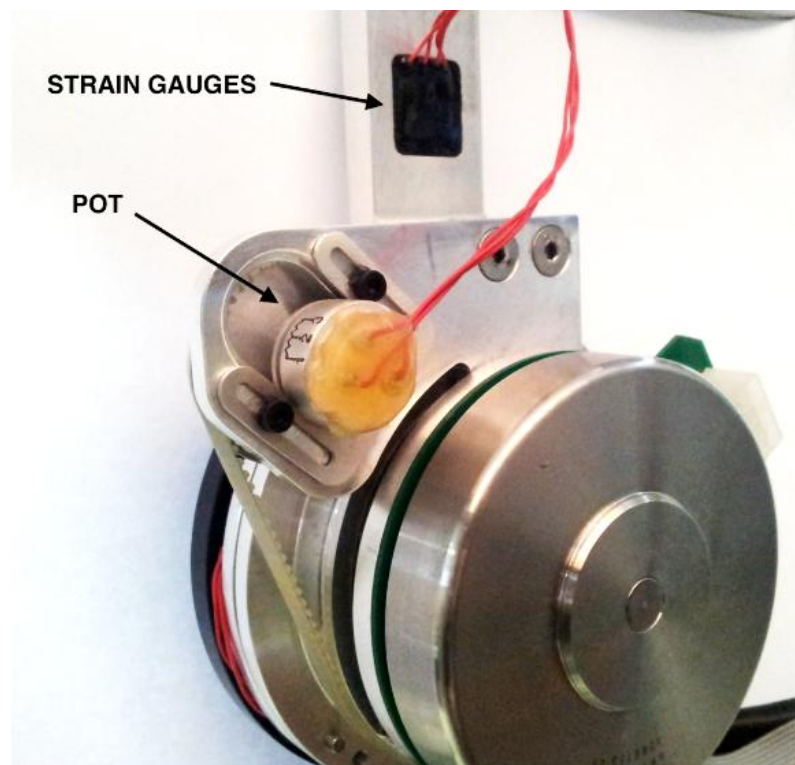


Figure 2.18: Position and interaction torque sensors in the hip joint (Bortole, 2013)

2.7.2.3 Control system

The control system of the LLE serves as the central processing unit, where data is received and processed to ensure optimal balance, stability, motion, safety, and sensibility of the walking robot (Bosscher et al., 2016).

The control system of the LLE comprises three distinct layers as defined previously under the sensors subheading. When utilising an LLE to aid with movement, it is essential to implement a force/torque control system. This control system plays a fundamental role in regulating the exoskeleton's motion while considering the applied force. This is particularly crucial for effective exoskeleton operation (Li et al., 2015).

The exoskeleton is equipped with sensors that collect valuable data, which is then transmitted to the controller. The controller then provides precise, coordinated instructions to the actuators, enabling smooth, efficient movement. The force/torque control system plays an important role in ensuring the successful operation of the device, as it facilitates the careful orchestration of the movements of the LLE (Chen et al., 2020).

a. Walking controller and gait cycle

It has been observed that by utilising classical control theory and drawing on the reference trajectory, also known as the gait cycle, the walking controller for an LLE can be effectively obtained (Thakur et al., 2017). The walking process of an LLE should be like that of humans. It involves two phases: the single-supporting phase and the dual-supporting phase. In the single-supporting phase, one foot remains still while the other swings forward and backwards. In the dual-support phase, both feet of the exoskeleton support the robot and its user. During this phase, the heel of the forward foot remains in contact with the ground, while the toe of the rear foot is lifted off the ground (Huang et al., 2001; Viswanadh et al., 2019).

The gait cycle of a human being comprises two distinct phases, namely stance and swing, which are key components of the walking cycle. The stance phase accounts for most of the walking cycle, constituting 60% of it, while the swing phase occupies the remaining 40%. Figure 2.19 presents an illustration of the gait cycle of a healthy human being (Vaughan et al., 1999).

Di Gregorio & Vocenas (2021) describe the gait cycle as being divided into eight subphases, with their timing expressed as a percentage of the complete walking gait cycle. The stance phase consists of four subphases: 10% for the loading response, 20% for midstance, 20% for terminal stance, and 10% for pre-swing. In contrast, the swing phase comprises three subphases: 13% for the initial swing, 14% for the midswing, and 13% for the terminal swing.

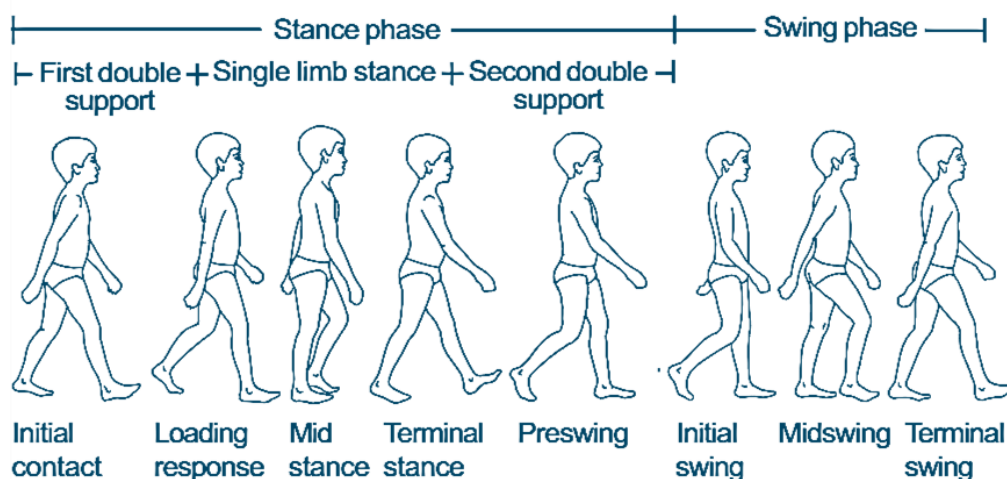


Figure 2.19: Human gait cycle (Vaughan et al., 1999)

b. Trajectory generation and path planning

In robotics, implementing a walking controller is important to ensure a smooth, collision-free motion of an LLE. To accomplish this, advanced algorithms are employed to manage the walking process effectively (Strandberg, 2004).

Wee (2014) outlines three distinct approaches for generating a suitable path for the robot. These approaches include rhythm-based, capturing human movement, or a combination of both. The computer-generated trajectory can be produced by any one of these methods.

When a walking robot moves, its path can be influenced by factors such as the distance between its footsteps, variations in its amplitude, and its ability to keep track of time. The trajectory of a walking robot is determined by several critical elements, including inverse dynamics, forward dynamics, and impedance control (Li et al., 2015). By adding a trajectory generator to the control system, movements of the LLE can be pre-defined to ensure collision-free motion (Zhao, 2016).

2.7.2.4 Power supply

Exoskeleton technology is currently facing several challenges, including identifying a suitable power source to actuate each joint. While portable batteries are available in the market, their size, safety concerns, and limited running time remain significant obstacles (Alshatti, 2019). It is noteworthy that the electric exoskeletons rely on electrical batteries to operate. Lithium-ion batteries are the preferred choice for most existing exoskeletons due to their recognised safety and widespread availability. These batteries can power LLEs for up to five hours of continuous use. However, the desire to increase power supply capacity necessitates the use of multiple or larger batteries, which would significantly increase the weight of the exoskeleton and subsequently impact its overall functionality (Young & Ferris, 2017).

2.8 Conclusion

In Chapter 2, all relevant information necessary for the successful implementation of this research centred on the LLE was gathered. This included reviewing the current literature, analysing existing technologies, conducting a gap analysis, compiling design specifications, and exploring the scientific principles underlying LLE.

CHAPTER 3: METHODOLOGY OF LLE 3D MODELING

3.1 Introduction

To ensure the success of this research, six essential points were identified: literature reviews, design specifications, 3D modelling, FEA simulation, walking controller simulation, and the construction of a scaled-down LLE prototype. These elements are visually illustrated in Figure 3.1. The implementation of this methodology ensured a research process which played a pivotal role in the research's success.

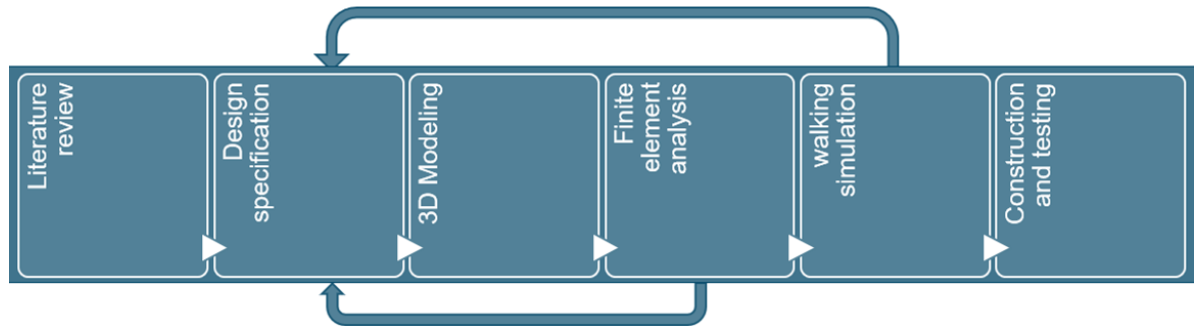


Figure 3.1: Schematic diagram of research methodology

3.2 Review of previous work

Thorough examinations were conducted on the current LLEs to identify their advantages and limitations. The literature review highlighted the various challenges these LLEs still face, which helped to define the problem statement of this research. Once the problem to be addressed was identified, research objectives were established to answer the research questions posed.

3.3 Design specifications

3.3.1 Frame specifications

The LLE has been designed with an anthropomorphic-centred approach to emulate the structure and movements of human legs. This approach ensures that the robot's frame and functionality align with the anatomy and kinematics of the human lower limbs, thereby enhancing its performance and usability. It is imperative to consider the user's potential muscular weakness during the design phase of the proposed exoskeleton, as these impairments can affect balance, stability, and metabolic expenditure. Therefore, incorporating these factors during the design process guarantees the effectiveness and usability of the final exoskeleton.

The LLE was designed to accommodate individuals of varying sizes, including both males and females. It was tailored to meet the 5th, 50th, and 95th percentile measurements, as presented in Table 3.1, which shows the percentile values for both women and men. These values served as a reference for defining the LLE's height constraints.

Table 3.1: The percentile male and female statures (Pheasant, 1996)

Percentile	Stature of female (cm)	Stature of male (cm)
5 th	150,5	162,5
50 th	161,0	174,0
95 th	171.0	185,5

To ascertain the weight-bearing capacity of the LLE, a thorough analysis was performed using the 5th, 50th, and 95th percentiles from the data collected by Stephen Pheasant. The results suggested that the exoskeleton should be designed to accommodate users weighing up to 81 kg. The weight limit was established based on the 5th and 95th percentiles for women, as this also encompasses the 5th and 50th percentiles related to males. For additional clarification, please refer to Table 3.2, provided.

Table 3.2: The percentile male and female weights (Pheasant, 1996)

Percentile	Weight of female (Kg)	Weight of man (Kg)
5th	44	55
50th	63	75
95th	81	94

The LLE's design features adjustable height, ensuring accessibility for individuals of varying heights. It has been specifically designed to accommodate users with heights ranging from 150.5 cm to 174.0 cm and to support weights of up to 81 kg. This design accounted for the 5th, 50th, and 95th percentiles of the female population, as well as the 5th and 50th percentiles of the male population, to facilitate its usability by a diverse range of individual.

The dimensions of the lower limb, including its height and circumference, were measured based on the anthropometric data of the human model depicted in Figure 2.4 under the literature review. The waist circumference, hip height, knee height, ankle height, and foot length measurements were determined using $y_i = x_i * s$. In the formula used, the ratio of each part of the lower limb is represented by x_i ; the height of the user is defined by s ; and the size of each part of the lower limb is represented by y_i . Additionally, the weight of each segment was determined using the percentage assigned to that segment. x_{if} , F_h , x_{im} , and M_h stand respectively for ratios of each part of the lower limb of the female, female stature, ratios of each part of the lower limb of the female, and male stature. Table 3.3 provides the measurements of the lower limb.

Table 3.3: Measurements of the lower limb

	x_{if}	F_s	x_{im}	M_s	$y_{if} = x_{if} * F_s$	$y_{im} = x_{im} * M_s$
Hip height (H_h)	0,524	150,5	0,53	174,0	79,7528	92,22
Knee height (K_h)	0,282		0,285		42,9204	49,59
Ankle height (A_h)	0,048		0,043		7,3056	7,482
Waist circumference	0,219		0,200		33,3318	34,8
Foot length	0,151		0,152		22,9822	26,448

The distance between the hip joint and the knee, known as the thigh, was determined by subtracting the knee height from the hip height. By subtracting the ankle height from the knee height, the shank length was calculated, defined as the distance between the knee joint and the ankle. The results are presented in Table 3.4.

Table 3.4: Thigh and shank lengths

Thigh length (T_l) = $H_h - K_h$ in mm	Shank length (S_l) = $K_h - A_h$ in mm
426.3 = 922.2 - 495.9	412.38 = 495.9 - 83.52

Table 3.5 provides a detailed analysis of the distribution of weight across the segments of the lower limb of the human body. The obtained weights of 11.398 kg, 5.189 kg, and 1.387 kg correspond to the thigh, shank, and foot, respectively, when the 81 kg user, combined with the 16 kg of the batteries, was used to determine the weight of each segment as described. These defined weights were important because they were used to determine torque per joint.

Table 3.5: Weight of lower limb segment (Plagenhoef et al., 1983)

Segment	Fraction * Mass of user
Foot	0.0143*M
Shank	0.0535*M
Thigh	0.1175*M
Pelvis	0.1596*M

3.4 Preliminary actuator specification

To identify the appropriate motor and harmonic drive for each relevant joint, torque calculations for the related joints were conducted by analysing the moment at each joint. Once the torques were determined, these values were used to preselect actuators for each joint. The subsequent sections, 3.4.1 to 3.4.4, will provide detailed information on these calculations.

3.4.1 Hip torque

The moment of point Ph was measured to determine the hip torque of the exoskeleton in conjunction with its user. In Figure 3.2, an FBD of the lower limb is presented, where τ_h represents the hip torque calculated using Equation 3.1. Note that the detailed calculations for the hip, knee, and ankle torques are also provided in Appendix H.

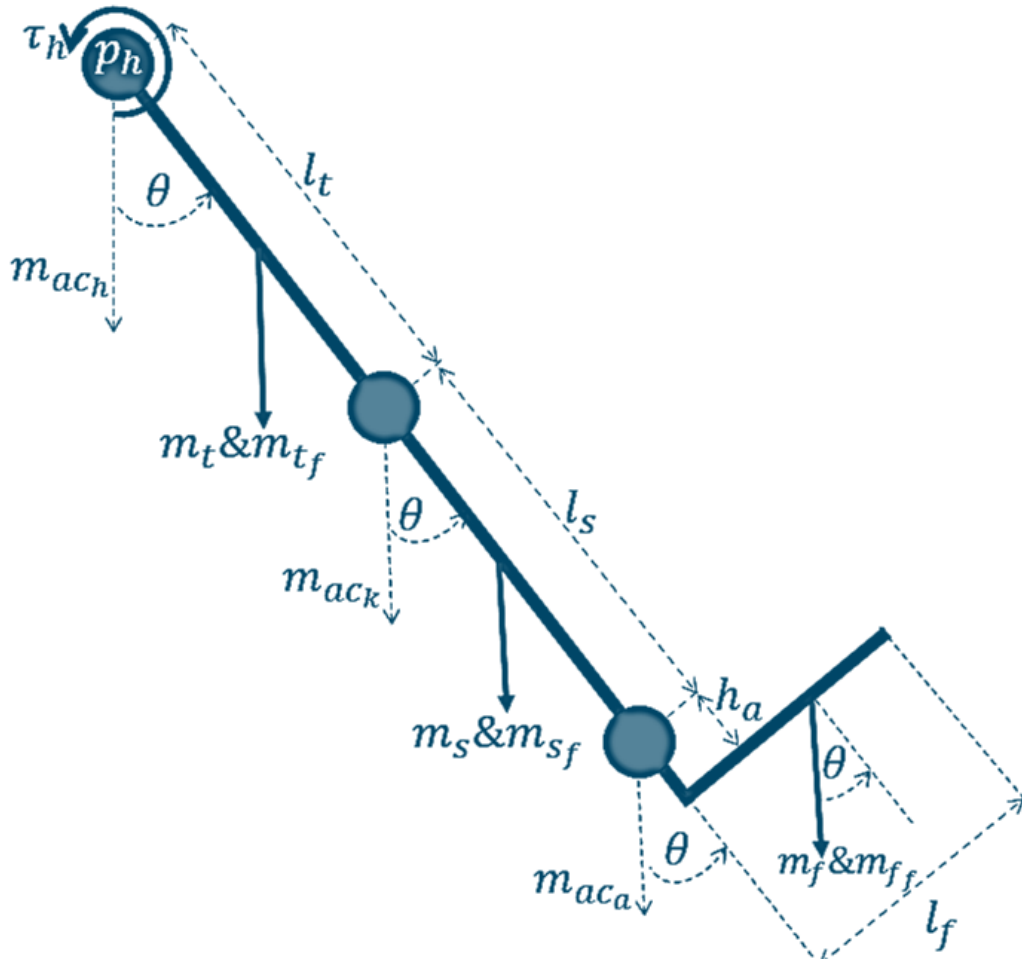


Figure 3.2: Free body diagram for hip torque calculation (Latif Shaari et al., 2015)

The variables are defined as follows: m_{ac_h} is the mass of the hip actuator, m_t is the mass of the user's thigh, m_{t_f} is the mass of the LLE thigh, m_{ac_k} is the mass of the knee actuator, m_s is the mass of the user's shank, m_{s_f} is the mass of the LLE shank, m_{ac_a} is the mass of the ankle actuator, m_f is the mass of the user's foot, m_{f_f} is the mass of the LLE foot, l_t is the length of

the thigh, l_s is the length of the shank, h_a is the height of the ankle, and l_f is the length of the foot. Lastly, θ signifies the maximum angle that the hip is designed to reach.

To accurately calculate torque at the hip, knee, and ankle joints, all relevant masses were included, including those of the frame segments, the human lower limb segments, and the actuators. In addition to the user's weight of 81 kg, an extra 29.338 kg was added, representing the combined weight of the frame, actuators, batteries, and control system, which was considered in the calculations.

Below is a list of parameters that details the essential numerical values necessary for the calculations:

- m_t (mass of thigh) = 11.397 Kg
- m_{tf} (mass of thigh frame) = 0.709 Kg
- m_s (mass of shank) = 5.189 Kg
- m_{sf} (mass of shank frame) = 0.645 Kg
- m_f (mass of foot) = 1,387 Kg
- m_{ff} (mass of foot frame) = 0.208 Kg
- $m_{ac.h}$ (mass of hip actuator) = 1.584 kg
- $m_{ac.k}$ (mass of knee actuator) = 1.584 kg
- $m_{ac.a}$ (mass of ankle actuator) = 1.03 kg
- l_t (length of thigh link) = 0.4263 m
- l_s (length of shank link) = 0.41238 m
- l_f (length of foot link) = 0.26448 m
- h_a (ankle height) = 0.0852 m
- g is the gravitational acceleration = 9.81 m/s^2
- θ is the angle of inclination of the lower limbs while walking = 33°

$$\tau_h = g * \sin(\theta) * \left[l_t * \left(\frac{1}{2} * (m_t + m_{tf}) \right) + (l_t + l_s/2) * (m_s + m_{sf}) + (l_t + l_f) * (m_{ac.a}) \right] + g * (m_f + m_{ff}) * [(l_t + l_s + h_a) * \sin(\theta) + l_t/2]$$

3.1

The calculated hip torque of 51.321 Nm was determined to be sufficient for generating the necessary force to effectively facilitate the movement of both the robotic lower limb and the user interacting with it. This capability ensured that the robotic system could perform its intended functions, providing movement and support as needed.

3.4.2 Knee torque

In reference to the FBD of the lower leg illustrated in Figure 3.3, torque τ_k was calculated by taking the moment of point pk . The resulting equation, denoted as Equation 3.2, was utilised to quantify the torque required to facilitate movement of the shank and foot.

$$\begin{aligned} \tau_k = & g * l_s * \sin(\theta) * \left[1/2 * (m_s + m_{sf}) + m_{aca} \right] \\ & + \left[g * (m_f + m_{ff}) * ((l_s + h_a) * \sin(\theta) + l_t/2 * \cos(\theta)) \right] \end{aligned} \quad \mathbf{3.2}$$

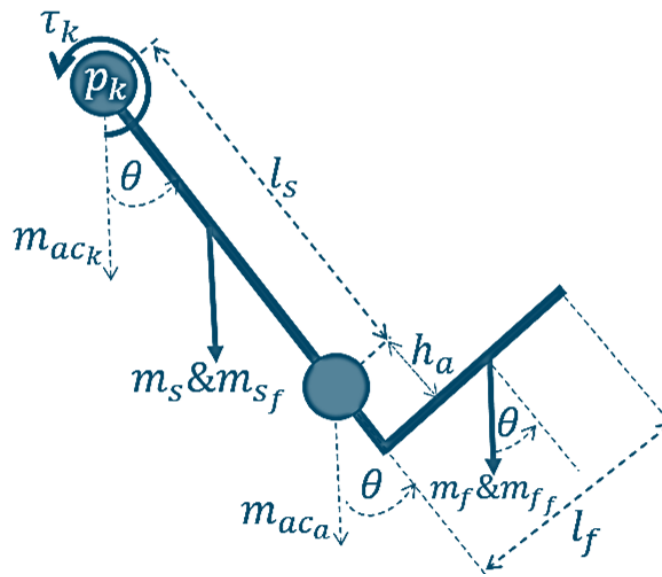


Figure 3.3: FBD for knee torque calculation (Latif Shaari et al., 2015)

A torque of 14.659 Nm was obtained using Equation 3.2. This calculated torque represents the minimum torque needed to produce flexion and extension movements of the knee.

3.4.3 Ankle torque

The torque required to rotate the foot, encompassing both the LLE foot and the user's foot, was calculated by taking moments of the pitch ankle joint, denoted as pa , as illustrated in Figure 3.4. This process led to the formulation of Equation 3.3. Subsequently, the numerical values were substituted into the equation to obtain an ankle torque of 2.447 Nm.

$$\tau_a = g * (m_f + m_{f_f}) * [h_a * \sin(\theta) + l_t/2 * \cos(\theta)] \quad \mathbf{3.3}$$

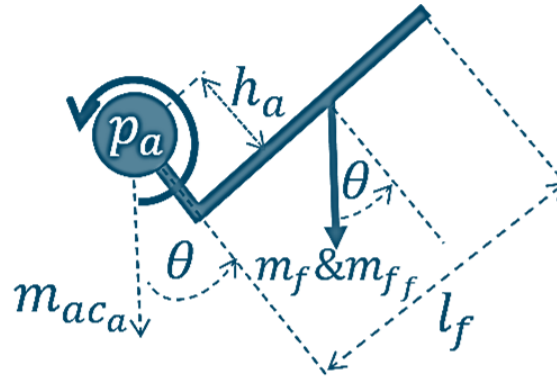


Figure 3.4: FBD for ankle torque calculation (Latif Shaari et al., 2015)

3.4.4 Gravity compensation torque for the knee and ankle joints

After calculating the torques required to move the LLE joints, it was necessary to determine the torque needed to support the lower body's weight. As shown in Figure 3.5, the thigh bears the weight of the lower body, including the lower torso, pelvis, and right leg, represented by the point load M_{wt} on p_h . The combined weight of the left thigh of both the user and LLE must also be considered in the calculation, excluding the left lower leg, as it is behind the point p_k . Equation 3.4 was used to calculate the torque required to support the thigh segment under applied loads, yielding a torque of 58.5167 Nm.

$$\tau_{k1} = g * \sin(15) * \left[(M_{wt} * l_t) + \left(m_{t+tf} * \frac{l_t}{2} \right) \right] \quad 3.4$$

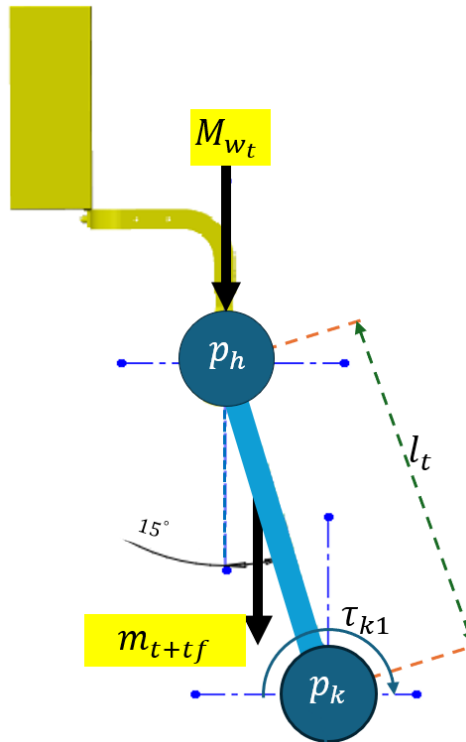


Figure 3.5: FBD for the knee supporting torque

In Equation 3.5, the torque of 69.024 Nm was calculated to prevent movement of the shank segment under the influence of the gravitational load, determined by accounting for the weight of the user's lower body and the LLE; the weight of the foot in stance was excluded. This torque was calculated by taking moments about p_a , as illustrated in Figure 3.6. In this torque calculation, the combined weights of the user's lower body and the left leg were considered, excluding the weight of the left foot since it is behind the point of rotation p_a .

$$\tau_{a1} = g * \sin(15) * \left[(M_{w_s} * l_s) + \left(m_{s+sf} * \frac{l_s}{2} \right) \right] \quad 3.5$$

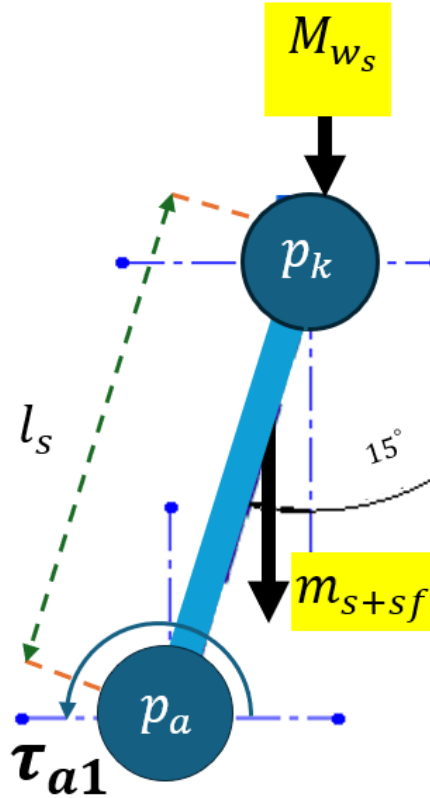


Figure 3.6: FBD for the ankle supporting torque

3.4.5 Actuator selection (Motor and harmonic drive)

The researcher selected electric motors for their numerous advantages. According to Chen et al. (2019), electric motors feature a high power-to-weight ratio, are engineered with precision, and exhibit remarkable reliability. They can accommodate multiple degrees of freedom across the three planes of motion, including transverse, frontal, and sagittal planes, and are easy to control and install. Unlike hydraulic and pneumatic actuators, electric motors do not face problems with fluid leakage (Aliman et al., 2024).

Maxon Group is a Swiss company specialising in high-precision electric drive systems. For over six decades, Maxon Group has served industries like automation, robotics, medical technology, and aerospace, where its products have proven dependable (Maxon, 2021).

The Maxon brushless motors, often referred to as electronically commutated (EC) Maxon motors, were chosen for their impressive torque and high power. Maxon (2019) states that their brushless DC motors deliver a broad speed range and are engineered for longevity.

The hip joint exhibited the highest torque during the robot's walking motion, followed by the knee joint, while the ankle joint recorded the lowest torque, as calculated in subsections 3.4.1 to 3.4.3. These figures were crucial for selecting the appropriate actuator for each joint. Furthermore, specific power (power-to-weight ratio) should be considered alongside factors such as size, power, and portability.

3.4.5.1 Hip joint actuator

For the hip joint application, a Maxon EC 90 flat motor was selected, featuring a rated power of 220 watts, as illustrated in Figure 3.7. This motor has a diameter of 90 mm and a height of 27.4 mm, excluding the shaft height. With these dimensions, the selected motor falls into the category of a compact and portable actuator, weighing 624 g. The motor boasts a specific power of 352.564 W/kg and a nominal torque of 0.729 Nm, making it a suitable choice for the hip joint of the LLE, given the calculated torque requirement of 51.321 Nm. Given that the nominal torque of the Maxon motor alone is insufficient to meet the computed torque, a harmonic drive gear reduction from Zhejiang Laifual Drive was added to the Maxon EC 90 motor to achieve the desired torque for the hip joint.



Figure 3.7: EC 90 flat Ø90 mm, brushless, 220 Watt (Maxon Group, 2012)

Laifual Drive is a Chinese company specialised in manufacturing planetary gears and precision-engineered harmonic speed reducers (Laifual, 2023). From the company's catalogue, the Laifual LSG-20-160-U-1 model was selected, as depicted in Figure 3.8, due to its notable features, including a reduction ratio of 160:1, a compact and lightweight design, zero backlash, and excellent positioning and rotational accuracy.



Figure 3.8: LSG High Torque Harmonic Drive (Laifual, n.d.)

With the harmonic drive gear reduction and the selected motor's nominal torque, the resulting torque was 116.64 Nm, exceeding the calculated torque, indicating that the actuator is appropriate for the task. The hardware datasheets for the chosen motors and harmonic drives are included in Appendix G.

3.4.5.2 Knee joint actuator

A 220-watt Maxon EC 90 flat motor was selected for the knee joint of the LLE. This motor shares many features with the one used for the hip joint, but its nominal torque is 0.715 Nm. By integrating the Laifual LSG-20-160-I-1 harmonic drive with the chosen motor, a torque of 114.4 Nm was achieved, making this combination ideal for the knee joint, as both knee flexion-extension torque and gravity compensation torque remain within the torque limits of the knee actuator.

3.4.5.3 Ankle joint actuator

The ankle actuator was chosen based on the torque requirements for this joint. According to calculations, the ankle necessitated the lowest walking torque among all joints; however, its gravity-compensation torque was determined to be a substantial value, as detailed in subsection 3.4.4. For this reason, the Maxon EC 60 flat motor in Figure 3.9, with a rated power of 200 watts, was chosen for the ankle joint. The motor has a standard torque of 0.577 Nm and weighs 360 g. Its compact design makes it ideal for the ankle joint, with a 60 mm diameter and a 45.5 mm height, excluding the shaft. This motor offers a specific power of 555,555 W/kg. Additionally, a Laifual LSG-17-120-U-I was selected for this application, providing a final torque of 69.24 Nm.



Figure 3.9: EC 60 flat Ø60 mm, brushless, 200 Watt (Maxon Group, 2012)

Appendix G presents the specifications and design details for the selected Maxon motors and Laifual harmonic drive used in each joint mechanism. This section provides insights by offering detailed information on the functionality and effective performance of the actuator components.

3.5 Conceptual designs of LLE

In this section, two concepts were examined, with one ultimately chosen as the final concept due to its advantages over the other. The first concept is presented under the first heading, followed by the coverage of the second concept in the subsequent heading. This comparative analysis facilitated decision-making about which concept to select.

3.5.1 Concept 1: LLE with Motor and an epicyclic gear system

Concept 1, as shown in Figure 3.10, was constructed from carbon fibre. The actuation system of this concept consisted of a Maxon motor paired with an epicyclic gear system. The designed gear reduction provides a 56:1 ratio. Figure 1 shows that the LLE frame weight in Concept 1 is 16.93 kg, excluding the Maxon motors, battery, and control system.

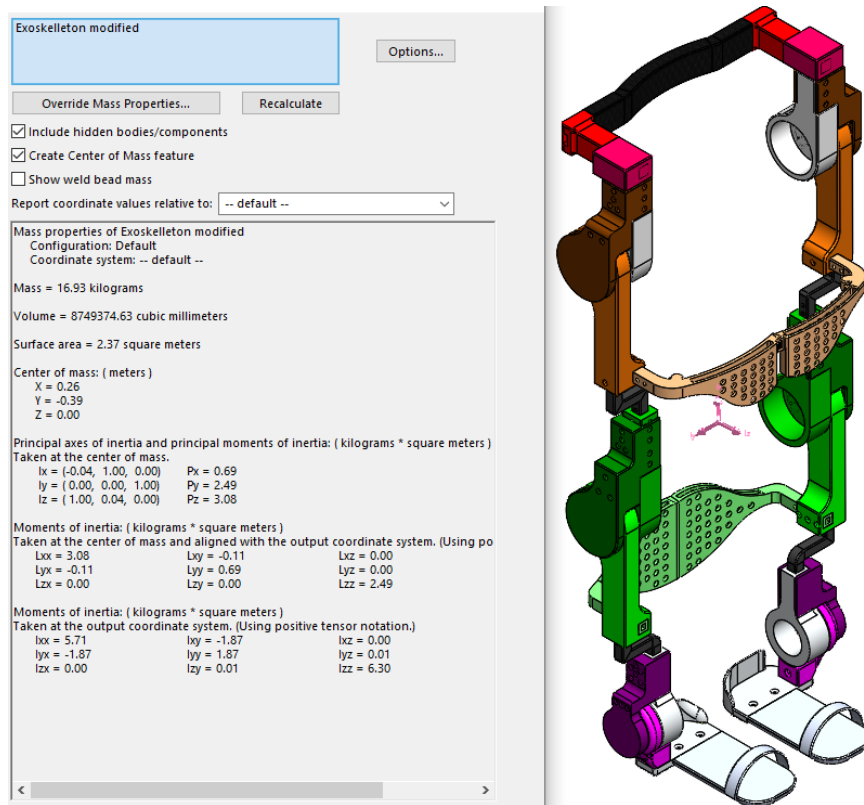


Figure 3.10: Concept 1 (LLE with epicyclic gear system)

Figure 3.11 illustrates the epicyclic gear system designed for LLE in Concept 1. By integrating a Maxon motor with torque values of 0.729 Nm for the hip and knee, and 0.577 Nm for the ankle, using a 56:1 epicyclic gear ratio, the hip, knee, and ankle were able to produce torque outputs of 40.84 Nm, 40.84 Nm, and 32.312 Nm, respectively. According to the torque results obtained for this concept, the robot's hip torque capacity is insufficient to sustain the required 51.321 Nm at the hip joint.

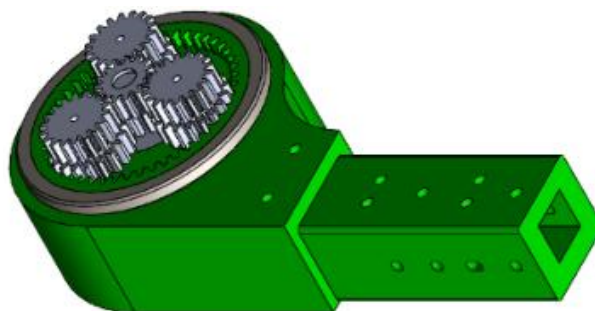


Figure 3.11: Concept 1 (actuator system)

3.5.2 Concept 2: LLE with motor and harmonic driver

This section aimed to minimise the weight, volume, and surface area of the LLE by simplifying its design. The LLE was constructed from aluminium, as shown in Figure 3.12, with its motors featuring harmonic drives with reduction ratios of 160:1 for the hip and knee joints and 120:1 for the ankle joint. The frame's weight under concept 2 is 7,17 kg, not including the actuation system. By adding the weight of the actuators, which totals 8.396 kg across all six harmonic drives, the overall weight of the LLE reaches 15,566 kg, excluding the motors, control system, and battery.

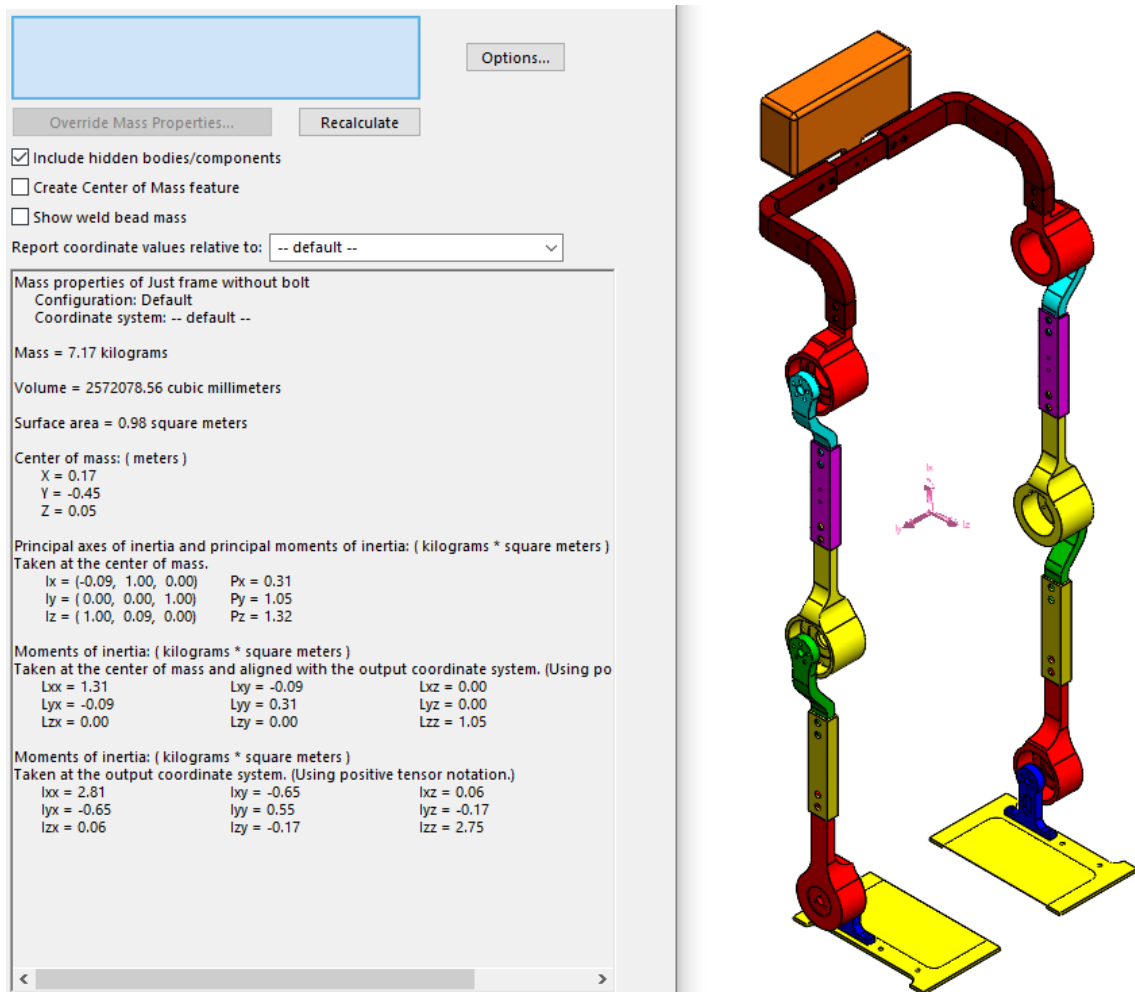


Figure 3.12: Concept 2 (LLE with motor and harmonic driver)

Figure 3.13 depicts the connection between the Maxon motor, labelled 1, and the Laifual harmonic drive, labelled 2. These components are interconnected via a shaft coupling, indicated as 3, which is affixed to the motor shaft using two grub screws, denoted as 4. The combination of the selected Maxon motors and the Laifual harmonic drives can produce torque values of 116.62 Nm, 114.4 Nm, and 69.24 Nm for the hip, knee, and ankle joints, respectively.

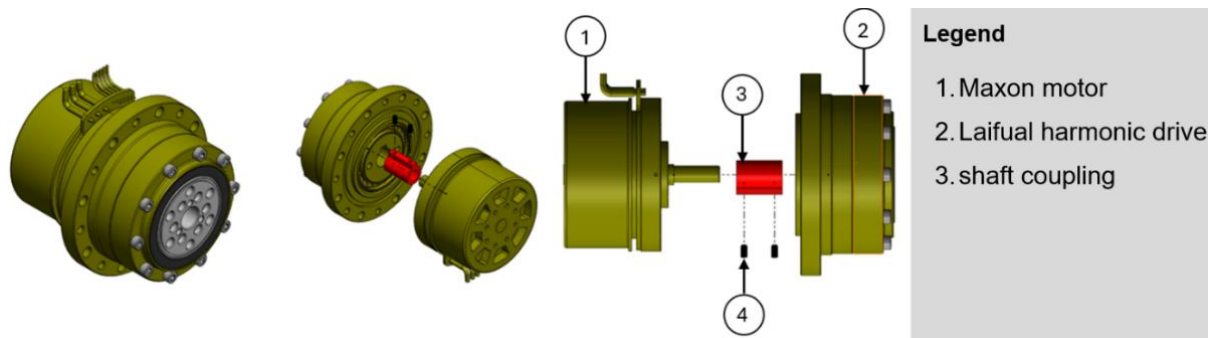


Figure 3.13: Concept 2 (Actuation system)

3.6 Selection of final concept

After analysing the two concepts and gathering the necessary information, the researcher was tasked with selecting the most suitable option. This selection process involved comparing the concepts based on various factors, including weight, actuation output torque, volume, material availability, design simplicity, ease of manufacturing and assembly, anthropomorphic design, and low-cost maintenance. The concept that scored highest in this evaluation was selected as the final concept; as shown in Table 3.6, Concept 2 received a score of 10/10, while its counterpart received 3/10. Based on these results, Concept 2 was selected as the final choice.

Table 3.6: Criteria for selecting

	Concept 1	Concept 2
Anthropomorphic	1	1
volume	0	1
weight	0	1
Actuation output torque	0	1
Material fidelity	1	1
Easy for manufacturing	0	1
Easy for assembly	1	1
The simplicity of the design	0	1
material availability	0	1
Low-cost maintenance	0	1
Total	3/10	10/10

3.6.1 Optimisation of the final Concept

While the selected concept's weight was found to be quite satisfactory, it was decided to pursue weight optimisation for specific components of the LLE to achieve a lighter exoskeleton. To do this, a structural optimisation was performed on the hip, knee, and ankle connectors, the thigh and shank sliders, and the back supporter. Figure 3.14 illustrates the steps taken to generate an optimised hip connector. ANSYS and SolidWorks were utilised for this optimisation, which was conducted under load conditions simulating the weight of the human body acting on the hip connector, with the hip connector fixed at the bottom bolt hole. Based on the results of the initial structural optimisation, the student identified areas where material could be removed to achieve a greater weight reduction. In the second step, SolidWorks was used to remodel the hip connector accordingly.

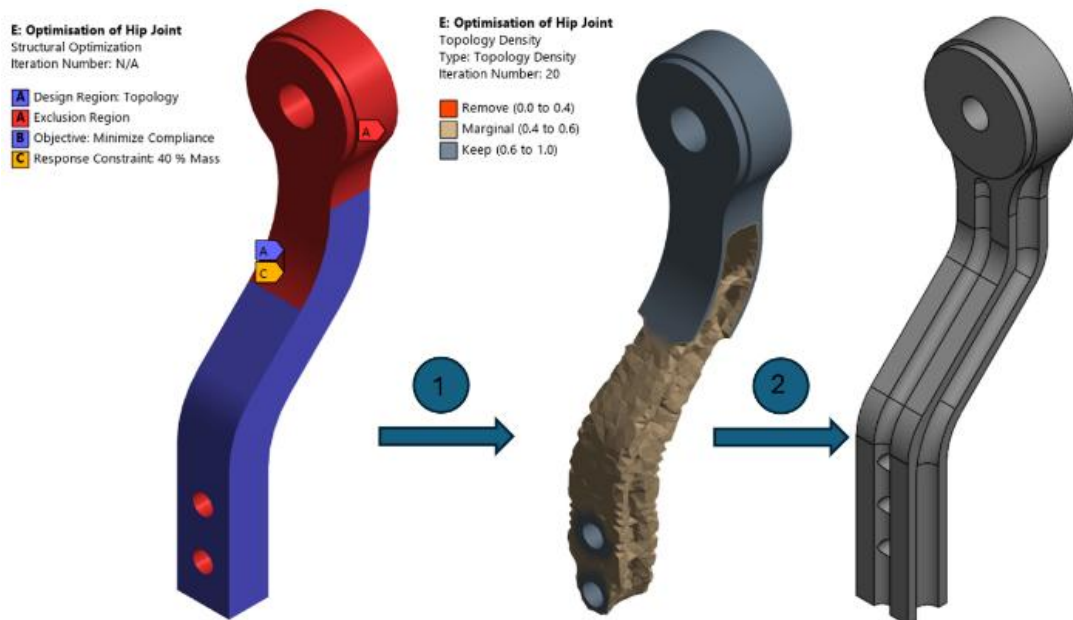


Figure 3.14: Optimisation process

3.6.1.1 Exploring the optimised selected concept

The subsequent sections are dedicated to the assembly of the optimised LLE, detailing the integration of various components, including the back supporter, waist-to-hip, thigh, shank, and foot. This discussion will culminate in an examination of how these segments are interlinked to form the LLE. Technical drawings of each part of the LLE are provided in Appendix B, while Appendix A contains the overall dimensions of each segment and the full LLE.

a. Back supporter/ lower torso

The back supporter of the LLE, as illustrated in Figure 3.15, consists of two waist supports, one back connector, and a control system box, also referred to as a battery pack. The waist

supports frame the user's lower limbs, while the back connector facilitates lateral movement, enabling an adjustable fit for various waist sizes. M6x25.3mm and M4x21mm bolts and nuts are used to secure the connection between these components. Additional details regarding the selection of the bolts are provided below. The waist supports were constructed from Aluminium 6061-T6, and the back connector was made from Aluminium 6082-T6.

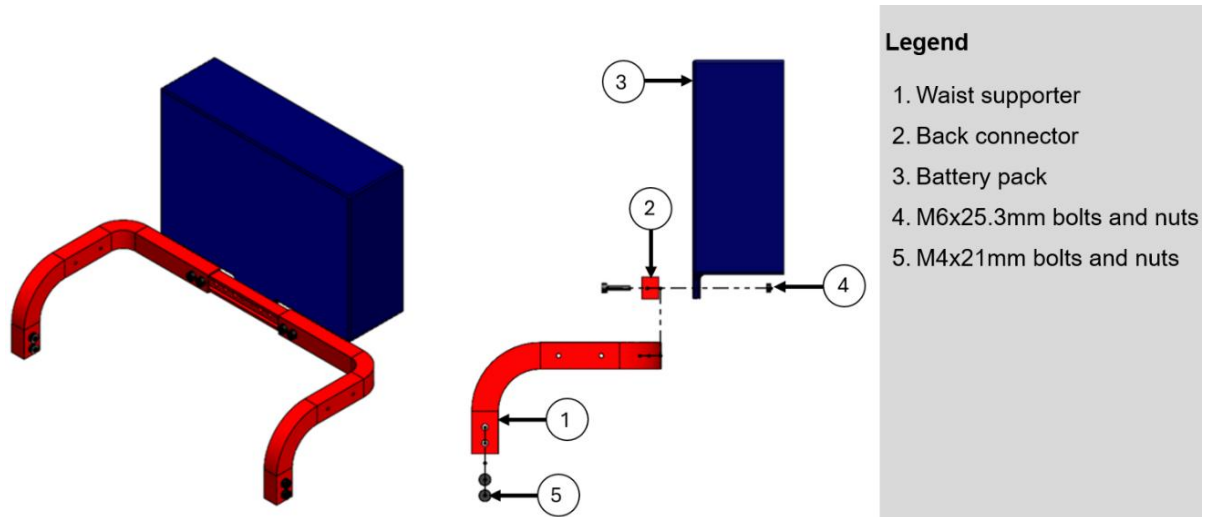


Figure 3.15: Back support

The LLE frame is constructed from 6000-series aluminium alloys due to their reliability, durability, and strength-to-weight ratio. Wong et al. (2010) selected aluminium 6061 for the construction of the powered exoskeleton due to its lightweight nature and machinability capability.

The LLE components, which include hip and knee connectors, hip and knee sliders, ankle connectors, waist-to-hip, and the foot, require precise manufacturing using Computer Numerical Control (CNC) technology. This method guarantees that each part is an exact replica of its CAD design. CNC technology excels at producing the intricate geometries typically found in LLE components. Aluminium 6082 is preferred for its excellent CNC machinability, making it a popular choice in the industry. Engineering Product Design (2023) Indicated that the 6061-T6 and 6082-T6 grades are the most used aluminium alloys within the 6000 series, and they are highly suitable for CNC machining. This manufacturing approach enables the creation of a diverse array of parts and components, ranging from intricate, small designs to extensive, complex structures.

In conclusion, the 6061-T6 and 6082-T6 alloys were chosen for their low-density of 2700 kg/m^3 , making them lighter than materials such as steel, zinc, iron, and copper. This property makes them ideal for constructing a lightweight LLE.

Determining the appropriate bolt length is important for the safe and effective operation of the LLE. Proper engagement between the bolt and the nuts is essential to prevent potential failures during use. Equation 3.6 was used to calculate the correct bolt length.

$$H > l + h \quad 3.6$$

In the equation above, H denotes the defined length of the bolt, ensuring it accommodates all necessary components. The variable l represents the grip length, the distance between the bolt and the nut when the bolt is tightly secured. This measurement is important because it directly affects the assembly's overall strength and stability. Lastly, h represents the height of the nut, contributing to the total stack-up of components.

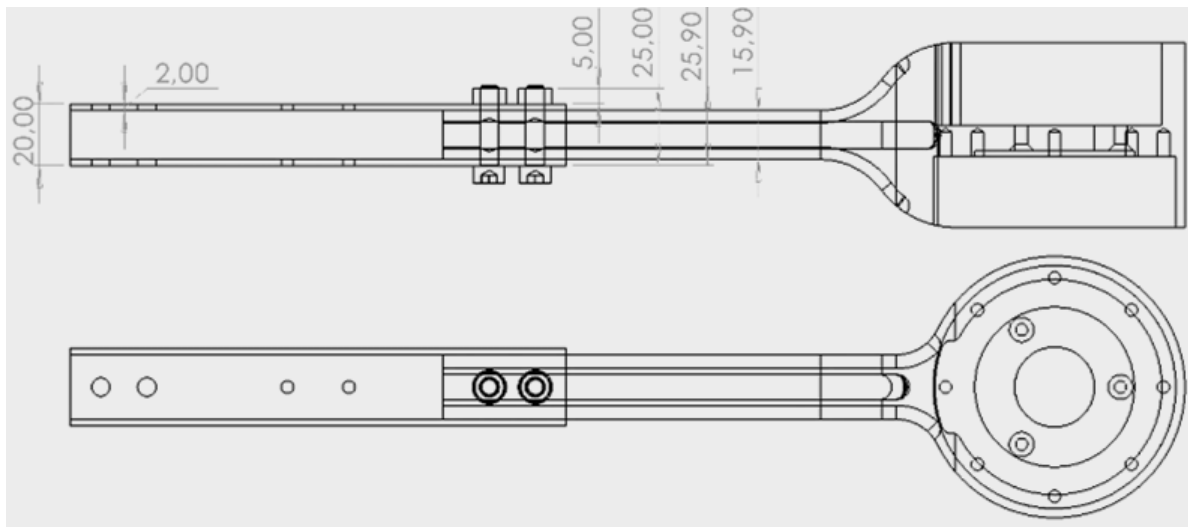


Figure 3.16: Fastener connection

Using the numerical values in Figure 3.16 and substituting them into Equation 3.6, where l represents the grip and h denotes the height of the nut, a total of 25 mm was calculated as the sum of l and h . Since the bolt shank must exceed 25 mm, a bolt measuring 25.9 mm was chosen, as this allowed the nut to fully engage the bolt.

M6 bolts and nuts were used to secure the right and left waist supporters to the back connector, and M4 bolts and nuts were used to attach the control system box to the back connector. It is worth noting that the same approach was taken to select suitable bolts and nuts to connect the remaining parts of the LLE.

b. Waist-to-Hip

The waist-to-hip connector is an essential component that connects the waist supporter to the hip connector. Manufactured from 6082 T6 aluminium, it is securely attached to the waist support using M6 x 25.9 mm bolts and nuts. The waist-to-hip was designed to accommodate the EC 90 flat brushless motor (ø90 mm, 220 Watts) and the 160:1 LSG-17-160-U-I harmonic

drive from Laifual. Both the motor and the harmonic drive are integrated within the waist-to-hip and secured in place with three M5x10 mm socket countersunk head screws for the motor and eight M5x15 mm socket head cap screws for the harmonic drive.

A shaft coupling is secured to the motor shaft using an M2.5x7 mm grub screw. This connection ensures that the motor shaft is firmly held within the hollow section of the harmonic drive via the shaft coupling, thus preventing any separation during operation. Additionally, a clearance closer fills the gap between the harmonic drive and its opening in the waist-to-hip connector. An isometric view and an exploded side view of the waist-to-hip connector are illustrated in Figure 3.17.

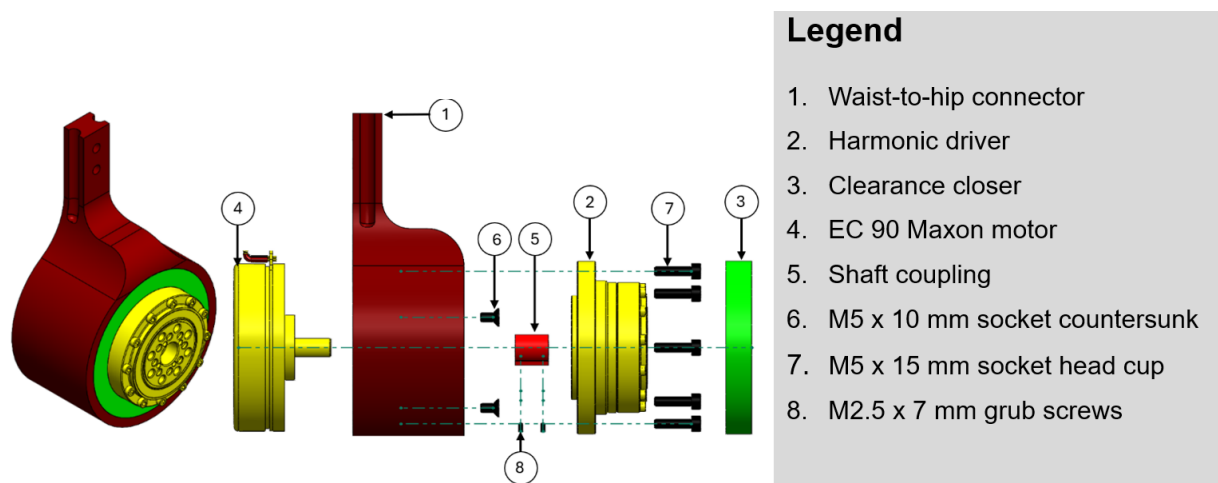


Figure 3.17: Wait-to-hip front isometric view

c. Thigh and Shank

This section focuses on describing the LLE's thigh and shank by showing how they are connected. It is worth noting that these two parts are almost identical in design.

In Figure 3.18, the LLE thigh is illustrated, showing the hip connector and thigh slider attached to a rectangular tube known as the thigh tube. The design emphasises the connection between these components and their functionality within the overall system. The thigh components are linked by a passive prismatic joint mechanism that allows movement within the channel and provides flexibility during assembly. This design enables adjustments to both the hip connector and the thigh slider via the thigh tube, accommodating a range of thigh lengths. To secure these components, M6 x 25.9 mm bolts and nuts made from low-carbon steel 4.8 are used to fasten these parts together. The lower section of the thigh slider features a brushless EC 90 flat motor with a 90 mm diameter and a power rating of 220 watts. This motor is securely integrated into its housing on the thigh slider using three M5 x 10 mm socket countersunk head screws. In addition, the thigh slider features a 160:1 LSG-20-160-U-I harmonic drive from Laifual, which is rigidly attached with eight M5 x 15 mm socket head cap screws. The coupling

shaft functions as a connector between the motor and the hollow shaft of the harmonic drive, facilitating proper interaction between the two components. It is tightened to the motor shaft using two M2.5 x 7mm grub screws. Then, a clearance closer is used to seal the gap between the harmonic drive and the thigh slider.

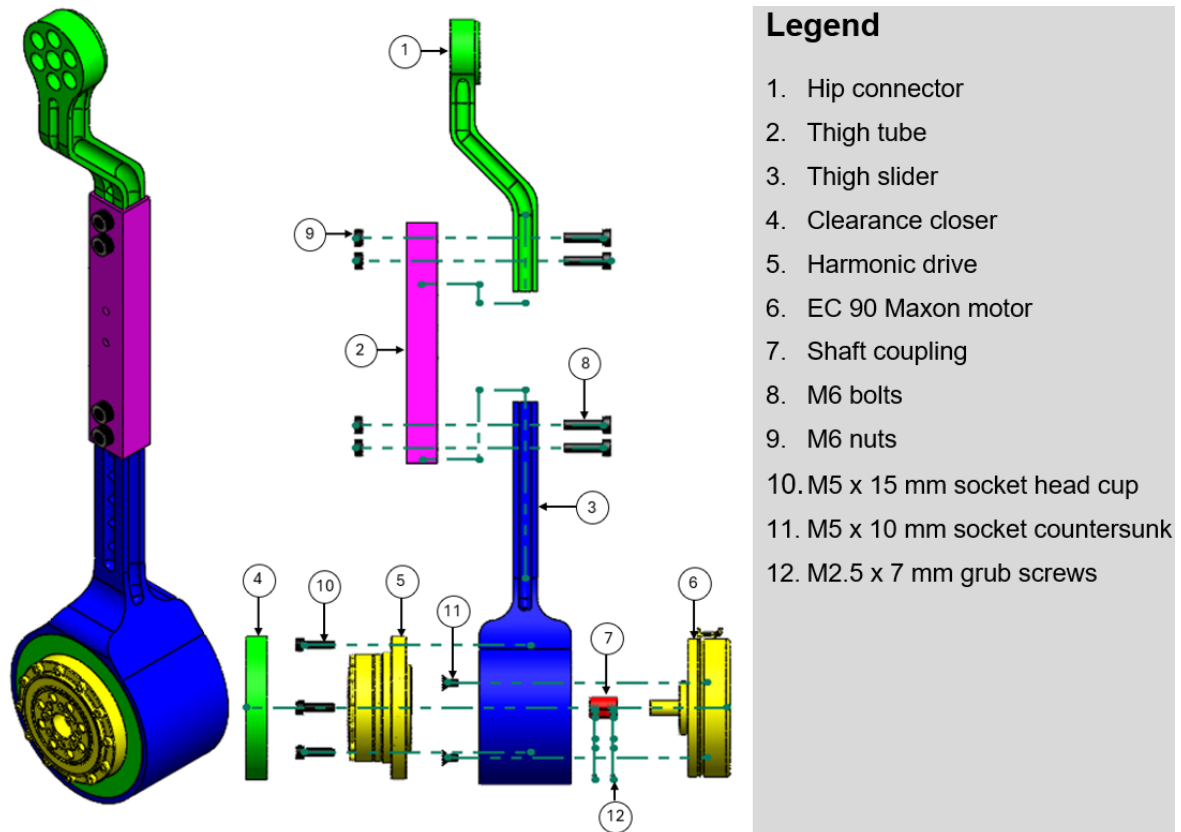


Figure 3.18: Thigh assembly

In Figure 3.19, the shank was assembled using the same methods and features as those used for the thigh segment. The components are labelled as follows: knee connector, shank tube, shank slider, clearance closer, harmonic drive LSG-17-120-U-I, and an EC 60 flat brushless motor with a 60 mm diameter and a 200-watt power rating. A shaft coupling connects the motor shaft using two M2.5 x 7 mm grub screws. M6 x 25.9 mm bolts and nuts secure the knee connector and shank slider to the shank tube. Additional fasteners include three M3 x 10 mm screws and eight M4 x 15 mm screws, which are used to rigidly attach the motor and harmonic drive to the shank slider. Furthermore, the motor clearance closer is used to close the gap between the motor and its surroundings.

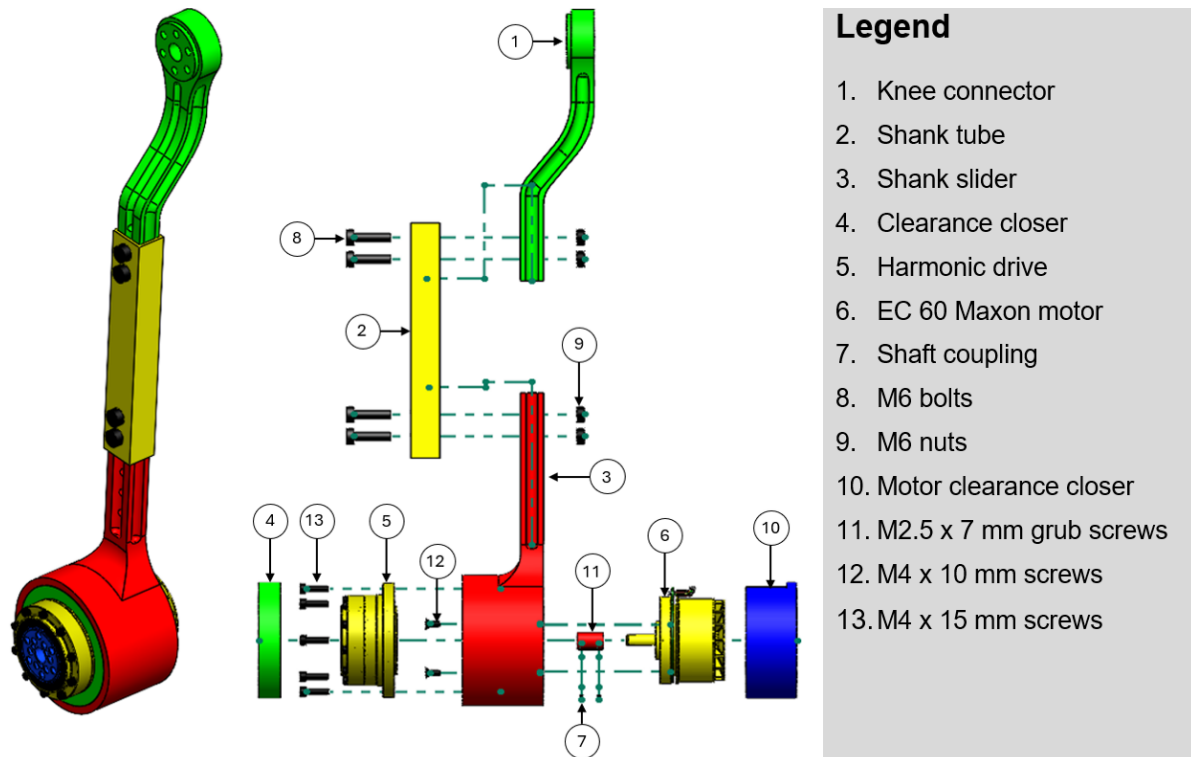


Figure 3.19: Shank assembly

d. Foot subassembly

The ankle connector is an important link between the shank slider and the foot base, serving as a key component of the mechanical structure. The foot base is made from aluminium 6061-T6, while the ankle connector is made from aluminium 6082-T6, ensuring durability and stability. Four low-carbon steel M4 x 8 mm bolts secure the ankle connector to the foot base. Six low-carbon steel M5 x18 mm bolts are used to connect the ankle connector to the shank slider. The foot subassembly is represented in Figure 3.20.

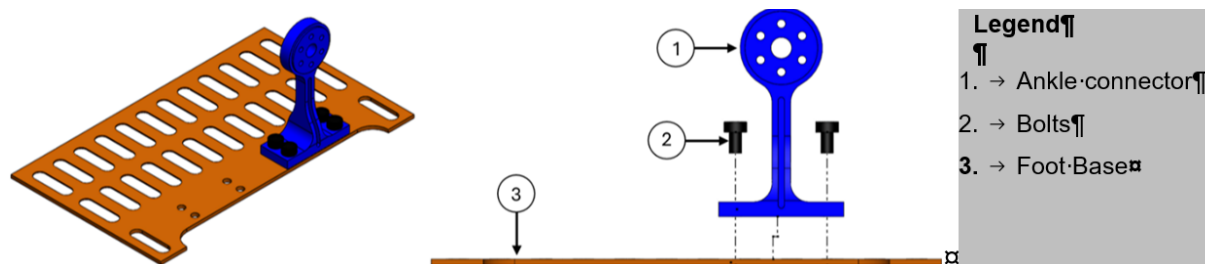


Figure 3.20: Foot subassembly

e. Prototype Assembly

Figure 3.21 presents a depiction of the LLE assembly, along with an exploded view that specifically highlights the connection between one leg and the waist supporter. The waist-to-hip component features a prismatic connection to the waist support. This connection is secured

using M6 x 25.9 mm bolts and nuts. Notably, the waist-to-hip apparatus features an adjustable range of 15 mm, allowing the LLE height to elongate or shorten around the lower torso to accommodate varying user needs.

The thigh component connects to the harmonic drive on the waist-to-hip through bolt fittings that interface with the hip connector. This joint architecture facilitates revolute motion, which is crucial for enabling the thigh to perform a range of movements that mimic natural flexion and extension, effectively allowing the user to replicate a walking gait. The shank segment is connected to the harmonic drive situated on the thigh slider. The revolute joint mechanism used in this section allows the shank to flex and extend.

Finally, the foot component of the LLE is intricately linked to the harmonic drive within the shank slider. A revolute joint mechanism is used in this context, enabling the foot to perform specified flexion and extension movements, which are essential for natural locomotion. The integration of M5x 18mm bolts reinforces this joint, ensuring the foot remains securely attached to the harmonic drive.

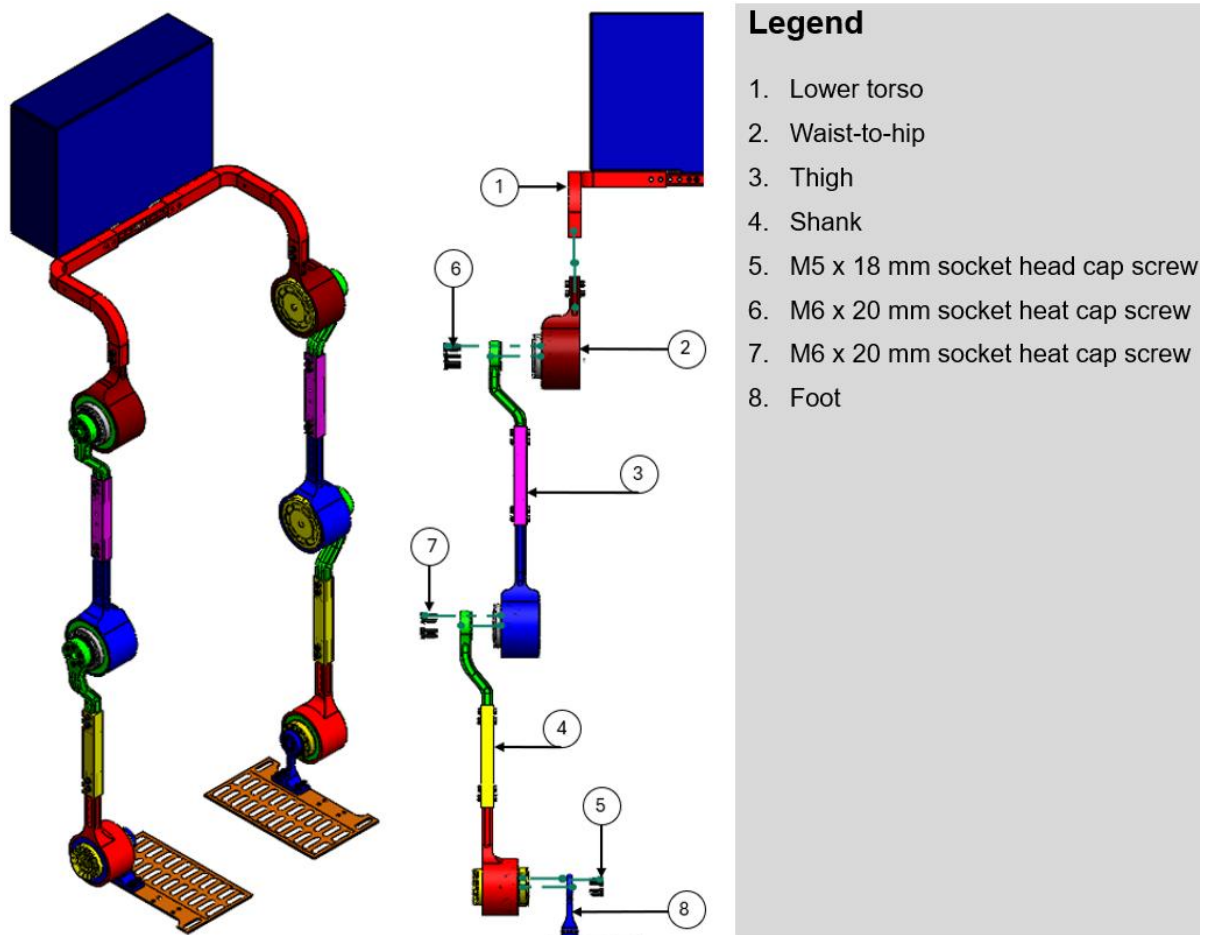


Figure 3.21: LLE 3D model

The optimised LLE frame in Figure 3.22 weighs 4.942 kg, which is 31.074% lighter than its initial design in concept 2 in Figure 3.12. By combining the weight of the LLE frame and its

selected actuators, the LLE's total weight reaches 13.338 kg, excluding the battery and control system, which weigh 16 kg. When these components are combined, the LLE's overall weight is 29.338 kg.

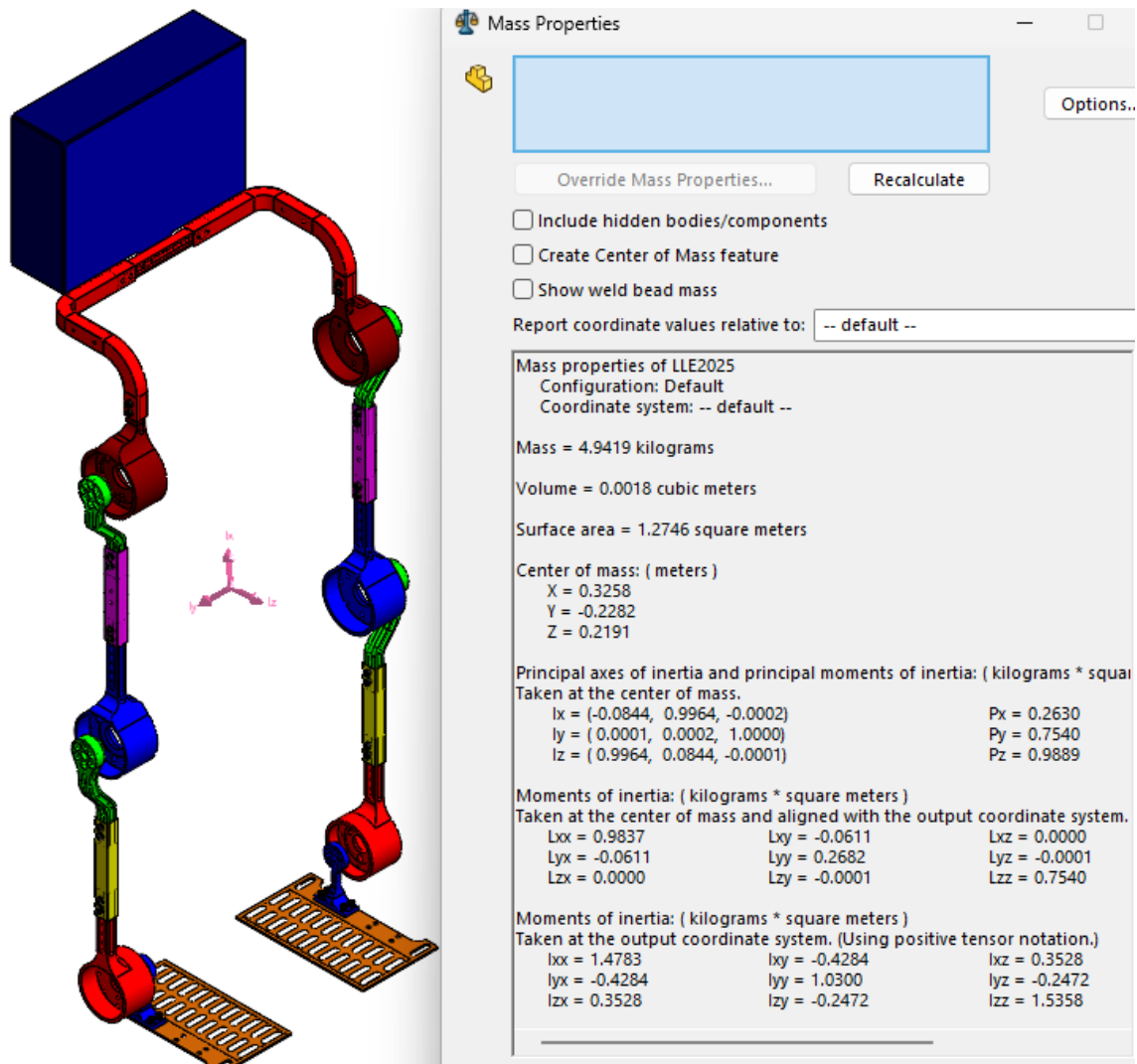


Figure 3.22: LLE 3D model frame mass properties

3.7 Finite Element Analysis (FEA) Methodology

This section focused on establishing the structural fidelity of the LLE through static structural and linear buckling analyses. These analyses aimed to determine whether the LLE design could support its own weight and the user's weight simultaneously. The static structural analysis was intended to evaluate the exoskeleton's capability to withstand applied loads, while the linear buckling analysis had to predict the load at which the robot's structure would buckle.

Figure 3.23 illustrates the steps the researcher took to perform the FEA of the walking robot, beginning with geometry selection and culminating in solutions. The methodology presented

in this section provides a framework for achieving one of the objectives of this research: identifying potential weaknesses in the exoskeleton design and implementing necessary improvements to enhance its overall performance and the safety of the LLE.



Figure 3.23: The FEA methodology diagram (Pawar et al., 2015)

The FEAs of the LLE segments were conducted under two distinct scenarios. In the first scenario, the left leg was in the stance phase; in other words, this leg was analysed in a vertical position, with a point load applied axially. In the second scenario, the robot components were analysed during a swing phase, with the leg positioned horizontally and subjected to two transverse loads during the simulation. Additionally, a linear buckling analysis was performed on the segment in the stance phase to identify the load at which the robot components could become unstable. Both scenarios are illustrated in Figure 3.24.

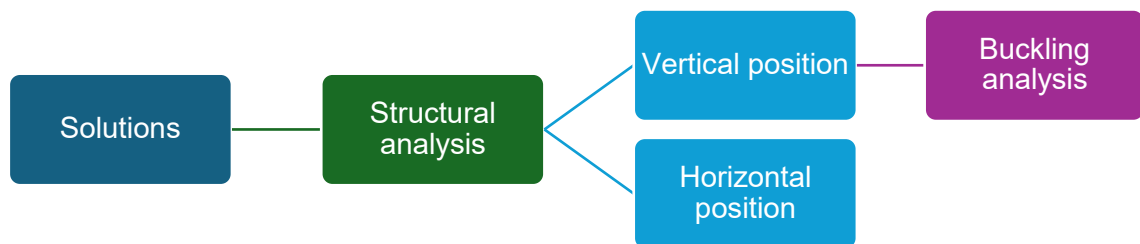


Figure 3.24: Sub-solution diagram

3.7.1 Geometry selection

The 3D model of the LLE segment, including the lower torso, upper and lower thighs, upper and lower shanks, and the foot, was individually imported into ANSYS for structural and buckling analysis, evaluating its performance under defined loads and boundary conditions.

Effective structural analysis requires careful decisions regarding material selection, segment connections, mesh quality, and boundary conditions. The following sections will explore these key considerations in more detail.

3.7.2 Material assignments

Upon importing the step file into the FEA software, the next step was to assign the appropriate materials to each component. As illustrated in Figure 3.25, aluminium 6061-T6 (green) was designated for the Thigh tube, Aluminium 6082-T6 (yellow) for the hip connector, and low

carbon steel 4.8 (red) for the bolts and nuts. Similar procedures were followed to assign materials to the remaining segments of the LLE.

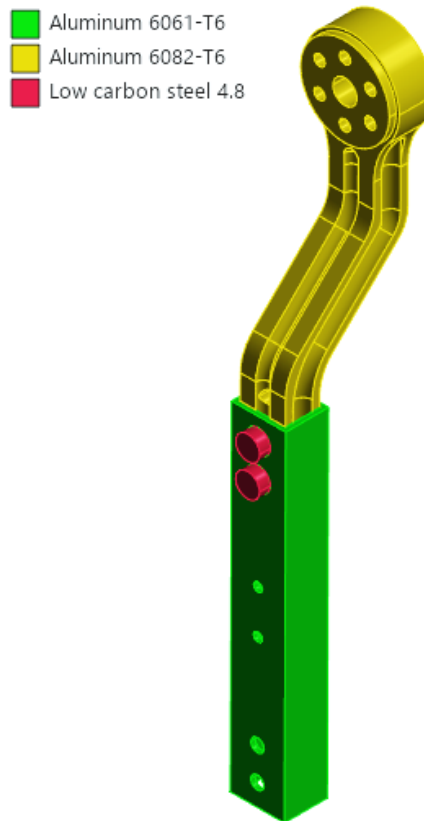


Figure 3.25: Vertical Upper thigh

3.7.3 Connections

Özgün (2021) Defined that the bonded contact used in ANSYS is a type of contact used in FEA when the interacting parts are intended to stick together and not separate or slide relative to each other during the simulation. This type of contact is useful for modelling situations in which parts are joined, such as adhesively bonded joints or welded connections.

To ensure the results were accurate, the default bonded contact was modified to comply with the appropriate specification. This adjustment directly influenced both the static structural and buckling analysis conducted on the upper thigh and other components of the LLE. Utilizing default contacts could result in potential inaccuracies; therefore, implementing these adjustments was essential to ensure the precision of the results. Figure 3.26 illustrates the types of contacts considered in this scenario to ensure accurate results.

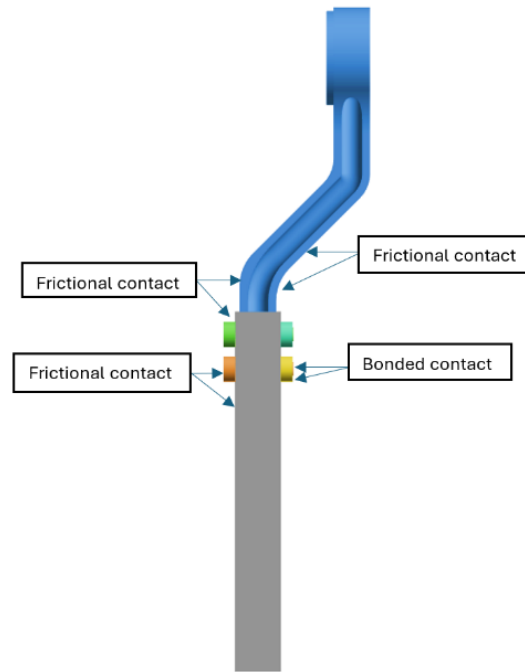


Figure 3.26: Upper thigh under established connections

To analyse the behaviour of the hip connector and the thigh tube, a frictional contact model was employed to simulate their interaction. This frictional contact was also implemented between the bolt shanks and the bolt holes on the thigh tube, the lower surfaces of the bolt heads and their imprints on the outer surfaces of the thigh tube, as well as the lower surfaces of the nuts and their imprints on the outer surfaces of the thigh tube, and between the bolt shanks and the bolt holes on the hip connector. However, the contact between the nut hole and bolt shank was modelled using bonded contact to prevent separation during the simulation. Table 3.7 presents the coefficients of friction to be considered during the FEA conducted for this research.

Table 3.7: Friction coefficient (Naval Surface Warfare Centre, 2007)

Material	Static friction coefficient	
	c_{μ} dry	c_{μ} lubricated
Aluminium on steel	0.6	0.5
Aluminium on Aluminium	1.1	-
Steel on steel	0.8	0.5

3.7.4 Meshing

The method used to size these elements had to be consistent throughout to ensure uniformity. The sizes of the elements varied depending on the area they were applied to; for example, in

this simulation, the bolts, bolt holes, and nuts were the smallest, reducing the likelihood of errors and ensuring the accuracy of the results. Minor features, such as small fillets and extraneous holes, were simplified in this simulation to mitigate potential inaccuracies. A close-up illustration of the meshes applied to the bolt holes and bolts is presented in Figure 3.27.

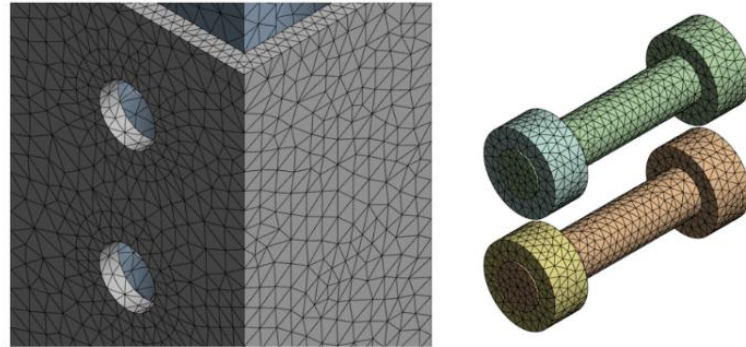


Figure 3.27: Close-up on the applied meshes

In ANSYS Mechanical, there is a tool that assesses the quality of the applied mesh. This tool visualises mesh quality through a metric graph. In Figure 3.28, an element quality analysis was generated, consisting of 66966 elements and 120160 nodes, by defining the mesh size for each part of the upper thigh.

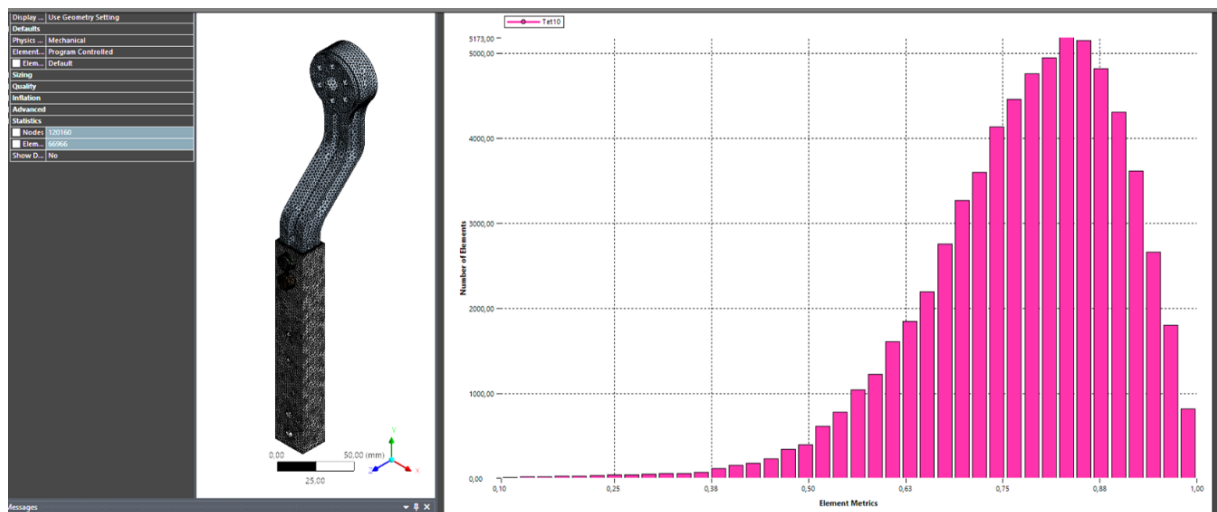


Figure 3.28: Mesh metrics

Element quality is measured on a scale of 0 to 1 in ANSYS. A score of 0 indicates inferior quality, while a score of 1 signifies perfect quality. This quality assessment is crucial for determining the reliability and accuracy of simulations and analyses performed using ANSYS software (ANSYS, n.d.).

Upon reviewing the definition of element quality according to ANSYS, it was determined that the upper thigh's element quality met the satisfactory criteria, as most elements exhibited a

quality index above 0.5. This indicates that the components of the upper thigh are of acceptable quality and meet the required standards for the analysis.

3.7.4.1 Mesh sensitivity

The mesh sensitivity analysis was performed manually by adjusting the mesh size, providing enhanced control over variations in mesh elements and nodes. Given the limitations of the Ansys student version, which restricts the total number of elements and nodes to fewer than 130 thousand, this approach was essential to ensure that the results were independent of the mesh configuration. Furthermore, it ensured that the mesh elements and nodes conformed to the constraints of the finite element analysis software used. A total of six iterations were conducted, with a 1 mm decrement in the mesh element size, as illustrated in Table 3.8.

Table 3.8: Mesh sensitivity study

	Iteration(I) in mm					
	I1	I2	I3	I4	I5	I6
Body sizing	7.5	6.5	5.5	4.5	3.5	2.5
Body sizing 2	7.3	6.3	5.3	4.3	3.3	2.3
Face sizing	6.3	5.3	4.3	3.3	2.3	1.3
Body sizing 3	6.5	5.5	4.5	3.5	2.5	1.5
Body sizing 4	6.5	5.5	4.5	3.5	2.5	1.5
	Results					
	Re1	Re2	Re3	Re4	Re5	Re6
Tot def (mm)	3.001	2.676	3.053	2.109	2.985	2.931
Von Mises (MPa)	134.5	1572.8	141.7	5521.4	157.6	171.4010.
Safety Factor	2.105	0.172	1.948	0.062	1.752	2.105
	Refine					
Tot def (mm)						Refine 6
Von Mises (MPa)						2.308
Safety Factor						168.62
						1,637

Based on results i5 and i6 in Table 3.8, the difference in stress values is not statistically significant; the disparity between these two figures is 8.052%. Given the proximity of these results, mesh refinement was implemented in regions with stress concentrations. After applying the refinement to i6, the stress decreased to 168.62 MPa, a change of 1,623%. From this variation, it was concluded that the stress result was not dependent on the mesh.

3.7.5 Boundary conditions and loads

To examine the behaviour of the LLE segments under specific load conditions, it was assumed that the walking robot was in the mid-swing phase. In this scenario, one leg of the LLE is in complete contact with the ground, while the other leg is entirely off the ground, as illustrated in Figure 3.29.

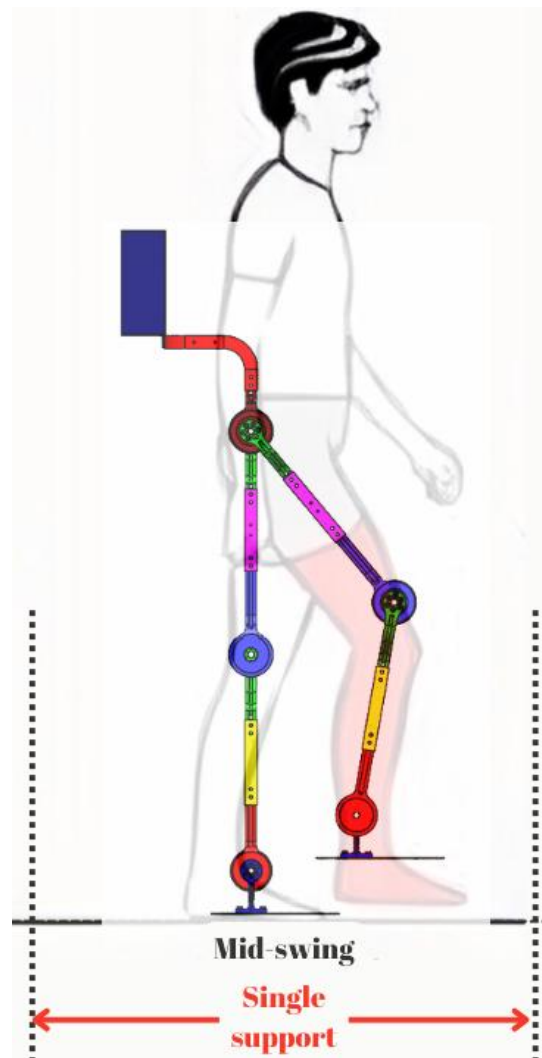


Figure 3.29: Swing phase of the gait cycle (Pirker & Katzenschlager, 2016)

For illustrative purposes, the upper thigh was used as a sample to demonstrate how the boundary conditions were applied to it; the same approach was used for the remaining segments of the LLE, as presented in Appendix C.

In the first scenario, analyses were conducted on the leg in contact with the ground, as shown in Figure 3.30, where the upper thigh was subjected to the applied axial load. The axial load, marked as (C), was applied to the head of the hip connector. This segment was fixed at the last bolt hole marked as (A). The standard earth gravity (B) was applied to the upper thigh to account for the segment's weight throughout the simulation.

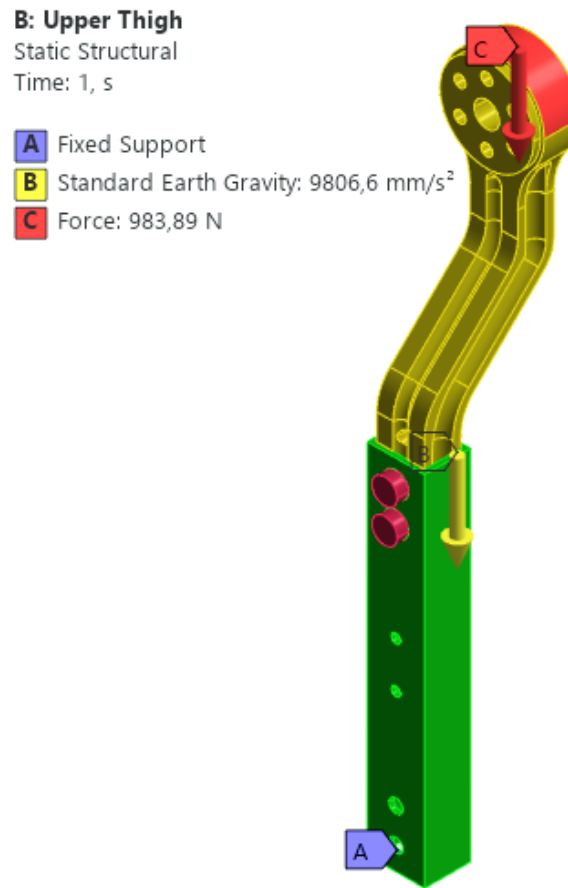


Figure 3.30: Upper thigh under axial loading boundary conditions

Table 3.9 details how the acting axial load on the upper thigh was determined. The weight of the lower leg is excluded from this calculation due to its position relative to the fixed support. The 29,338 kg LLE was designed to support up to 81 kg of its user. By combining these two weights, the total weight required to perform the FEA of the LLE was determined to be 110.338 kg. In the table 3.9, the combined weight is referred to as C-weight, along with the C-weight of the lower leg.

Table 3.9: Acting load on the horizontal thigh

C-weight (Kg)	lower leg's C-weight (Kg)	Applied weight (kg)	Applied weight (N)
110.338	10.0433	$110.338 - 10.0433 = 100.295$	983.891

In the second scenario, the upper thigh was subjected to transverse applied loads as illustrated in Figure 3.31. The simulation was conducted on the leg that was elevated from the ground. During this simulation, the weight of the leg and the surrounding LLE composite were accounted for, whereas the remaining weights of the user and the LLE were disregarded. This exclusion is justified by the fixed point's position, which renders the remaining weights inactive. The analysis of this segment under transverse loading was conducted under a worst-case scenario, assuming the thigh was aligned with the ground. Under this condition, the elevated leg will experience the highest point loads and stresses.

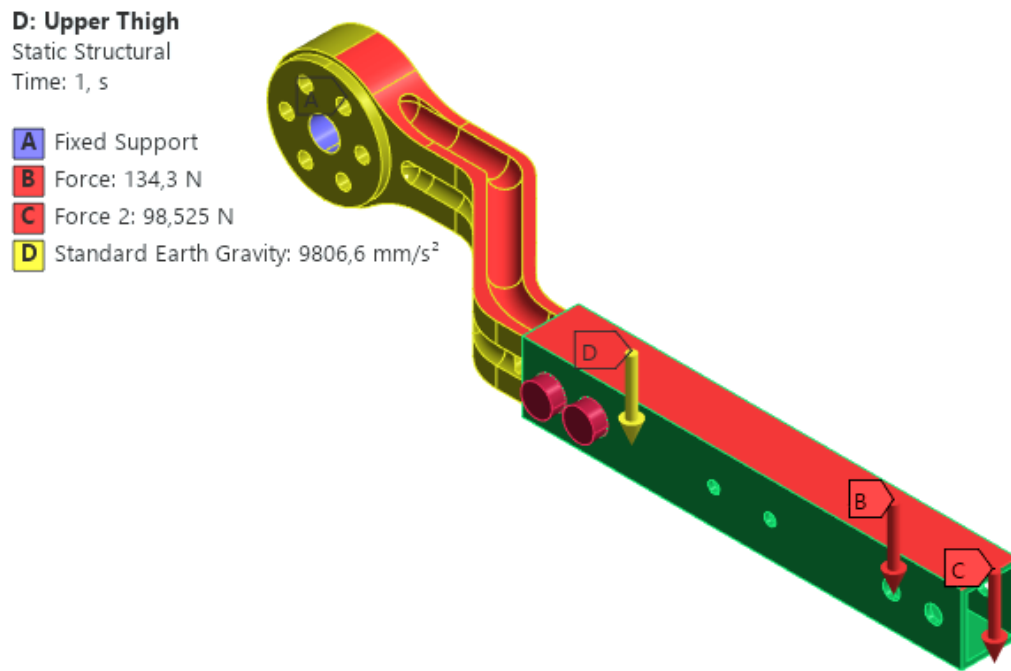


Figure 3.31: Upper thigh under transverse loading boundary conditions

Under transverse loading, the upper thigh was fixed via a hole in the hip connector head; this boundary condition was indicated as A in Figure 3.31. Two transverse loads were applied, with B denoting the load on the upper thigh and C representing the load on the lower leg, applied at the edge of the thigh tube of the LLE. Additionally, the standard earth gravity, labelled as D, was applied to the upper thigh to account for the segment's weight during the simulation. Table 3.10 shows the forces acting on the upper thigh under transverse loading.

Table 3.10: Acting load on the horizontal thigh

Thigh weight in Kg	Thigh weight in N	Lower leg weight in Kg	Lower leg weight in N
13.69049	134.304	10.0433	98.5251

The results of the FEA for each concerned segment of the LLE will be presented and discussed in Chapter 4, focusing on total deformation, Von Mises stress, factor of safety, bending stress, and linear buckling.

3.7.6 Consolidation of key assumptions

This subsection summarises the assumptions concerning the boundary conditions, followed by an examination of the contact interface between the mating components of the LLE segment. Upon completing the setup, material selection will be made to assess the walking robot's feasibility under various load conditions.

The analysed segments were subjected to gravitational and applied forces. The gravitational force accounted for the segment's weight. To ensure that the simulations were conducted in accordance with the human gait cycle, the main phases, known as stance and swing, were considered; these two scenarios are illustrated in Figures 3.30 and 3.31. The segments were supported at their ends with fixed supports.

In this research, two types of contacts were considered. Firstly, a frictional contact was established between two mating parts, enabling the segment to emulate the behaviour of a physical segment by accounting for the friction between the components. Secondly, a bonded contact was applied between the nut and the end of the bolt shank, a contact type selected to prevent dislocation of both the nut and the bolt.

3.8 MATLAB Methodology

In this section, all steps are detailed, starting with setting up the environment and importing the LLE into Simulink. This is followed by the configurations made to the blocks and actuators. The LIPM is introduced to simplify the walking robot's complexity. Throughout this process, the research explores the benefits of the ZMP, the COM, and inverse kinematics in achieving the desired motion. In Appendix E, snippets of code related to this portion are presented for illustrative purposes.

It is worth mentioning the following two points:

- The walking gait algorithm discussed in this section is adapted from the program developed by the MATLAB team.
- It is assumed that users of the designed LLE are accustomed to using crutches, as the LLE is limited to the sagittal plane. These crutches aim to improve the walking robot's balance by providing support during movement. These crutches also support the user's upper body weight.

3.8.1 Setting the environment

The initial step was to define the environment in which the LLE would be simulated. This environment consisted of a world plane linked to a configuration setup that included a solver configuration, a world frame, and a mechanism configuration. The dimensions of the world plane were defined as 8 m in length, 1 m in width, and 0.0012 m in height. In the simulation, the joints of the LLE rotate around the x-axis, which extends out of the plane. The y-axis aligns with the direction of motion from one point to another, while the z-axis is aligned with the

direction of gravity. This arrangement of the world frame was established by adjusting the mechanism's configuration. A six-degree-of-freedom joint was used to position the lower torso of the LLE within the environment; this joint enables the LLE to move freely in any direction. Conn1 serves as the connection port for the lower torso, conn2 is designated for connecting the LLE's feet to the world plane, and conn3 is a port through which the transform sensor is connected to the mechanism configuration. The described Simulink simulation environment is illustrated in Figure 3.32.

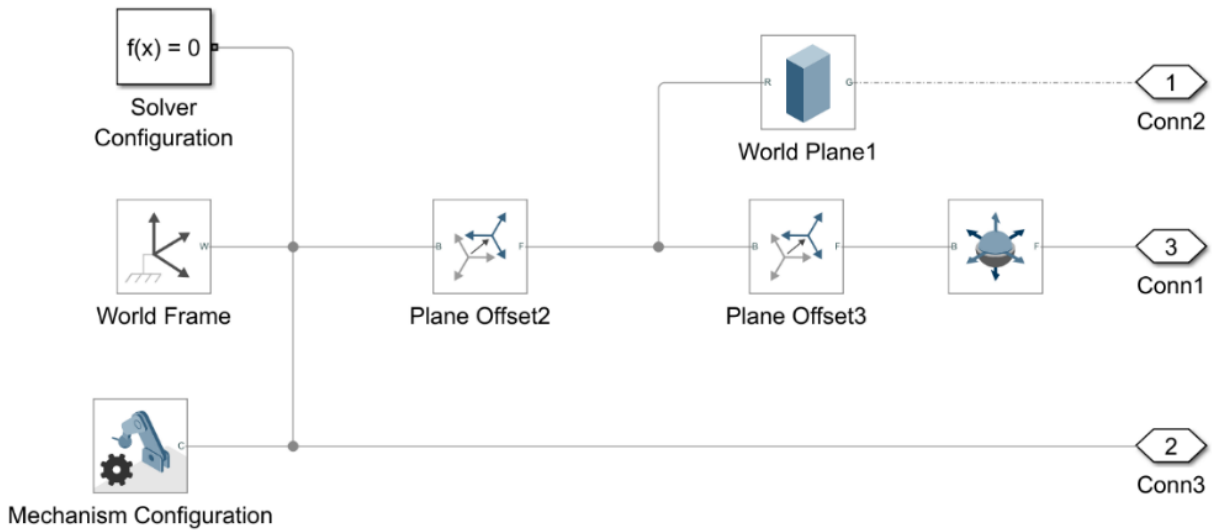


Figure 3.32: Configuration of the simulation environment

3.8.2 Simulink preliminary assembly

After establishing the environment for the LLE's operation, the next step was to import each LLE segment into Simulink individually. This was accomplished using a block called the Solid file block, which is available in Simulink. This block was selected for its flexibility and control. It is important to note that the files being imported were in the STEP file format. Once all seven files were successfully imported into Simulink, they were connected to their respective counterparts. Where actuation was necessary, it was integrated into the segments of the two counterparts. To ensure that the revolute joints of the LLE function as intended, the assembled segments were connected to a preliminary world frame. Figure 3.33 illustrates the schematic of the LLE and its assembly in Simulink's multibody environment.

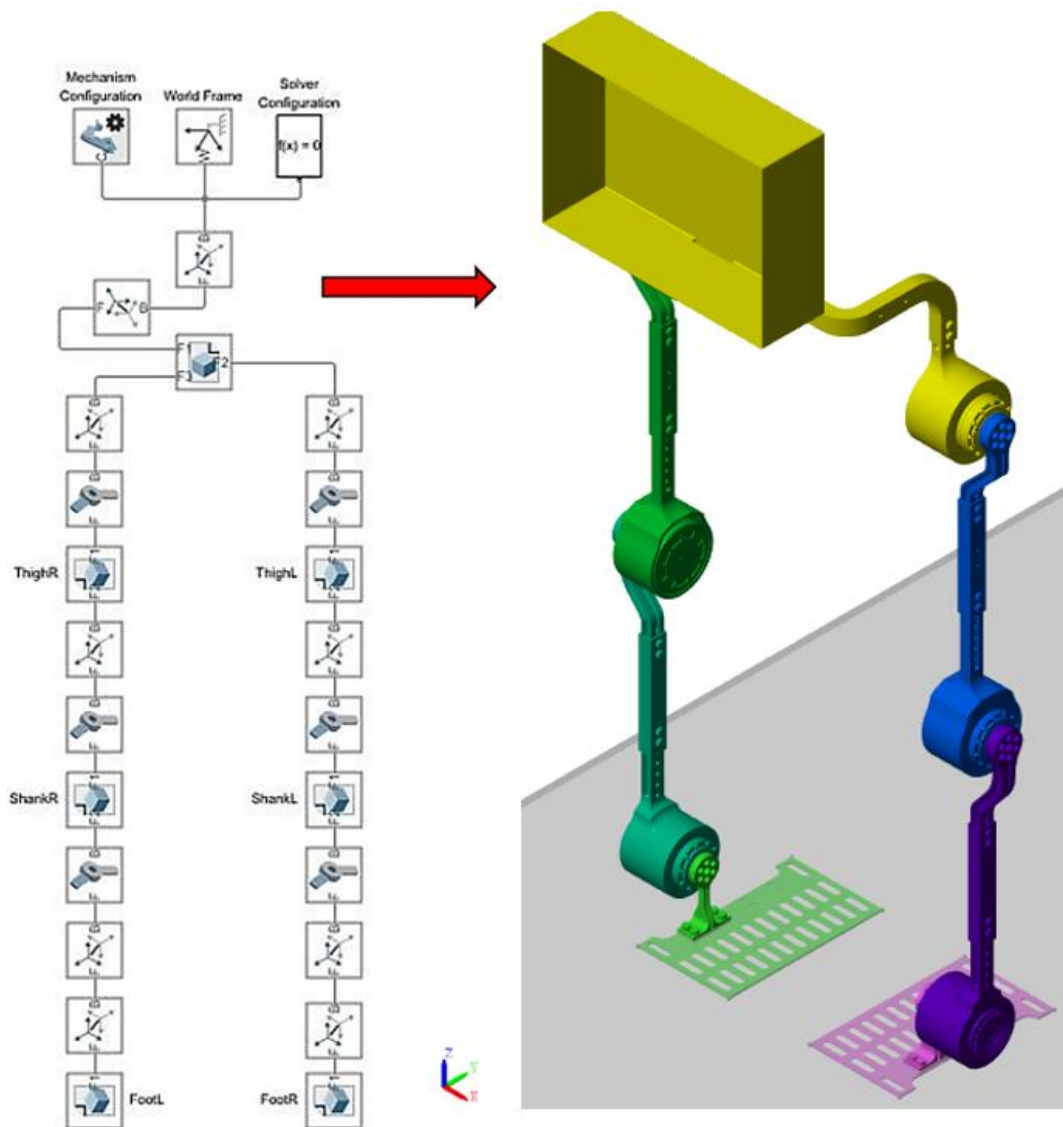


Figure 3.33: LLE preliminary Simscape schematic and assembly

3.8.3 Connecting the Modified LLE to the World Frame

The configuration of the imported and assembled segments of the LLE was modified to enhance its connection to the world frame subsystem. As previously established in the environment setup, Conn1, Conn2, and Conn3 were used to connect the segments to their respective ports. Conn1 was connected to the lower torso via F1, while Conn2 was linked to G1 and G2, where G represents ground. This connection effectively establishes a link between the LLE's feet and the world plane (ground). Furthermore, the sensor transform was connected to the world frame subassembly through Conn3. In Figure 3.34, find the illustration of the modified LLE in the world frame.

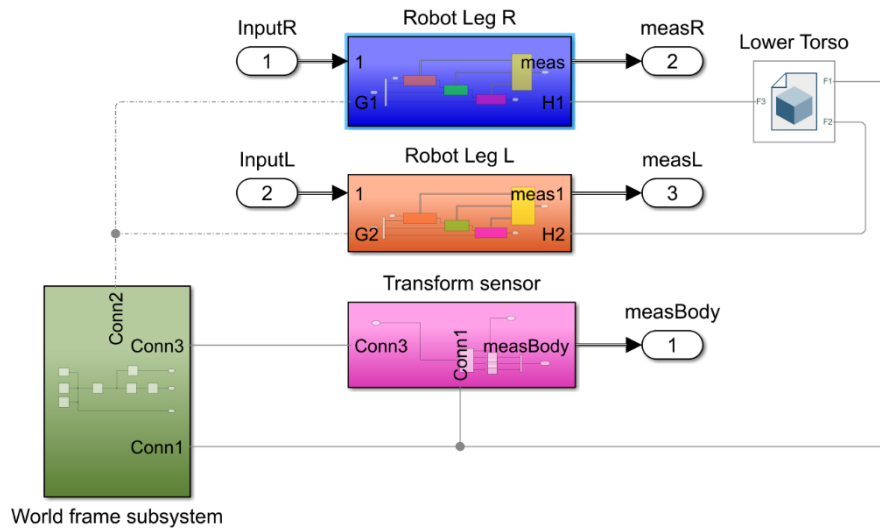


Figure 3.34: Modified LLE connected to the world frame (adapted from Castro, 2019)

In Simulink, a transform Sensor block measures the relative spatial relationship between two frames. The base frame is connected to the follower frame through their designated ports, B and F. The output of this block includes quantities such as pose, velocity, and acceleration, all of which can be represented as time-varying physical signals (MathWorks, 2012).

3.8.4 Leg assembly

Given that the right leg of the LLE is an exact replica of the left leg, the Simulink assembly process explanation focused solely on the left leg. The left leg comprises links, rigid transformations, and an actuation system. Unlike the thigh and shank, the foot is influenced by an additional factor: contact force. The three subassemblies of the left leg are connected to the package measurements, and the data for each joint were processed as illustrated in Figure 3.35.

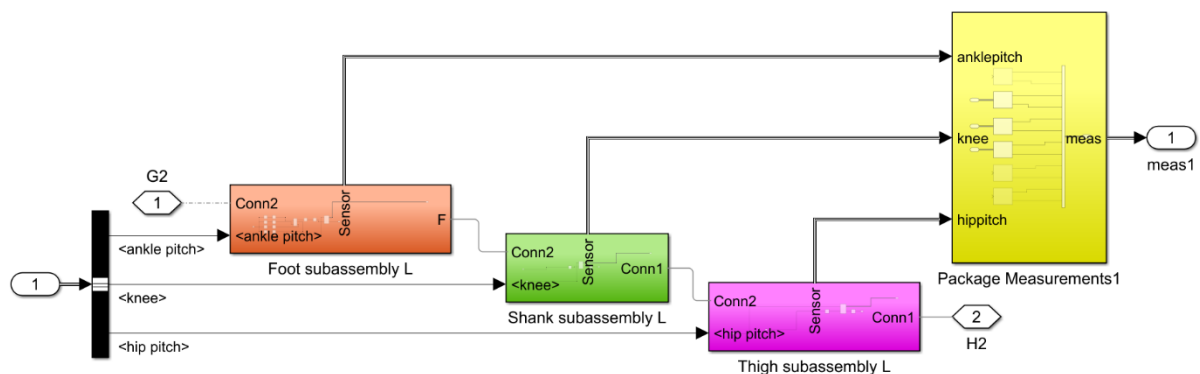


Figure 3.35: Lower limb exoskeleton left leg (adapted from Castro, 2019)

3.8.4.1 Thigh and shank system

As previously discussed, the thigh system consists of a link, an actuation mechanism, and a rigid transform. The thigh was connected to the lower torso via the actuator at the hip pitch

joint. To accurately position this joint, a rigid transform was used to adjust the connection of the hip pitch actuator until proper alignment with the hip connector was achieved, thereby obviating the need for an additional rigid transform for the hip joint. Conn1 functions as a connecting port between the thigh subassembly and the lower torso frame F1, whereas Conn2 connects the thigh subassembly to the shank subassembly. The *<hip pitch>* input port enables the transfer of inverse kinematics data to the actuator, facilitating the attainment of desired joint angles. An illustrative schematic of the thigh segment is provided in Figure 3.36.

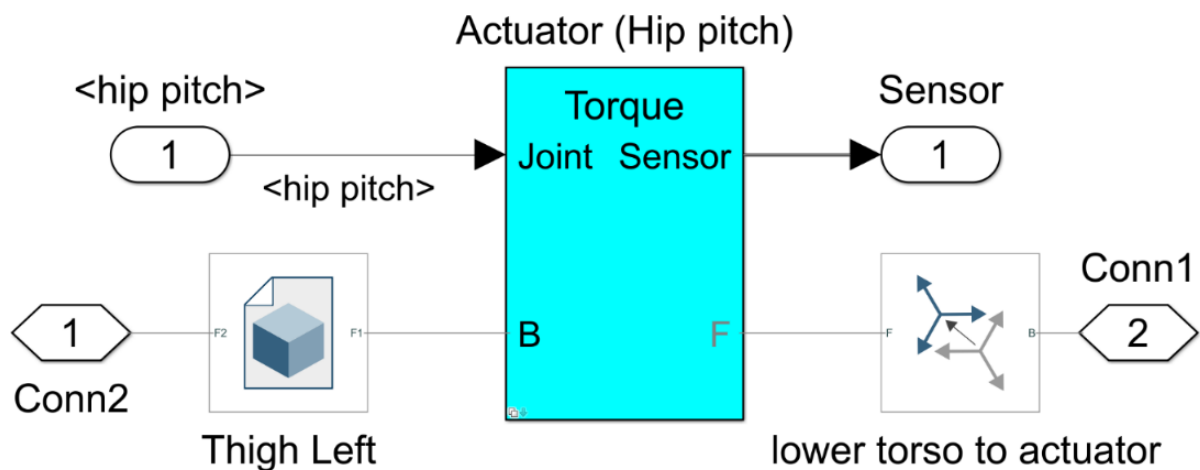


Figure 3.36: Thigh system (adapted from Castro, 2019)

In Figure 3.37, local coordinate frames are defined for both the hip connector and the thigh slider, with the z-axis representing the rotation axis for each segment. This detail is important because it determines how the thigh segment rotates relative to the hip joint at which they are connected.

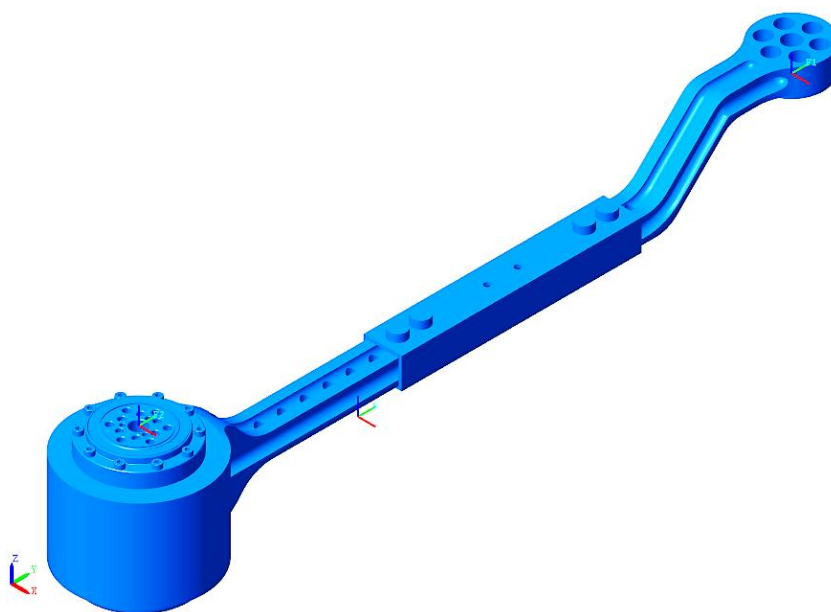


Figure 3.37: Thigh system with local coordinate

The *<knee>* acts as the motor's receptor, enabling it to receive the knee inverse kinematic data, while Conn2 serves as the connection port between the shank subassembly and the foot subassembly. In the shank segment, the knee connector F1 and the shank slider F2 both have their axes of rotation aligned along the z-axis within the local coordinate frame; consequently, the pitch joints are aligned accordingly. The shank system is illustrated in Figure 3.38.

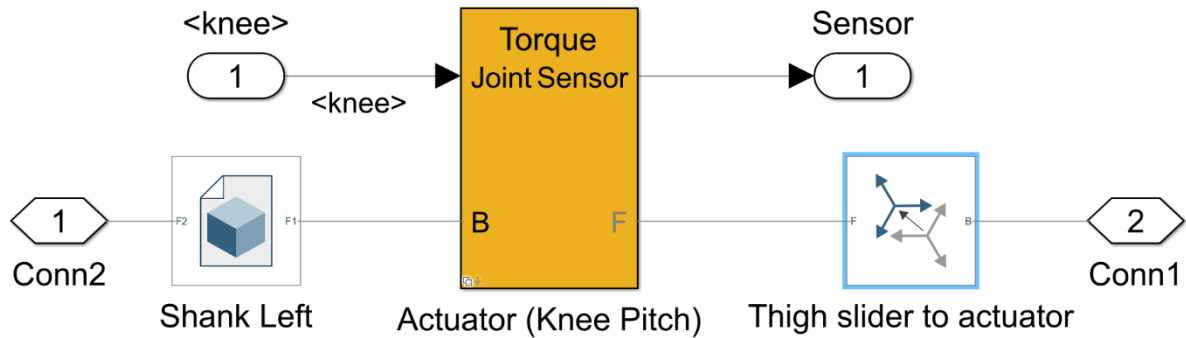


Figure 3.38: Shank system (adapted from Castro, 2019)

3.8.4.2 Foot and contact forces

Contacts are essential in simulations to establish interactions between two or more parts. The forces used during this operation are crucial because they reproduce the robot's realistic physical performance. In Simscape Multibody, there are two types of contact forces: normal and frictional. Both contacts are essential because they serve specific roles in the interactions between bodies. Below are the definitions of these two concepts.

MathWorks (2019) states that the Normal forces prevent two bodies from penetrating each other and are directed perpendicular to the contact surface. In Simulink, a spring-damper system is used to model the normal force. During this process, the damping and stiffness parameters are used to calculate the force magnitude as a function of relative velocity and penetration depth. On the other hand, while normal forces act perpendicular to the contact surface, frictional forces oppose the motion between two bodies and act parallel to the contact surface. These frictional forces can be calculated using the coefficients of static and dynamic friction, along with the normal force acting between the two bodies, to define their mathematical value (Al-Bender, 2005). Figure 3.39 illustrates the contact force used in the Simscape.

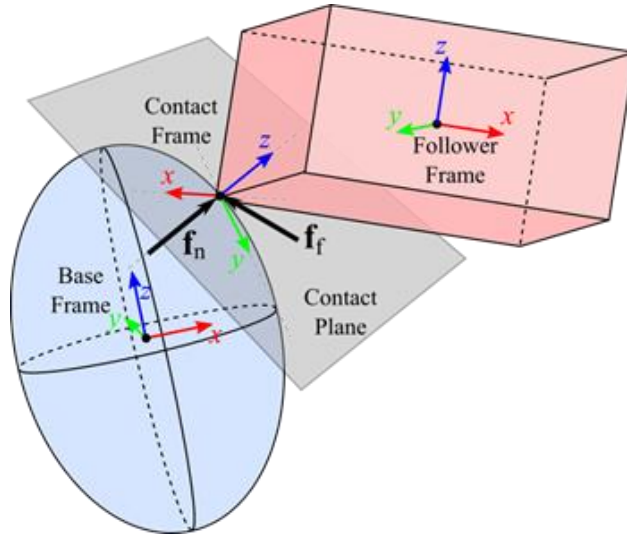


Figure 3.39: Detailed spatial contact force (MathWorks, 2019)

In this research, contacts were introduced to enable the LLE's feet to interact with the ground. This approach was beneficial because it allowed the robot to move in relation to the ground while also defining ground forces and reactions. As previously defined, the spatial contact force block shown in Figure 3.39 was used in Simulink to model contact forces, as illustrated in Figure 3.40. This block applied Newton's third law by exerting equal and opposite forces along the chosen contact plane. The contact block can be adjusted by modifying various parameters, including frictional force, normal force, and separation distance (MathWorks, 2019).

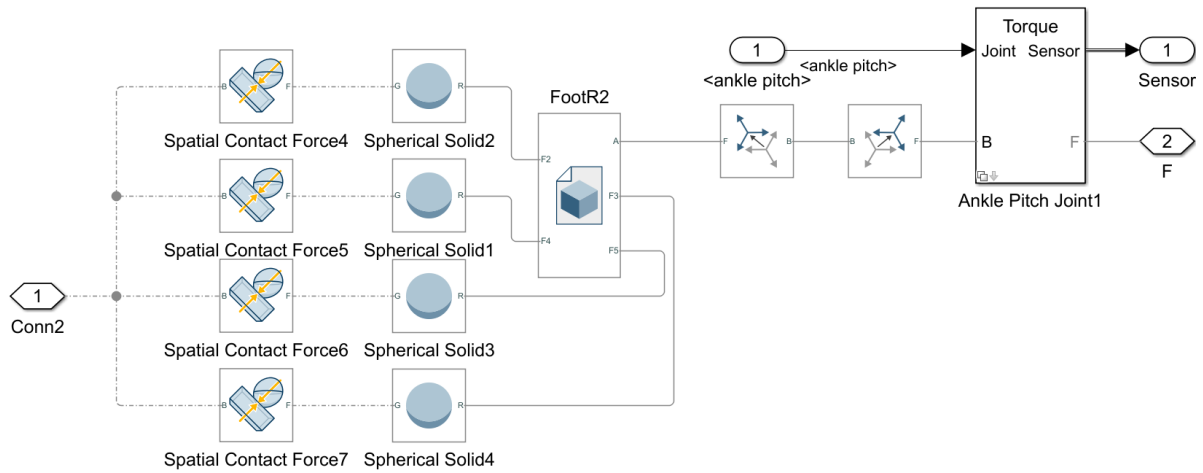


Figure 3.40: Simulink contact schematic (adapted from Castro, 2019)

Each foot of the LLE features four spherical contacts at each corner of the foot. To ensure proper contact with the ground, the radius of each spherical contact was chosen to be 0.0025m. Figure 3.41 shows the contact generated in relation to the Simulink contacts in Figure 3.40. Although there is no strict limit on the number of spherical contacts that can be used, it is recommended not to overuse them, as they may make the simulation complex.

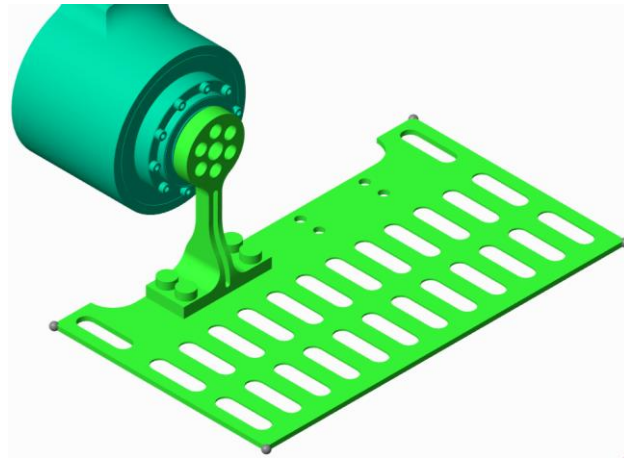


Figure 3.41: Generated contacts on the foot

3.8.4.3 Actuation system

During the transition from conceptual design to final design of the LLE actuation system, three configurations were available, as illustrated in Figure 3.42. Initially, during the conceptual design stage, the motion configuration was considered, in which the robot followed a predefined trajectory. This approach allowed for the observation of the kinematic behaviour of the relevant joint without emphasising actuation efforts. In the preliminary design phase of the actuation system, the torque actuation configuration emerged as a crucial focus, utilising the specified torque of the joint to facilitate motion. Finally, during the final design stage of the joint actuator, the actuation system was developed in greater detail, with an examination of the electrical aspects of the motor, such as current and voltage, to evaluate actuator performance at the drive system level.

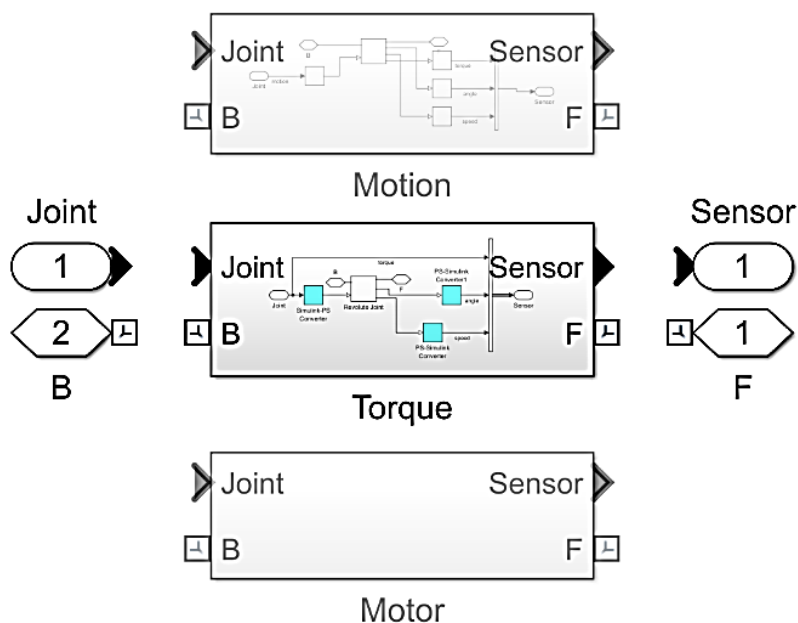


Figure 3.42: Actuation system (adapted from Castro, 2019)

In this design, the joint trajectories of the hip, knee, and ankle were first conceptualised and subsequently implemented through a torque actuation configuration. The aim was to assess whether the joint torques were sufficient to achieve a consistent walking motion. At this stage, the configuration of the motor actuators was not considered, as the focus was on the actuation system's preliminary phase. The primary goal of the torque actuation system was to validate the feasibility of the gait and quantify the required joint torques, rather than specifying the current and voltage needed to produce the desired output torque.

Figure 3.43 provides a detailed view of the torque actuation system. Based on the joint connection characteristics, a revolute joint actuator was selected to enable rotational motion. The Simulink-PS converter was used to transfer data from an input Simulink signal to a Physical System Representation in Simscape. Conversely, the PS-Simulink converter operated as an output converter, enabling the flow of information from a physical signal into a Simulink output signal.

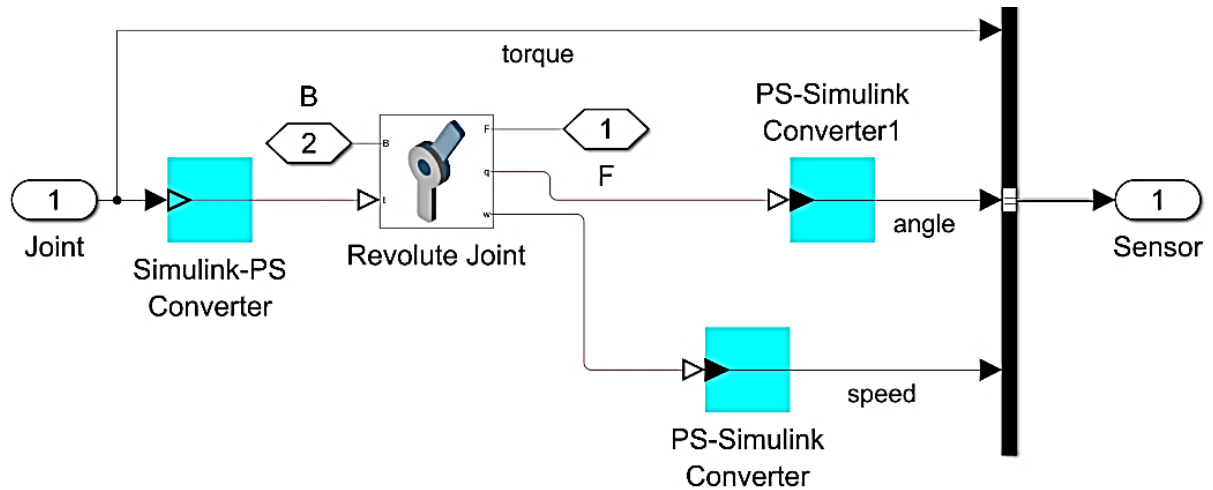


Figure 3.43: The torque actuation system (adapted from Castro, 2019)

3.8.4.4 Controller parameter

The robot parameters listed in Table 3.11 specify the controller settings for the LLE simulation. These include the PID gains and torque limits derived from the inverse dynamics is discussed in subsection 3.10. These parameters were used to calibrate the LLE's walking gait until satisfactory locomotion was achieved with minimal disturbance. Further details of this process are provided in Appendix E.

Table 3.11: Controller parameter

Parameter	Hip	knee	Ankle	Units
Proportional gain (K_p)	9000	7000	5000	Nm/rad
Integral gain (K_i)	2500	2000	1500	$Nm/(rad \cdot s)$
Derivative gain K_d	200	150	100	$Nm \cdot (s/rad)$
Derivative filter coefficient	45	33	22	
Max torque	116.6	114.4	69.2	Nm
Min torque	-113.3	-108.1	-67.1	Nm
Joint damping	18.9			$Nm \cdot s/rad$
World-rot-damping	1130			$Nm/(rad/s)$
World damping	14.7			N/m
Motion time constant	0.0005			sec

3.8.5 The LIPM of the LLE system

3.8.5.1 Overview

Three steps were undertaken to simulate the LLE system using Simulink, listed in the order of their completion.

First, the COM trajectory was analysed, along with the positions of the swing leg and the grounded leg within the polygon region; this step establishes the robot's balance. In the next step, the foot-movement trajectory was generated from the data on the COM and the feet. This trajectory served as the foot's pathway during each phase of the walking cycle, ensuring coordinated movements. The third step involved calculating the robot's joint trajectories using inverse kinematics. This technique is useful for determining the joint angles and configurations needed to achieve the desired foot positions at each point in the walking cycle. Finally, these detailed steps culminated in a cohesive simulation of the walking robot, showcasing the interactions between the COM, foot trajectories, and joint movements. This approach was developed by Castro and Kim from MathWorks in 2019, contributing to progress in robotics simulation.

3.8.5.2 Define the COM

The first step in this task was to simplify the LLE system to its inverted pendulum model (IPM), as illustrated in Figure 3.44. Two main phases can be observed when the LLE system is in motion: the double-support phase and the single-support phase. Considering that the LLE

system is in a single-support phase, its dynamics can be modelled as an IPM, with a weightless link connecting the foot to the LLE's COM. The positions of the point mass p in the x , y , and z directions can be determined using the Equations 3.7 to 3.9, when considering the state variables represented by r , β_r , and β_p as suggested by (Kajita et al., 2001). Note that S_p and S_r stand for $\sin\beta_p$ and $\sin\beta_r$. This simplification makes the equations of motion more straightforward, enabling analysis of the walking robot's dynamics using an IPM.

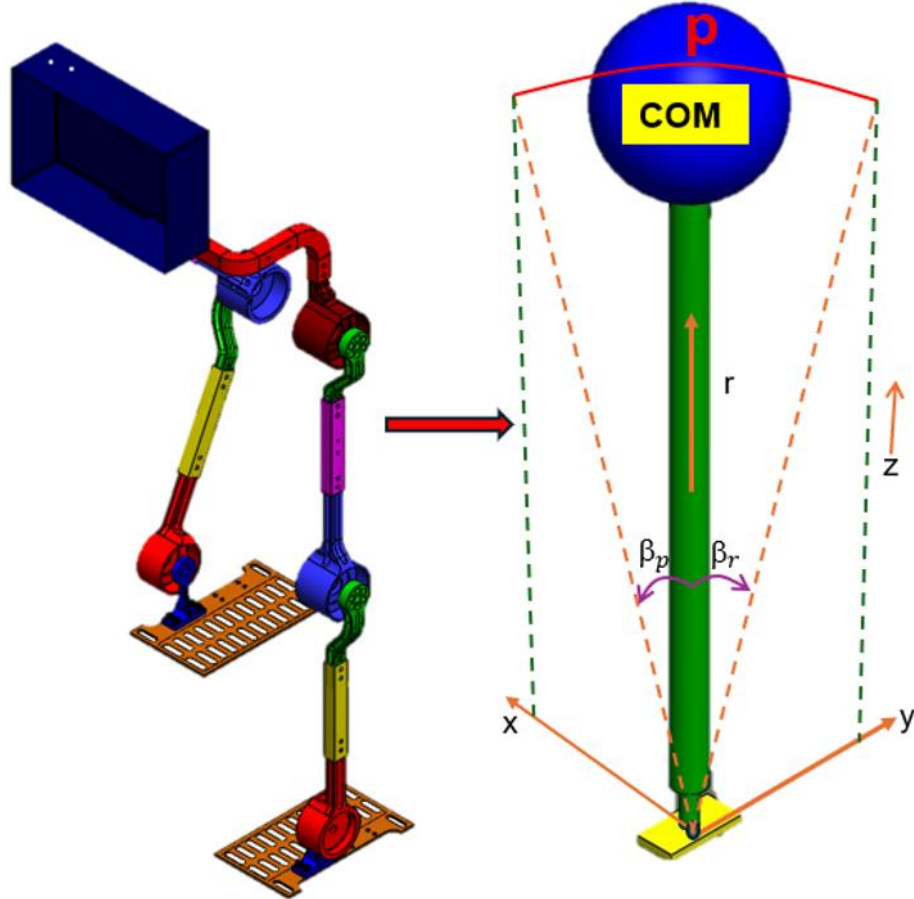


Figure 3.44: Simplification of the LLE to the IPM

$$x = rS_p \quad 3.7$$

$$y = -rS_r \quad 3.8$$

$$z_{com} = r \left(\sqrt{1 - S_r^2 - S_p^2} \right) \quad 3.9$$

The COM of this model followed an arc path during its motion, allowing the dynamic equation of motion for the IPM to be straightforward. However, since the IPM is a nonlinear system, further simplification of its analytical solution was required. Additionally, the IPM could not generate the trajectory during its double-support phase (Li et al., 2021). In addition to not being

a nonlinear system, the IPM has a complex mechanism for generating walking motion (Van Dalen, 2012).

To transition from the IPM to the LIPM, a constrained transverse plane was introduced to the LLE system, represented as a green straight line in the sagittal plane in Figure 3.45; As described by Kajita et al. (2003), the COM of the LIPM is restricted to move exclusively along the defined transverse plane at a constant height above the ground. By applying control constraints to the inverted pendulum, the mass is permitted to move only along a predefined plane, thereby facilitating the transition to the LIPM, also known as a 3-D LIPM.

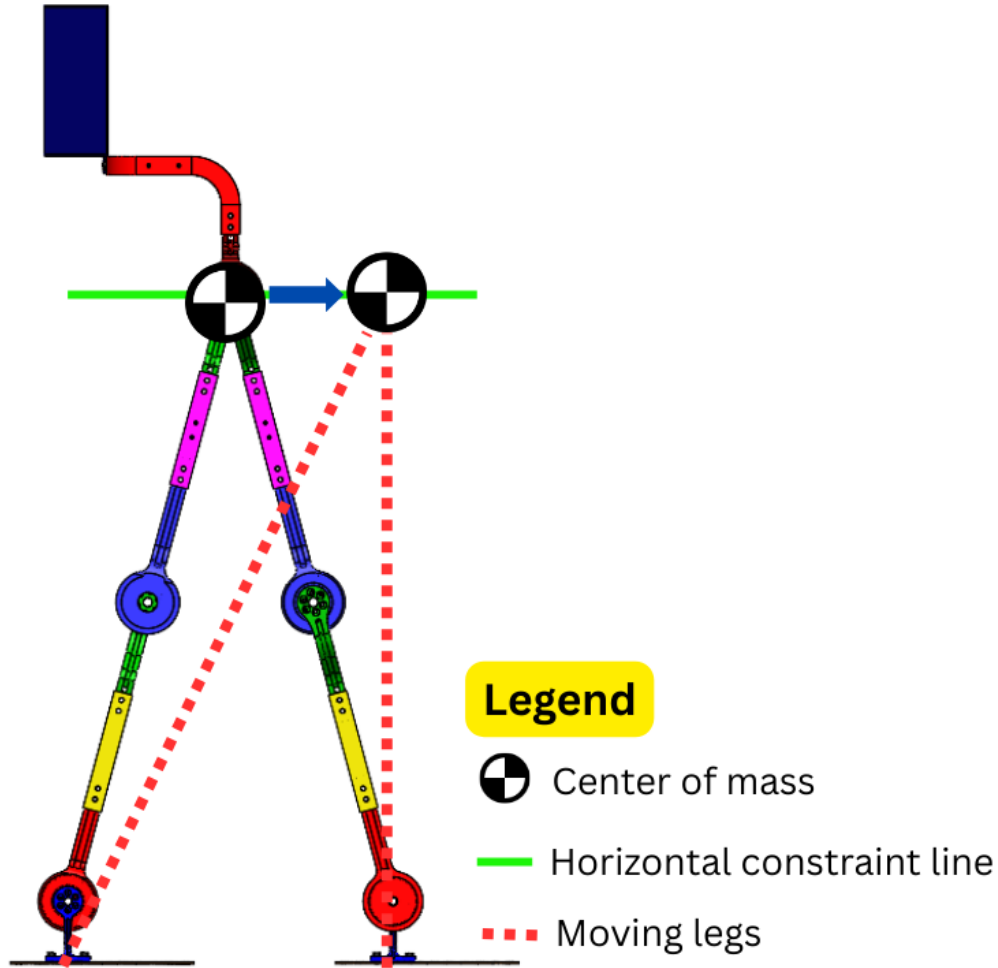


Figure 3.45: Further simplification of the IPM to the LIPM LLE system

Equation 3.10, introduced by Kajita et al. (2001), defines the constraint that maintains the motion of the COM in a specified plane, where its normal vector is $(k_x, k_y, -1)$ and z_c is the z-intercept. The constrained plane is considered horizontal if $k_x = k_y = 0$.

$$z = k_x x + k_y y + z_c \quad 3.10$$

Equation 3.11 was used to determine the kinematic acceleration of the COM in relation to the y-axis. This equation can be further articulated as a representation of the dynamics of the

LIPM, illustrating how the acceleration of the COM is influenced by its displacement from the support point, while being scaled by the effects of gravity and a constant defined height.

$$\ddot{y} = \frac{g}{z_{com}} y \quad 3.11$$

Equation 3.11 is independent of any physical parameters of the body, such as mass and moment of inertia, but solely focuses on the position of the COM and its gravitational acceleration.

Since the COM of the LIPM moves along the defined transverse plane, its energy is related to the conservative potential energy of the orbit. Kajita further developed Equation 3.11 to derive the LIPM orbital energy, as defined in Equation 3.12.

$$E = \frac{1}{2} \dot{y}^2 - \frac{g}{2z_{com}} y^2 \quad 3.12$$

In the LIPM orbital energy (E) equation, y represents the horizontal displacement from the foot relative to the COM, z_{com} denotes the constant height of the COM, and g is the constant acceleration due to gravity.

To ensure that the LLE system followed stable and symmetric trajectories within the LIPM framework, it was fundamental to establish initial conditions, which included step length, the desired lateral velocity of the COM, the initial forward position of the COM, a constant height of the COM, gravitational acceleration, and the time constant.

The application LIPM.mlapp, developed by Castro & Kim (2019) from MATLAB, was used to visualise the behaviour of the LIPM under initial conditions; its use was essential, as it helped to understand LIPM dynamics. Notably, the application defined the LIPM's height as the distance from the ground to the LLE's hip and set the simulation time to 1 second.

Figure 3.46 illustrates the simulated results, showing the pendulum's initial conditions. Here, the variables $x0$ and $y0$ indicate the starting positions of the LIPM on the coordinate plane, while $dx0$ and $dy0$ correspond to the initial velocities along the x and y axes, respectively. This information is important for understanding how the pendulum initiates its motion.

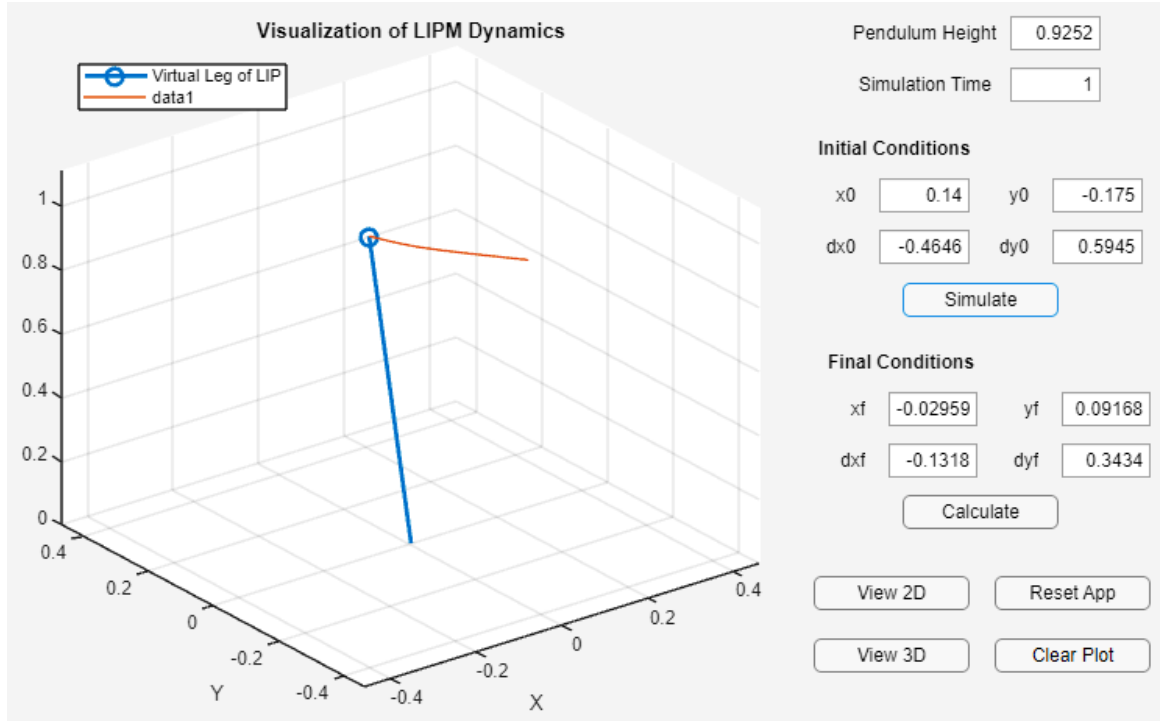


Figure 3.46: Initial conditions and LIPM

3.8.5.3 Trajectory of the COM of the LLE with respect to the world frame

The robot, in its initial state, had a height of 0.9525 meters, denoted as Z_{robot} . To maintain stability during locomotion, the robot slightly bent its knees, resulting in a walking height of 0.8700 meters, designated as Z_{model} . At this stage, its COM was aligned with this walking height, thereby enhancing its stability during movement. The COM simulation of the LLE was created using a discrete-state-space model of the LIPM, as described in Equations 3.13 to 3.16.

$$A = \begin{bmatrix} 0 & 1 & 0 & 0 \\ g/z_{Model} & 0 & 0 & 0 \\ 0 & 0 & 0 & 1 \\ 0 & 0 & g/z_{Model} & 0 \end{bmatrix} \quad 3.13$$

$$B = \begin{bmatrix} 0 & 0 \\ 1 & 0 \\ 0 & 0 \\ 0 & 1 \end{bmatrix} \quad 3.14$$

$$C = \begin{bmatrix} 1 & 0 & 0 & 0 \\ 0 & 0 & 1 & 0 \end{bmatrix} \quad 3.15$$

$$D = \begin{bmatrix} 0 & 0 & 0 & 0 \end{bmatrix} \quad 3.16$$

In matrix A, the first two rows describe the behaviour of the LLE robot with respect to the x-axis. The first row indicates how the position of the robot's link can influence its velocity. In contrast, the second row shows how the robot's position (represented by x) affects its acceleration at different locations. The last two rows follow the same explanation as those for the x-axis, but the focus shifts to the y-axis this time.

In summary, A, B, C, and D are defined as the state, input, output, and feedthrough matrices, respectively. In Equation 3.17, the initial state, referred to as state0, was defined, as this was essential for determining a symmetric trajectory. Furthermore, since the LIPM simulation aimed to replicate the human walking gait cycle, at this stage, the aim was to ensure that the input vectors exhibited a natural dynamic quality.

$$\mathbf{State0} = \begin{bmatrix} x_0 \\ \dot{x}_0 \\ y_0 \\ \dot{y}_0 \end{bmatrix} \quad 3.17$$

Subsequently, the simulation was run after modifying the animateLIPM.mlx file to align with the established LLE parameter. During this process, it became evident that the LIPM's motion was symmetrical. The robot began the simulation by lowering its Zrobot height to match the model height, which represents the target height at which the LLE's LIPM COM is situated during the robot's walking cycle. Figure 3.47 illustrates the unactuated motion of the LIPM.

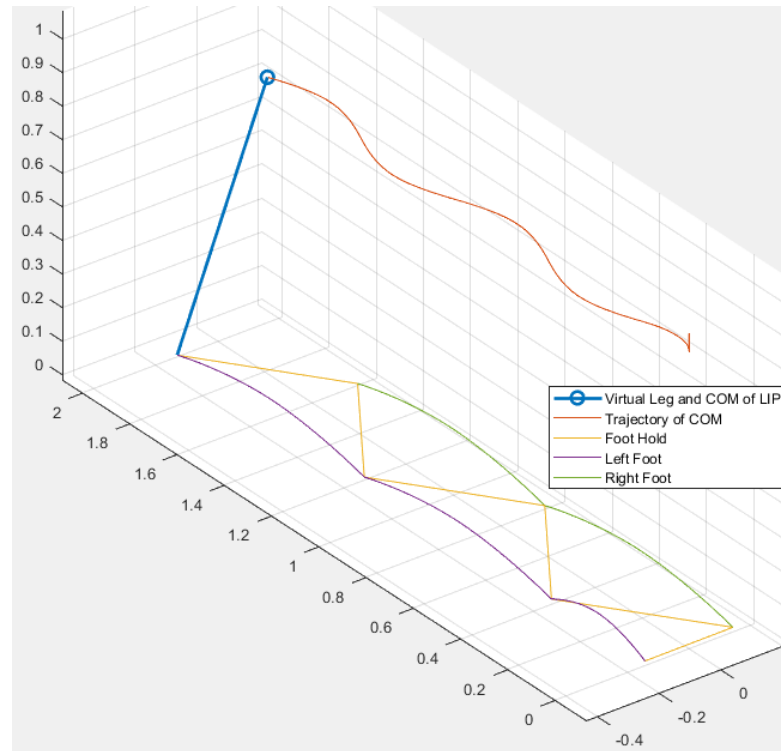


Figure 3.47: LIPM unactuated motion

In the robot simulation presented above, the robot first lowered its height, then moved its left leg halfway along the specified step length. This adjustment enabled the robot to reach its starting position. Subsequently, the robot's movements continued in a consistent pattern. Notably, the trajectories of both swinging feet and the COM were defined with respect to the world frame.

3.8.5.4 Foot (end-effector) trajectory relative to the robot

The foot trajectories were initially defined in the world frame. To clarify foot movements within the LLE context, these trajectories were then transformed to express the positions of the feet relative to the robot. This transformation enabled a more precise analysis of the robot's movement dynamics and its interactions with the surrounding environment. Figure 3.48 shows the foot trajectories.

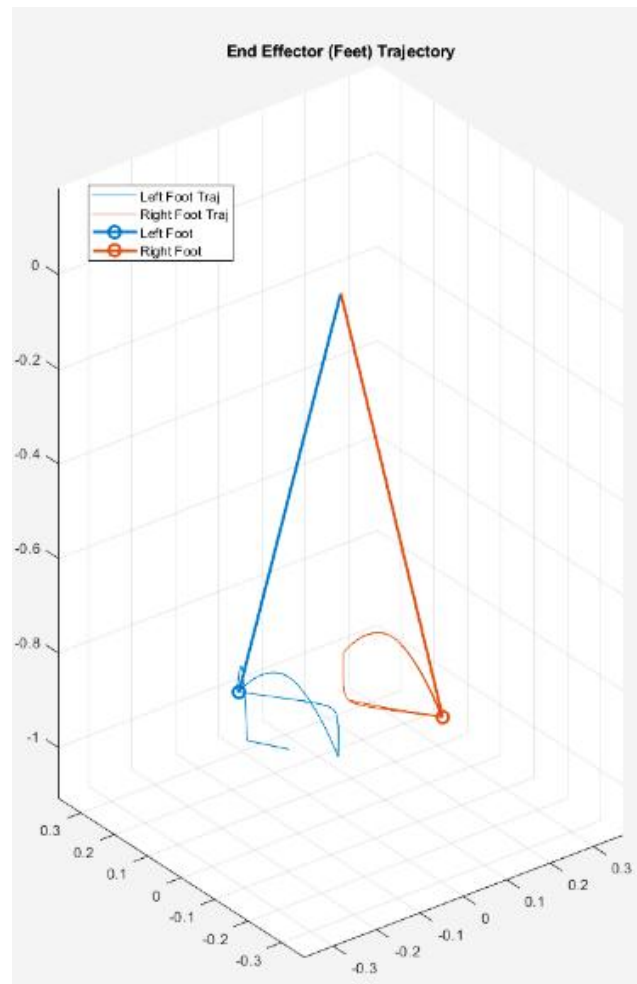


Figure 3.48: 3D end-effector (feet) trajectory

3.8.5.5 Determination of the trajectory of the Joints

After identifying the foot trajectories, the next step was to determine their positions relative to the joint angles. This required determining the forward kinematics first, followed by calculating the trajectories of the remaining joints using inverse kinematics.

3.8.6 Kinematics of the LLE

3.8.6.1 Forward kinematics

An analytical approach was used to perform the forward kinematics calculations. To determine the final position of the robot leg, a combination of three variables, defined as x_1 , x_2 , and x_3 , was used, where these variables represent the robot's movement along the x-axis and correspond to specific joints or segments of the walking robot's limb, contributing to its overall horizontal positioning. A similar approach was used to calculate the y-axis positions of the walking robot's feet, representing the robot's position with respect to the vertical plane. In Equations 3.18 to 3.24 for forward kinematics, L_1 represents the length of the thigh segment, L_2 denotes the length of the shank segment, and L_3 indicates the length from the ankle joint to the ground. Refer to Figure 2.12, where the forward kinematics of a 3R open chain manipulator is given; the same principle was used to define the forward kinematics of the LLE.

$$x_1 = L_1 \cos \theta_1 \quad 3.18$$

$$x_2 = L_1 \cos \theta_1 + L_2 \cos(\theta_1 + \theta_2) \quad 3.19$$

$$x_3 = L_1 \cos \theta_1 + L_2 \cos(\theta_1 + \theta_2) + L_3 \cos(\theta_1 + \theta_2 + \theta_3) \quad 3.20$$

$$y_1 = L_1 \sin \theta_1 \quad 3.21$$

$$y_2 = L_1 \sin \theta_1 + L_2 \sin(\theta_1 + \theta_2). \quad 3.22$$

$$y_3 = L_1 \sin \theta_1 + L_2 \sin(\theta_1 + \theta_2) + L_3 \sin(\theta_1 + \theta_2 + \theta_3) \quad 3.23$$

$$\emptyset = \theta_1 + \theta_2 + \theta_3 \quad 3.24$$

Additionally, the angle at the end effector, where the robotic lower limb interacts with its environment, was determined by summing up the three angles: θ_1 , θ_2 , and θ_3 . Each of these angles represents a rotation at a joint along the LLE and influences the orientation of the end effector.

3.8.6.2 Inverse Kinematics

Since the foot's trajectory relative to the robot was already defined, the remaining joint trajectories were determined by treating the hip joint as the new end-effector of the walking robot. To define the inverse kinematics of the joints, an analytical method was used to

determine the orientations of the hip, knee, and ankle joints. Equations 3.25 and 3.26 are used to define phi located at the wrist.

$$w_x = p_x - L_1 \cos \phi \rightarrow \cos \phi = \frac{w_x - p_x}{-L_1} \quad 3.25$$

$$w_y = p_y - L_1 \sin \phi \rightarrow \sin \phi = \frac{w_y - p_y}{-L_1} \quad 3.26$$

Equations 3.27 and 3.28 are employed to ascertain the orientation of the knee joint, designated as θ_2 . Initially, $\cos \theta_2$ was calculated, followed by the application of the fundamental Pythagorean identity to define $\sin \theta_2$. Once both sine and cosine were established, the $\arctan2$ function, as presented in Equation 3.29, was used to determine θ_2 . For a solution to be considered valid, $\cos \theta_2$ must be less than or equal to 1; otherwise, the end-effector of the walking robot would remain outside its accessible workspace.

$$\cos \theta_2 = \frac{(w_x^2 + w_y^2 - L_1^2 - L_2^2)}{2L_1L_2} \quad 3.27$$

$$\sin \theta_2 = \pm \sqrt{1 - \left(\frac{w_x^2 + w_y^2 - L_1^2 - L_2^2}{2L_1L_2} \right)^2} \quad 3.28$$

$$\theta_2 = \arctan2 \left(\frac{\pm \sqrt{1 - \left(\frac{w_x^2 + w_y^2 - L_1^2 - L_2^2}{2L_1L_2} \right)^2}}{\frac{(w_x^2 + w_y^2 - L_1^2 - L_2^2)}{2L_1L_2}} \right) \quad 3.29$$

After determining θ_2 , the next step was to identify the orientation of the ankle, referred to as θ_1 . Given that both θ_2 and θ_1 produced two practical solutions, it was essential to use the $\arctan2$ function. The expressions for $\sin \theta_1$, $\cos \theta_1$, and $\arctan2$ are presented in equations 3.30 to 3.32.

$$\cos \theta_1 = \frac{w_x(L_1 + L_2 \cos \theta_2) + w_y L_2 \sin \theta_2}{w_x^2 + w_y^2} \quad 3.30$$

$$\sin \theta_1 = \pm \sqrt{1 - \left(\frac{w_x(L_1 + L_2 \cos \theta_2) + w_y L_2 \sin \theta_2}{w_x^2 + w_y^2} \right)^2} \quad 3.31$$

$$\theta_1 = \arctan2 \left(\pm \sqrt{1 - \left(\frac{w_x(L_1 + L_2 \cos \theta_2) + w_y L_2 \sin \theta_2}{w_x^2 + w_y^2} \right)^2}, \frac{w_x(L_1 + L_2 \cos \theta_2) + w_y L_2 \sin \theta_2}{w_x^2 + w_y^2} \right) \quad 3.32$$

θ_3 was derived by subtracting θ_2 and θ_1 from ϕ , as outlined in Equation 3.33.

$$\theta_3 = \phi - \theta_2 - \theta_1 \quad 3.33$$

Using the formulas described earlier, the inverse kinematics for each joint of the LLE were determined. This involved identifying the required angles and positions each joint must adopt to achieve the exoskeleton's intended movements.

The script titled `animateInverseKinematics.mlx` by Castro & Kim (2019) was executed within the MATLAB environment. This script enabled simulations of the exoskeleton's kinematic behaviour, providing visualisation of its responses under defined inverse-kinematic conditions. The simulation yielded valuable insights into the exoskeleton's functioning and dynamics during different movements.

Figure 3.49 depicts the inverse kinematics simulation of the walking robot, illustrating the joint configurations throughout the animation, which mimics the human walking gait cycle.

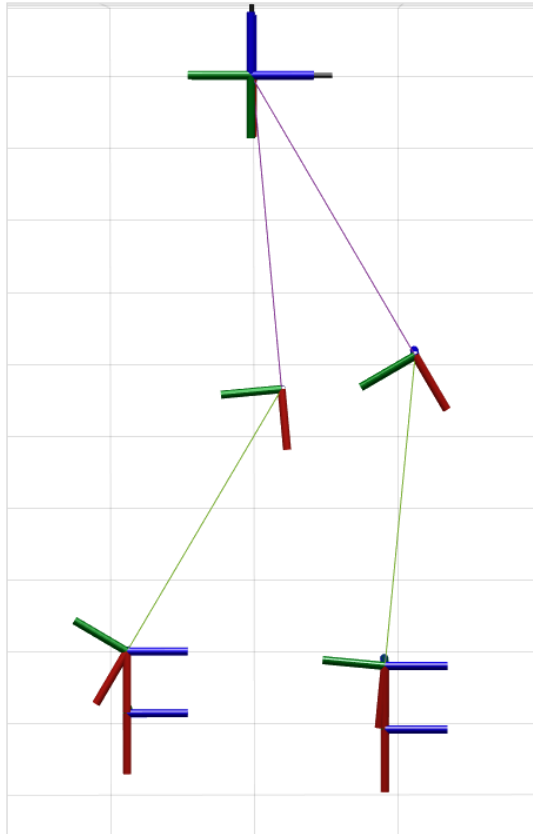


Figure 3.49: 2D walking pattern

Following the completion of the inverse kinematics animation, it was integrated with the Simulink model of the walking robot to verify that the robot accurately exhibited the behaviour predicted by the inverse kinematics simulation.

3.9 Dynamic modelling methodology of the LLE leg

Given the complex morphology of the LLE segments, their COMs are not situated at the midpoint along the y-axis. The COM values for each segment, derived from the material properties obtained through CAD software, will be utilised.

3.9.1 Thigh, shank, and foot mass properties

In this section, the LLE was treated as a two-dimensional object, specifically focusing on the sagittal plane. Due to its symmetrical characteristics along the x-axis, the COM for both the thigh and shank was determined to lie along this axis.

In Figure 3.50, the COM of the thigh along the y-axis is located at -0.2887 m, while its position along the z-axis is 0.0369 m, measured from the origin point. The moment of inertia (MOI) of a mechanical structure affects its stability and rotational dynamic behaviour. Understanding how MOI influences each segment of the LLE provides insights into the dynamics of the thigh, shank, and foot. The mass properties from the CAD file of the thigh segment were used to determine its MOI. p_x , p_y , and p_z are the MOIs associated with the corresponding axis; the notation of these ones will be defined as I_{xx} , I_{yy} , and I_{zz} .

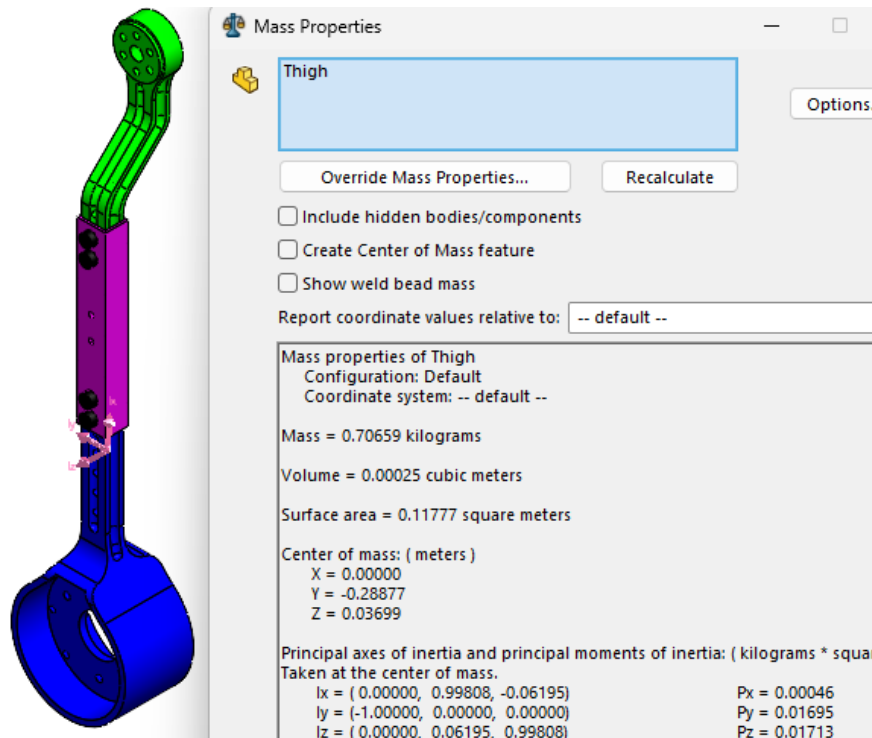


Figure 3.50: Mass properties of the thigh

From the mass properties of the thigh, it is evident that the lowest MOI value occurs along the X-axis. This indicates that the thigh is less resistant to rotation around the X-axis, due its I_{xx} , which is equal to 0.0005 kg.m^2 , meaning only a small amount of torque is needed to start the rolling motion. Conversely, the thigh shows greater stiffness along its Y-axis compared to the X-axis, due to a higher MOI, where I_{yy} is equal to 0.016 kg.m^2 , associated with this axis. As a result, yaw movements of the thigh face more opposition than rolling motions around the X-axis. The greatest resistance is encountered during flexion and extension when rotating around the Z-axis, since this movement involves the largest MOI, where its $I_{zz} = 0.017 \text{ kg.m}^2$.

Mass properties of the shank, as illustrated in Figure 3.51, indicate that the shank exhibits its greatest resistance to rotation during flexion and extension. This resistance is attributable to its I_{zz} of 0.0148 kg.m^2 . The COM for this segment is defined at the intersection where COM_y is -0.2727 m and COM_z is 0.0448 m . It is important to note that COM_x is symmetric, meaning it aligns with the X-axis of symmetry of the shank. Additionally, the MOIs of the shank are lower than those of the thigh, indicating that the torque required at the knee pitch joint is less than that at the hip pitch joint.

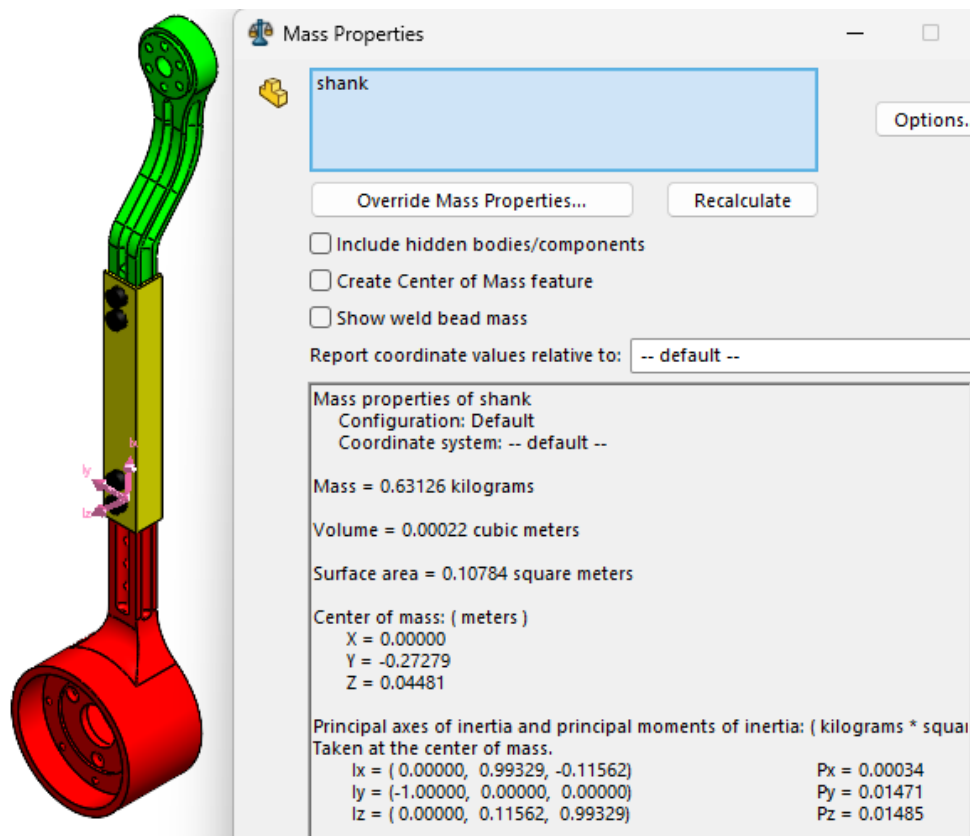


Figure 3.51: Mass properties of the shank

In Figure 3.52, the mass properties of the ankle connector and the base foot are depicted in detail. Notably, the COM relative to the X-axis, Y-axis, and Z-axis are measured as 0.061 , -0.068 , and 0.0255 meters, respectively, from the origin point. Like the thigh and shank, the foot

exhibits the greatest resistance to rotation around the z-axis, as evidenced by its highest MOI when compared to the values associated with the y-axis and x-axis.

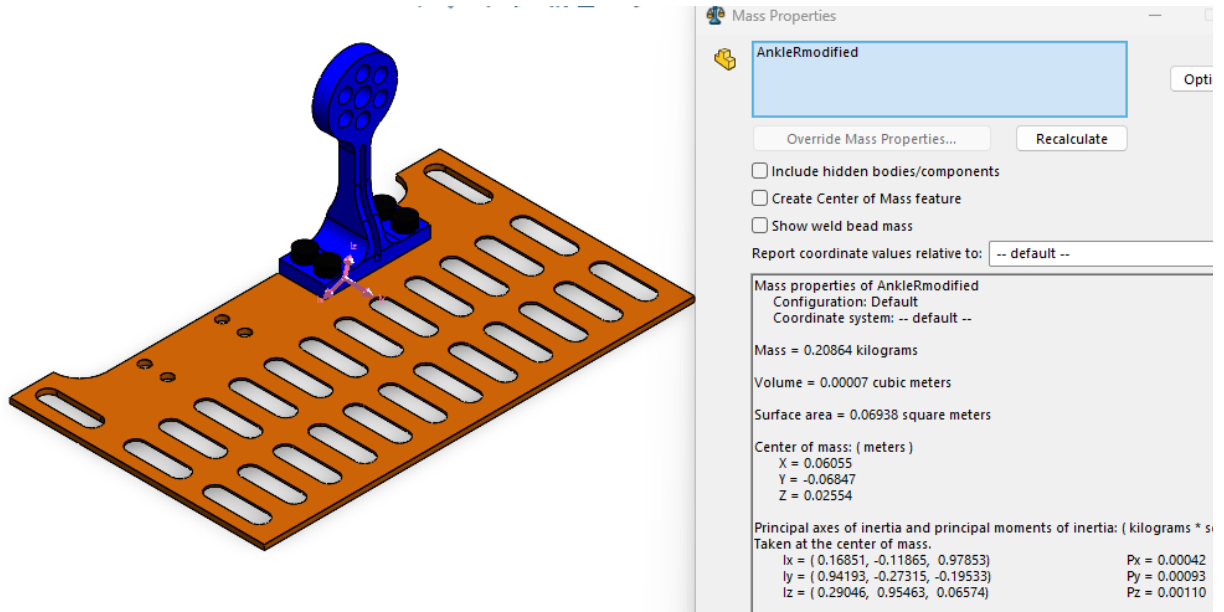


Figure 3.52: Mass properties of the ankle connector

The specified COMs and MOIs are useful variables in the equations of motion and the Lagrangian formulation. Incorporating the obtained COMs and MOIs into the associated equations helps to determine the torque related to each joint.

3.9.2 Inverse dynamics of the LLE

Figure 3.53 presents the Inverse dynamic model of the leg of the walking robot. While the primary focus of this research is not solely on the robot's stability, it was essential to define the LLE's dynamics, as this information is pivotal for designing its control system.

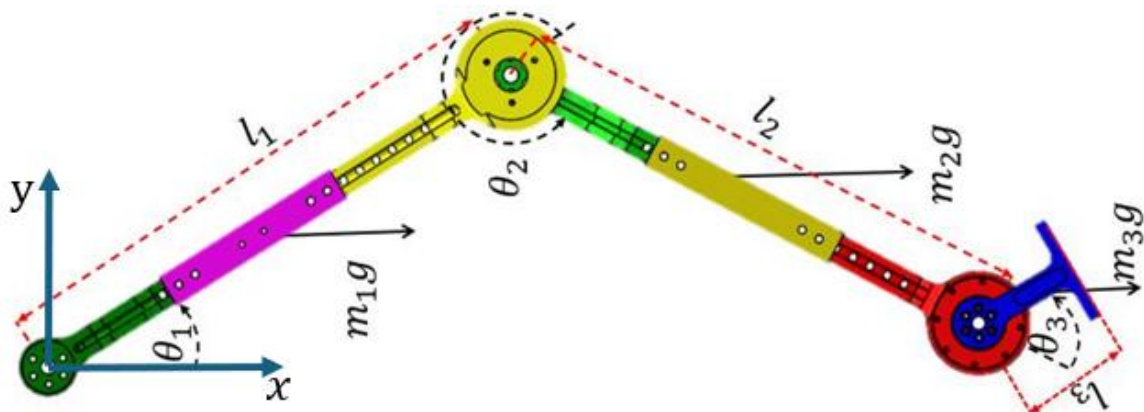


Figure 3.53: Dynamic modelling

The Lagrangian is defined as the difference between the kinetic energy and the potential energy of the LLE leg. Once the Lagrangian was established, it was incorporated into the Euler-

Lagrange equation to determine the required torque for each joint segment. The Euler-Lagrange equation is examined in the following paragraph. However, prior to that discussion, the location of the COM for each segment was identified along with its corresponding velocity, which was obtained by differentiating the position of the COM. In Equations 3.34 to 3.45 and 3.48 to 3.50, the COMs of the thigh, shank, and foot are denoted as $x_1, x_2, x_3, y_1, y_2,$ and y_3 respectively, while $\dot{x}_1, \dot{x}_2, \dot{x}_3, \dot{y}_1, \dot{y}_2,$ and $\dot{y}_3$ represent the velocities of the COM of the thigh, shank, and foot, respectively. In these equations, $l_{c1}, l_{c2},$ and l_{c3} denote the distances to the COM from the hip, knee, and ankle, respectively.

- For the thigh

$$x_1 = l_{c1} \cos(\theta_1) \quad 3.34$$

$$y_1 = l_{c1} \sin(\theta_1) \quad 3.35$$

$$\dot{x}_1 = -l_{c1} \sin(\theta_1) \dot{\theta}_1 \quad 3.36$$

$$\dot{y}_1 = l_{c1} \cos(\theta_1) \dot{\theta}_1 \quad 3.37$$

$$v_1^2 = \dot{x}_1^2 + \dot{y}_1^2 \quad 3.38$$

- Shank

$$x_2 = l_1 \cos(\theta_1) + l_{c2} \cos(\theta_1 + \theta_2) \quad 3.39$$

$$y_2 = l_1 \sin(\theta_1) + l_{c2} \sin(\theta_1 + \theta_2) \quad 3.40$$

$$\dot{x}_2 = -l_1 \sin(\theta_1) \dot{\theta}_1 - l_{c2} \sin(\theta_1 + \theta_2) (\dot{\theta}_1 + \dot{\theta}_2) \quad 3.41$$

$$\dot{y}_2 = l_1 \cos(\theta_1) \dot{\theta}_1 + l_{c2} \cos(\theta_1 + \theta_2) (\dot{\theta}_1 + \dot{\theta}_2) \quad 3.42$$

$$v_2^2 = \dot{x}_2^2 + \dot{y}_2^2 \quad 3.43$$

- Foot

$$x_3 = l_1 \cos(\theta_1) + l_2 \cos(\theta_1 + \theta_2) + l_{c3} \cos(\theta_1 + \theta_2 + \theta_3) \quad 3.44$$

$$y_3 = l_1 \sin(\theta_1) + l_2 \sin(\theta_1 + \theta_2) + l_{c3} \sin(\theta_1 + \theta_2 + \theta_3) \quad 3.45$$

Let us say:

$$o_3 = -l_1 \sin(\theta_1)\dot{\theta}_1 - l_2 \sin(\theta_1 + \theta_2)(\dot{\theta}_1 + \dot{\theta}_2) \quad 3.46$$

$$k_3 = l_1 \cos(\theta_1)\dot{\theta}_1 + l_2 \cos(\theta_1 + \theta_2)(\dot{\theta}_1 + \dot{\theta}_2) \quad 3.47$$

Equations 3.46 and 3.47 are parts of Equations 3.48 and 3.49; the researcher did this for simplicity purposes.

$$\dot{x}_3 = o_3 - l_{c3} \sin(\theta_1 + \theta_2 + \theta_3)(\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) \quad 3.48$$

$$\dot{y}_3 = k_3 + l_{c3} \cos(\theta_1 + \theta_2 + \theta_3)(\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) \quad 3.49$$

$$v_3^2 = \dot{x}_3^2 + \dot{y}_3^2 \quad 3.50$$

In Equations 3.34 to 3.45 and 3.48 to 3.50, the length of the thigh is denoted by l_1 , the shank length is identified by l_2 , and the l_3 indicates the distance from the ankle joint to the ground. Once the COM and their relevant velocities were established, the following step involved determining the potential energy and kinetic energy for each link.

3.9.2.1 Potential energy (V)

The potential energy of the leg is influenced by the mass of each segment (the thigh, knee, and foot) and their respective COM along the y-axis. The total potential energy of the leg is the sum of the potential energies of the thigh, knee, and foot, as described in Equation 3.51. Please note that indices 1, 2, and 3 refer to the thigh, shank, and foot.

$$V = m_1 g y_1 + m_2 g y_2 + m_3 g y_3 \quad 3.51$$

3.9.2.2 Kinetics energy(T)

To ascertain the kinetic energy of the leg in an LLE, it is essential to consider both the translational and rotational kinetic energies. The total kinetic energy of the exoskeleton leg, which encompasses the thigh, shank, and foot, is illustrated in Equation 3.52. In this equation, m signifies the mass of each segment, v indicates the translational velocity of the COM, I represents the moment of inertia of the segment, and $\dot{\theta}$ denotes the angular velocity at the joint. In the given formula below, I_1 represents the MOI I_{zzt} , I_2 represents I_{zzs} , and I_3 represents I_{zzf} .

$$T = \frac{1}{2} m_1 v_1^2 + \frac{1}{2} I_1 \dot{\theta}_1^2 + \frac{1}{2} m_2 v_2^2 + \frac{1}{2} I_2 (\dot{\theta}_1 + \dot{\theta}_2)^2 + \frac{1}{2} m_3 v_3^2 + \frac{1}{2} I_3 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3)^2 \quad 3.52$$

3.9.2.3 Lagrangian (L)

After defining the potential and kinetic energies, the next step was to establish the Lagrangian by subtracting the potential energy from the kinetic energy, as defined in Equations 3. 53 and 3.54.

$$L = T - V \quad 3. 53$$

$$L = \left(\frac{1}{2}m_1v_1^2 + \frac{1}{2}I_1\dot{\theta}_1^2 + \frac{1}{2}m_2v_2^2 + \frac{1}{2}I_2(\dot{\theta}_1 + \dot{\theta}_2)^2 + \frac{1}{2}m_3v_3^2 + \frac{1}{2}I_3(\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3)^2 \right) - (m_1gy_1 + m_2gy_2 + m_3gy_3) \quad 3. 54$$

3.9.2.4 Equation of motion

Once the Lagrange was defined, the Euler-Lagrange equation was used to determine the torque of each link of the robot. The Euler-Lagrange equation was already described in the literature review in Equation 2.10, and it is repeated below here as Equation 3.55. τ_i is the torque with respect to the concerned joint i . Equation 3.56, previously defined in the literature review as Equation 2.8, represents the equation of motion for the LLE.

$$\frac{d}{dt} \left(\frac{\partial L}{\partial \dot{\theta}_i} \right) - \frac{\partial L}{\partial \theta_i} = \tau_i, i = 1,2,3 \quad 3. 55$$

$$M(\theta)\ddot{\theta} + C(\theta, \dot{\theta})\dot{\theta} + G(\theta) = \tau \quad 3. 56$$

The mass matrix $M(\theta)$, Coriolis and centrifugal terms $C(\theta, \dot{\theta})$, and the gravity vector $G(\theta)$ used in MATLAB are detailed in Appendix E, and the complete derivation of the Euler-Lagrange equation related to the LLE is presented in Appendix I. After defining these variables and incorporating them into the equations of motion outlined in Equation 3.56, the torques associated with each joint were defined as follows:

- Hip torque: 104.516 Nm
- Knee torque: 32.428 Nm
- Ankle torque: 2.500 Nm

These obtained torques were found satisfactory and confirmed that the selected motors were up to the task; this is discussed in depth in subheading 3.10.4.

3.10 Conceptual Methodology for Control System

3.10.1 Introduction

This section discusses the conceptual design of a control system for the LLE. The control system plays a key role in operating the LLE by ensuring harmonised and synchronised operation.

The assist-as-needed (AAN) control strategy is the foundational principle of the LLE's control system. This approach supports users by using a pressure sensor to detect applied forces. During operation, the motor adjusts the robot's torque in response to torque requirements detected by the sensor.

3.10.2 Control architecture

The control system was based on an established three-level control principle to improve the smoothness of the LLE's walking motion. In the lower-level control, the emphasis was on torque and current regulation. Proportional-integral-derivative (PID) controllers were used for torque control, enabling precise regulation of the torque generated by each motor for its specific joint. Current control uses motor drives to produce the necessary current to achieve the desired torque.

The mid-level control incorporated an inverse kinematics algorithm, which was essential for determining the position and velocity of each joint and link of the LLE. This approach also considered the robot's desired gait pattern.

At the high-level control, the researcher defined trajectory planning and balance control. Trajectory planning involved setting parameters such as step length, swing height, step time, step width, swing phase duration, joint angles, ground clearance, ZMP, and COM. Balance control used pressure sensors to maintain stability during the LLE's walking motion.

3.10.3 Microcontroller Selection

The Arduino platform is renowned for its user-friendly interface and extensive community support, making it an ideal choice for beginners or simple applications. However, its processing speed and memory limitations can be constraints in more complex scenarios, such as LLE (Monk, 2016). Consequently, the microcontroller selected for the LLE was required to exhibit superior performance relative to the Arduino microcontroller, while maintaining compatibility with the Arduino system. Benjamin (2024) states that the Teensy 4.1 is a 32-bit microcontroller device operating at a 3.3V logic level. It is considered more powerful than an 8-bit Arduino microcontroller. Compared to other 32-bit boards, the Teensy offers two distinct advantages. Firstly, it uses the same IDE as Arduino through an add-on called Teensyduino. Secondly, there

is a vast collection of libraries specifically designed for the Teensy microcontroller, optimised to enhance performance.

For this project, the Teensy 4.1 was chosen over the Arduino not only for its advanced capabilities but also for its optimal performance in real-time applications, thereby ensuring a seamless experience for LLE users. Williams et al. (2024) Used a Teensy microcontroller to control the development of the OpenExo, as it offers a good analogue-to-digital converter and features a fast clock. In the development of the assistive lower limb exoskeleton (ALEXO), Yu et al. (2023) selected Teensy for the developed exoskeleton, which processes data from its knee and hip joints.

3.10.4 Motors and harmonic drive

Based on the torque calculated for each joint of the LLE and the MOI implications for each segment in the sagittal plane, suitable motors and harmonic drives were selected for each joint, as their combination provides sufficient torque for the task.

Since the thigh has the highest MOI and a calculated torque of 104.526 Nm, the selected motor for this joint should be capable of overcoming the MOI, which resists rotation of the thigh in the sagittal plane. Consequently, a Maxon EC 90 flat brushless motor, each with a power output of 220 W, coupled with a 160:1 LSG Laifual harmonic drive, was selected for this task. With a generated torque of 116.6 Nm, this configuration provided the hip actuator with a 10.356% safety margin.

The same motor and harmonic drive used for the hip joint were selected for the knee joint, albeit with a different torque specification. This configuration delivered a torque of 114.4 Nm, yielding a safety margin of 71.654% relative to the calculated torque of 32.428 Nm.

Finally, the ankle joints are powered by EC 60 flat brushless motors, each with a diameter of 60 mm and a power output of 200 W. These motors were paired with a 120:1 LSG and a Laifual harmonic drive, resulting in a combined torque of 69.24 Nm, providing a safety margin of 96.389%. This torque was determined to be sufficient to overcome the moment of inertia and ensure safe rotation of the foot.

One key advantage of these motor selections is that they come with integrated encoders, enabling precise control and feedback on joint position and movement. This integration enhances the exoskeleton's overall functionality and responsiveness, contributing to a more intuitive and supportive user experience.

3.10.5 Motor drives

The motor drives were selected based on their power ratings to guarantee optimal performance with the motors. Drives rated up to 250 W were chosen since they effectively support both the

220 W and 200 W motors designated for each joint. The Maxon ESCON module 50/5, depicted in Figure 3.54, met these requirements and was therefore chosen for the task.



Figure 3.54: ESCON 50/5 (Maxon, 2015)

3.10.6 Force sensitive resistor (FSR)

Dynamic and static forces can be measured using an electronic device known as a force-sensing resistor (FSR), which detects the reaction force between the ground and the applied force. The device's response depends on variations in its electrical resistance. Two well-known FSR brands available on the market are Interlink and Flexiforce (Sadun et al., 2016). The FSR sensors are compatible with the LLE's functionality. They can be attached to the foot sole of the LLE to define the ground reaction forces exerted by the user and the LLE system (Yang et al., 2020). Tekscan is an American company based in Norwood, Massachusetts, which has demonstrated excellence in the research and development of ultra-thin force and pressure-sensing technology since its founding in 1981. The company offers a range of sensors and systems suitable for various applications, including medical, dental, and test-and-measurement. These sensors are designed to function as either embedded technologies or stand-alone solutions, effectively distinguishing their products in the market (Tekscan, 2018). In their project, Jung et al. (2015) used a set of eight Tekscan Flexiforces on each foot of the LLE robot. These sensors were used to determine if the foot was in contact with the ground.

In this conceptual design for the LLE control system, each foot was equipped with five A301 FSR sensors from Tekscan. These advanced devices accurately detect which leg is in contact with the ground during movement. When a leg is identified as grounded, the control system processes this information quickly and sends a command to initiate movement in the opposite

leg. This synchronised operation facilitates smooth, coordinated locomotion, ensuring the exoskeleton responds effectively to the user's movements and enhancing overall mobility. The FSR sensor attached to a foot insole is illustrated in Figure 3.55.



Figure 3.55: Flexiforce A301 sensors incorporated into an insole (Tekscan, 2023)

A load cell amplifier, designated HX711, was chosen to improve the sensitivity of the FSR sensor measurements, resulting in more accurate results. This device operates within a voltage and current range of 2.7 to 5V and 1.5A; HX711 is shown in Figure 3.56.

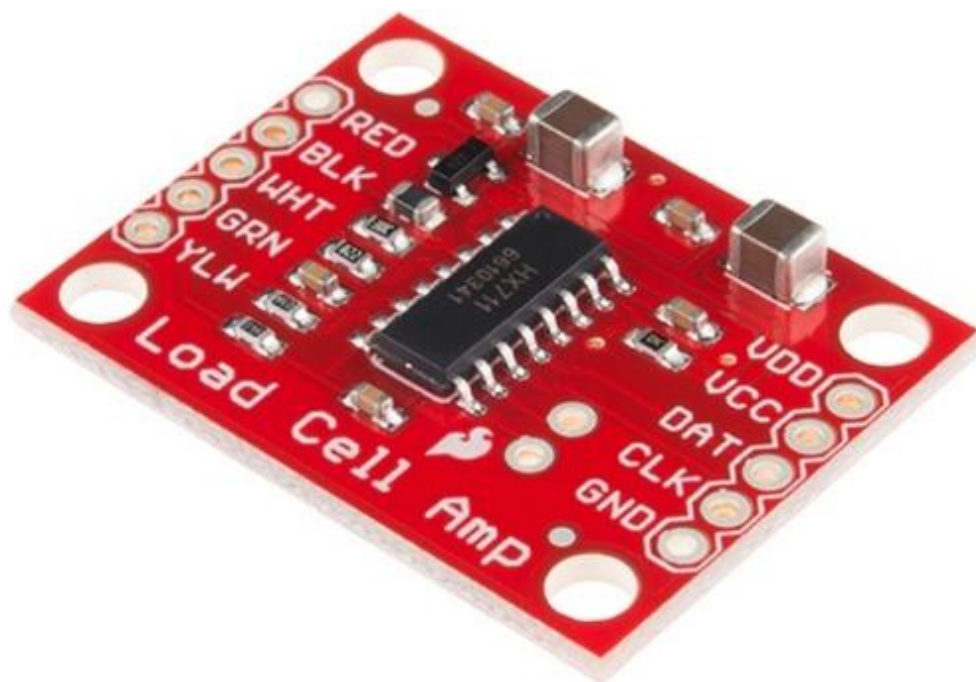


Figure 3.56: Load cell amplifier, HX711(Alex & Sarah, 2015)

3.10.7 Battery selection

To determine the proper battery to sustain the LLE's power for a specified duration, the initial step was to calculate the total electrical power for each leg of the system. The sum of these

two legs was then used to establish the input power required to ensure the LLE's motion. Table 3.12 summarises the power and voltage requirements for each device used in the LLE.

Table 3.12: The LLE power requirement for battery selection

Quantity	Device	Power (W)	Voltage (V)	R&L power (W)
2	Hip motors	220	24	440
2	Knee motors	220	36	440
2	Ankle motors	200	48	400
2	HX711	7.5	7.5	15
2	ESCON Module 50/5 (Hip)	220	50	440
2	ESCON Module 50/5 (Knee)	220	50	440
2	ESCON Module 50/5 (Ankle)	200	50	400
2	Teensy 4.1	0.33	3.3	0.66
12	Flexiforce A301	0.25	1.25	2.5
	Total			2578.16

Using Equation 3.57, the energy required of 2.57816 kWh was determined by multiplying the input power of 2.57816 kW by one hour.

$$E_{required} (w) = p_{required} * t \quad 3.57$$

Each leg of the walking robot required 1289.08 kW to operate; therefore, the calculation will be conducted per leg. The necessary battery capacity (Ah) was calculated using Equation 3.58. In this equation, V represents the voltage of the selected battery, DOD denotes the depth of discharge, and η refers to the battery's efficiency.

$$Capacity_{robot} = \frac{W_{robot}}{V \times DoD \times \eta} \times safety\ margin \quad 3.58$$

$$Capacity_{robot} = \frac{1289.08}{64.2 \times 0.9 \times 0.9} \times 1.2 = 29.471\ Ah/leg$$

The calculations above indicated that the robot has a capacity of 29.471 Ah; therefore, a higher-capacity battery was chosen. A Maxamps 30000mAh 18S6P 64.8V lithium-ion battery pack was selected, providing an additional margin of 1.763% for full coverage.

The discharge rate (C-rate) of the selected battery was calculated using Equations 3.59 and 3.60. The current related to the robot was obtained by dividing the required energy by the battery voltage, and the resulting current was divided by the selected battery's capacity.

$$\text{Current } (I_{robot}) = \frac{W_{robot}}{V_{battery}} \quad 3.59$$

$$\text{Current } (I_{robot}) = \frac{1289.08}{64.8} = 19.893 \text{ A per leg}$$

$$C - \text{rate} = \frac{\text{Current } (I_{robot})}{\text{Capacity } (Ah_{battery})} \quad 3.60$$

$$C - \text{rate} = \frac{19.893}{30} = 0.6631C \text{ per leg}$$

The discharge rate indicated that the selected battery per leg can support each leg of the walking robot for 1.508 hours (1 hour 30.484 minutes) at full capacity. The operation cycle duration was determined by dividing 1 hour by the discharge rate.

Since the LLE's electronic components operate at lower voltages than the selected battery, buck converters would be used to step down from 64.8 V to the lower voltages required by the other devices. Figure 3.57 illustrates the Maxamps 30000mAh 18S6P 64.8 V lithium-ion battery pack. It is important to note that the selected battery weight is 7.714 kg per leg. In conclusion, the selected battery is adequate to power the LLE for 1 hour and 30.484 minutes; more details on this battery are available in Appendix G.



Figure 3.57: Li-ion 30Ah 18S6P 64.8 v battery (MaxAmps Lithium Batteries, 2022)

3.10.8 Conceptual control system schematic

Following the establishment of the control architecture and the selection of necessary components for the LLE's control system, which is currently in the conceptual design phase, a computer equipped with MATLAB and Simulink will be used, as shown in Figure 3.58. Two Teensy microcontrollers act as intermediaries between the computer and the other components of the control system, enabling the execution of code from the computer. Six ESCON module 50/5 Maxon motor drives are used to connect a Maxon motor to the Teensy 4.1. Two load cell amplifiers, designated as HX711, are employed to increase the sensitivity of the connected sensors, A301. Two 30 Ah, 64.8 V batteries are used to power the walking robot, with each leg connected to its own battery.

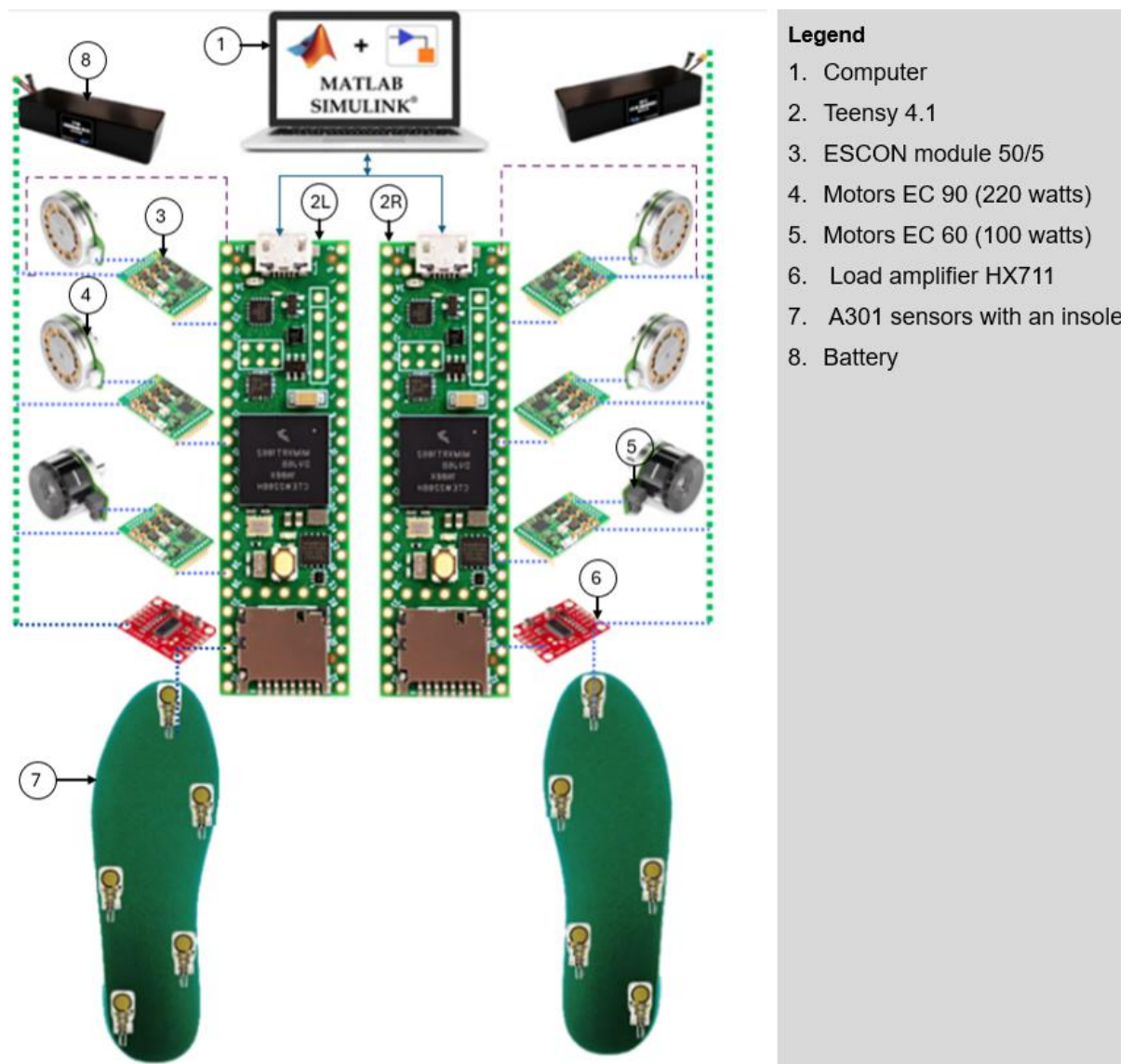


Figure 3.58: Conceptual control system

CHAPTER 4: DISCUSSION OF RESULTS

This chapter discusses the results from LLE simulations, focusing on two aspects. First, a static structural analysis assesses component integrity. Second, MATLAB and Simulink simulations evaluate LLE walkability by examining constraints on the robot's pitch joints, analysing angles and torques at the hip, knee, and ankle.

4.1 Structural analysis

The results are analysed in two gait cycle scenarios, considering Von Mises stress, safety factor, and total deformation. The first scenario examines the left leg in mid-stance under compressive axial load, using buckling analysis. The second scenario evaluates the same leg during terminal swing, allowing the thigh and shank to reach a 90-degree angle parallel to the ground, with transverse forces applied as needed. This scenario focuses on bending stress assessment.

It is important to emphasise that the discussion of the FEA results in this section will focus solely on the lower-limb segments, with the subsequent section of this subheading addressing this further. Appendix D presents the relevant FEA results for the lower torso.

4.1.1 Compressive Axial Loading (mid-stance)

The thigh and shank were divided into two segments for simulation purposes: the upper and lower segments. An axial compressive force of 983.89 N was applied to the thigh components, and 1056.7 N to the shank. Conversely, the foot was analysed as a single entity subjected to an axial compressive load of 1082.4 N. The Von Mises stress, safety factor, and total deformation of each part were derived from the simulation results to evaluate potential material behaviour. The buckle load (P_b) for each component was calculated by multiplying the applied load by the load multiplier obtained from the simulation, whereas the critical buckling load (P_c) was determined using Equation 4.1, where, E represents Young's modulus, I pertains to the moment of Inertia of the cross-section, L signifies the length, and K Indicates the effective length factor. These calculations were performed to evaluate the risk of buckling failure in each component. It is important to note that this leg is in mid-stance.

$$P_c = \frac{\pi^2 * E * I}{(K * L)^2} = \frac{\pi^2 * E * I}{4L^2} \quad 4.1$$

4.1.1.1 Thigh

a. Upper thigh

In Figure 4.1, the illustration, arranged from left to right, shows the Von Mises stresses on the frame and bolts, along with the FOS for the upper thigh. The analysis displays that the maximum Von Mises stress of 189.01 MPa, occurring under a compressive axial load of 983.89 N, was found at the bolt hole on the thigh tube, where fixed boundary conditions were applied. The components were analysed independently to assess the maximum Von Mises stress experienced by the fasteners and the frame. The investigation revealed that one of the bolts connecting the hip connector to the thigh tube experienced a Von Mises stress of 149.47 MPa. Notably, the Von Mises stresses remained below the yield strengths of 276 MPa for Aluminium 6061-T6 and 340 MPa for low-carbon steel. The maximum Von Mises stress on the upper thigh was 189.01 MPa, providing a key reference for assessing the integrity of this segment. Under these conditions, the minimum safety factor for the upper thigh was calculated to be 1.460, which was deemed acceptable. This indicated that the frame was designed not only to support the user's weight and the LLE but also to endure an additional 46.03% of the applied load. Since the applied load on the upper thigh was axial, the bolts connecting the hip connector to the thigh tube were subject to shear stress, as the applied load pushes these two parts against the bolts.

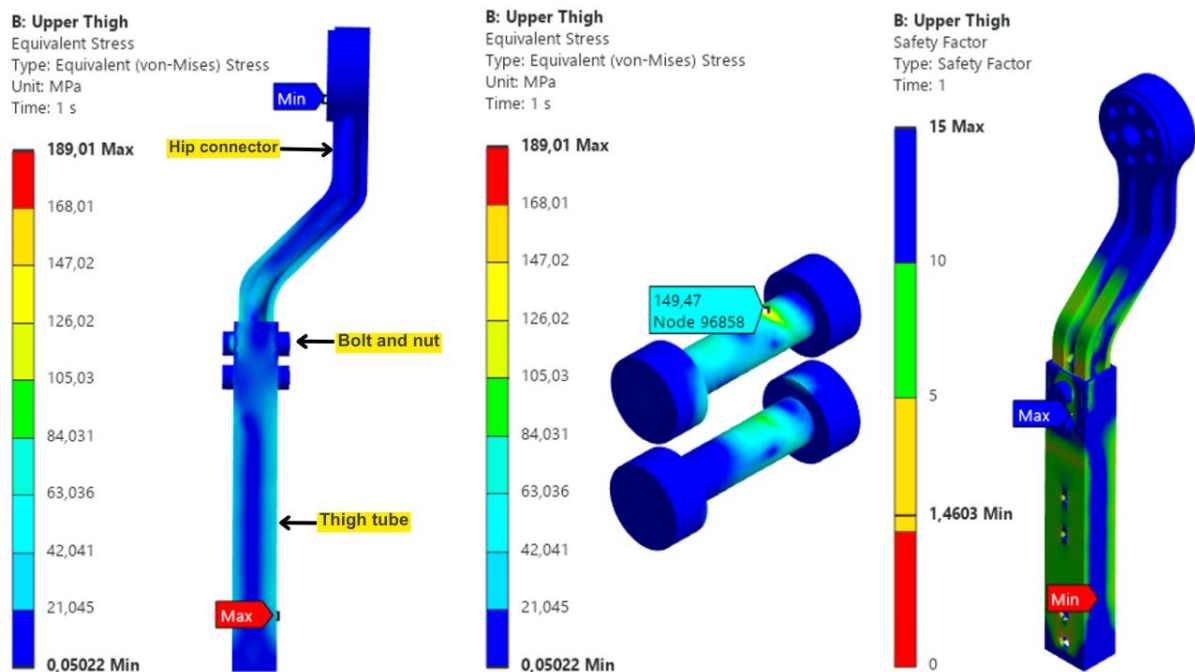


Figure 4.1: Von Mises stress and FOS of the upper thigh related to Axial loading

The factor of safety was further validated using Equation 4.2. This equation compares the material's yield strength (specifically, aluminium 6061-T6) with the Von Mises stress acting on

it by calculating their ratio. This comparison confirmed that the thigh segment met the safety criteria, demonstrating that the applied load remained within the material's elastic limit.

$$FOS = \frac{S_y}{\sigma} \quad 4.2$$

The total deformation and the linear buckling of the upper thigh are shown in Figure 4.2, from left to right. The upper segment of the thigh underwent a total deformation of 3.231 mm, localised at the hip connector. Due to the complex shape of the hip connector, the applied load on it acted as both an axial and a transverse load.

The linear buckling analysis of the upper thigh indicated that it did not buckle under the applied load. The researcher multiplied the load multiplier obtained for the simulated part by the acting load to determine the buckling load for each part. The axial acting load on the upper thigh was 983.89 N. By multiplying this load by its respective load multiplier of 15.094, the buckling load, P_b , of 14.851 kN is obtained.

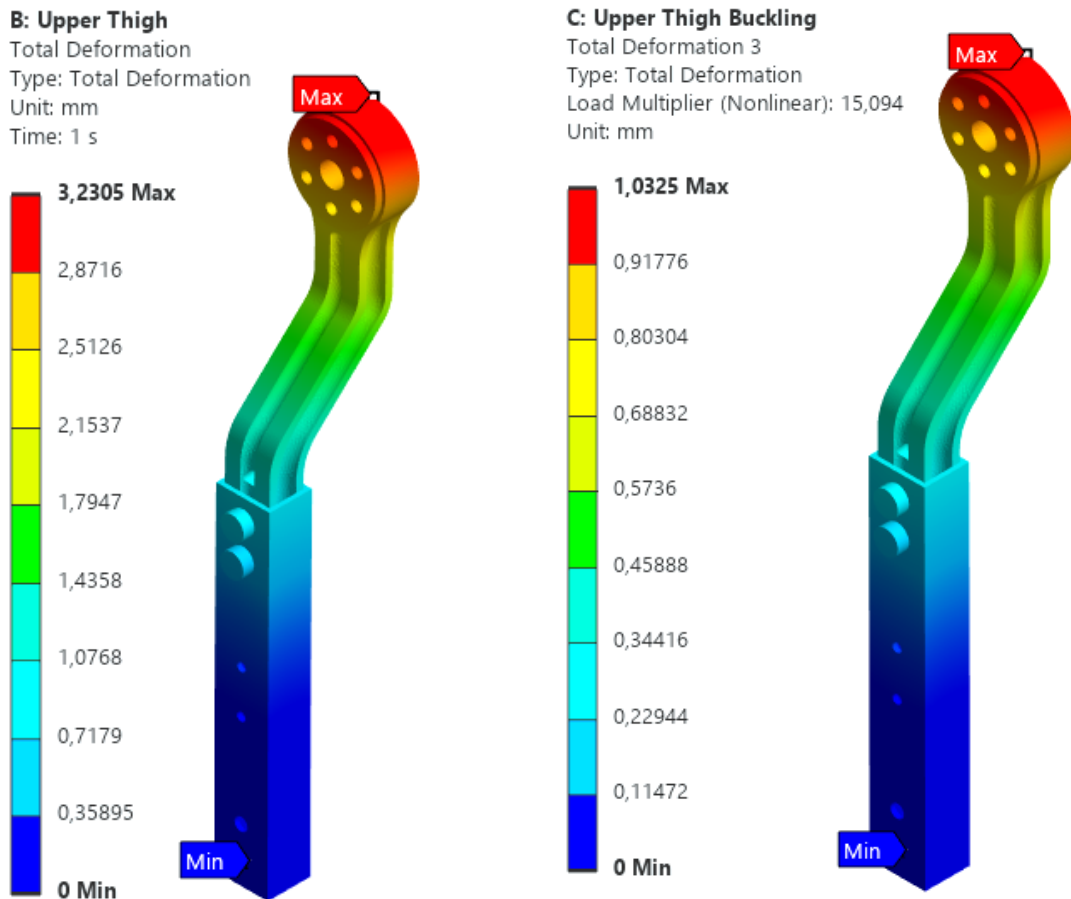


Figure 4.2: Deformation and linear buckling of the upper thigh related to axial loading

A critical buckling analysis was performed to determine whether the calculated buckling load was within its limit. The variables in the critical buckling load equation for the upper thigh were defined as $I_{xx} = 26.581e^{-9} m^4$ and $l = 0.0.257 m$; where I_{xx} was the axis about which the

upper thigh buckled most easily. By substituting the respective values into Equation 4.1, the critical buckling load was calculated as 69.435 kN. By comparing P_b and P_c , it was concluded that the upper thigh does not buckle under the applied load, as P_b does not exceed P_c .

b. Lower thigh

Figure 4.3 presents two types of results. The initial pair, from left to right, displays Von Mises stress, while the last one illustrates the factor of safety associated with the maximum Von Mises criterion. The lower thigh experienced its peak Von Mises stress on the bolt shank, in contrast to the upper thigh. This stress was attributable to the applied axial load of 983.89 N on the thigh tube, which did not cause any translational motion of the lower thigh. According to the simulation results, the maximum Von Mises stress of 112.28 MPa was deemed acceptable, as it remained within the elastic region of low-carbon steel. The calculated safety factor of 2.803 indicated that the lower thigh was adequate for the task, providing a safety margin of 180.3% for the lower thigh segment. In addition to the maximum Von Mises stress occurring on the bolt, the lower thigh also experienced a maximum Von Mises stress of 98.452 MPa at the bolt hole. These results demonstrate that the lower thigh maintained its structural integrity under the specified applied load.

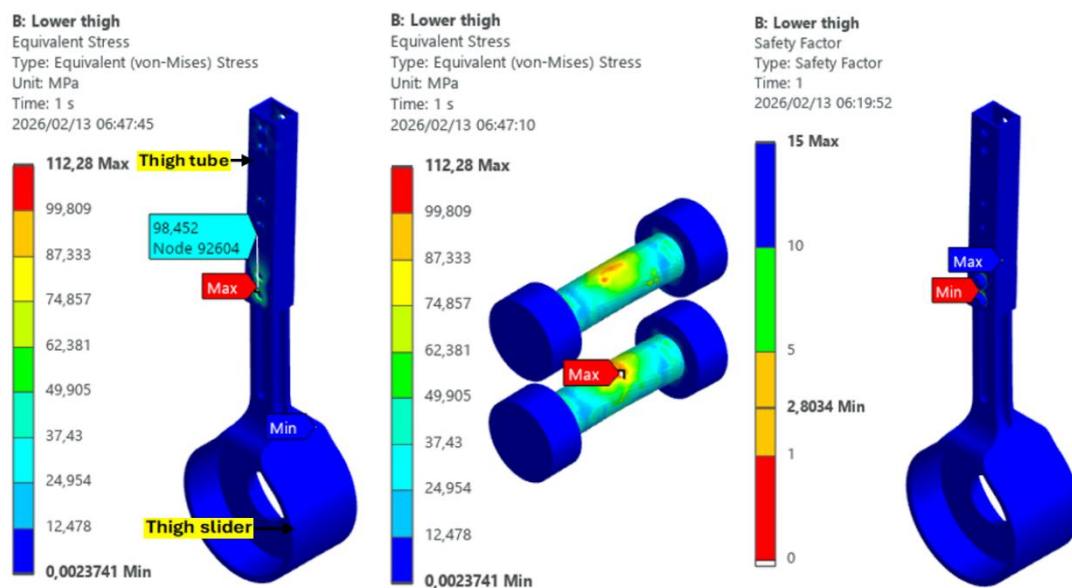


Figure 4.3: Von Mises and FOS of the lower thigh related to an applied axial load

Both the total deformation and the eigenvalue buckling (also referred to as linear buckling) of the lower thigh, as depicted in Figure 4.4, were within acceptable limits. The total deformation of the lower thigh was 0.0253 mm, which was 99.217 % smaller than that of the upper thigh under the same loading condition. The maximum total deformation occurred at the topmost section of the lower thigh segment, while the minimum deformation was observed at the bottom, near the actuation housing, where the lower thigh was fixed for simulation purposes. This magnitude was negligible compared with 0.316 m, the length of the lower thigh. This

suggested that the component design was adequately capable of withstanding the applied loads.

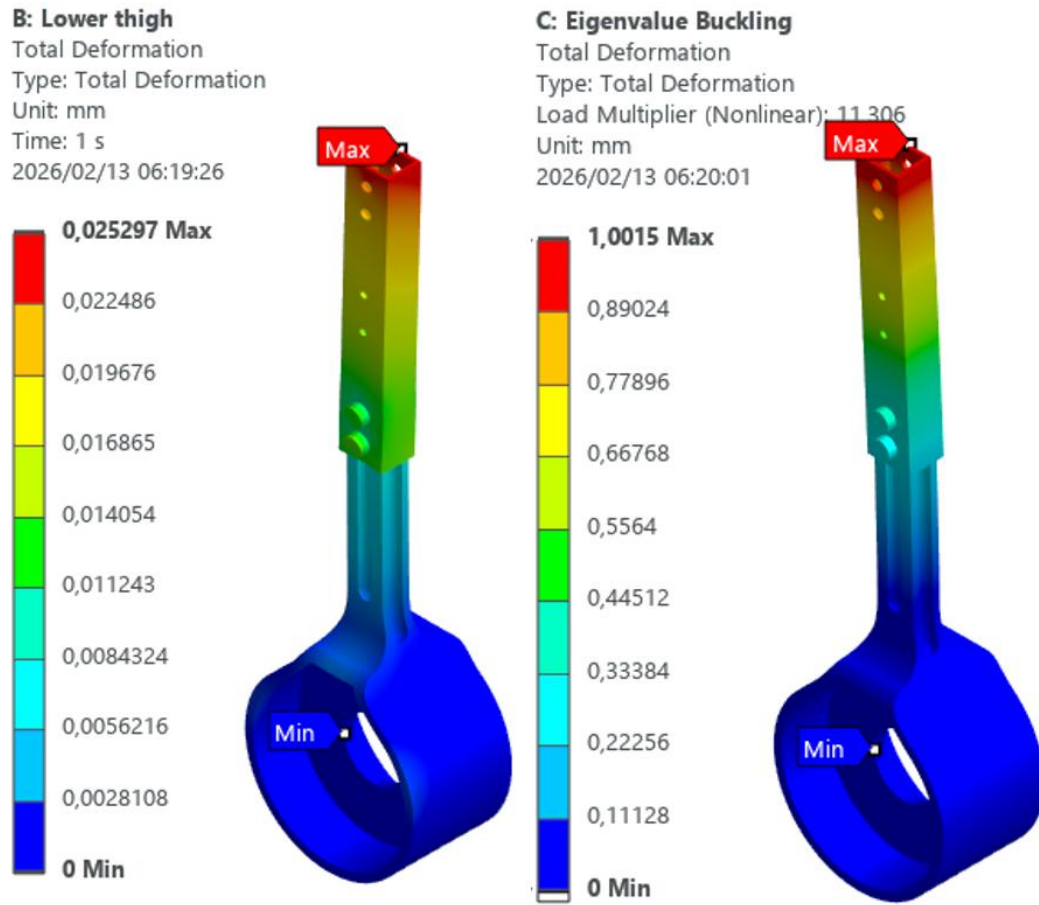


Figure 4.4: Total deformation and buckling of the lower thigh under applied axial load

A p_b of 11.124 KN was obtained by multiplying the load multiplier of 11.306 by the applied axial load of 983.89 N. In the process of defining the buckling behaviour of the lower thigh, its critical buckling load of 16.443 KN was determined by substituting the relevant segment's variable into the critical buckling equation, where $I_{yy} = 9.5e^{-9} m^4$ and $l = 0.316 m$. Since p_b is less than p_c it was concluded that the lower thigh segment was properly designed to withstand the applied load of 983.89 N without experiencing buckling.

4.1.1.2 Shank

As previously stated, the shank is divided into two sections: the upper and lower parts. These segments were subjected to an axial load of 1056.662 N. The subsequent discussion focuses on these two segments. It is essential to clarify that the bolts connecting the shank components were subjected to shear stresses, as the axial load pushed them against each part simultaneously.

a. Upper shank

The results shown in Figure 4.5, from left to right, are related to the upper shank's Von Mises stresses and FOS. The analysis indicates that this segment experienced a maximum Von Mises stress of 162.9 MPa and a minimum safety factor of 1.694 at the shank tube bolt holes, where the segment was constrained at the bottom section. The maximum Von Mises stress recorded for this segment confirmed that the design was acceptable, as the upper shank remained within the elastic range of aluminium 6061-T6, preventing permanent deformation. The safety factor of the upper shank demonstrated its capacity to support not only the weight of the exoskeleton and its user, but also an additional 69.43% of the combined load of both the user and the LLE, while maintaining safety and preventing failure.

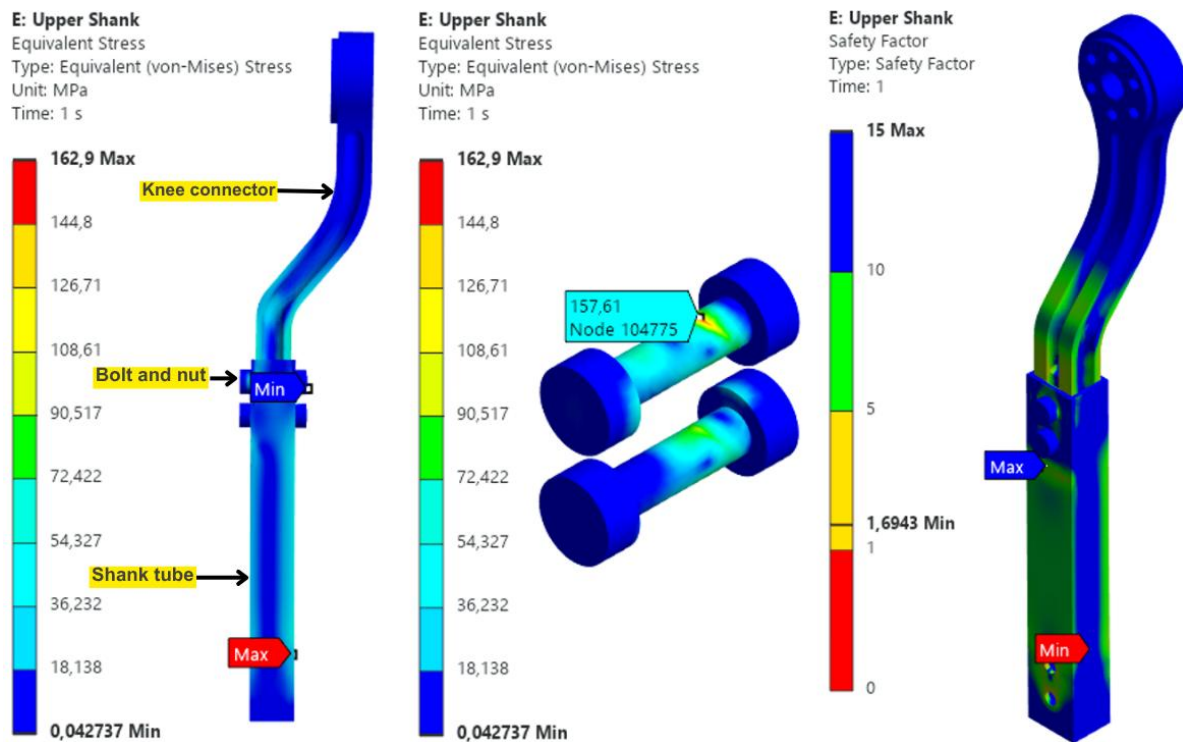


Figure 4.5: Von Mises and FOS of the upper shank under applied axial load

To determine the maximum Von Mises stress on the upper bolt, the knee connector and the shank tube were isolated, revealing a Von Mises stress of 157.61 MPa. With this result, it was proven that the selected M6 bolt, made of low-carbon steel 4.8, did not fail under an applied load of 1058.70 N applied at the top of the knee connector head. The upper shank experienced a total deformation of 3.026 mm and a load multiplier of 12.608, as shown in Figure 4.6 from left to right. The upper shank underwent a deformation of 0.890% of its length, which was deemed negligible.

A P_b of 13.408 kN and a P_c of 62.597 kN, pertaining to the upper shank, were determined using the same methodology as for the previous components. As P_b is less than P_c , it was concluded that the structure can withstand the applied axial load, indicating that the segment does not buckle.

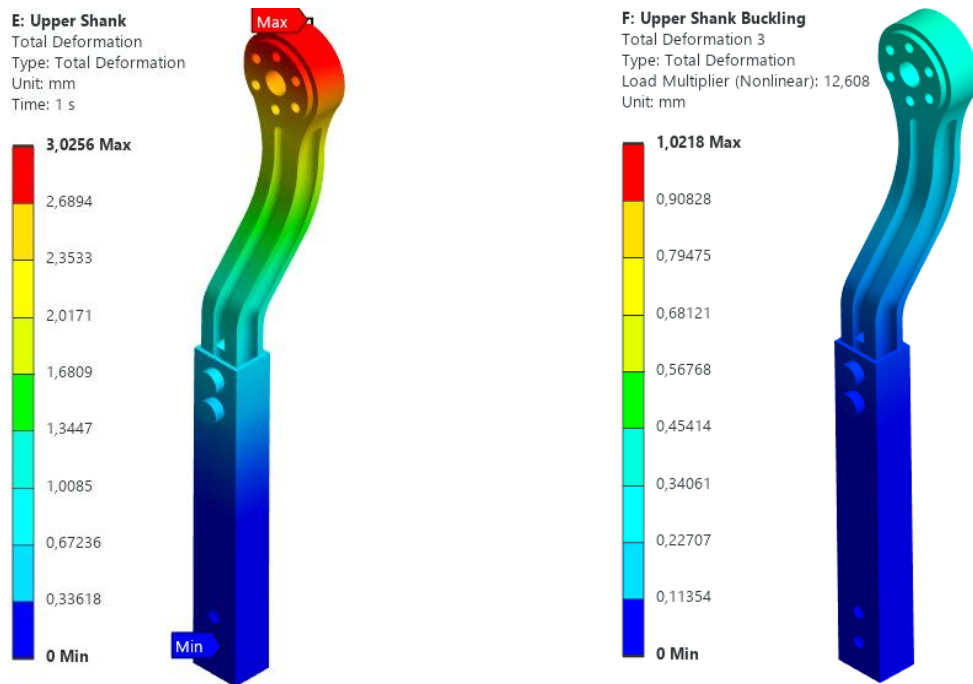


Figure 4.6: Total deformation and buckling of the upper shank under applied axial load

b. Lower shank

Under the same applied load as for the upper shank, the lower shank experienced a maximum Von Mises stress of 157 MPa, which occurred at the upper bolt hole on the shank tube; with a safety factor of 5.731, as illustrated in Figure 4.7, from left to right. These results indicate that the structural integrity of the lower shank was satisfactory, as the maximum Von Mises stress on the shank tube remained within the elastic limit of the aluminium 6061-T6. The safety factor further demonstrated the robustness of the concerned segment, confirming its ability to support the expected weight of both the user and the walking robot, with an additional 116,6% safety margin assured.

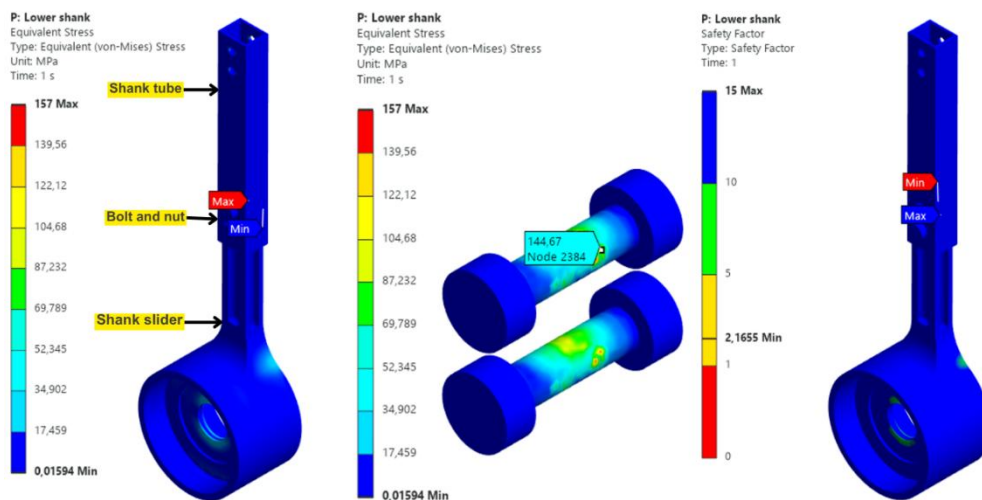


Figure 4.7: Von Mises and FOS of the lower shank under applied axial load

The total deformation of the lower shank was 0.609 mm under a load multiplier of 9.736, as illustrated in Figure 4.8. This total deformation represented 0.183% of the lower shank's initial length, which was considered negligible. In comparison with the obtained P_b of 9.997 kN and P_c of 19.868 kN for the lower shank, derived using the same methodology as the previous measurements, it was evident that, since P_b is smaller than P_c , the lower shank can support the applied axial load without buckling.

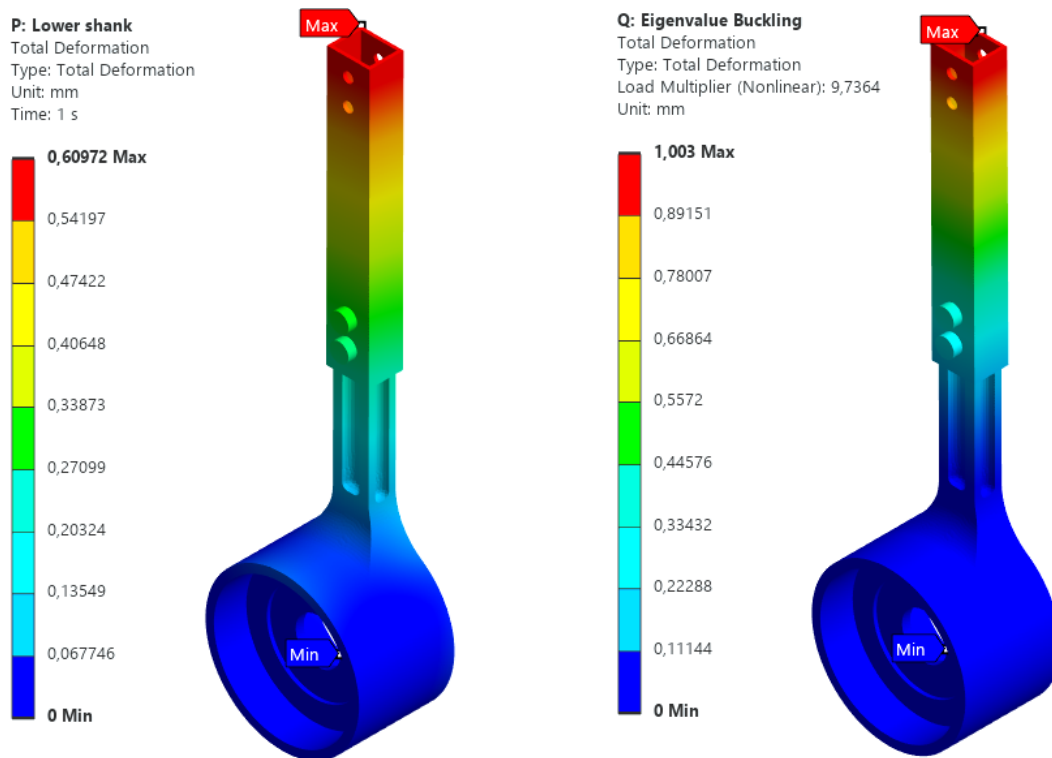
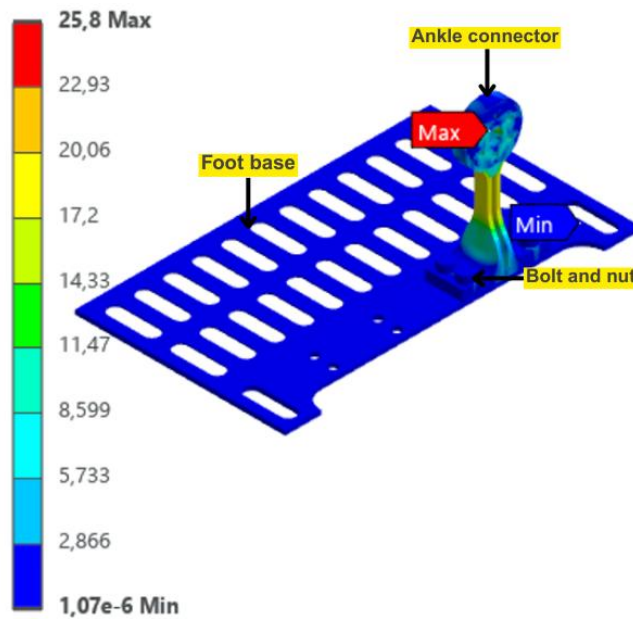


Figure 4.8: Total deformation and buckling of the lower shank under applied axial load

4.1.1.3 Foot

The Von Mises stress analysis of the vertical foot, as illustrated in Figure 4.9, showed that the foot could effectively support the applied load of 1082.414 N, with a maximum Von Mises stress of 25.797 MPa at the ankle connector. This indicated that the segment remained well within the elastic region of aluminium 6082-T6 and had not reached the plastic range. Additionally, the M4 bolt experienced a maximum Von Mises stress of 8.342 MPa. Based on these findings, it can be concluded that both the foot segment and its fasteners are adequately equipped to support the combined weight of the user and the walking robot without the risk of failure. Contrary to the thigh and shank, the bolts connecting the foot base to the ankle connector are under compressive stress.

H: Foot
 Equivalent Stress 2
 Type: Equivalent (von-Mises) Stress
 Unit: MPa
 Time: 1 s



H: Foot
 Equivalent Stress
 Type: Equivalent (von-Mises) Stress
 Unit: MPa
 Time: 1 s

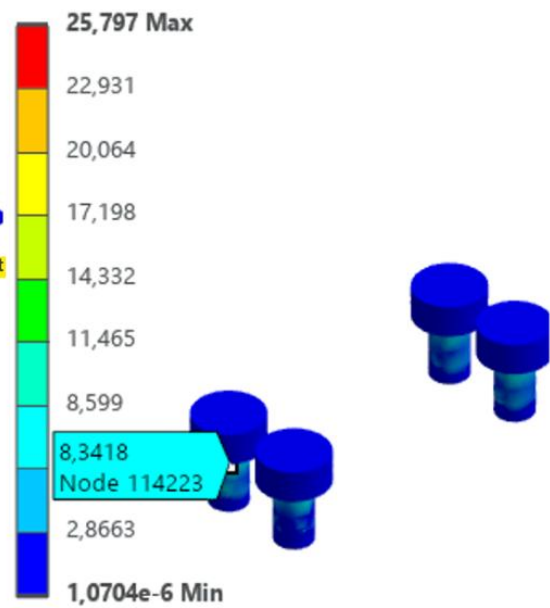
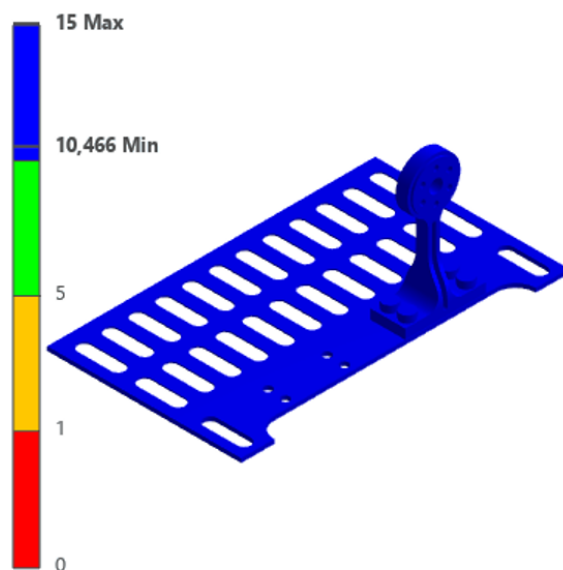


Figure 4.9: Von Mises stress of the foot under applied axial load

This segment demonstrated a safety factor of 10.466 and underwent a deformation of 0.0203 mm, as shown in Figure 4.10. The safety factor indicated that the vertical foot was designed to support the entire weight of the LLE and its user. Both the deformation measurement and the safety factor indicate that the vertical foot's performance is within acceptable limits.

H: Foot
 Safety Factor
 Type: Safety Factor
 Time: 1



H: Foot
 Total Deformation
 Type: Total Deformation
 Unit: mm
 Time: 1 s

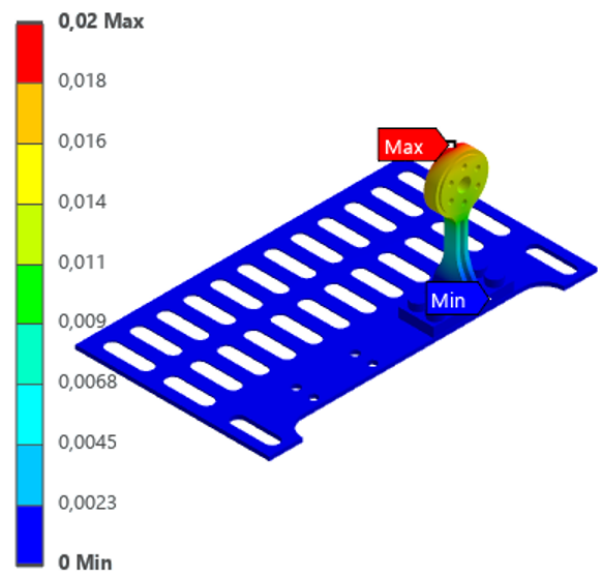


Figure 4.10: FOS and total deformation of the foot under applied axial load

The buckling value of 20.732 KN was determined for the foot, indicating its capacity to withstand the applied axial load of 981 N, as shown in Figure 4.11. This conclusion is based on comparing the obtained pb of 20.732 KN with the pc of 98.346 KN.

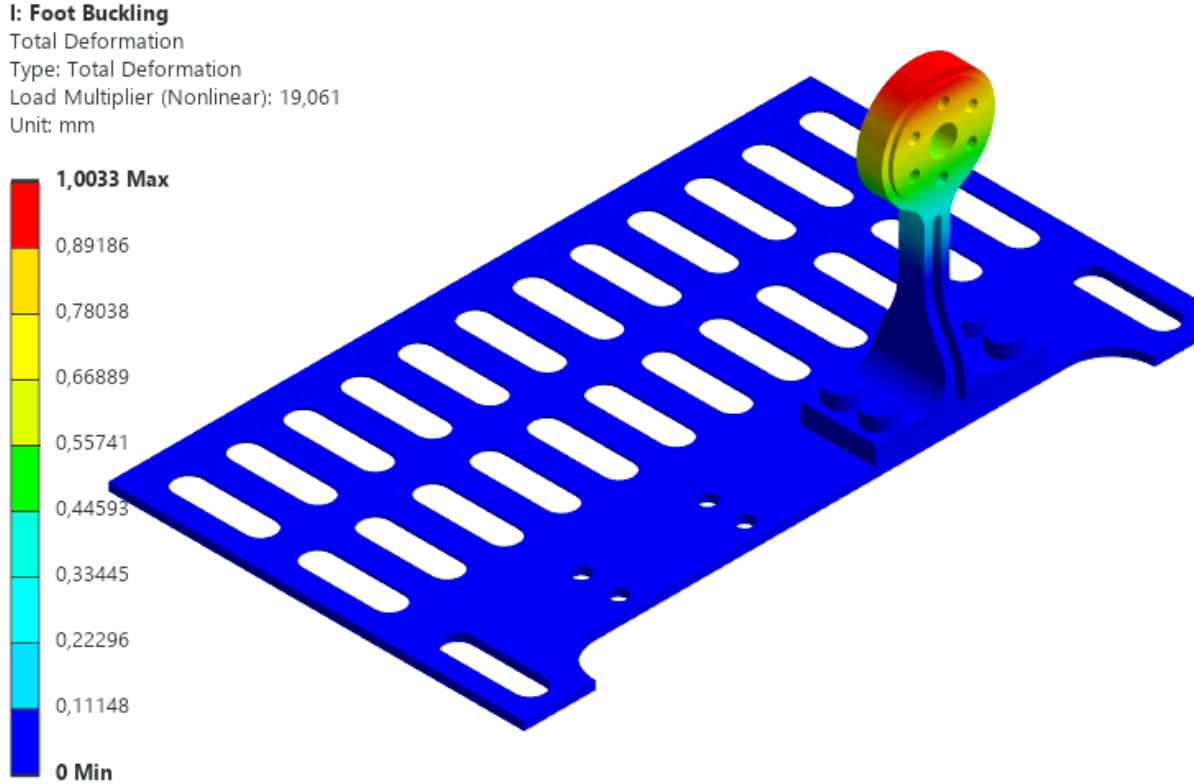


Figure 4.11: Buckling of the foot under applied axial load

4.1.2 Transverse Loading (Terminal swing)

The FEA simulation of the left leg during the terminal swing phase used the same settings as in the mid-stance phase. The thigh and shank were divided into upper and lower sections and subjected to transverse loads, raising concerns about bending stress. The thigh segments experienced forces of 134.304 N and 98.525 N, the shank 72.772 N and 25.753 N, and the foot 25 N. The discussion of the results focuses on total deformation, safety factors, and Von Mises stress. Once the simulation confirmed that all components were within stress limits, bending stresses were calculated using Equation 4.3 and compared with the allowable stress, defined as the yield strength divided by 1.5.

$$\sigma_{max} = \frac{M * C}{I} \quad 4.3$$

4.1.2.1 Thigh

a. Upper Thigh

Under transverse loads of 134.304 N and 98.525 N applied to the upper thigh, the maximum Von Mises stress of 90.893 MPa was recorded at the hip connector, which remained below the yield strength of aluminium 6082-T6. The M6 bolt experienced a maximum stress of 82.307 MPa, comfortably within the yield strength of low-carbon steel. Both the frame and the bolt operate within their elastic limits, as illustrated in Figure 4.12.

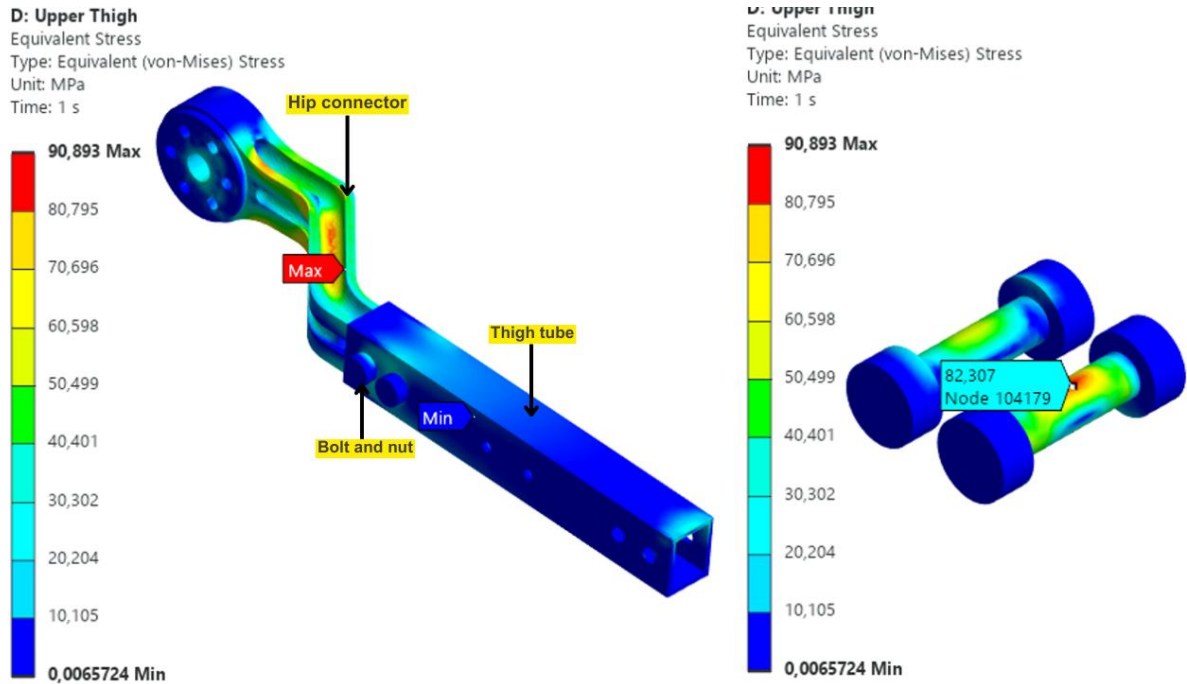


Figure 4.12: Von Mises stresses of the upper thigh under applied transverse load

The upper thigh underwent a maximum bending stress of 40.386 MPa, calculated by substituting the relevant variables into Equation 4.3. To verify that this maximum bending stress fell within acceptable limits, the yield strength was divided by 1.5 and compared to the calculated bending stress. Pendse & Deshmukh (2022) state that if the design exhibits a minimum safety factor of 2 on ultimate strength or 1.5 on yield strength, it is considered acceptable.

By comparing the maximum bending stress of 40.386 MPa with the allowable bending stress of 180 MPa, obtained by dividing 270 MPa by 1.5, it was concluded that the upper thigh remained within the specified stress limits.

Figure 4.13 shows the safety factor and total deformation for the upper thigh. A safety factor of 2.971 was obtained, indicating that the upper thigh, when subjected to transverse loads, could support the weight of the walking robot and its user. This finding was further verified by comparing the yield strength of the hip connector with the Von Mises stress, obtained by

dividing the latter by the former. The results confirm that the upper-thigh design met the required standards, with acceptable Von Mises stress and safety factor. The total deformation is also acceptable relative to its initial length.

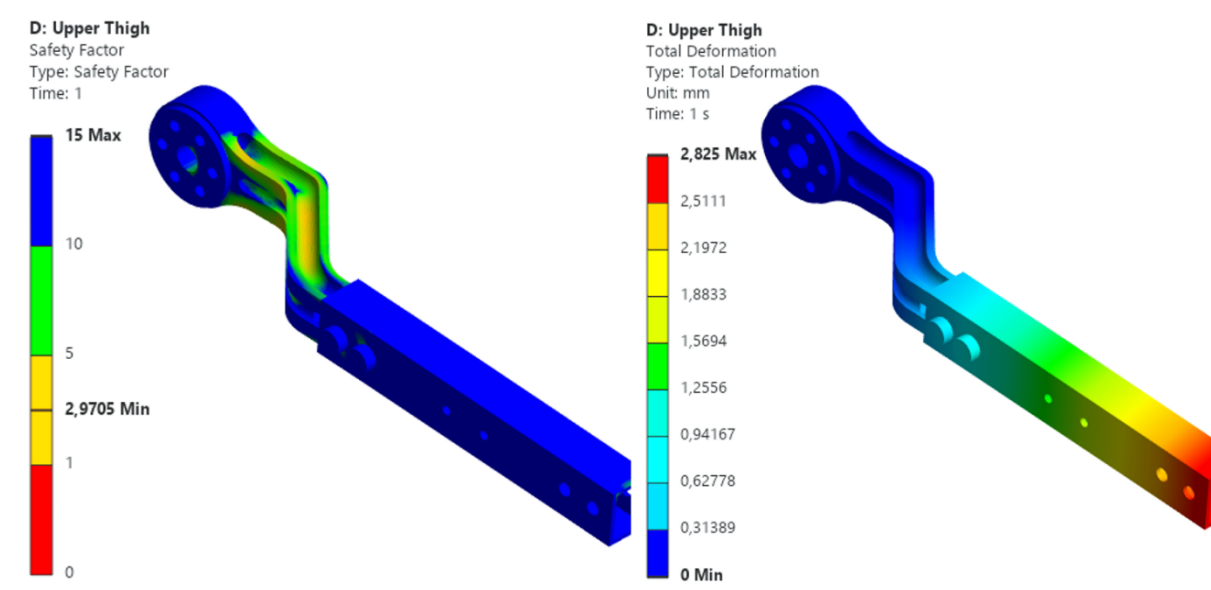


Figure 4.13: FOS and total deformation of the upper thigh under applied transverse load

b. Lower Thigh

The two main Von Mises stresses on the lower thigh are shown in Figure 4.14, with values of 117.41 MPa on the frame segment and 197.73 MPa on the M6 bolt. These stresses were considered acceptable relative to the yield strengths of the respective materials, as they remained within the elastic limits. Notably, the maximum Von Mises stress on the lower thigh was observed at the bolt shank.

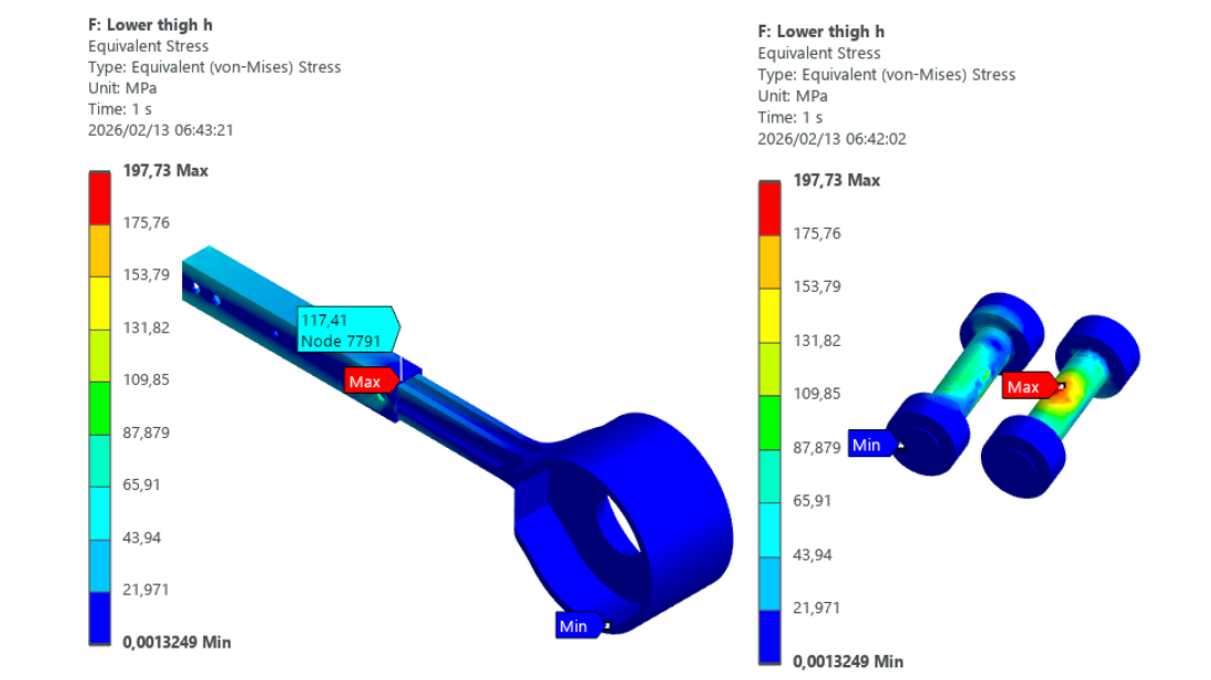


Figure 4.14: Von Mises stresses of the lower thigh under applied transverse load

The lower thigh experienced a maximum bending stress of 57.287 MPa. By comparing this value with the allowable bending stress of 180 MPa related to aluminium, it was concluded that the design of the lower thigh was satisfactory, as the maximum bending stress remains well within the defined limits of the allowable bending stress.

Figure 4.15 illustrates the FOS and the total deformation of the lower thigh, where the safety factor of 1.719 confirms that this segment design is appropriate for the task, as it can sufficiently support the load of the thigh and lower leg for both the LLE and its users.

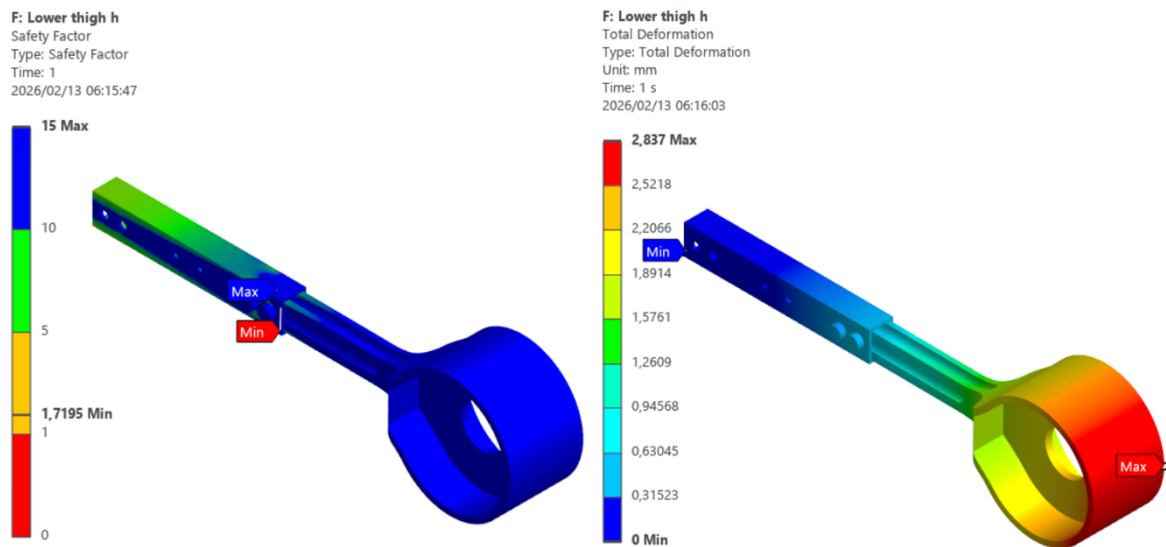


Figure 4.15: FOS and total deformation of the lower thigh under applied transverse load

The total deformation of the lower thigh was measured at 0.814% of its initial length of 358.428 mm. This assessment was made under the condition that the maximum Von Mises stress of 197.73 MPa remained within the elastic limits of low-carbon steel, with a safety factor to ensure the robustness of the lower thigh. Therefore, it was concluded that the total deformation of this segment is acceptable.

4.1.2.2 Shank

a. Upper shank

In the scenario depicted in Figure 4.16, the Von Mises stresses for both the frame and the bolt were closely matched, measuring 30.193 MPa and 29.557 MPa, respectively, under the influence of the transverse loads applied to the shank segment. The maximum Von Mises stress was recorded at the knee connector of the segment. The stress levels within the segment were considered satisfactory, remaining below the yield strengths of 270 MPa for aluminium 6082-T6 and 340 MPa for low-carbon steel.

The maximum bending stress of 16.268 MPa, associated with the upper shank, demonstrated that this segment's bending stress was acceptable, as it remained within the defined allowable stress limit of 180 MPa.

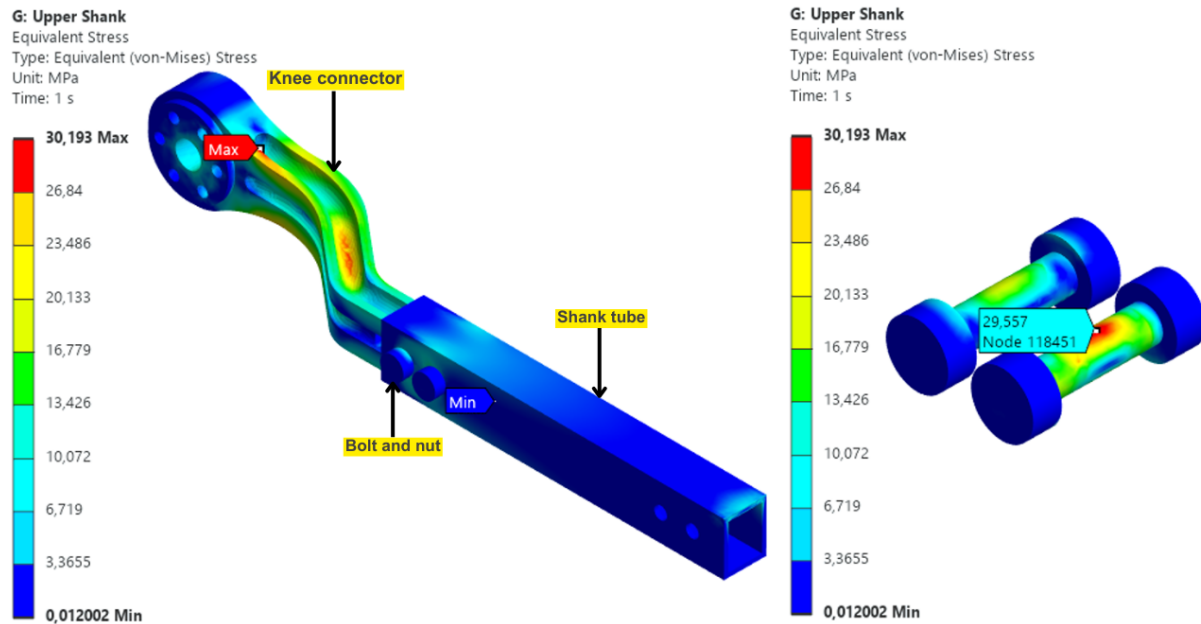


Figure 4.16: Von Mises stresses of the upper shank under applied transverse load

As illustrated in Figure 4.17, a safety factor of 8.942 and a total deformation of 0.992 mm indicate that the upper shank can fully support the transverse loads on the shank segment. Further reduction in the upper shank size is not recommended, as it could adversely affect results under axial loading. Such modifications might increase the maximum Von Mises stress and reduce the FOS from 1.460 to near 1 or below. The total deformation of 0.992 mm under the specified maximum Von Mises stress conditions, together with its corresponding safety factor, is considered acceptable.

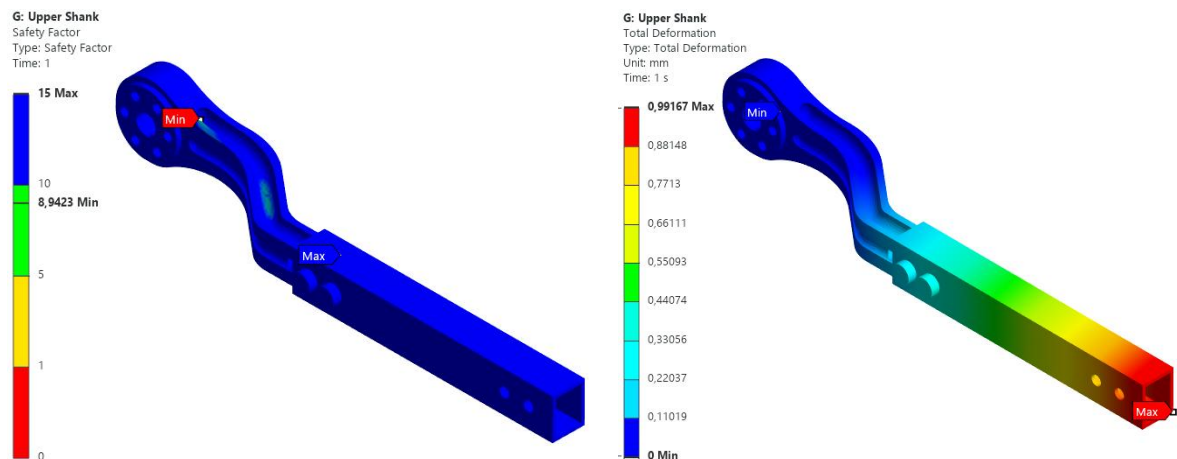


Figure 4.17: FOS and total deformation of the upper shank under applied transverse load

b. Lower shank

Figure 4.18 illustrates the Von Mises stresses experienced by the lower shank segment and the bolts. Under applied transverse loads of 72.772 N and 25.753 N, the maximum Von Mises

stress observed on the bolt was 162.41 MPa, while the stress at the bolt hole on the lower shank was 173.51 MPa. These stress values, when compared with the yield strengths of low-carbon steel and aluminium, confirm that the design of this segment is sufficiently robust. Additionally, the materials employed for the shank tube, shank slider, and fasteners remained within their elastic limits, as their stress levels did not surpass the yield strengths of Aluminium 6061-T6, Aluminium 6082-T6, and low-carbon steel.

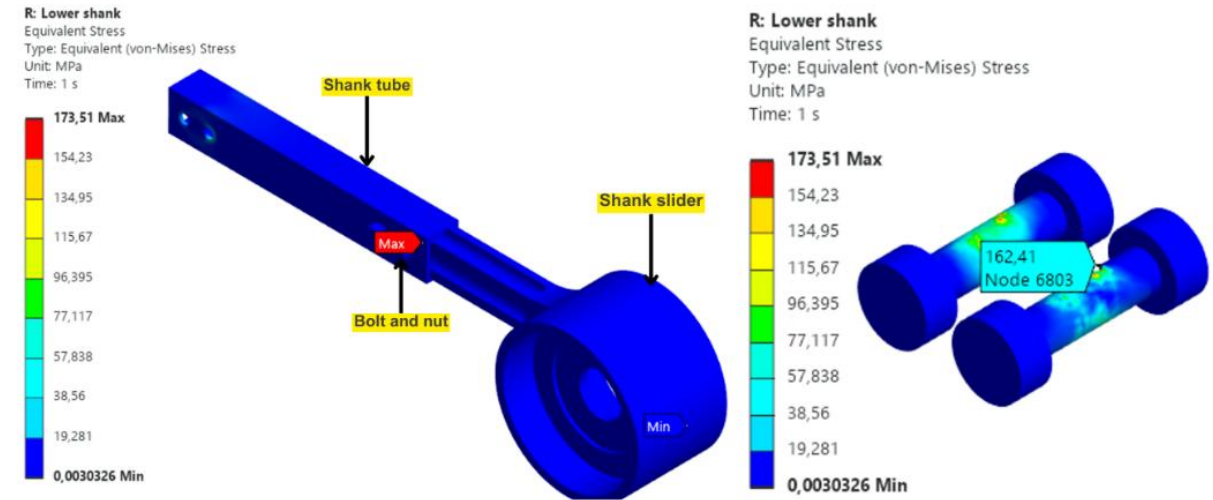


Figure 4.18: Von Mises stresses of the lower shank under applied transverse load

The shank design was verified as acceptable, as the maximum bending stress of 17.226 MPa is below the established allowable bending stress of 180 MPa, thereby meeting the bending stress criteria.

The factor of safety of the lower shank under the applied transverse loads was 1.959, confirming that both the frame and the bolt have sufficient capacity to support the weight on this segment. The total deformation of the lower shank in this scenario was 0.251% of the shank segment's initial length, measured from the fixed support to the centre of the actuator housing. Both the FOS and the total deformation are illustrated in Figure 4.19.

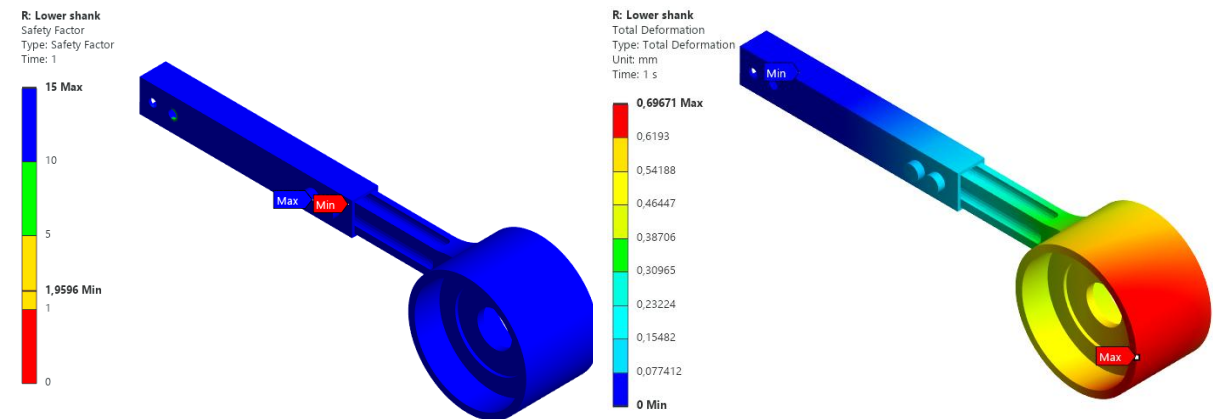


Figure 4.19: FOS and total deformation of the lower shank under transverse load

4.1.2.3 Foot

Under the applied transverse load shown in Figure 4.20, the foot experienced a maximum Von Mises stress of 38.774 MPa. This level indicates that the foot's design is sufficiently robust to fulfil its intended functions, as it remained within the elastic range of Aluminium 6082-T6, not exceeding its yield strength of 270 MPa. The component was analysed independently to assess the effect of the applied force on the bolt; the bolt's Von Mises stress was found to be 19.382 MPa. Since both the frame and the bolts exhibited Von Mises stresses below their respective yield strengths, it was concluded that the horizontal foot has successfully passed the simulation. The maximum bending stress of 1.275 MPa at the foot indicated that the segment's design is appropriate, as it stayed below the allowable bending stress of 180 MPa, complying with the established limit.

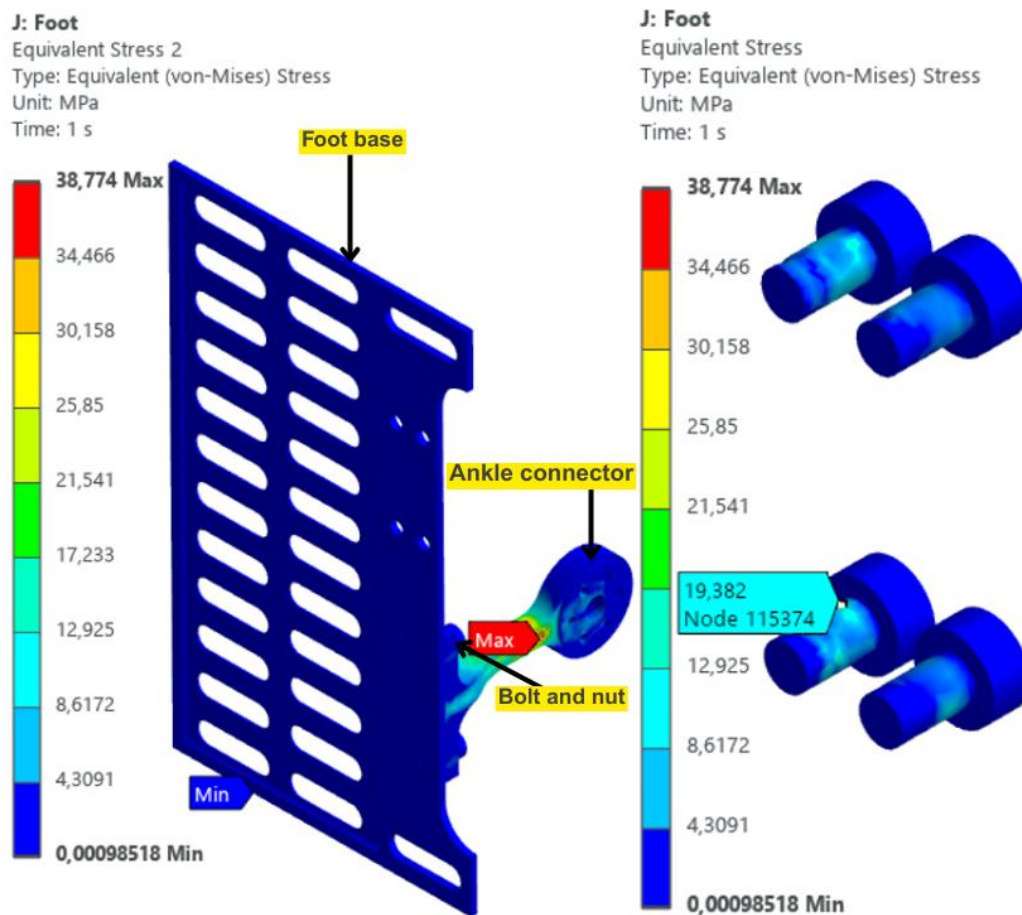


Figure 4.20: Von Mises stresses of the foot under applied transverse load

The safety factor and total deformation of the foot are illustrated in Figure 4.21. The FOS was determined by dividing the yield strength of aluminium 6082-T6 by the maximum Von Mises stress at the ankle connector, yielding a value of 7.165. This finding is consistent with the FOS obtained through simulation. This demonstrates that the foot can support its weight and that of its user. In this scenario, the segment experienced a total deformation of 0.697 mm, which is deemed satisfactory.

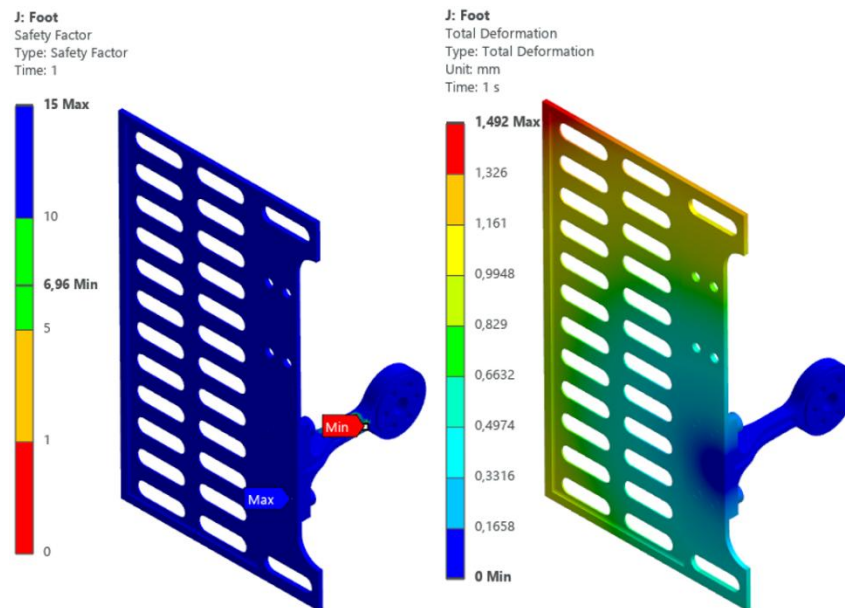


Figure 4.21: FOS and total deformation of the foot under transverse load

In Figure 4.22, a transverse load was applied to the side of the ankle connector, as illustrated. In this scenario, the leg was in mid-stance and subjected to a lateral force of $1082.415\sin 17.75^\circ$, which tended to cause the robot to tip sideways. Under these conditions, the foot experienced a maximum Von Mises stress of 185.85 MPa at the ankle connector, while the bolt endured a maximum Von Mises stress of 75.908 MPa. The stress levels for both the bolt and the ankle connector indicate that the foot could withstand the lateral force attempting to destabilise it, as the computed Von Mises stresses remained below the foot's yield strength. In this scenario, the foot experienced a maximum bending stress of 7.782 MPa, which remains within the allowable bending stress of 180 MPa.

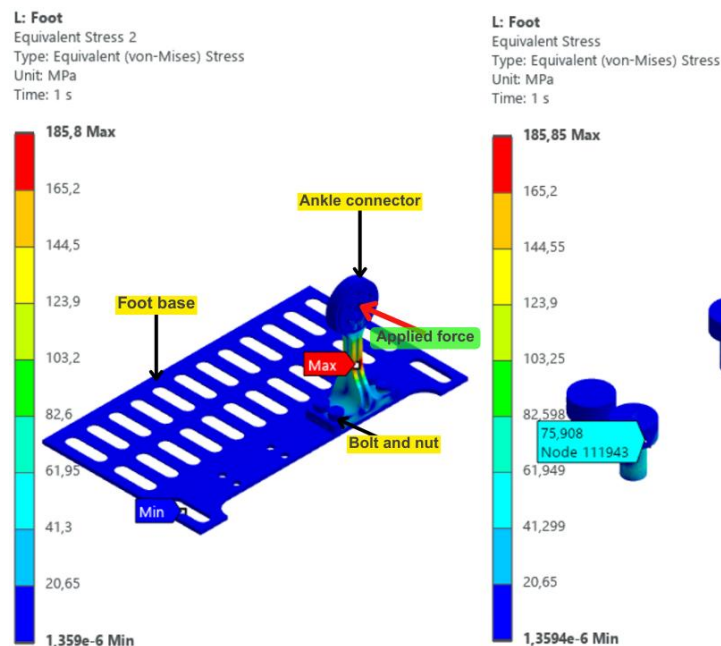


Figure 4.22: Von Mises stresses of the foot and bolts under lateral load

The FOS of 1.453 and the total deformation of 1.101 mm, as illustrated in Figure 4.23, demonstrate that the foot, under the applied load, ensured safe use with a capacity to resist the applied load by an extra 45.3 %, while the total deformation of this segment is 1.101 % of the initial length of 100.026 mm.

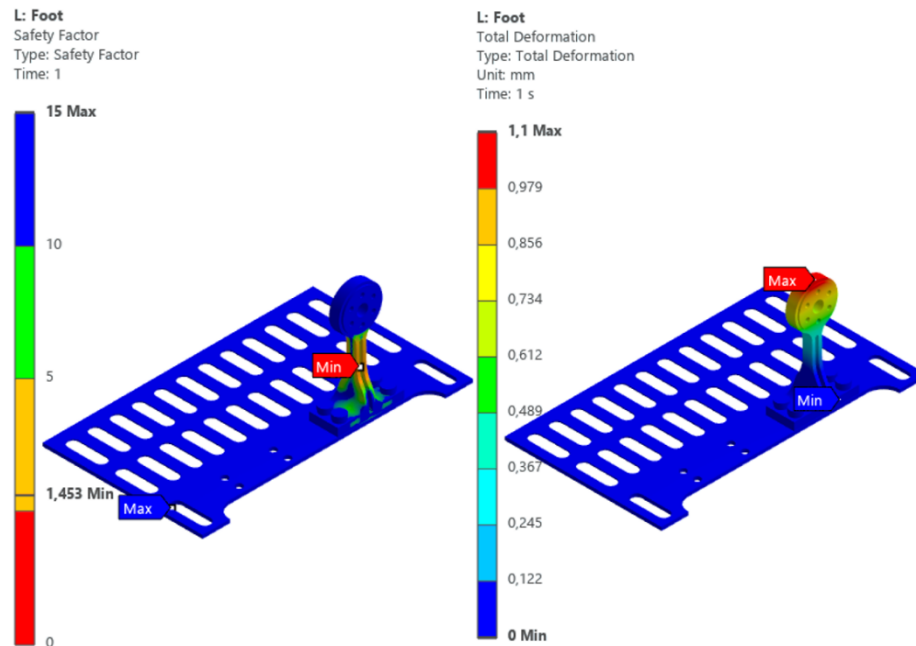


Figure 4.23: FOS and total deformation of the foot under lateral load

4.1.3 Load Sensitivity Analysis of the LLE Segments

After performing the FEA for each segment of the LLE under the design loads, the next step was to conduct a load-sensitivity analysis by varying the applied load by $\pm 10\%$. The purpose was to determine whether the walking robot's segment could withstand loads different from the original design load. In Table 4.1, %, seg, UT, LT, US, LS, VM, and SF represent percentage, segment, upper thigh, lower thigh, upper shank, lower shank, Von Mises stress, and safety factor.

Table 4.1: Load sensitivity analysis

	seg	Under Axial load			Under Transverse load			
		Load (N)	VM	SF	Load (N)		VM	SF
-10%	UT	-885.5	166.37	1.658	-120.87	-88.67	81.836	3.299
+10%		-1082.28	205.85	1.340	-108.38	-147.7	99,948	2.701
-10%	LT	-885.5	107.7	2.846	-120.87	-88.67	185.44	1.833
+10%		-1082.28	116.76	2.763	-108.38	-147.7	207.07	1.643
-10%	US	-951.03	146.63	1.882	-65.50	-23.18	27.731	9.736

+10%		-1162.37	179.27	1.539		-28.33	-80.05	33.714	8.009
-10%	LS	-951.03	156.77	2.169		-65.50	-23.18	172.18	1.975
+10%		-1162.37	157.28	2.162		-28.33	-80.05	175.67	1.935
-10%	Foot	-974.70	23.237	11.62			-23.18	36.231	7.452
+10%		-1191.30	28.408	9.504			-28.33	43.486	6.208

With a $\pm 10\%$ change in the applied load, the stress and safety factor change accordingly. The mean Von Mises stress and mean safety factor for the upper thigh, lower thigh, upper shank, lower shank, and the foot are presented in Table 4.2. The mean Von Mises stress values remained below the yield strengths of aluminium 6061-T6, 6082-T6, and low-carbon steel, which was acceptable, as the segments of the LLE remained in the elastic region. The safety factors of the walking robot's segments demonstrated the ability to withstand an additional 10% of the designed weight and applied load.

Table 4.2: Load sensitivity analysis (mean results)

seg	Under Axial load				Under Transverse load			
	VM (MPa)	Mean VM (MPa)	SF	Mean SF	VM (MPa)	Mean VM (MPa)	SF	Mean SF
UT	166.37	186.11	1.658	1.499	81.836	90.892	3.299	3
	205.85		1.340		99.948		2.701	
LT	107.70	112.23	2.846	2.805	185.44	196.255	1.833	1.738
	116.76		2.763		207.07		1.643	
US	146.63	162.95	1.882	1711	27.731	30.723	9.736	8.873
	179.27		1.539		33.714		8.009	
LS	156.77	157.025	2.169	2.166	172.18	173.925	1.975	1.955
	157.28		2.162		175.67		1.935	
Foot	23.237	25.823	11.62	10.562	36.231	39.858	7.452	6.83
	28.408		9.504		43.486		6.208	

4.1.4 Conclusion

Following FEA of the thigh, shank, and foot segments under various loading conditions, including axial and transverse applied loads, the results indicate that the Von Mises stresses for these segments are satisfactory for design. The stresses remain within the elastic limits of their respective materials.

The FOS confirm that each limb segment can effectively withstand the applied loads. The segment with the lowest maximum FOS is the foot, with a value of 1.453 under transverse (lateral) loading, whereas the lower shank exhibits the highest minimum FOS of 10.466 under transverse loading.

Additionally, the segments under axial loading were shown to resist buckling, as the buckling loads of each segment fell within the limits defined by its corresponding critical load. For transverse loading, the segments were deemed satisfactory because the maximum bending stresses were confirmed to be below the allowable bending stresses for each segment.

4.2 MATLAB-Simulink Walking Simulation

This section presents key findings derived from simulations performed using MATLAB Simulink. The analysis aims to provide insights into the gait cycle of the LLE across its phases and to assess the exoskeleton's performance and behaviour by examining the LLE's torso-based ambulatory capabilities along the XYZ axes. This provides an assessment of its spatial behaviour. Following the evaluation of the LLE's ambulatory performance, a detailed examination of the torques and angles at the respective joints will be conducted. While the discussion in this section focuses on the left lower limb, Appendix F presents results for the right leg gait cycle, with zoomed-in graphs for clarity. In addition, Figures F4 to F9 present the combined full simulation graphs with zoomed-in views for both legs.

4.2.1 The LLE gait cycle

The overlay of the human silhouettes and the LLE in Figure 4.24 confirms that the LLE gait cycle closely mimics human walking ability. Through the initial configuration, the LLE transitioned from its initial position to the walking position. During this process, the LLE underwent calibration to ensure the neutral angle of each joint was defined.

A heel strike characterises the leading foot of the LLE during the stance phase, while the contralateral leg begins toe-off; this behaviour introduces a phase in which both feet are in contact with the ground, known as the double support phase, and it correlates with the support phase of the human gait cycle, as illustrated in the overlay silhouettes.

During the loading response, the LLE's weight shifts to the leading leg. In this scenario, the hip and knee joints generate substantial torques to stabilise them and ensure a controlled descent.

During the single support phase, the LLE mimics the human gait cycle by progressing through mid-stance and terminal stance. The stance leg continues to bear the LLE's weight, while the swing leg lifts off the ground. Throughout this phase, the COM shifts forward to maintain balance. In terminal stance, the heel of the walking robot lifts off the ground as the trailing limb prepares for toe-off.

The double support phase is reintroduced during the pre-swing phase, with the left leg as the lead limb making initial contact with the ground while the trailing leg continues to push off. During this stage, rapid flexion of the knee and plantarflexion of the ankle joints occur to provide sufficient momentum for transitioning into the swing phase with the left leg.

The trailing limb lifts completely off the ground due to the propulsive force generated during toe-off. Next, the mid-swing phase occurs, during which the foot moves forward while clearing the ground. Finally, the trailing limb enters the terminal swing, preparing for the subsequent heel strike.

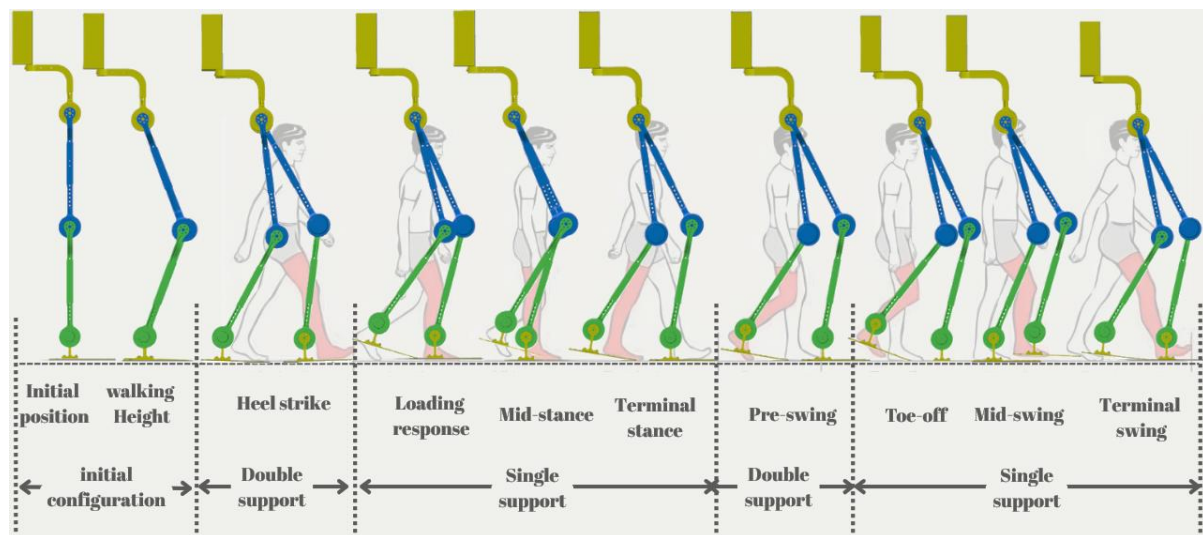


Figure 4.24: Walking phases for human and LLE (Pirker & Katzenschlager, 2016)

In Figure 4.25, a close-up of the left leg is shown, highlighted in green, illustrating the robot's ability to imitate these two phases. This colour highlight simply serves to distinguish between the two legs.



Figure 4.25: Heel strike and toe-off

4.2.1.1 X-y-z Displacements

In Figure 4.26, it can be observed that during the first seconds, the LLE experiences a transient drop from its initial Zrobot height of 0.9252 m to the torso walking height of 0.870 m, which aligns with the desired Zmodel walking height; this initial phase is referred to as the system's settling phase. From 1.035 seconds to 1.500 seconds, the LLE begins the walking cycle with a shorter step than the subsequent steps during the defined simulation period. During this initial step, the LLE's torso height temporarily rises to 0.872 meters due to the initial heel strike of the foot with the ground. This height subsequently drops to 0.870 meters at 1.588 seconds, during the loading response. This indicates an improvement in the robot's walking stability. The z-oscillations observed result from cyclic stance and swing phases, with vertical movement driven by ground reaction forces and leg orientation. In this walking simulation, it is evident that the Zmodel walking height slightly deviates from the desired Zmodel height, with an increase of +0.004 meters (+0.476%) at the peak and a decrease of -0.0028 meters (-0.332%) at the valley. These deviations can be attributed to the walking gait cycle, resulting from the shock produced when the foot contacts the ground.

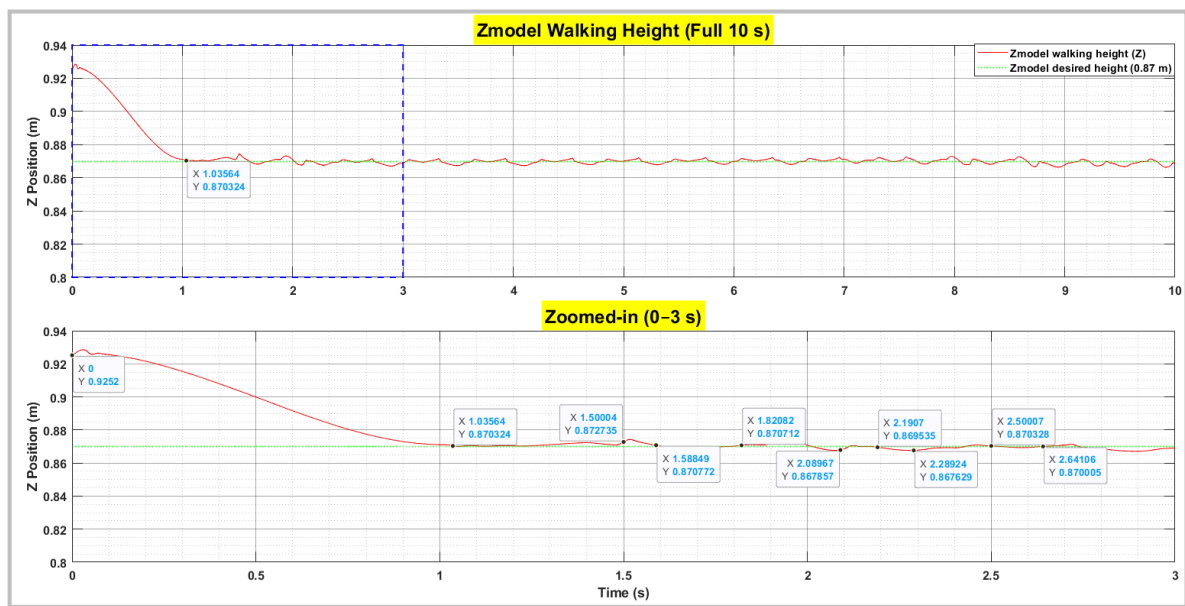


Figure 4.26: Walking height of the robot

Throughout the simulation, the walking robot achieved a forward walking velocity of 0.563 meters per second, successfully covering 5.63 meters in 10 seconds, as illustrated in Figure 4.27. This consistent movement emphasises the effectiveness of the LLE in generating sufficient propulsive motion to promote the robot's forward progression. The trajectory of the walking robot resembles a quasi-straight line, characterised by gentle curves with peaks and valleys that represent the robot's gait cycle.

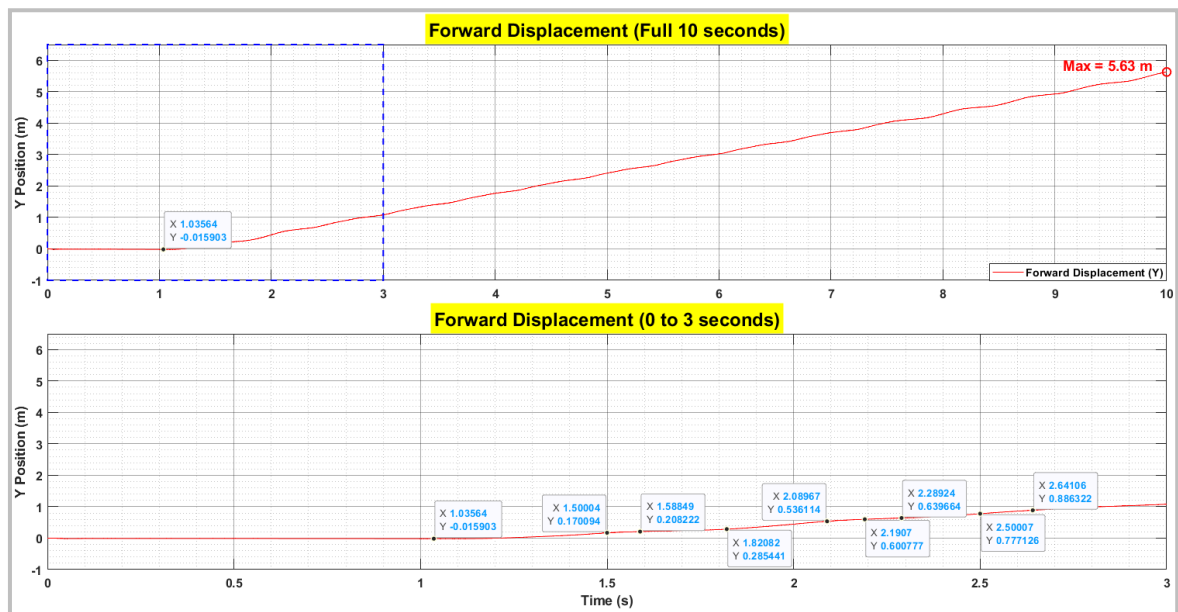


Figure 4.27: Travelled distance

Throughout the walking cycle of the LLE, the lateral movement of the torso was influenced by both the stance and swing phases. A detailed analysis of the torso behaviour is illustrated in Figure 4.28, which shows that the maximum peak value is approximately 0.017 m at heel strike, while the maximum valley value is around 0.022 m during the pre-swing phase. The greatest difference between the peak and valley measurements is 0.005 m, representing a deviation of 22.727%. This variation suggests that the LLE exhibits a slight imbalance, tilting toward the left leg during the swing phase.

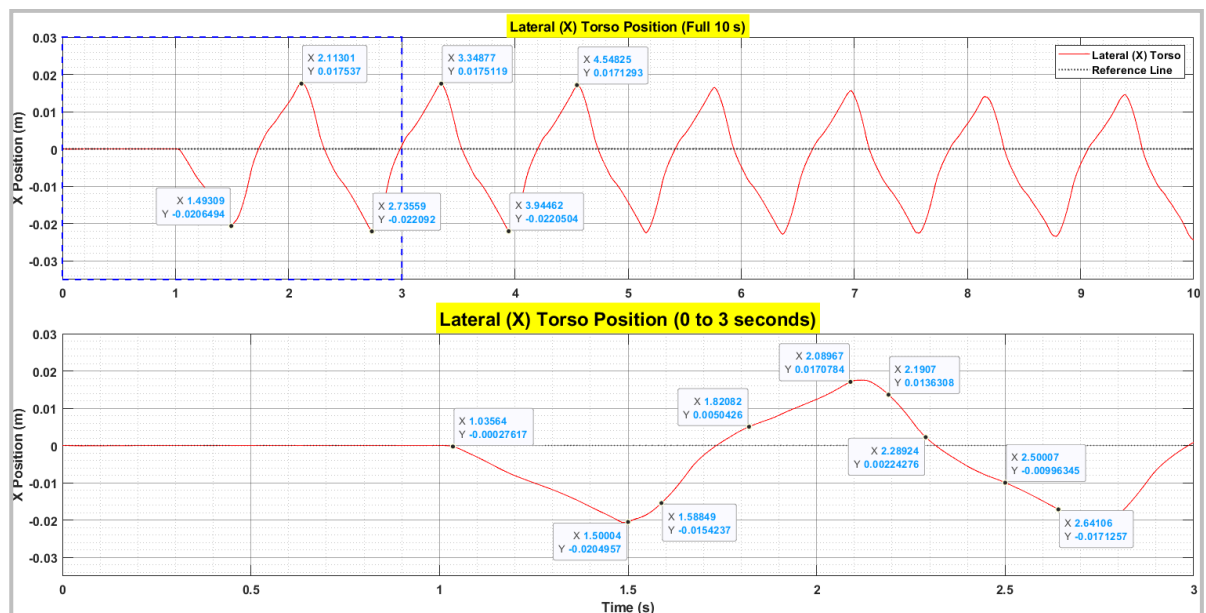


Figure 4.28: Lower torso x-position

a. Influence of speed variations on travel distance and torso height

The joints' speed was adjusted by ± 0.100 m/s to examine how the LLE would respond at speeds different from its designated maximum of 0.555 m/s. In the initial scenario, the LLE walked 4.86 m at 0.455 m/s, followed by 4.01 m at 0.355 m/s. As walking speed increased, the walking distance decreased, while the torso height remained approximately constant at the model's standard walking height of 0.87 m. In the second scenario, the LLE's speed was increased by 0.100 m/s, enabling it to cover 6.88 m at 0.655 m/s, and subsequently 7.91 m at 0.755 m/s. With increased velocity, the LLE walked farther; however, this also led to greater instability relative to the model's walking height. In conclusion, the mean walking distance covered by the LLE was 6.807 m in 10 seconds.

Figures illustrating variations in speed with travel distance and torso height are provided in Appendix J, from J.5 to J. 8.

4.2.1.2 LLE walking gait cycle

This section focuses on analysing the gait cycle of the walking robot during simulations conducted in MATLAB. It discusses the two main phases and their seven subphases related to the LLE. Given the correlation between these subphases and the associated torques and angles, the analysis also emphasises the torques and angles experienced by the hip, knee, and ankle in the sagittal plane. The figures presented below are divided into five distinct areas, each shaded in a different colour: grey for the setting phase, orange for the initial swing phase, blue for the stance phase, and red for the swing phase, with the subsequent gait beginning in the white area. The discussion below is centred on the blue- and red-shaded areas.

a. Stance phase

The stance phase of the walking robot at the hip joint, illustrated in the blue-shaded area of Figure 4.29, begins at 1.500 seconds with the heel-strike subphase. At this moment, the hip joint flexes by 30.206 degrees under a peak flexion torque of 116.6 Nm. Subsequently, the torque shifts to an extension torque of -113.3 Nm, which remains dominant for over 50% of the stance phase. This period includes the midstance subphase, during which the extension angle decreases from 30.042° to 3.861° . By the end of the terminal stance, the flexion torque reaches 76.774 Nm and continues to rise until it attains a maximum flexion torque of 116.6 Nm at an angle of 2.634 degrees during the pre-swing subphase.

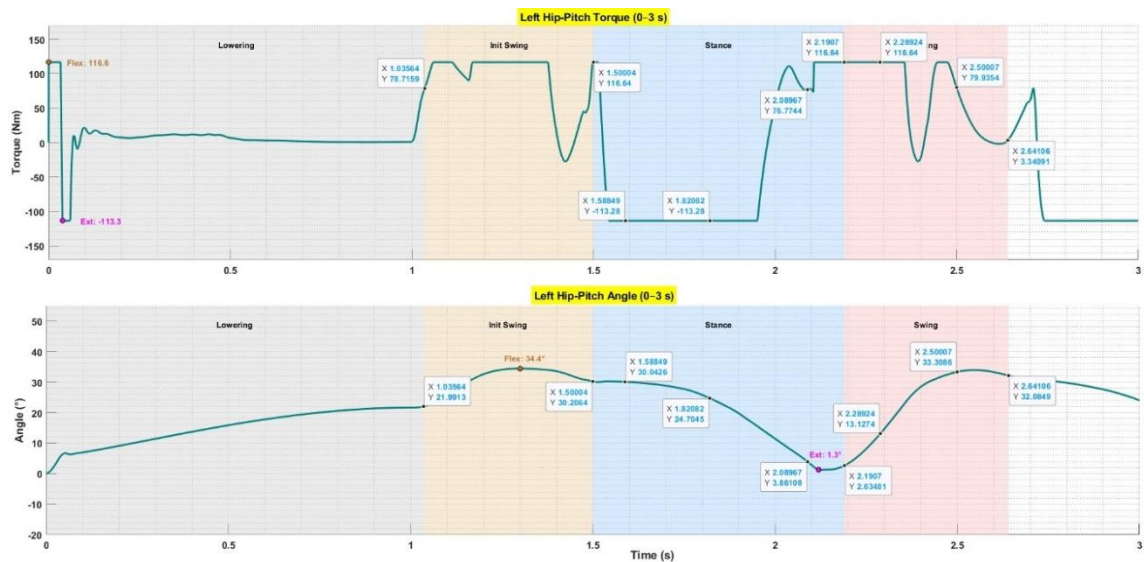


Figure 4.29: Hip torques and angles during the gait cycle

At this moment, the simultaneous behaviour of the stance phase associated with the knee joint of the LLE, as illustrated in Figure 4.30, exhibits flexion and extension performance related to the knee joint.

During the transition from heel strike to the midstance subphases, the knee joint flexes, with angles ranging from 34.293 degrees to 38.848 degrees during the loading response. As motion progresses from the loading response to midstance, the knee flexes further to 42.187 degrees. Throughout this interval, torque fluctuates from a maximum of 114.4 Nm at full flexion down to 43.342 Nm. Subsequently, from the midstance subphase to the pre-swing subphase, the knee joint undergoes both extension and flexion, achieving a peak extension angle of 34.2 degrees. Following this peak, the knee flexes to 39.614 degrees, reaching a torque of -19.553 Nm, thereby concluding the stance phase.

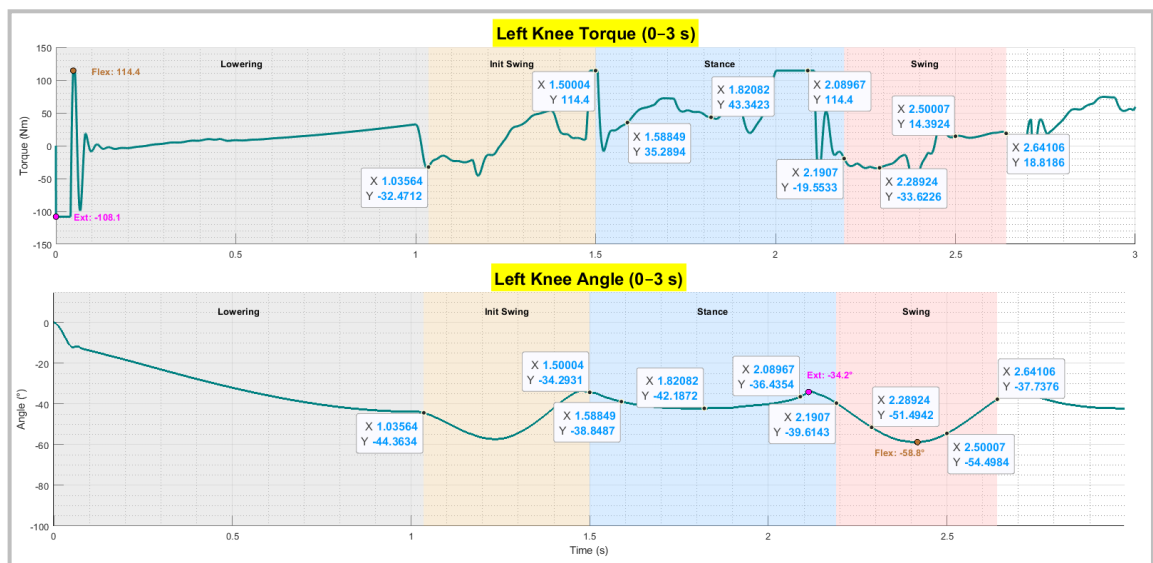


Figure 4.30: Knee torques and angles during the gait cycle

In analysing the ankle joint during the stance phase, this region is characterised by 85.929% flexion motion, with a peak flexion angle of 30.5 degrees, as illustrated in Figure 4.31.

The ankle flexion angle during the stance phase increases from 3.228 degrees at heel strike to 8.780 degrees during the loading response subphase, then to 17.848 degrees by midstance. Throughout the transition from heel strike to loading response, the torque fluctuates from an extension torque of 67.06 Nm to the maximum flexion torque of 69.2 Nm, which is sustained throughout midstance. After reaching a peak angle of 30.5 degrees during the swing phase, the flexion angle decreases to 28.929 degrees as the stance phase concludes with the pre-swing subphase, which features a maximum flexion torque of 69.2 Nm.

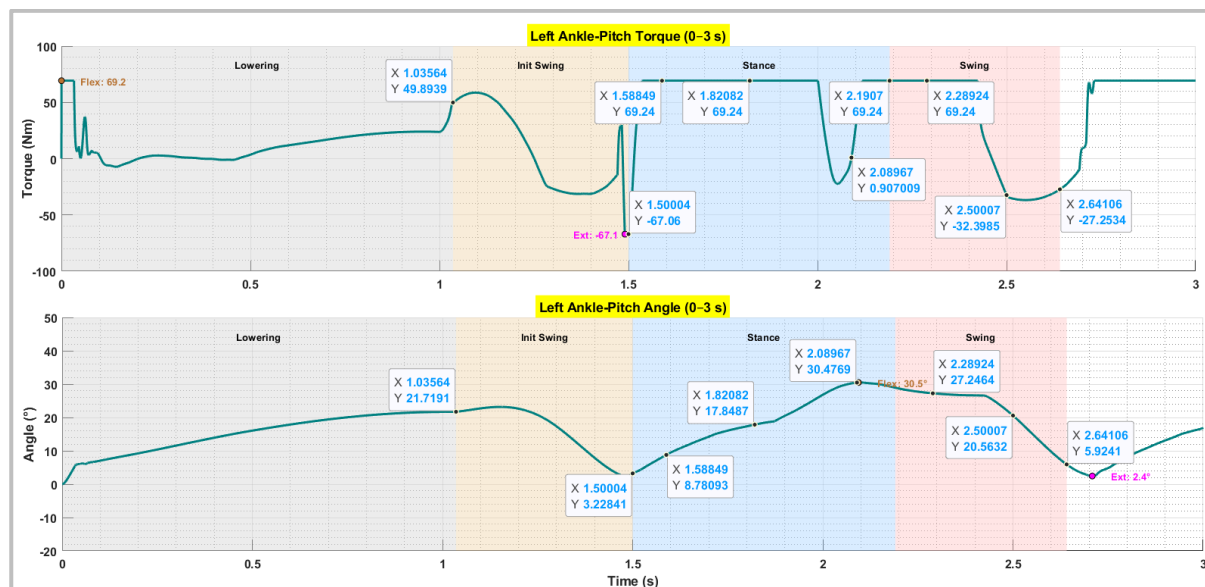


Figure 4.31: Ankle torques and angles during the gait cycle

The torques generated during the stance phase, as illustrated in Figures 4.29 to 4.31, indicate the capacity to support flexion and extension motion of the LLE. The simulation results reveal that the left leg can generate sufficient torque to support both the weight of the LLE and its user during the stance phase. Minor fluctuations in the torques at the hip, knee, and ankle were observed, which may be attributed to slight mechanical backlash or saturation of the hip joint's actuation, potentially leading to transient instability.

b. Swing phase

The figures below, related to the swing phase, provide a detailed view derived from the three-second gait cycle discussed earlier in the stance phase. This swing phase is further divided into three main subphases: toe-off, midswing, and terminal swing, which will be the focus of the upcoming analysis. During this process, the study examines the ability of the joints of the LLE to produce sufficient torque to move the swing leg.

The toe-off subphase commences at the conclusion of the pre-swing subphase, with the hip angle at 2.634 degrees, as illustrated in Figure 4.32. During this phase, a maximum flexion torque of 116.6 Nm is generated, providing sufficient torque to lift the thigh as the foot is prepared to leave the ground. By the conclusion of this subphase, the angle increases to 13.127 degrees.

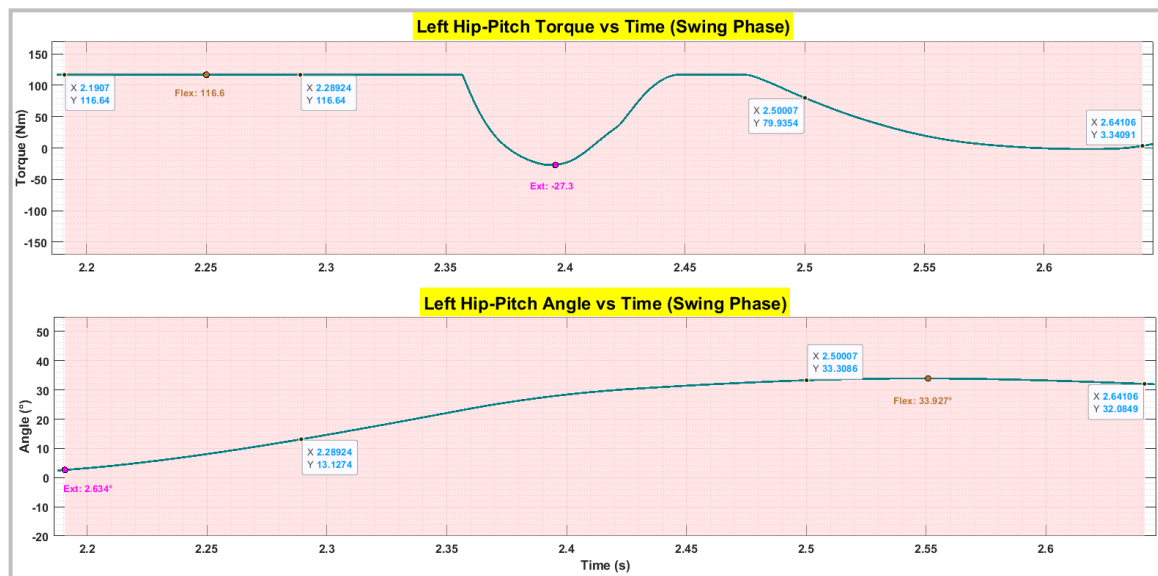


Figure 4.32: Hip torques and angles during swing phase

From the end of the toe-off to the midswing, the torque remains constant before fluctuating to -27.3 Nm. It then rises to its maximum value, with the LLE completing the midstance at a torque of 79.935 Nm and an angle of 33.308 degrees, close to the flexion peak angle of 33.927 degrees. The terminal swing phase concludes at an angle of 32.084 degrees. In summary, during the swing phase of the LLE, the torques vary between 116.6 Nm and -27.3 Nm, with the latter reflecting fluctuations throughout this phase. The angles associated with the swing phase range from 2.634 degrees to 32.084 degrees.

During the swing phase, the knee joint undergoes both flexion and extension. Flexion is used to lift the foot off the ground, while extension movement aids in propelling the lower leg forward, as illustrated in Figure 3.33.

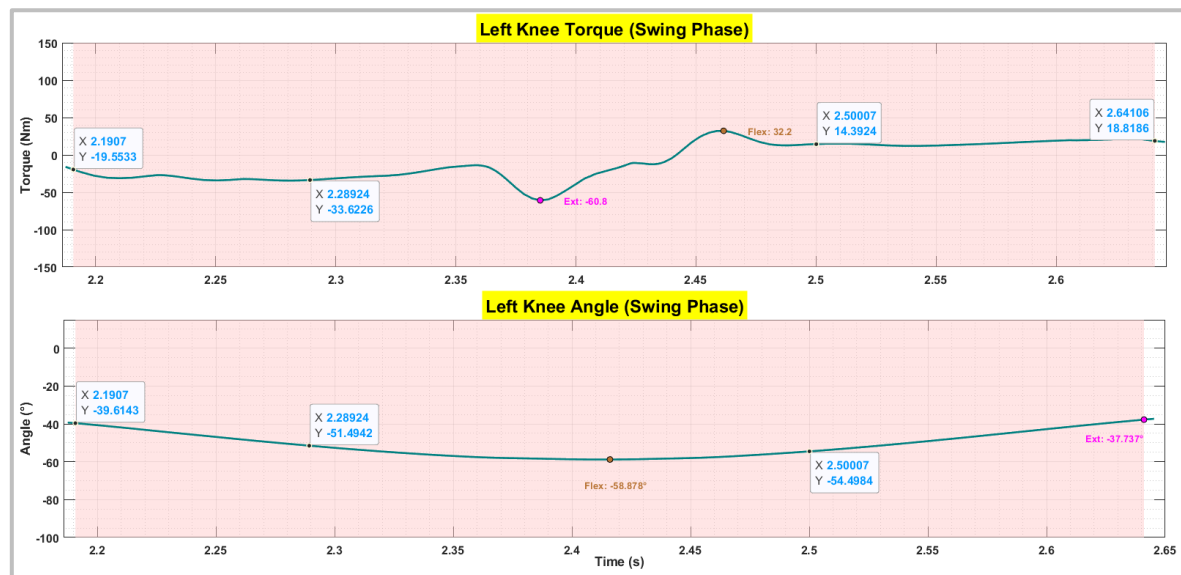


Figure 4.33: Knee torques and angles during swing phase

From the pre-swing subphase to toe-off, the knee joint reaches a maximum flexion torque of -33.622 Nm at an angle of -51.494 degrees. The joint continues to flex until the midswing subphase at -54.498 degrees, with an extension torque of 14.392 Nm. As the knee extends further, it reaches -37.737 degrees under an extension torque of 18.818 Nm. Throughout the swing phase, the knee attains a maximum flexion angle of -58.8 degrees, marking the valley. Subsequently, during the terminal swing, the joint extends to its maximum angle of 37.737 degrees, with torque fluctuating between -60 Nm and 32 Nm.

During the swing phase of the LLE in relation to the ankle joint, it is noted that the ankle primarily exhibits flexion movement. The angle in this phase varies from 27.9 degrees to 6.9 degrees, while the corresponding torque ranges between 69.2 Nm and -27.253 Nm, as illustrated in Figure 4.34.

A constant maximum dorsiflexion torque of 69.2 Nm facilitates the transition from the end of the pre-swing phase to the conclusion of the toe-off subphase. This torque is essential for generating sufficient force to lift the foot off the ground. It is briefly sustained before fluctuating, reaching -32.398 Nm at an angle of 20.563 degrees. Due to these fluctuations, the torque dips to -36.8 Nm. After this, until the end of the terminal swing subphase, the joint continues to dorsiflex in preparation for the upcoming heel strike of the next gait cycle, reaching an angle of 6.9 degrees, supported by a torque of 27.253 Nm just before heel strike.

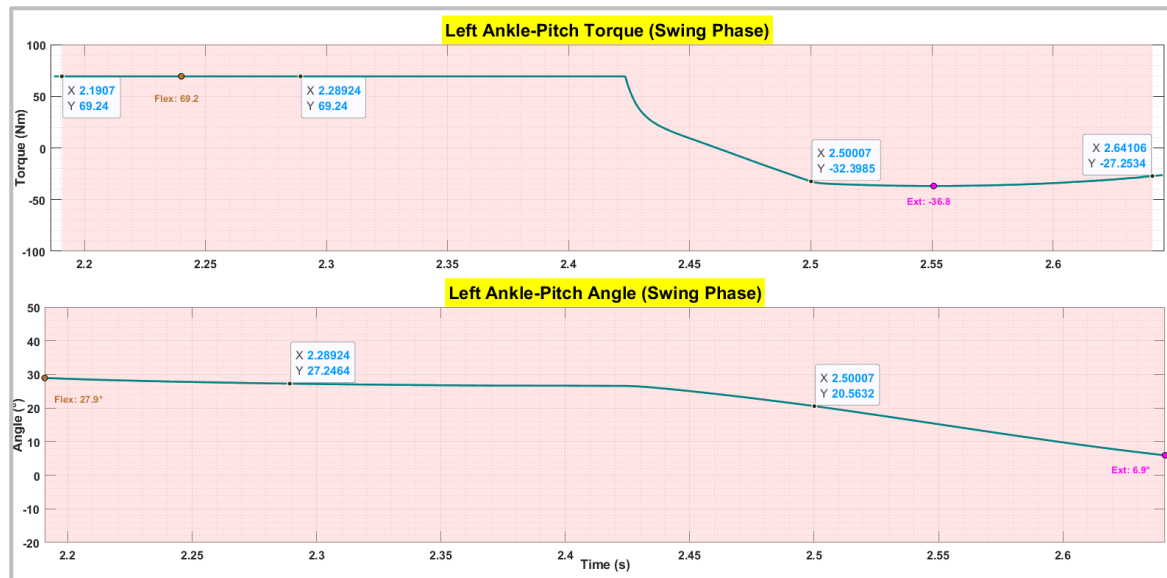


Figure 4.34: Ankle torques and angles during swing phase

c. Impact of torque changes on travel distance and torso height

This section examines how variations in torque may affect the robot's walking distance and torso height, following an examination of the LLE's behaviour at different speeds. Initially, the LLE torque was reduced by 10% in consecutive steps; after that, the same procedure was applied, increasing the initial torque by 10%. Table 4.3 outlines the torque manipulations for both scenarios. It is important to note that during this process, the velocity remains constant at 0.555 m/s. (+) and (-) denote flexion and extension, respectively.

Table 4.3: Torque values associated with displacement

Joint	Torque (-10%) Nm	Torque (+10%) Nm
Hip (+)	104.976	128.304
	94.478	141.134
Hip (-)	101.952	124.608
	91.575	137.069
Knee (+)	102.96	125.84
	92.664	138.424
Knee (-)	97.908	118.932
	87.577	130.825
Ankle (+)	62.136	76.164
	56.084	83.780

Ankle (-)	60.354	73.766
	54.132	81.143

The analysis of travel distance and torso height outcomes in relation to torque variations shows that reduced torque is associated with longer travel distances. However, this decrease in torque adversely affects torso height, causing the LLE to fall further below the optimal walking height of 0.87 metres, thereby leading to instability. In contrast, increasing the torque results in shorter travel distances, with a minimal effect on torso height compared with the previous scenario. This indicates that the robot demonstrates a stable walking pattern when torque is elevated. Figures J.1 to J.4 in Appendix J illustrate the effects of torque on both walking distance and torso height.

4.2.2 Conclusion

The joints of the LLE exhibit a slight amount of backlash; however, the hip, knee, and ankle joints can generate sufficient torque for both forward and backward movements, enabling the necessary angles throughout the gait cycle. During the stance phase, the torque produced by these joints supports the weight of both the LLE and its user. Conversely, during the swing phase, each joint of the swing leg effectively manoeuvres the leg. This capability allows the robot to navigate the seven subphases of the gait cycle successfully. Notably, it covered 5.63 metres in just 10 seconds while maintaining a stable and balanced gait.

CHAPTER 5: CONSTRUCTION AND VALIDATION

Chapter five outlines a process to validate whether the LLE's walking algorithm adheres to the principles of the human gait cycle. This is accomplished through two primary steps. First, various components of the LLE are manufactured specifically for this purpose, followed by the assembly of both the electronic and fabricated parts. The second step involves experimentation to ensure that the walking algorithm mimics the phases of the human gait cycle. An intermediate step, known as the coding step, is situated between the construction stage and the experimental process. During this task, it was necessary to ensure that the actuator synchronisation was aligned with the human gait cycle.

5.1 Manufacturing process and assembly of LLE

5.1.1 Scaling down process

A preliminary step before advancing to the manufacturing process was to scale down the designed LLE model. This reduction was accomplished using the similarity formula. Zhou et al. (2024) state that the scaling factor is instrumental in establishing the relationship between the prototype and its scaled models, where l_m and l_p are lengths related to the model and prototype, and l_s is the scaling factor, as defined in Equation 5.1.

$$\frac{l_p}{l_m} = L_s \quad 5.1$$

The scaling factor of 0.504 was determined by calculating the ratio of the desired scaled length of the LLE thigh to the length of the prototype thigh. This scaling factor was subsequently applied in the CAD software to resize the LLE. In Equation 5.2, the scaling factor of 0.205 related to torque was defined using the shaft diameters of the prototype (D_p) and the model (D_m). This operation uses the cubic rule as described by Saerens et al. (2019).

$$\left(\frac{D_p}{D_m}\right)^3 = L_s \quad 5.2$$

It is essential to note that the primary objective of this experiment is to assess the LLE's walking algorithm, particularly its ability to replicate the human gait cycle as simulated in MATLAB-SIMULINK. To this end, the scale-down process disregards the user's mass and focuses only on geometric and torque similarities. The comparative analysis in Table 5.1 highlights key characteristics of both the prototype and the model, including dimensions, materials, and torque. In this instance, the model's torque calculations focus on the LLE's frame and motor,

as the experiment excludes humans. At this stage, the experiment intended to confirm the walking algorithm. See Figure H.2 in Appendix H for torque calculations related to this model.

Table 5.1: Comparison between the LLE prototype and model (only frame and motors)

name	Model	Prototype
Hip torque	12.461 Nm	2.559 Nm
Knee torque	3.757 Nm	0.772 Nm
Ankle torque	0.320 Nm	0.065 Nm
LLE height	1.117 m	0.562 m
Thigh length	0.426 m	0.214 m
Shank length	0.412 m	0.207 m
Ankle height to ground	0.084 m	0.042 m
Mass of LLE	13.338 Kg	1.069 Kg
Material	Aluminum 6061T6 and 6082 T6	Plastic (PLA)
Density	2700 kg/m^3	1250 kg/m^3
strength	276 MPa (6061 T6) and 270MPa (6082 T6)	70 MPa

The estimated prototype torque values in the above table were used to select the servomotor for each joint of the LLE.

5.1.1.1 Modification and assembly

With the relationship between the prototype and model established, the model design underwent minor modifications to better accommodate the selected servomotor for experimental use. This adjustment was essential to maintain the simplicity of the LLE structure, thereby minimising the model's design complexity. Figure 5.1 illustrates the revised lower torso, thigh, shank, and foot.

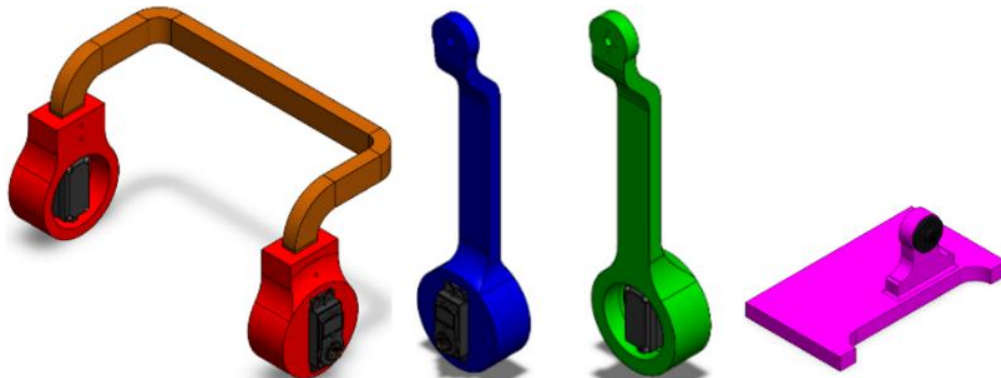


Figure 5.1: From left to right are the lower torso, thigh, shank, and foot

These components were assembled to ensure proper connection and interaction. For an overview of how the parts were mated, see Figure 5.2, which shows the complete assembly of the LLE.

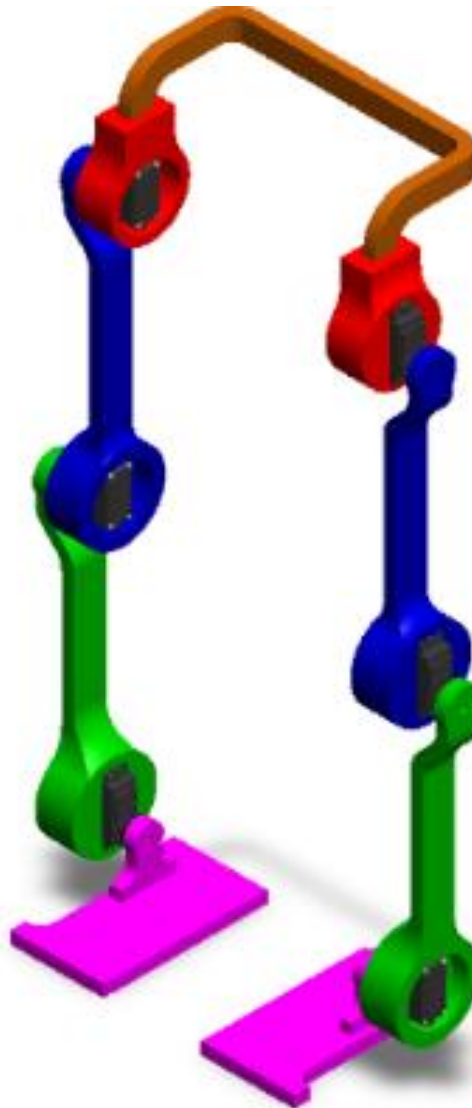


Figure 5.2: LLE prototype

5.1.2 Material selection

In reducing the aluminium LLE model to its scaled prototype, selecting an appropriate material was essential to achieve an optimal balance between manufacturability and cost efficiency. Dickinson (2023) chose Polylactic Acid (PLA) as an effective alternative to traditional materials such as aluminium, primarily for its cost-effectiveness and sufficient strength for prototype fabrication.

Its lower density compared with aluminium makes it an ideal choice to prevent overloading of the selected servomotor. The leg shown in Figure 5.3 is constructed from PLA; the same applies to the remaining segment of the frame, which is also constructed from PLA.



Figure 5.3 Leg of the LLE prototype made of PLA in white colour

5.1.3 Mechatronics

This section discusses the electronic components used to regulate the mechanical parts of the prototype, including servo motors, a motor driver, a microcontroller, a buck converter, a power supply, and a controller.

5.1.3.1 Motor and motor drive selection

To determine the appropriate servomotor for each joint in the prototype, the researcher evaluated the torque requirements using the moment method. Unlike the model's joint torque calculations, the prototype considered only the frame and motor weights. The findings revealed that the hip joint required the highest torque, followed by the knee and ankle joints, with corresponding torque values of 2.559 Nm (26.094 kg/cm), 0.772 Nm (7.872 kg/cm), and 0.064 Nm (0.6525 kg/cm), respectively. Based on these torque values, a servomotor with a torque rating of 30 kg/cm is ideal for the hip joint, 11 kg/cm for the knee joint, and 1.2 kg/cm for the

ankle joint. However, due to the availability of 40 kg/cm and 11 kg/cm servomotors in the research labs, the researcher chose to utilise these alternatives. Notably, these motors exceed the required torque specifications by 34.765% for the hip joint, 28.436% for the knee joint, and 94.068% for the ankle joint, making them suitable for experimentation.

Figure 5.4 shows the 40 kg/cm DS servo motor selected for the hip joint. This servo motor is recognised as a high-torque unit, offering a rotation range of 270 degrees, enabling extensive movement. The torque output varies from 36 kg/cm to 45 kg/cm, depending on the operating voltage, which ranges from 4.8 to 6.8 volts (Micro Robotics, 2024).



Figure 5.4: DS Servo 40kg/cm 270 Degree (Micro Robotics, 2024)

The TowerPro Servo Motor, identified as MG996R, features a design torque of 9.4 kg/cm at 4.8V, with a maximum torque of 11 kg/cm at 6V. This motor offers a range of motion from 0 to 180 degrees, making it ideal for a variety of robotic and mechanical applications, with the neutral axis located at 90 degrees (TowerPro, 2023). Figure 5.5 presents a visual representation of the MG996R.



Figure 5.5: Tower Pro MG996R 180 Degree 11 kg.cm (Makers, n.d.)

An Arduino device is designed to support components with a maximum voltage of 5 volts, which restricts its capability to accommodate the selected servo motors in this experiment. To address this limitation, a PCA9685 16-channel, 12-bit PWM (Pulse Width Modulation) servo driver was incorporated into the system, enhancing its functionality. This driver enables precise control of up to 16 servos or PWM-compatible devices simultaneously. By connecting the PCA9685 to the Arduino, the management of motors and other electronic components becomes more streamlined, facilitating more complex projects and smoother operation. Figure 5.6 shows the PCA9685, a 16-channel, 12-bit PWM controller used to control the hip, knee, and ankle motors.

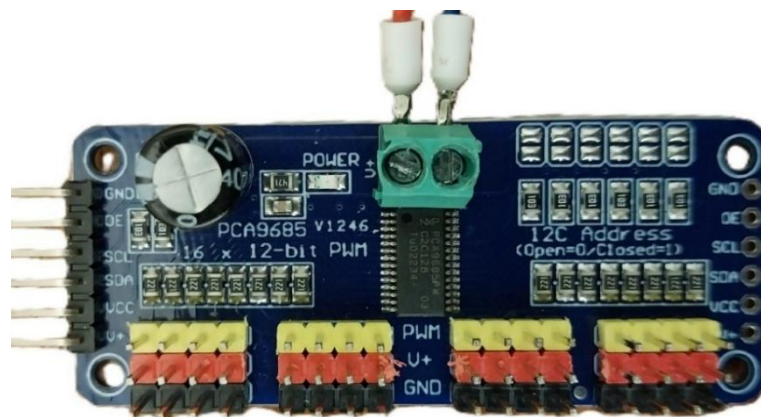


Figure 5.6: PCA9685, which is a 16-channel, 12-bit PWM controller

5.1.3.2 Microcontroller

The Arduino Uno Rev 3 is built around the ATmega328 8-bit microcontroller (MCU) from Microchip Technology. It features 14 digital input/output pins, six of which can be configured as PWM outputs, and six analogue inputs. A 16 MHz quartz crystal drives the device, which operates at 5 volts. The Arduino Uno is ideal for prototyping and educational purposes, as well as for robotics, wearable technology, and industrial automation, among other applications (Mouser electronics, 2024).

For this experiment, an Arduino microcontroller was selected to enable the robot to operate in accordance with its walking algorithm; the Arduino Uno was chosen as the primary platform for developing, transferring, and testing the walking code intended for the LLE prototype. This choice was made because the Arduino Uno is versatile and user-friendly, enabling the implementation and adjustment of the walking algorithms, as necessary. The testing phase involved iterative modifications to the code until it achieved optimal performance, at which point the LLE's walking behaviour mimicked a human walking gait cycle. The selected Arduino Uno is illustrated in Figure 5.7.



Figure 5.7: Arduino Uno (Arduino, 2024)

5.1.3.3 Buck converter and power supply

The step-down voltage regulator is engineered to convert an input voltage ranging from 5V to 36V DC, with a maximum current of 9A, to an output voltage between 1.2V and 34V. This flexibility allows the device to support a diverse array of electronic devices and power supplies, ensuring their operation remains within specified power parameters (Diy Electronics, n.d.).

A 24V, 350W power supply powered the LLE prototype, and the voltage was then stepped down to 5V using an adjustable voltage regulator. This regulated 5V output powered the hip

servomotor, which operated at 5-6.8V, as well as the knee and ankle servomotors, which operated at 4.8-6.6V. This configuration ensured that all motors remained within their specified voltage limits. Furthermore, the 5V output was sufficient to power both the Arduino and the motor drive, which also operated at 5V. Figure 5.8 shows the selected buck converter.

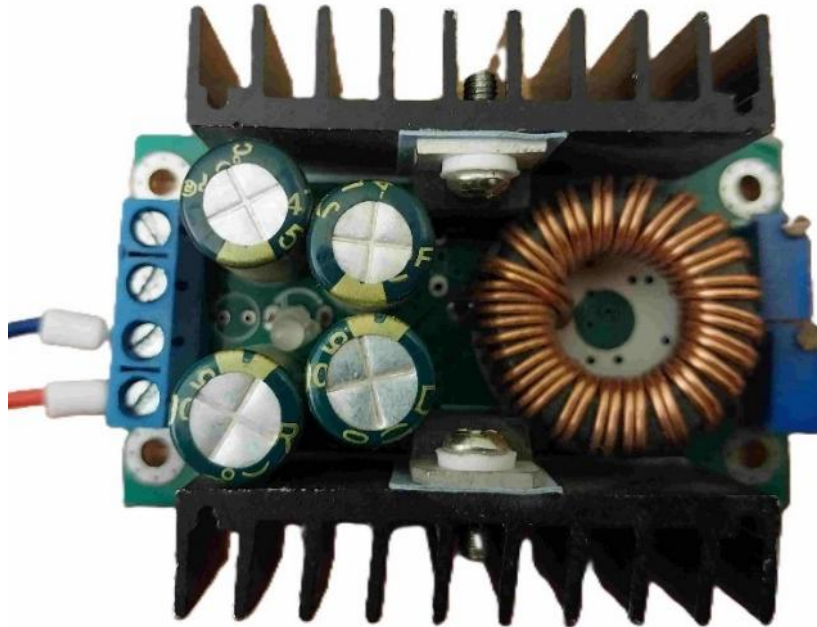


Figure 5.8: Buck converter

Figure 5.9 shows a 24 V power supply operating over a current range of 0 to 14.6 A. This power supply was selected to power the LLE prototype for experimental purposes. The primary considerations for its selection were its adequate voltage and current capacity to support all interconnected electronic components, including servomotors and the control system.

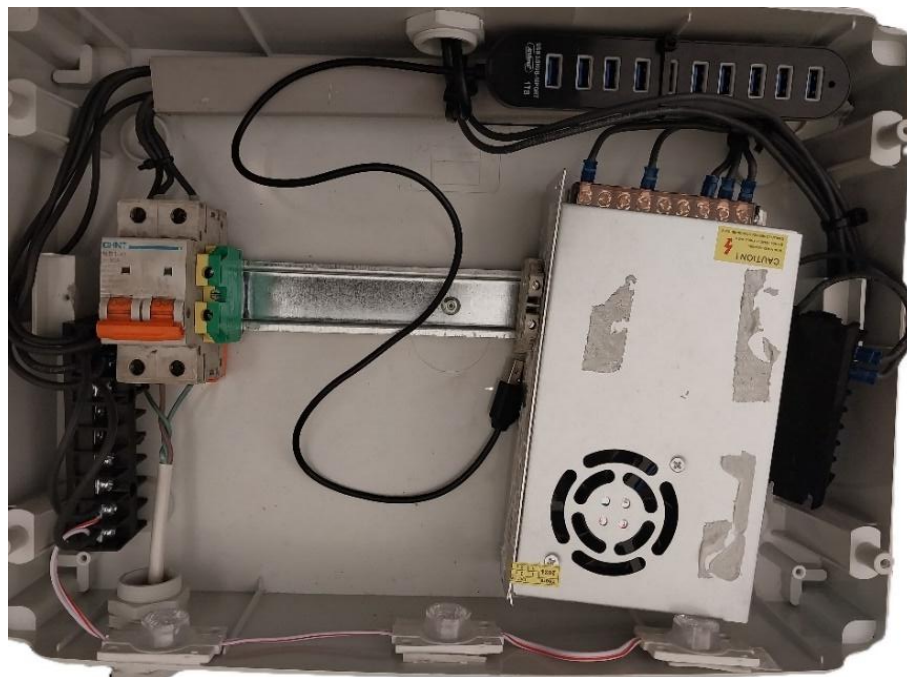


Figure 5.9: Power supply

5.1.3.4 Controller

The control system of the walking robot prototype is made of an Arduino Uno, a motor driver, a buck converter, and two 15A glass fuses. The walking algorithm was uploaded to and stored on the Arduino Uno, which, via the PCA9685 motor driver, instructed each joint of the LLE prototype to move, starting with the left leg. The buck converter ensured the servomotors operated at the correct voltage. The fuses provided stability and protection for the controller's electronic components. The assembly of the control system is illustrated in Figure 5.10. The microcontroller (Arduino Uno) and motor driver synchronise the actuators at each joint, ensuring smooth transitions between joints and minimising jerking.

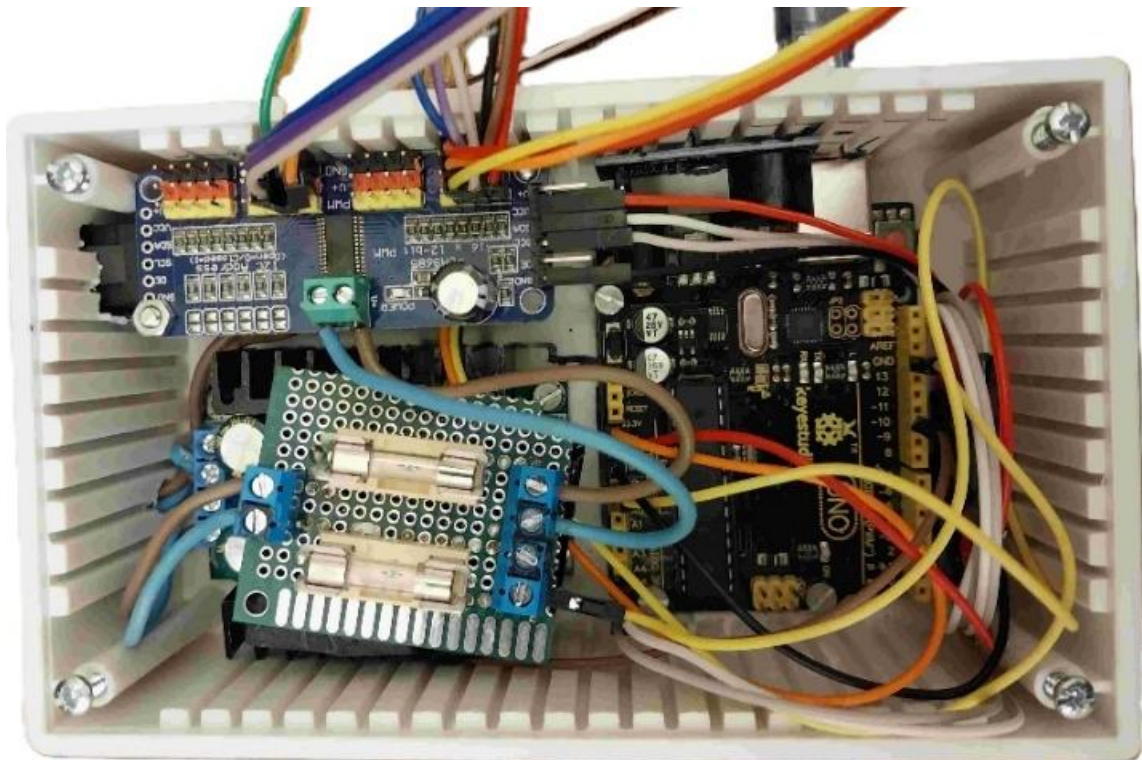


Figure 5.10: Control system

5.1.4 Assembly of the 3D-printed model

The electronic components were expertly integrated with the mechanical structure to develop a functioning LLE prototype. Motors were strategically placed within their respective joints. Each hip pitch joint is equipped with a 40 kg·cm motor that connects the waist to the hip and thigh. At the knee joint, an 11 kg·cm motor links the thigh to the shank, while another 11 kg·cm motor located at the ankle pitch joint connects the shank to the foot. Figure 5.11 illustrates the assembly of a portion of the prototype, showing, from left to right, the lower waist-to-hip joint coupled to the thigh, and the shank attached to the foot.



Figure 5.11: Right to left, the thigh connects to the hip, and the shank to the foot.

Following the assembly of the LLE segments, the components were joined to ensure the complete assembly aligned with the CAD design, with both legs connected to the lower torso, as illustrated in Figure 5.12.



Figure 5.12: Walking robot prototype

After completing the assembly of the walking robot prototype, it was suspended on a stand, as shown in Figure 5.13, for experimental purposes. This setup allowed the robot's legs to move freely in space, enabling observation of whether the robot followed the phases of the gait cycle. The wires from each joint servomotor were connected to the controller, which, via a buck converter, was connected to the power supply. A computer was used to upload the walking algorithm to the Arduino Uno and modify it as needed until it effectively mimicked the human gait cycle.

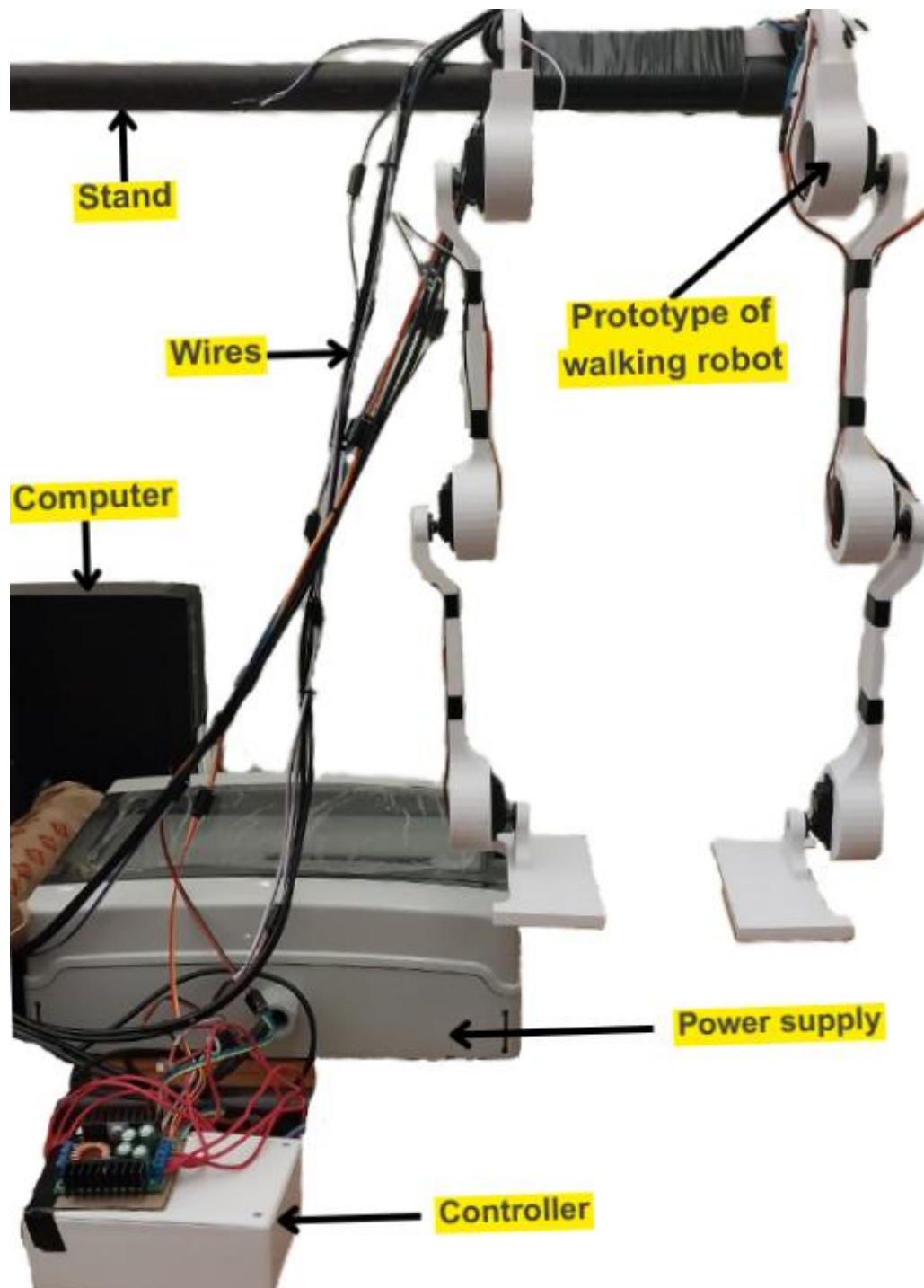


Figure 5.13: Suspended prototype with controller and power supply

5.1.5 Experimental protocol

The selected motors for this experiment were calibrated to reach a desired position, ensuring their initial positions aligned with the robot's neutral position. This step established the motors' neutral position in the Arduino code, facilitating a smooth walking motion. The walking gait cycle time was adjusted from 1500 to 3500 milliseconds (ms) in 500 ms increments until the optimal time was obtained and a smooth walking cycle was attained.

5.2 Experiment

The experimental section begins by outlining the objective of the experiment. Once this is established, the next step involves detailing the experimental setup. The walking algorithm was developed utilising the Arduino Integrated Development Environment (IDE). The final phase of the experiment focused on observing the walking gait cycle of the robot prototype. However, before reaching this stage, each joint in the robot's lower limb was individually assessed for responsiveness to ensure that each one moved according to the specified angles.

5.2.1 Purpose

The purpose of this experiment was to determine whether the walking robot replicated the phases of the human gait cycle through its walking algorithm. Given the nature of the experiment, the robot was evaluated while suspended on an experimental rig.

5.2.2 Experimental setup

The purpose of this experiment was to determine whether the walking configuration generated by the defined algorithm aligned with the simulation results. Therefore, in the absence of lateral supports, the prototype was suspended on a test rig during the experiment. The robot was powered by a 24V, 350W power supply, which was stepped down to 5V by a DC-DC buck converter and then directly connected to the PCA9685 motor controller to power the Arduino Uno and the six servo motors, which were connected to it and powered simultaneously.

5.2.2.1 Testing the motors

Figure 5.14 illustrates a leg of the LLE, where its actuated joints underwent preliminary testing to confirm their responsiveness to commands from the Arduino microcontroller. In this initial test, before the experiment, the motors at each joint were assessed under the load of the supported limb for 10 seconds, and the results were successful.

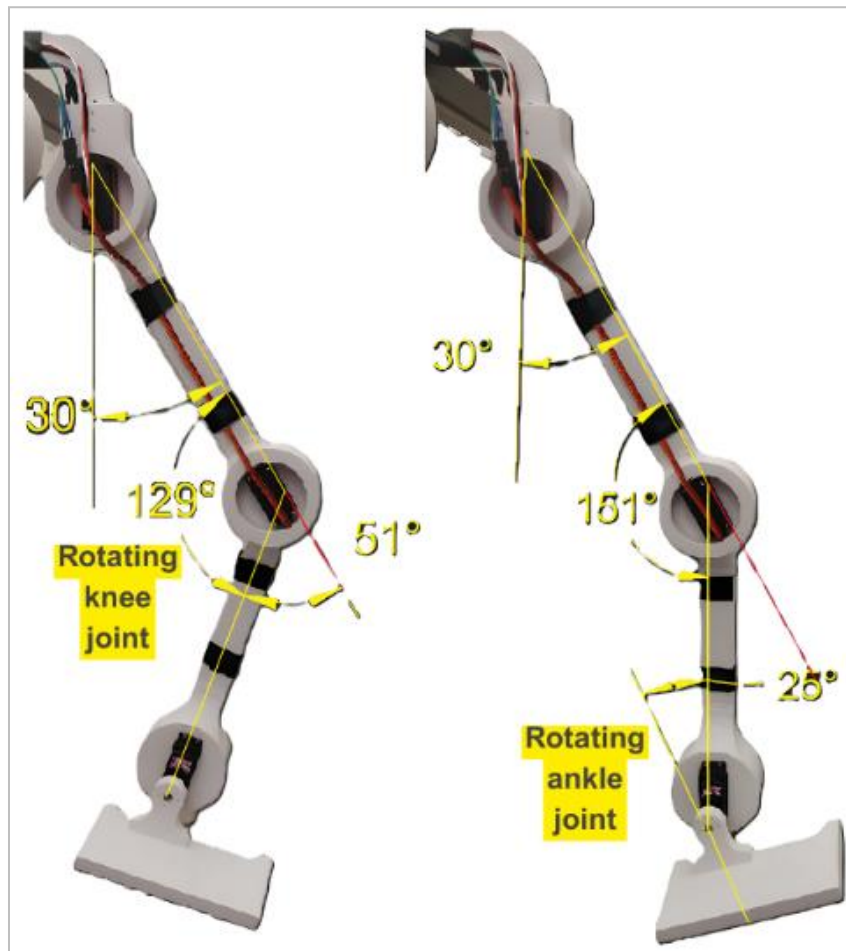


Figure 5.14: Test of actuated joints

5.2.3 Gait cycle walking algorithm

A section of the walking gait cycle code for the LLE prototype is illustrated in Figure 5.15. In this code, the neutral position of each robot joint was established after reaching its Zmodel position. The neutral positions for the hip, knee, and ankle were defined as 100° , 105° , and 90° , respectively. The stance phase was segmented into portions: 10% for the loading phase, 20% for the midstance phase, 20% for the terminal phase, and 10% for pre-swing. Conversely, during the swing phase, the walking robot experienced 13% in the initial phase, 14% in the midswing phase, and 13% in the terminal swing phase. Overall, the stance phase accounted for 60% of the walking gait cycle, while the swing phase comprised the remaining 40%. These percentage distributions for the subphases were derived from the literature review in Chapter 2.

To define a smooth walking gait cycle, the parameters were iterated several times until a satisfactory point was reached. The forward step length of the LLE prototype was defined as 30° hip joint flexion, while the step height was determined as 20° knee joint flexion. In the code, the gait cycle time was set to 2500 milliseconds, from the loading phase to the terminal swing. The smoothness of the gait cycle was determined by an update interval of 25 milliseconds.

```

1  #include <Wire.h>
2  #include <Adafruit_PWMServoDriver.h>
3  #include <math.h>
4
5  // Initialize PCA9685
6  Adafruit_PWMServoDriver pwm = Adafruit_PWMServoDriver();
7
8  // Servo configuration
9  #define SERVO_FREQ 50 // Analog servos run at ~50 Hz updates
10 #define SERVOMIN 150 // Minimum pulse length count (0 degrees)
11 #define SERVOMAX 600 // Maximum pulse length count (180 degrees)
12
13 // Joint mapping
14 const uint8_t R_HIP = 0; const uint8_t R_KNEE = 2; const uint8_t R_ANKLE = 5;
15 const uint8_t L_HIP = 7; const uint8_t L_KNEE = 8; const uint8_t L_ANKLE = 11;
16
17 // Neutral positions (degrees)
18 const float NEUTRAL_R_HIP = 100; const float NEUTRAL_R_KNEE = 105; const float NEUTRAL_R_ANKLE = 90;
19 const float NEUTRAL_L_HIP = 100; const float NEUTRAL_L_KNEE = 105; const float NEUTRAL_L_ANKLE = 90;
20
21 // Gait parameters
22 const float STEP_LENGTH = 30.0; // Degrees of hip movement
23 const float STEP_HEIGHT = 20.0; // Degrees of knee flexion
24 const float GAIT_CYCLE_TIME = 2500; // ms for complete gait cycle
25 const int UPDATE_INTERVAL = 25; // ms between updates (smoother motion)
26
27 // Gait phase percentages (total = 100%)
28 const float PHASE_LOADING = 10.0;
29 const float PHASE_MID_STANCE = 20.0; const float PHASE_TERMINAL_STANCE = 20.0; const float PHASE_PRE_SWING = 15.0;
30 const float PHASE_INITIAL_SWING = 13.0; const float PHASE_MID_SWING = 14.0; const float PHASE_TERMINAL_SWING = 13.0;

```

Figure 5.15: The structure of the Arduino walking gait cycle code.

Table 5.2 summarises the duration of each subphase of the walking robot along with its corresponding percentage.

Table 5.2: Gait cycle repartition

Phases	Percentage	Stance phase (time in ms)	Swing phase (time in ms)
Heel strike	0	0	
Loading phase	10	250	
Midstance	20	500	
Terminal stance	20	500	
Pre-swing	10	250	
Initial swing	13		325
Midswing	14		350
Terminal swing	13		325

5.2.4 Experiment results

Upon observing the walking robot during its gait-cycle experiment, it was apparent that the LLE prototype accurately replicated the seven subphases of the human gait cycle, advancing from the loading response to the terminal stance. Figures 5.16 and 5.17 depict 60% of the stance phase, illustrating the heel strike (initial contact), loading response, and midstance in Figure 5.16, as well as the terminal stance and pre-swing subphases in Figure 5.17.

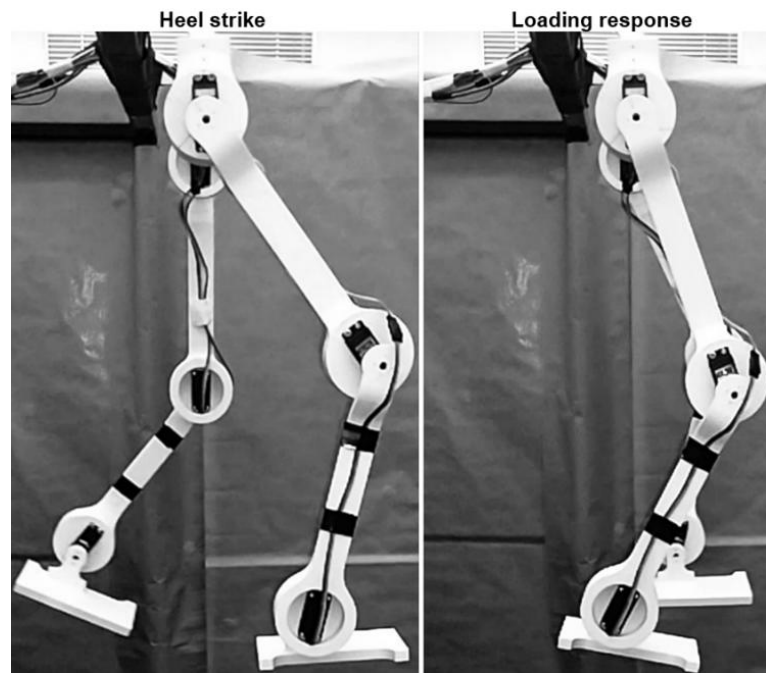


Figure 5.16: Stance phase (heel strike and loading response)

The heel-strike subphase was the initial stage of the stance phase, during which the left heel contacted the ground. During this subphase, the robot briefly entered a double-support phase, preparing to transition into a single-support phase characterised by the loading response, in which only the left leg supported the robot's weight. This position prepared the robot for the next subphase, the mid-stance. In the mid-stance, the exoskeleton's weight was expected to be borne by its left leg, indicating its intended contact with the ground. In the terminal stance subphase, the right leg had reached its final position, while the left leg continued to support the robot's weight, poised to lift off the ground.

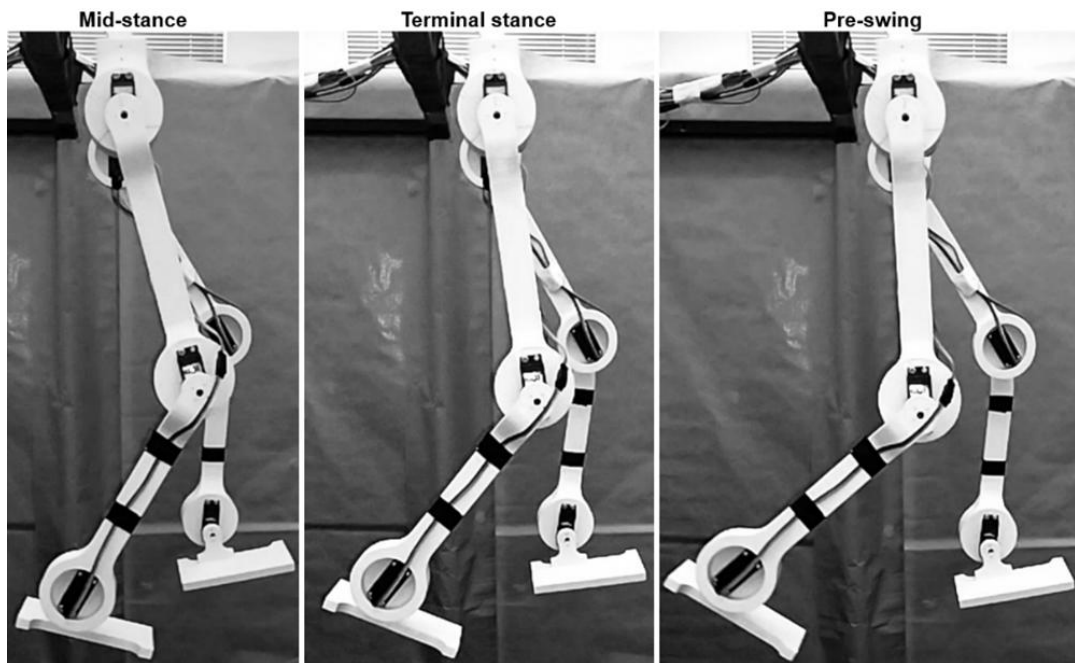


Figure 5.17: Stance phase (mid-stance, terminal stance, and pre-swing)

In the pre-swing subphase, the left leg began to lift off the ground; during this phase, the knee and ankle joints flexed and plantarflexed, respectively. However, minor irregularities, such as slight backlash in the knee, were noted. Despite these issues, the left leg effectively initiated and completed the stance phase.

In Figure 5.19, the experimental and simulation results are overlaid for comparative analysis. Figure 5.18 presents the torque results from the virtual-prototype simulation, with backlash at the hip joint highlighted in yellow. The results in this figure show that both the virtual prototype and the physical model successfully fulfil the stance phase of the gait cycle.

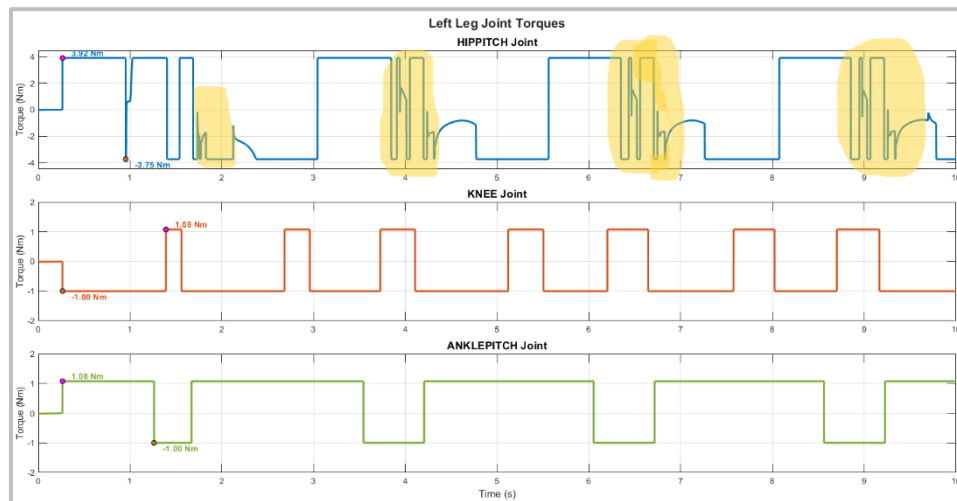


Figure 5.18: Simulation results of the hip joint demonstrating the occurrence of backlash

However, the presence of these backlashes in both the experimental and virtual prototypes led to discrepancies between the simulation and experimental outcomes, affecting the walking stability of the experimental robot compared to its virtual counterpart.

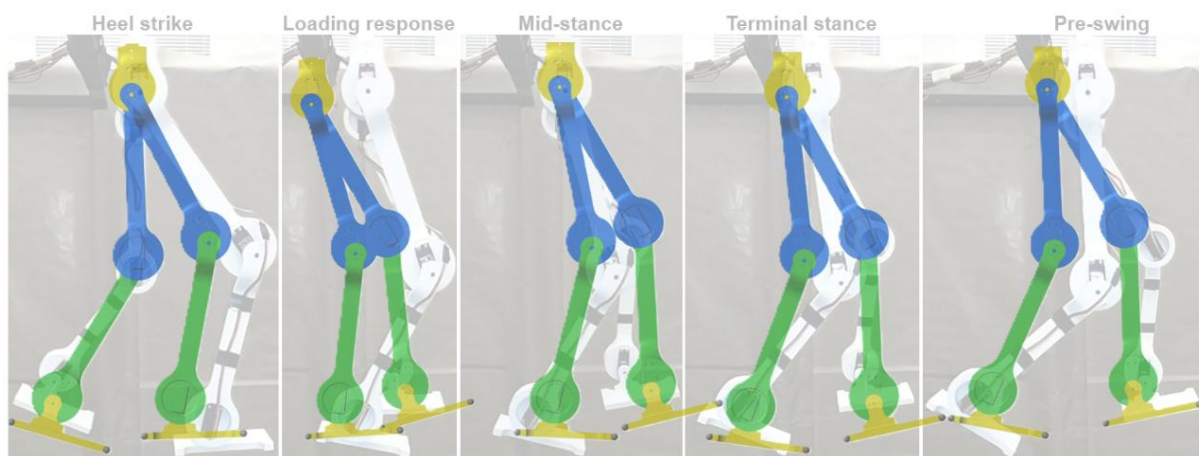


Figure 5.19: Stance phase (comparison of simulation and experimental results)

Figure 5.20 depicts the final 40% of the gait cycle, the swing phase. In this phase, the robot's weight was supported by the right leg while the left leg was elevated and swinging. During the initial swing, the left leg accelerated forward after being lifted off the ground. The exoskeleton's

knee joint flexed to maintain clearance between the ground and the left foot during its motion, while the hip joint facilitated the forward movement of the left leg.

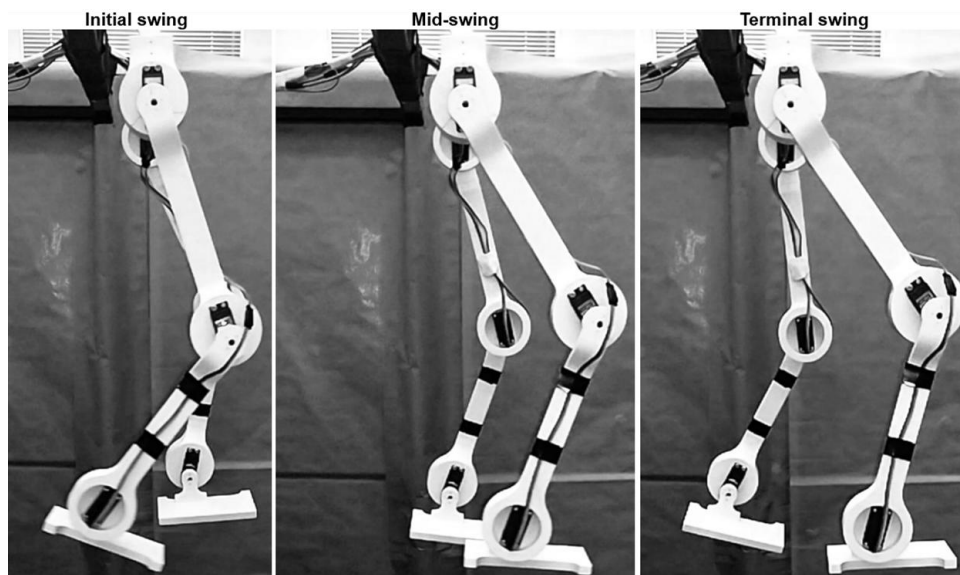


Figure 5.20: Swing phase (initial swing, mid-swing, and terminal swing)

During the midswing subphase, the left leg of the prototype continued its forward motion, distinct from the initial swing. Throughout this phase, the knee joint extended, while the hip joint remained flexed. As the leg prepared for heel strike, it gradually decelerated until the hip joint attained its maximum flexion during the terminal swing subphase. At this stage, the knee joint was fully extended, ensuring stable ground contact that transitioned into the heel strike.

In Figure 5.21, the overlaid images of the LLE prototype show similarities between the experimental and virtual prototypes. Both prototypes completed 40% of the walking gait cycle, although there were some instances of backlash.

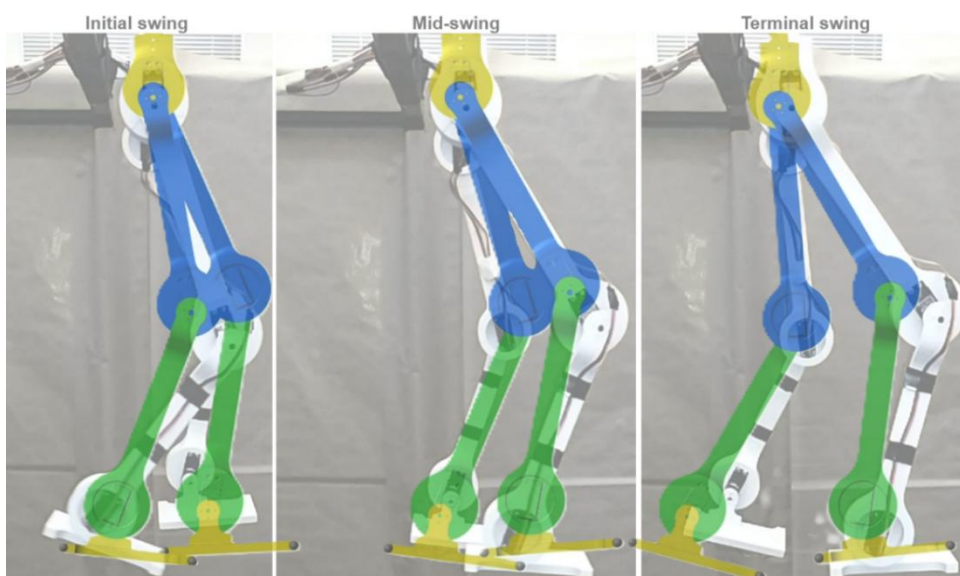


Figure 5.21: Swing phase (comparison of simulation and experimental results)

5.2.4.1 Conclusion

The conducted experiment demonstrates that the LLE prototype successfully completed the eight subphases of the human gait cycle. The motion of the affected joint was influenced by the backlash present in that joint. Notably, the motion of the virtual prototype was more fluid compared to that of the experimental prototype. During the stance phase, the experimental robot demonstrated its ability to support the walking robot's weight from heel strike through the pre-swing subphase. Meanwhile, during the swing phase, the selected motors effectively lifted and propelled the intended segment of the swinging leg forward, with minimal backlashes.

CHAPTER 6: CONCLUSIONS

The objective of this research was to develop a lightweight and compact LLE to support elderly and paraparesis individuals in their daily walking activities while ensuring that it does not increase metabolic cost. To achieve this goal, six primary research questions were formulated. The first question focused on identifying the most effective approach for designing a lightweight LLE. The second question centred on how an LLE of 30 kg could help an elderly person, with a height of 1.5-1.73 metres and a weight of up to 80 kg, walk. The third question examined the LLE's potential to reduce the user's metabolic cost. The fourth question focused on determining the structural integrity of the walking robot frame. The fifth question examined how changes in joint torque and velocity influence the robot's gait cycle. Finally, the sixth question examined how an assist-as-needed control system influences the interaction between the walking robot and its user.

To address these research questions, a set of eight objectives was established. This includes developing a kinematic and dynamic model for the LLE and creating a size-adaptable 3D model. Subsequent objectives focused on conducting FEA simulations and simulating the walking gait of the LLE, as well as designing an appropriate control system. Finally, an experimental model of the LLE was constructed and tested.

6.1 Achievement

The achievements of this research are outlined below, along with their respective levels of accomplishment, from the 3D modelling of the LLE to its experimental validation:

1. The primary objective of this research was achieved through the development of a 3D model of the LLE, which was calibrated to align with specific anthropometric characteristics of the human body. The LLE's size adjustability was designed with consideration for the 50th and 5th percentile male populations, while also accommodating the 5th, 50th, and 95th percentile female demographics. These anthropometric constraints were integral to ensuring an ergonomic and comfortable design for the LLE. Throughout the modelling process, decisions regarding materials and actuators were systematically made, as these factors directly influence the robotic device's overall weight.
2. After completing the 3D model of the LLE, its structural integrity was confirmed through FEA. The simulation evaluated the LLE segments under two scenarios: the first involved axial loading, while the second involved transverse loading.
3. The formulation and validation of the kinematic and dynamic modelling of the LLE were completed. The kinematic objective was divided into two sections: forward and inverse

- kinematics for each joint of the LLE. Analytical methods were utilised to establish the relationships between the positions and orientations of the joints and the end effector.
4. The dynamic modelling was derived using the Euler-Lagrange formulation, where, in this research, emphasis was placed on inverse dynamics to determine the torques required to actuate the thigh, shank, and foot. The calculated torques in this formulation were related to the position, velocity, acceleration, moment of inertia, and gravitational acceleration for each segment of the LLE.
 5. The simulation of the walking gait cycle was carried out to validate the kinematic and dynamic models of the LLE. The goal of this task was to generate a human-like gait cycle, enabling in-depth analysis of each joint's trajectory. This objective was deemed successful, as the LLE effectively mimicked the human gait cycle, with the generated motion pattern exhibiting consistency.
 6. The control system design was completed; however, it could not be further validated because the model's control system had not yet been built, as the design remains conceptual.
 7. A prototype was successfully built and tested to confirm the LLE's walking ability by ensuring that the robot generated a human-like gait cycle.

6.2 Findings summary

The achievements of this research served as tools for answering the six research questions defined in this thesis. The findings related to the LLE are described below as follows:

The first question concerned how to design a lightweight LLE effectively. This was achieved by selecting appropriate materials and components, which were key to developing a lightweight LLE for this project. Although many lightweight materials are available, it was essential to choose those with an optimal strength-to-weight ratio to meet the assistive device's requirements. Aluminium 6061-T6 and Aluminium 6082-T6 were selected over alternatives such as carbon fibre and titanium due to their availability and cost-effectiveness. The LLE design was refined to achieve the desired weight, with individual parts modified to meet the target weight specifications. Based on the selected material, the LLE frame weighs 4.942 kg. When combined with the chosen actuator, the total weight of the LLE reached 13.338 kg. Adding the weight of the two batteries increased the LLE's weight to 29,338 kg.

The centre of interest of the second question was to know whether a sagittal-plane LLE with a design weight of 30 kg and its modular ability to accommodate users with heights from 1.5 m to 1.73 m and weighing up to 81 kg, could reproduce the human gait cycle. Based on the simulated walking gait cycle obtained in MATLAB, the LLE, designed according to the user-defined criteria, could theoretically reproduce a human gait cycle. The results showcase the

walking robot's ability to travel from point A to point B, successfully covering 5.63 meters in 10 seconds.

The subsequent research findings regarding how an LLE can potentially reduce metabolic cost for its users and improve walking ability in both older adults and individuals with paraparesis demonstrate that a lightweight LLE design is recommended to effectively decrease metabolic expenditure. This efficiency is enhanced when the joint actuators can generate sufficient torque to compensate for the limited torque available from the user's muscles. With minimum torques of 104.516 Nm, 32.428 Nm, and 2.500 Nm for the hip, knee, and ankle, respectively, as determined by the Euler-Lagrange formulation, the device could support individuals with reduced muscle strength, such as the elderly or those with paraparesis, and could accommodate a maximum weight of 81 kg.

The question concerning the structural integrity of the LLE in the event of variations in the applied load demonstrates that the thigh, shank, and foot, which were subjected to axial loads during the mid-stance phase and transverse loads during the swing phase, were designed to withstand load variations of $\pm 10\%$. This is supported by the FEA results presented in Tables 4.1 and 4.2 under subsection 4.1.3, known as the load sensitivity analysis. The results indicate that the segments of the walking robot remained within the elastic region of the materials, despite high Von Mises stresses, with the lowest safety factor being 1.340 for the upper thigh, subjected to a 10% increase in its initial axial load. Conversely, the foot exhibits the highest safety factor of 11.62 when subjected to a 10% decrease in its initial axial load.

The subsequent question focused on how changes in torque or speed could impact the stability of the LLE. Simulation results indicated that both torque and speed variations significantly influenced the walking stability of the LLE. When the joints operated at lower torque, as shown in Table 4.3, the robot was able to cover a greater distance; however, this increased the LLE's walking instability. On the contrary, increasing joint torque enhanced the walking robot's stability, but at the expense of a shorter walking distance. Additionally, reducing the walking joint speed of the robot's actuators improved stability, though it also reduced the walking distance. Conversely, when the walking speed was increased, the robot could travel longer distances, but stability was compromised. In conclusion, achieving a stable walk for the LLE requires careful calibration of both the joint actuators' speed and torque.

The last research question addressed how an assist-as-needed control system contributes to the interaction between the LLE and its user. Only the theoretical section of this was achieved by making a conceptual design of the assist-as-needed control system. Theoretically, the control system was designed to ensure synchronisation between the robot joints and follow a defined trajectory. The assist-as-needed control system is designed to ensure both safety and

functionality, supporting users when needed while promoting quasi-independence during operation. By recognising the user's context and actions, the system can provide timely interventions, enhance the overall effectiveness of the robotic interaction, and foster a more intuitive user experience.

6.3 Experimental findings

The LLE model was designed to replicate the human walking gait cycle, specifically confined to the sagittal plane. The walking robot was required to follow a defined trajectory while maintaining a consistent gait. The experiment confirmed the ability of the LLE's joints to adhere to predetermined trajectories, allowing the LLE's end-effector (foot) to reach its target position. In contrast to the LLE model, the prototype presented some minor issues. Although the selected servomotors enabled effective movement across the LLE's leg segments, they exhibited slight backlash. This backlash adversely affects the gait cycle time and step length due to the resulting delays. Additionally, the backlash leads to jerky motion in the LLE, contributing to an unstable walking gait cycle with minor inconsistencies.

6.4 Contributions

This section outlines the contributions made throughout this research, particularly in 3D modelling of LLE, kinematic and dynamic modelling, simulation, and prototype experimentation. The contributions are listed below:

1. An anthropometric 3D design of the LLE was developed based on the 5th, 50th, and 95th percentiles of the female population, as well as the 5th and 50th percentiles of the male population. Incorporating human body characteristics into the design ensured comfort and ergonomic suitability during use. The LLE was engineered to accommodate individuals with heights ranging from 1.501 meters to 1.740 meters, facilitated by its adjustable feature.
2. A three-revolution (3R) planar kinematic model of the LLE was modified from the MATLAB program developed by the MATLAB team. An analytical approach was used to enable the robot's end-effector (foot) to attain a designated position while following a predetermined trajectory. This objective was achieved through both forward and inverse kinematics. Additionally, a walking trajectory resembling the human gait cycle was designed, allowing the hip, knee, and ankle joints to effectively reproduce flexion and extension movements, thereby facilitating the forward locomotion of the LLE in the sagittal plane.
3. Based on the Euler-Lagrange formulation, a dynamic model of the LLE was developed in MATLAB. This approach enabled the definition of the torques at the three joints. In

this formulation, the moment of inertia was incorporated, which proved valuable for assessing the joint torques required for walking.

4. A MATLAB and Simulink simulation was adapted from sample code created by the MATLAB team, focusing on the kinematic and dynamic modelling of the LLE. This simulation enables the observation of LLE walking behaviour. Furthermore, the walking trajectory was verified based on the principles of the LIPM.
5. The experiment illustrates that the performance of the walking robot is intricately linked to the quality of the selected actuator, as even minor issues within the system can impact the walking pattern and trajectory.

6.5 Limitations

Five limitations were identified in this research, which impeded the designed LLE from achieving its full potential and from being fully validated. These limitations are listed below:

1. The nature of this experiment prevented the incorporation of pressure sensors on the prototype, limiting the validation of how the LLE could react with the ground.
2. Actuator torque constraints, including limited torque and physical size, affected the capacity of the LLE hip, knee, and ankle joints to generate additional torque. While this issue could be mitigated by selecting larger motors, this approach was dismissed due to concerns that it might result in a heavier, bulkier LLE.
3. To facilitate this LLE experiment in this research, a scaled-down version of the model was developed; however, this approach limited the direct applicability of the experiment for testing the full capabilities of the LLE in use, thereby necessitating validation of the full-size LLE in future investigations.

6.6 Closing statement

This research makes a significant contribution to the field of LLE by offering kinematic and dynamic modelling of the LLE, along with a walking trajectory. The findings illustrate how the LLE could enhance walking abilities among the elderly population. Additionally, the importance of selecting appropriate materials and components was highlighted, as these choices directly affect the robot's weight.

Furthermore, creating a prototype that precisely mirrored the dimensions, actuators, control system, and sensor characteristics of the LLE model was not feasible due to the substantial budget required for such an endeavour. Consequently, a modified prototype was developed in its place.

CHAPTER 7: Recommendations

Due to the complexity of the LLE field, there are still areas that require improvement. This section outlines six recommendations based on the findings, contributions, and limitations.

1. Since the control system of the LLE was conceptually designed, it is recommended that this task be advanced by conducting experiments to confirm whether the designed control system is suitable.
2. Due to the increased weight of the LLE attributable to its associated battery, future research should consider selecting a lightweight battery with adequate power to operate the designated electrical devices on the walking robot. Optimisation action must be considered to ensure that the selected battery is lighter and capable of running for more than 1h 30 minutes, as calculated in this thesis.
3. Although the weight of the LLE in this study was measured at 4.942 kg using aluminium material known for its favourable strength-to-weight ratio, further optimisation employing a lighter material, such as carbon fibre, is anticipated to enhance the LLE frame's weight reduction by approximately 40%, when considering the density reduction from 2.7 Kg/m^3 to 1.6 Kg/m^3 .
4. Even if the walking gait cycle and trajectory simulation of the LLE demonstrated satisfactory results, minor fluctuations were observed. For this reason, further damping refinement is recommended to minimise these fluctuations, as they could affect the walking robot's stability.
5. On the hip and ankle joints, additional roll DOF should be considered to control the side balance of the LLE, as this will improve the lateral balance of the LLE and its user.
6. Work on the improvement of the actuator by ensuring a suitable combination of motor and harmonic drive capable of generating sufficient torque with no backlash and remaining a lightweight component.
7. To ensure that this LLE is suitable for elderly people, a clinical trial must be considered and supervised by experienced personnel in this area.
8. Implement methods to enhance the connection between the user and the LLE by incorporating soft materials, such as foam, around the frame where it is in direct contact with the user, to improve comfort in those regions. Additionally, define an appropriate brace to securely attach the thigh, shank, and waist to the LLE frame, minimising movement and ensuring stability.

REFERENCES

- Abdullah, N., Ibrahim, Z., Ramesh, S., Muhammad, M.A., Ya'Akub, S.R. & Hamid, M.Z.B.S.A. 2023. Design and Simulation of Low-Cost Lower Limb Exoskeleton for Rehabilitation Exercise. In *AIP Conference Proceedings*. American Institute of Physics Inc.
- Alcobendas-Maestro, M., Esclarín-Ruz, A., Casado-López, R.M., Muñoz-González, A., Pérez-Mateos, G., González-Valdizán, E. & Martín, J.L.R. 2012. Lokomat Robotic Assisted Versus Overground Training Within 3 to 6 Months of Incomplete Spinal Cord Lesion: Randomized Controlled Trial. *Neurorehabilitation and Neural Repair*, 26(9): 1058–1063.
- Al-Bender, F. 2005. On the definition and laws of friction. In *Proceedings of the World Tribology Congress III - 2005*. American Society of Mechanical Engineers: 871–872.
- Alex, T.G. & Sarah, A.-M. 2015. Load Cell Amplifier HX711 Breakout Hookup Guide. *SparkFun*. <https://learn.sparkfun.com/tutorials/load-cell-amplifier-hx711-breakout-hookup-guide/all> 7 February 2025.
- Aliman, N., Ramli, R. & Amiri, M.S. 2024. Actuators and transmission mechanisms in rehabilitation lower limb exoskeletons: A review. *Biomedizinische Technik*, 69(4): 327–345.
- Alshatti, A.K. 2019. *Design and Control of Lower Limb Assistive Exoskeleton for Hemiplegia Mobility*. Masters. Sheffield: University of Sheffield.
- ANSYS. Element Quality. ANSYS. https://ansyshelp.ansys.com/public/account/secured?returnurl=///Views/Secured/corp/v242/en/wb_msh/msh_Element_Quality_Metric.html?form=MG0AV3 16 September 2024.
- Arduino. 2024. Arduino Uno Rev3. *Arduino*. https://store.arduino.cc/products/arduino-uno-rev3?srltid=AfmBOorRCKuA6F-R1hA4QzKIXnKjz9_5gnjFCAiCPwQcmLbzQCNygvDI 22 November 2024.
- Atmaja, R., Munadi & Tauviquirrahman, M. 2017. Stress Analysis of Lower Limb Exoskeleton for Walking Assistance using Finite Element Method. *International Journal of Applied Engineering Research*, 12: 3864–3866. <http://www.ripublication.com>.
- Benjamin. 2024. Arduino and Teensy microcontroller boards. *AvionicsDuino*. <https://avionicsduino.com/index.php/en/arduino-and-teensy-microcontroller-boards/> 31 January 2025.
- Bortole, M. 2013. *Master thesis design and control of a robotic exoskeleton for gait rehabilitation*. Masterr. Madrid: Universidad Carlos Iii De Madrid.

- Bosscher, I.P.M., Melbourne, W. & Summer, M.D. 2016. Robotic exoskeleton with fall control and actuation., 2(12): 13.
<https://patentimages.storage.googleapis.com/ef/8e/fd/9218045d505332/US9474632.pdf>
- Carpentier, J., Benallegue, M., Mansard, N. & Laumond, J.P. 2015. A kinematics-dynamics based estimator of the center of mass position for anthropomorphic system - A complementary filtering approach. In *IEEE-RAS International Conference on Humanoid Robots*. Seoul: IEEE: 1121–1126. <https://ieeexplore-ieee-org.libproxy.cput.ac.za/document/7363493>.
- Castro, S. 2019. Walking Robot Control: From PID to Reinforcement Learning. *MathWorks*.
<https://blogs.mathworks.com/student-lounge/2019/04/24/walking-robot-control/> 10 March 2024.
- Castro, S. & Kim, B. 2019. Model-Based Control of Humanoid Walking. *MathWorks*. Model-Based Control of Humanoid Walking 5 November 2024.
- Centers for Disease Control and Prevention. 2017. Important Facts about Falls. *National Center for Injury Prevention and Control*: 1–3.
<https://www.cdc.gov/homeandrecreationalsafety/falls/adultfalls.html> 20 April 2021.
- Chen, B., Ma, H., Qin, L.Y., Gao, F., Chan, K.M., Law, S.W., Qin, L. & Liao, W.H. 2016. Recent developments and challenges of lower extremity exoskeletons. *Journal of Orthopaedic Translation*, 5: 26–37.
- Chen, B., Zi, B., Qin, L. & Pan, Q. 2020. State-of-the-art research in robotic hip exoskeletons: A general review. *Journal of Orthopaedic Translation*, 20: 4–13.
<https://www.sciencedirect.com/science/article/pii/S2214031X19302104>.
- Chen, B., Zi, B., Wang, Z., Qin, L. & Liao, W.H. 2019. Knee exoskeletons for gait rehabilitation and human performance augmentation: A state-of-the-art. *Mechanism and Machine Theory*, 134: 499–511.
- Chevallereau, C., Djoudi, D. & Grizzle, J.W. 2008. Stable Bipedal Walking with Foot Rotation Through Direct Regulation of the Zero Moment Point. *IEEE transactions on robotics*, 24(2): 390–401.
- Chittar, O.A. & Barve, S.B. 2022. Waist-Supportive Exoskeleton: Systems and Materials. *Materials Today: Proceedings*, 57: 840–845.
- Cumplido-Trasmonte, C., Molina-Rueda, F., Puyuelo-Quintana, G., Plaza-Flores, A., Hernández-Melero, M., Barquín-Santos, E., Destarac-Eguizabal, M. & García-Armada, E. 2023. Satisfaction analysis of overground gait exoskeletons in people with neurological pathology. a systematic review. *Journal of NeuroEngineering and Rehabilitation*, 20(1).

- Cursi, F., Bai, W., Yeatman, E.M. & Kormushev, P. 2022. GlobDesOpt: A Global Optimization Framework for Optimal Robot Manipulator Design. *IEEE Access*, 10: 5012–5023.
- Cyberdyne. 2014. HAL for medical use (Lower limb type). *Cyberdyne*.
https://www.cyberdyne.jp/english/products/LowerLimb_medical.html 24 December 2024.
- Van Dalen, S. 2012. *A Linear Inverted Pendulum Walk Implemented on TULip Committee: prof. dr.* Master. Eindhoven: Eindhoven University of Technology.
- Department of Economic and Social Affairs. 2019. *World Population Prospects 2019: Data Booklet*. New York.
- Desai, K.R. & Patel, P. 2017. Design of Walking Assistance Lower Limb Exoskeleton for Paraplegic Patients. *International Research Journal of Engineering and Technology (IRJET)*, 4(5): 1143–1146. <https://www.irjet.net/archives/V4/i5/IRJET-V4I5224.pdf>.
- Dickinson, M. 2023. Preliminary Prototyping and Simulations to Explore Mechanical Properties of 3D Printed Materials for Supporting the Head.
- Diy Electronics. DC-DC Adjustable Step-Down Module - 9A 1.2V-35V Output.
https://www.diyelectronics.co.za/store/switch-mode-modules/2487-dc-dc-adjustable-step-down-module-9a-12v-35v-output.html?srsId=AfmBOormzagyZV_dyO7psiUfidtzb5K79X0Dt1u0HbDiGREL6j9m7WCN 21 July 2025.
- Duburcq, A., Chevaleyre, Y., Bredeche, N. & Bo  ris, G. 2021. Online Trajectory Planning Through Combined Trajectory Optimization and Function Approximation: Application to the Exoskeleton Atalante., 2020. <https://hal.sorbonne-universite.fr/hal-03175242v1>.
- Engineering Product Design. 2023. Aluminium Alloy for CNC machining. *Engineering Product Design*. <https://engineeringproductdesign.com/knowledge-base/aluminium-alloy-for-cnc-machining/> 11 May 2024.
- Esquenazi, A., Talaty, M., Packel, A. & Saulino, M. 2012. The Rewalk powered exoskeleton to restore ambulatory function to individuals with thoracic-level motor-complete spinal cord injury. *American Journal of Physical Medicine and Rehabilitation*, 91(11): 911–921.
- Farooq mir, S. & Anbarasi, M. 2024. Dynamics equation for 2R planar manipulator using Lagrange dynamics. *IJIRT 169226 INTERNATIONAL JOURNAL OF INNOVATIVE RESEARCH IN TECHNOLOGY*, 850.
- Farris, R.J., Quintero, H.A. & Goldfarb, M. 2012. Performance evaluation of a lower limb exoskeleton for stair ascent and descent with Paraplegia. In *Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*. 1908–1911.

- Flynn, B. 2016. *Synthesizing an Adjustable Lower-Limb Exoskeleton Device for Clinical Rehabilitation*. Project. Worcester Polytechnic Institute.
- Fosch-Villaronga, E. & Özcan, B. 2020. The Progressive Intertwinement Between Design, Human Needs, and the Regulation of Care Technology: The Case of Lower-Limb Exoskeletons. *International Journal of Social Robotics*, 12(4): 959–972.
<https://doi.org/10.1007/s12369-019-00537-8>.
- Ganesan, V. & Gu, E.Y.L. 2021. Fall Protection Framework of Lower Extremity Exoskeleton Walking System Based on Differential Motion Planning. *International journal of social robotics*, 13(5): 969–980.
- Ghaddar, R. & Mohammad, A.S.M. 2019. A Review of Lower Limb Exoskeleton Assistive Devices for Sit-to-Stand and Gait Motion. *International Journal of Current Engineering and Technology*, 9(1): 105–111. <http://inpressco.com/category/ijcet%0AA>.
- Gavrila Laic, R.A., Firouzi, M., Claeys, R., Bautmans, I., Swinnen, E. & Beckwée, D. 2024. A State-of-the-Art of Exoskeletons in Line with the WHO's Vision on Healthy Aging: From Rehabilitation of Intrinsic Capacities to Augmentation of Functional Abilities. *Sensors*, 24(7).
- Golabchi, A., Chao, A. & Tavakoli, M. 2022. A Systematic Review of Industrial Exoskeletons for Injury Prevention: Efficacy Evaluation Metrics, Target Tasks, and Supported Body Postures. *Sensors*, 22(7).
- Goo, A., Laubscher, C.A., Farris, R.J. & Sawicki, J.T. 2020. Design and evaluation of a pediatric lower-limb exoskeleton joint actuator. *Actuators*, 9(4): 1–16.
- Di Gregorio, R. & Vocenas, L. 2021. Identification of gait-cycle phases for prosthesis control. *Biomimetics*, 6(2).
- Gurriet, T., Finet, S., Boeris, G., Duburcq, A., Hereid, A., Harib, O., Masselin, M., Grizzle, J. & Ames, A.D. 2018. Towards Restoring Locomotion for Paraplegics: Realizing Dynamically Stable Walking on Exoskeletons. In *2018 IEEE International Conference on Robotics and Automation (ICRA)*. IEEE: 2804–2811.
<https://en.wandercraft.eu/offres/personal-exoskeleton> 16 June 2024.
- Hamza, M.F., Ghazilla, R.A.R., Muhammad, B.B. & Yap, H.J. 2020. Balance and stability issues in lower extremity exoskeletons: A systematic review. *Biocybernetics and Biomedical Engineering*, 40(4): 1666–1679.
<https://linkinghub.elsevier.com/retrieve/pii/S0208521620301145>.
- Heike, V., Martijn, W. & Herman, van der Kooij. 2013. Leg exoskeleton. *Delft University of Technology*. <https://www.tudelft.nl/me/over/afdelingen/biomechanical-engineering/research/dbl-delft-biorobotics-lab/leg-exoskeletons> 16 October 2023.

- Høifødt, H. 2011. *Dynamic Modeling and Simulation of Robot Manipulators: The Newton-Euler Formulation*. Thesis. Norwegian university of Science and Technology.
- Huang, Q., Yokoi, K., Kajita, S., Kaneko, K., Aral, H., Koyachi, N. & Tanie, K. 2001. Planning walking patterns for a biped robot. *IEEE Transactions on Robotics and Automation*, 17(3): 280–289. <https://ieeexplore-ieee-org.libproxy.cput.ac.za/stamp/stamp.jsp?tp=&arnumber=938385>.
- Innab, A. 2019. *Wearable robotic exoskeleton*. Bachelors. Nablus: An-Najah National University. <https://www.researchgate.net/publication/338257885>.
- Iris Dynamics Ltd. 2023. Kinematics relation. <https://irisdynamics.com/articles/forward-and-inverse-kinematics> 8 January 2024.
- Jung, J.Y., Heo, W., Yang, H. & Park, H. 2015. A neural network-based gait phase classification method using sensors equipped on lower limb exoskeleton robots. *Sensors (Switzerland)*, 15(11): 27738–27759.
- Kajita, S., Kanehiro, F., Kaneko, K., Fujiwara, K., Harada, K., Yokoi, K. & Hirukawa, H. 2003. Biped walking pattern generation by using preview control of zero-moment point. In *Proceedings - IEEE International Conference on Robotics and Automation*. 1620–1626.
- Kajita, S., Kanehiro, F., Kaneko, K., Yok, K. & Hiruka, H. 2001. The 3D Linear Inverted Pendulum Mode: A simple modeling for a biped walking pattern generation. In *IEEE/RSJ*. 239–245.
- Kalula, S.Z., Ferreira, M., Swingler, G.H. & Badri, M. 2016. Risk factors for falls in older adults in a South African Urban Community Physical functioning, physical health, and activity. *BMC Geriatrics*, 16(1).
- Kapsalyamov, A., Jamwal, P.K., Hussain, S. & Ghayesh, M.H. 2019. State of the Art Lower Limb Robotic Exoskeletons for Elderly Assistance. *IEEE Access*, 7(July 2019): 95075–95086. <https://ieeexplore.ieee.org/document/8759880/>.
- Kerdraon, J., Previnaire, J.G., Tucker, M., Coignard, P., Allegre, W., Knappen, E. & Ames, A. 2021. Evaluation of safety and performance of the self-balancing walking system Atalante in patients with complete motor spinal cord injury. *Spinal Cord Series and Cases*, 7(1).
- Kim, J.H., Han, J.W., Kim, D.Y. & Baek, Y.S. 2013. Design of a walking assistance lower limb exoskeleton for paraplegic patients and hardware validation using CoP. *International Journal of Advanced Robotic Systems*, 10.
- Kljuno, E. & Williams, R.L. 2010. Humanoid Walking Robot: Modeling, Inverse Dynamics, and Gain Scheduling Control. *Journal of Robotics*, 2010: 1–19.

- Kruthika, K., Kiran Kumar, B.M. & Lakshminarayanan, S. 2016. Design and development of a robotic arm. In *2016 International Conference on Circuits, Controls, Communications and Computing (I4C)*. IEEE: 1–4. <https://ieeexplore.ieee.org/document/8053274/>.
- Laifual. 2023. About Laifual. <https://www.laifual.net/about.php> 23 January 2025.
- Laifual. LSG High Torque Harmonic Drive. <https://www.laifualgroup.com/product/lsg-high-torque-harmonic-drive.html> 5 November 2024.
- Lam, P. 2002. *Walking Algorithm for Small Humanoid*. Master. University of Auckland. <https://robotics.ee.uwa.edu.au/theses/2002-Walking-Lam.pdf>.
- Latif Shaari, N.A., Md Isa, I.S. & Chee Jun, T. 2015. Torque Analysis of The Lower Limb Exoskeleton Robot Design. 10(19). www.arpnjournals.com.
- Lau, J.C.L. & Mombaur, K. 2024. Can lower-limb exoskeletons support sit-to-stand motions in frail elderly without crutches? A study combining optimal control and motion capture. *Frontiers in Neurorobotics*, 18.
- Lejeune, T., Nuic, D., Dehem, S., Previnaire, J.G., Cuenot, C., Debugne, T., Kaps, J., Paul, B., Pean, V., Perez, S.S., Juhel, F., Tatsidou, S. & Kerdraon, J. 2025. Hands-free Atalante exoskeleton in post-stroke gait and balance rehabilitation: a safety study. *Journal of NeuroEngineering and Rehabilitation*, 22(1).
- Lerner, K.L. 2001. Anatomical Nomenclature.
- Li, L., Xie, Z., Luo, X. & Li, J. 2021. Trajectory planning of flexible walking for biped robots using linear inverted pendulum model and linear pendulum model. *Sensors (Switzerland)*, 21(4): 1–26.
- Li, N., Yan, L., Qian, H., Wu, H., Wu, J. & Men, S. 2015. Review on lower extremity exoskeleton robot. *Open Automation and Control Systems Journal*, 7(1): 441–453.
- Li, N., Yan, L., Qian, H., Wu, J., Men, S. & Li, Y. 2014. Design and simulation analysis of an improved lower limb exoskeleton. *Journal of Vibroengineering*, 16(7): 3655–3664.
- Lifeward. 2015. Lifeward Personal 6.0 Exoskeleton for Spinal Cord Injury. *Lifeward*. <https://golifeward.com/products/rewalkpersonal-exoskeleton/> 10 December 2023.
- Lord, S.R., Ward John A., Williams, P. & Anstey, K.J. 1993. An epidemiological study of falls in older community-dwelling women: the Randwick falls and fractures study. 17(3): 230–245.
- Lynch, K.M. & Park, F. 2017. *Modern Robotics -Mechanics, Planning, and Control*. New York. <http://hades.mech.northwestern.edu/images/7/7f/MR.pdf>.

- Makers. MG996R Servo Motor Tower Pro 180 Degree (Full Metal Aluminum with Brass Gear). <https://makerselectronics.com/product/mg996r-servo-motor-tower-pro-180-2/?srsltid=AfmBOorHoSYMw8mAYDTRinupoGukPxqsjvMugPTMhFGOLuWNt4nIGC1Z>.
- MathWorks. 2019. Modeling Contact Force Between Two Solids. *MathWorks*.
- Mathworks. 2019. Modeling Contact Force Between Two Solids. *Mathworks*.
<https://www.mathworks.com/help/sm/ug/modeling-contact-force-between-two-solids.html> 26 March 2025.
- MathWorks. 2012. Transform Sensor.
<https://www.mathworks.com/help/sm/ref/transformsensor.html> 24 January 2025.
- MaxAmps Lithium Batteries. 2022. Li-ion 35000mAh 11S7P 39.6v Battery Pack. *MaxAmps Lithium Batteries*. https://maxamps.com/products/li-ion-35000-11s7p-39-6v-battery-pack?srsltid=AfmBOoqQPSd_XOpJEfbkh8O0qtAlPAgO0zUob1gU0c1Q1bL2wqOKXFN S 15 February 2025.
- Maxon. 2015. *ESCON Module 50/5 Hardware Reference*. Sachseln.
- Maxon. 2019. Maxon Drives & Systems - Brushless DC motors. *Maxon group*.
<https://www.maxongroup.com/en-in/drives-and-systems/brushless-dc-motors> 23 January 2025.
- Maxon. 2021. *Precision Drive Systems: Maxon*. Sachseln.
- Maxon Group. 2012. *maxon EC motor*. Bad Homburg.
- Menga, G. & Ghirardi, M. 2019. Estimation and closed-loop control of COG/ZMP in biped devices blending CoP measures and kinematic information. *Robotics*, 8(4).
- Mertz, L. 2012. The Next Generation of Exoskeletons. *IEEE Pulse*, 3(4): 1–1.
<http://ieeexplore.ieee.org/document/6248742/>.
- Micro Robotics. 2024. DS Servo 40kg.cm 270 Degree. *Micro Robotics*.
<https://www.robotics.org.za/DS3240-270> 20 May 2025.
- Miller, L.E., Zimmermann, A.K. & Herbert, W.G. 2016. Clinical effectiveness and safety of powered exoskeleton-assisted walking in patients with spinal cord injury: Systematic review with meta-analysis. *Medical Devices: Evidence and Research*, 9: 455–466.
- Minchala, L.I., Astudillo-Salinas, F., Palacio-Baus, K. & Vazquez-Rodas, A. 2017. Mechatronic Design of a Lower Limb Exoskeleton. In *Design, Control and Applications of Mechatronic Systems in Engineering*. InTech.
- Monk, Simon. 2016. *Programming Arduino: getting started with sketches*. McGraw-Hill Education.

- Mouser electronics. 2024. Arduino Uno Rev. 3 Microcontroller Board. *Mouser Electronics*.
<https://www.mouser.co.za/new/arduino/arduino-rev-3-boards/> 19 April 2025.
- Murray, S.A., Ha, K.H., Hartigan, C. & Goldfarb, M. 2015. An assistive control approach for a lower-limb exoskeleton to facilitate recovery of walking following stroke. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 23(3): 441–449.
- Murtagh, J. 2015. *ReWalk: Robotic Exoskeletons for Spinal Cord Injury*.
- Nacy, S.M., Ghaeb, N.H. & Abdallh, M.M. 2016. A Review of Lower Limb Exoskeletons. *Innovative Systems Design and Engineering*, 7(11): 1–9. www.iiste.org.
- Naval Surface Warfare Center. 2007. *Handbook of Reliability Prediction Procedures for Mechanical Equipment Logistics Technology Support CARDEROCK DIVISION*. Carderock. www.dt.navy.mil.
- Önen, Ü., Botsalı, F.M., Kalyoncu, M., Şahin, Y. & Tinkır, M. 2017. Design and Motion Control of a Lower Limb Robotic Exoskeleton. In *Design, Control and Applications of Mechatronic Systems in Engineering*. InTech.
- Orin, D.E., Goswami, A. & Lee, S.H. 2013. Centroidal dynamics of a humanoid robot. *Autonomous Robots*, 35(2–3): 161–176.
- özgün. 2021a. Ansys Contact Types and Explanations. *MECHEAD*.
<https://www.mechead.com/contact-types-and-behaviours-in-ansys/> 10 August 2024.
- özgün. 2021b. Difference Between Static and Transient Analysis? *MECHEAD*.
<https://www.mechead.com/difference-between-static-and-transient-analysis/> 14 May 2024.
- Pamungkas, D.S., Caesarendra, W., Susanto, S., Soebakti, H. & Analia, R. 2019. Overview: Types of lower limb exoskeletons. *Electronics (Switzerland)*, 8(11): 1–12.
- Paredes, V. & Hereid, A. 2020. Dynamic Walking on exoskeleton ATALANTE with virtual constraints regulation. In Columbus. <https://cyberboticslab.com/publication/paredes-2020-dynamic/paredes-2020-dynamic.pdf> 2 February 2026.
- Pawar, J.S., Chouksey, M. & Tripathi, K. 2015. Modeling and Stress Analysis of Single Lap Bolted Joint. *Universal Journal of Mechanical Engineering*, 3(1): 21–26.
- Pendse, A.K. & Deshmukh, B.B. 2022. Study of aluminum alloy as a material for mounting arm. *International Journal of Mechanical Engineering and Technology (IJMET)*, 13(11): 1–15. <https://doi.org/10.17605/OSF.IO/MFE74>.
- Pheasant, S. 1996. *Body space: Anthropometry, Ergonomics and The Design of Work*. 2nd ed. London: Taylor & Francis. https://books.google.co.za/books?id=1DrR_nskp8AC.

- Pirker, W. & Katzenschlager, R. 2016. Gait disorders in adults and the elderly. *Wiener Klinische Wochenschrift*, 129(3–4): 81–95.
- Pirker, W. & Katzenschlager, R. 2017. Gait disorders in adults and the elderly: A clinical guide. *Wiener Klinische Wochenschrift*, 129(3–4): 81–95.
- Plagenhoef, S., Gaynor Evans, F. & Abdelnour, T. 1983. Anatomical Data for Analyzing Human Motion. 54(2): 169–178.
- Qiao, L., Liu, Y., Fu, C., Ge, L., Li, Y., Rong, X., Chen, T. & Zhang, G. 2024. A Reactive Planning and Control Framework for Humanoid Robot Locomotion. *Advanced Intelligent Systems*.
- Qiu, S., Pei, Z., Wang, C. & Tang, Z. 2023. Systematic Review on Wearable Lower Extremity Robotic Exoskeletons for Assisted Locomotion. *Journal of Bionic Engineering*, 20(2): 436–469.
- Radar Africa. 2024. The revered role of elders in African society. *Radar Africa*.
<https://radarafrica.com/africa/the-revered-role-of-elders-in-african-society/> 22 December 2023.
- Raut, S., Hase, V., Kotgire, S., Dalvi, S. & Maske, Y. 2021. Design and Analysis of Lightweight Lower Limb Exoskeleton for Military Usage. *International Journal for Research in Applied Science and Engineering Technology*, 9(9): 896–907.
- Ren, H., Shang, W., Li, N. & Wu, X. 2020. A fast parameterized gait planning method for a lower-limb exoskeleton robot. *International Journal of Advanced Robotic Systems*, 17(1).
- Rodríguez-Fernández, A., Lobo-Prat, J. & Font-Llagunes, J.M. 2021. Systematic review on wearable lower-limb exoskeletons for gait training in neuromuscular impairments. *Journal of NeuroEngineering and Rehabilitation*, 18(22): 1–21.
<https://doi.org/10.1186/s12984-021-00815-5>.
- De Rossi, S.M.M., Vitiello, N., Lenzi, T., Ronsse, R., Koopman, B., Persichetti, A., Giovacchini, F., Vecchi, F., Ijspeert, A.J., Van Der Kooij, H. & Carrozza, M.C. 2010. Soft artificial tactile sensors for the measurement of human-robot interaction in the rehabilitation of the lower limb. In *2010 Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBC'10*. 1279–1282.
- Rosyada, Z.F., Sulardjaka, Munadi & Muslim, E. 2019. Anthropometric and Biomechanics Analysis of Lower Limb Exoskeleton for Indonesian Population. In *IOP Conference Series: Materials Science and Engineering*. Institute of Physics Publishing.

- Sadun, A.S., Jalani, J. & Sukor, J.A. 2016. Force Sensing Resistor (FSR): a brief overview and the low-cost sensor for active compliance control. In *First International Workshop on Pattern Recognition*. SPIE: 1001112.
- Saerens, E., Crispel, S., García, P.L., Verstraten, T., Ducastel, V., Vanderborght, B. & Lefeber, D. 2019. Scaling laws for robotic transmissions. *Mechanism and Machine Theory*, 140: 601–621.
- Sardain, P. & Bessonnet, G. 2004. Forces acting on a biped robot. Center of pressure - Zero moment point. *IEEE Transactions on Systems, Man, and Cybernetics Part A: Systems and Humans*. 34(5): 630–637.
- Schiele, A. & van der Helm, F.C.T. 2009. Influence of attachment pressure and kinematic configuration on pHRI with wearable robots. *Applied Bionics and Biomechanics*, 6(2): 157–173.
- Sebe, M. 2016. *Humanoid Arm Geometric Model*. Master. Gävle: University of Gävle.
- Sehgal, M., Jacobs, J., Healthcare System, M. & Wendy Biggs, F.S. 2021. *Mobility Assistive Device Use in Older Adults*.
www.aafp.org/afpwww.aafp.org/fpm/2021/0300/p15.html.
<https://www.aafp.org/afp/2021/0615/p737-s1.html>.
- Shin, H.K. & Kim, B.K. 2010. Energy-efficient reference gait generation utilizing variable ZMP and vertical hip motion based on inverted pendulum model for biped robots. In *ICCAS 2010*. IEEE: 1408–1413.
- Siva Suriya, B.A., Balaji, R.K., Pooja, S.S. & M, S.A. 2008. Structural Analysis of Lower Limb Exoskeleton During Walking. *International Journal of Advances in Engineering and Management (IJAEM)*, 3: 542–549. www.ijaem.net.
- Sørensen, F.A., Madsen, S.K. & Nielsen, T.J. 2020. *Dynamic Walk of a Biped Robot Using model predictive control and a linear inverted pendulum model*. Master's thesis. Aalborg: Aalborg University. www.es.aau.dk.
- Statistics South Africa. 2019. *Mid-year population estimates*. Pretoria.
<https://www.statssa.gov.za/publications/P0302/P03022019.pdf>.
- Statistics South Africa. 2016. *Mid-year population estimates*. Pretoria.
<https://www.statssa.gov.za/publications/P0302/P03022016.pdf>.
- Strandberg, M. 2004. *Robot Path Planning: An Object-Oriented Approach*. Phd thesis. Royal Institute of Technology (KTH). <https://www.diva-portal.org/smash/get/diva2:7803/FULLTEXT01.pdf>.

- Sun, Z. & Roos, N. 2018. Dynamically stable walk control of biped humanoid on uneven and inclined terrain. *Neurocomputing*, 280: 111–122.
- Tekscan. 2018. Company. *Tekscan*. <https://www.tekscan.com/company> 2 February 2025.
- Tekscan. 2023. FlexiForce A301 Datasheet. *Tekscan*.
<https://www.tekscan.com/resources/datasheets-guides/flexiforce-a301-datasheet> 7 February 2025.
- Thakur, A., Dhiman, A., Ahmad, O., Balgotra, A., Kumar, S. & Channi, H.K. 2017. Modeling and Designing of Bipedal Walking Robot. *International Journal of Scientific Research in Science and Technology (www.ijrst.com)*, 3(8): 244–247.
- TowerPro. 2023. MG996R. *TowerPro*. <https://towerpro.com.tw/product/mg996r/> 2 July 2025.
- Trombin, E. 2022. *Low Obstacles Avoidance for Lower Limb Exoskeletons*. Masters. Padua: University of Padua.
- Vaughan, C.L., Davis, B.L. & O'Connor, J.C. 1999. *Dynamics of Human Gait*. 2nd Edition. Cape Town: [Kiboho Publishers]. <http://analisedemarcha.com/papers/livro/GaitBook.pdf> 8 December 2023.
- Viswanadh, C., Sarkar, A. & Sreedharan, P. 2019. Kinematic and dynamic simulation of biped robot locomotion on multi-terrain surfaces. *IOP Conference Series: Materials Science and Engineering*, 577(1): 1–10. <https://iopscience.iop.org/article/10.1088/1757-899X/577/1/012094>.
- Vukobratovic, M., Branislav, B. & Dragoljub, S. 2001. zero-Moment Point-Proper Interpretation and New Applications. In *Proceedings of the IEEE/RAS International Conference on Humanoid Robots*. Yugoslavia.
- Walden, M. 2019. Anatomical Planes of Motion. *TeachPE*.
<https://www.teachpe.com/anatomy-physiology/planes-of-movement> 16 November 2023.
- Wandercraft. 2023. Personal exoskeleton. *Wandercraft*.
- Wang, S., Wang, L., Meijneke, C., Van Asseldonk, E., Hoellinger, T., Cheron, G., Ivanenko, Y., La Scaleia, V., Sylos-Labini, F., Molinari, M., Tamburella, F., Pisotta, I., Thorsteinsson, F., Ilzkovitz, M., Gancet, J., Nevatia, Y., Hauffe, R., Zanow, F. & Van Der Kooij, H. 2014. Design and Control of the MINDWALKER Exoskeleton. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 23(2): 277–286.
<http://ieeexplore.ieee.org/document/6940308/>.
- Wee, T.C. 2014. *Design and Control of a Humanoid Robot*. Doctor of Philosophy thesis. London: Imperial College London. <https://core.ac.uk/download/pdf/76995315.pdf>.

- Westervelt, E.R., Grizzle, J.W., Chevallereau, C., Choi, J.H. & Morris, B. 2007. *Feedback Control of Dynamic Bipedal Robot Locomotion*. S. S. G. and F. Lewis, ed. Florida: CRC Press.
- Williams, J.R., Cuddeback, C.F., Fang, S., Colley, D., Enlow, N., Cox, P., Pridham, P. & Lerner, Z.F. 2024. OpenExo: An Open-Source Modular Exoskeleton to Augment Human Function. <http://biorxiv.org/lookup/doi/10.1101/2024.10.02.616295>.
- Wilson, A.B. 1972. Transition. *Artificial Limbs*, 16(1).
- Wong, J., Song, H. & Piórkowski, J. 2010. Powered exoskeleton leg. *University of Waterloo*. <http://www.eng.uwaterloo.ca/~jpiorkow/fydp/design.html> 6 January 2025.
- Xing, H., Lv, G., Lin, J., Gregg, R.D. & Atkeson, C.G. 2020. A Task-Invariant Learning Framework of Lower-Limb Exoskeletons for Assisting Human Locomotion. In *2020 American Control Conference (ACC)*. AACC: 569–576.
- Yang, T., Zhang, W., Chen, X., Yu, Z., Meng, L. & Huang, Q. 2018. Turning gait planning method for humanoid robots. *Applied Sciences (Switzerland)*, 8(8).
- Yang, W., Zhang, J., Zhang, S. & Yang, C. 2020. Lower limb exoskeleton gait planning based on crutch and human-machine foot combined center of pressure. *Sensors (Switzerland)*, 20(24): 1–17.
- Yilmaz, M. 2010. *HUMANOID ROBOT OMNIDIRECTIONAL WALKING TRAJECTORY GENERATION AND CONTROL*. Master. Istanbul: Sabanci University.
- Young, A.J. & Ferris, D.P. 2017. State of the art and future directions for lower limb robotic exoskeletons. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 25(2): 171–182.
- Yu, L., Leto, H. & Bai, S. 2023. Design and Gait Control of an Active Lower Limb Exoskeleton for Walking Assistance †. *Machines*, 11(9).
- Yu, Z., Huang, Q., Ma, G., Chen, X., Zhang, W., Li, J. & Gao, J. 2014. Design and Development of the Humanoid Robot BHR-5. *Advances in Mechanical Engineering*, 2014(852937): 1–11. <https://journals.sagepub.com/doi/pdf/10.1155/2014/852937>.
- Zeghloul, S., Laribi, M.A. & Gazeau, J.-P. 2016. Robotics and mechatronics. In S. Zeghloul, M. A. Laribi, & J.-P. Gazeau, eds. *Mechanisms and Machine Science volume 37*. Switzerland: Springer US: 1–311.
- Zhang, L., Liu, Y., Wang, R., Smith, C. & Gutierrez-Farewik, E.M. 2021. Modeling and Simulation of a Human Knee Exoskeleton's Assistive Strategies and Interaction. *Frontiers in Neurorobotics*, 15.

- Zhao, R. 2016. *Trajectory planning and control for robot manipulations*. PHD these. UNIVERSITE DE TOULOUSE. <https://tel.archives-ouvertes.fr/tel-01285383v2/document>.
- Zheng, Y., Ma, C., Wu, Z., Shi, Y. & Xiao, M. 2023. Kinematic Characteristics Analysis of Wearable Lower Limb Rehabilitation Exoskeleton Robot based on ADAMS. *International Journal of Mechatronics and Applied Mechanics*, (13).
- Zoss, A., Kazerooni, H. & Chu, A. 2005. On the mechanical design of the Berkeley Lower Extremity Exoskeleton (BLEEX). In *2005 IEEE/RSJ International Conference on Intelligent Robots and Systems, IROS*. Berkeley: IEEE&RSJ: 3132–3139.
- Zhou, H., Okada, T., Kawamura, Y., Wang, D. & Hayakawa, G. 2024. Improved Similarity Law for Scaling Dynamic Responses of Stiffened Plates with Distorted Stiffener Configurations. *Applied Sciences (Switzerland)*, 14(14).

Appendix A: Overall dimensions of LLE segments and the LLE

Isometric View: Shows the assembly with dimensions: 348,00 (width), 385,20 (height), and 505,20 (total height). Callouts 1, 2, 3, and 4 point to the main arm, joints, and end effectors.

Front View: Shows the assembly with dimensions: 444,10 (width), 165,10 (height), and 263,50 (total height).

Side View: Shows the assembly with dimensions: 348,00 (width), 385,20 (height), and 505,20 (total height).

Parts List:

ITEM NO.	PART NUMBER	QTY.
1	Lower Torso	1
2	M6 x 25,9mm bolt	4
3	M6_nut	4
4	Waist-to-hip	2

Assembly Name: Lower torso
Part Number: LLE-sub-ass 001

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Figure A.2 presents comprehensive details about the thigh structure. It includes key dimensions, such as the total length of the thigh and the precise distance from the centre of the hip joint to the centre of the knee joint. Specifically, this distance measures 426.3 mm. This detailed information is crucial for ensuring accurate fit and proper alignment in various applications.

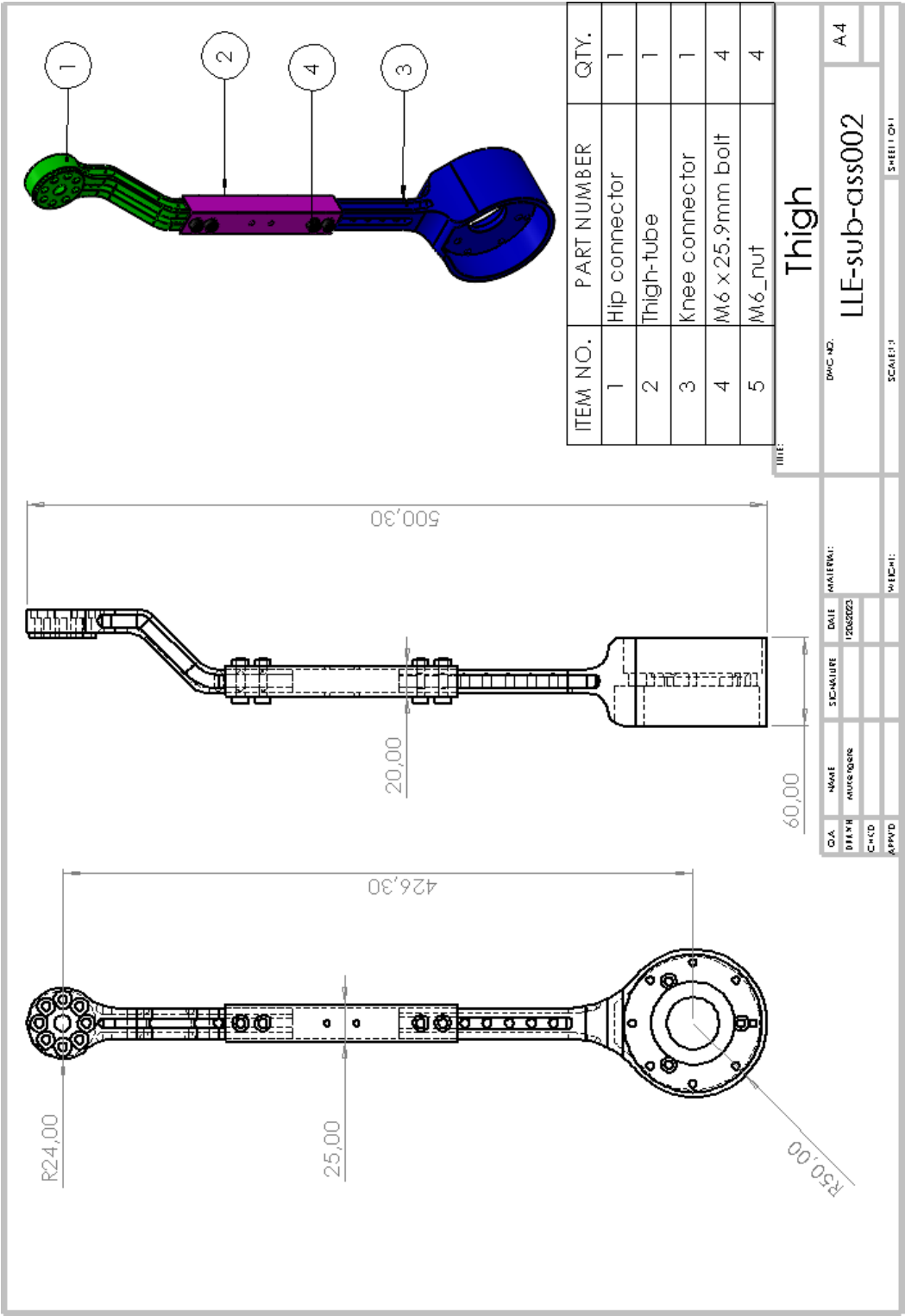


Figure A.2: Thigh front, side view, and 3D

The end-to-end distance between the knee joint and the ankle joint is 412.38 mm in its maximum extension, as illustrated in Figure A.3, related to the shank.

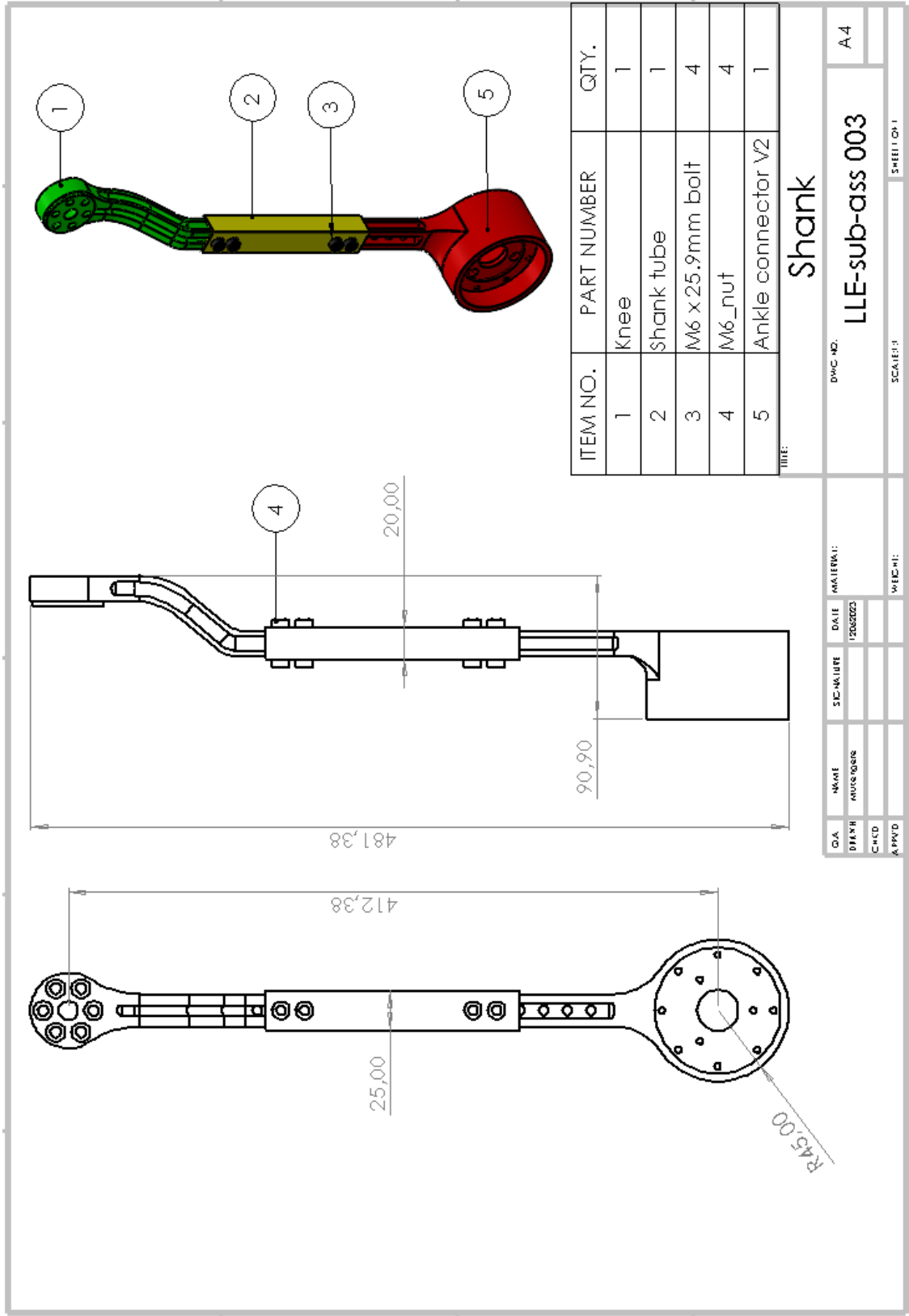


Figure A.3: shank front, side view, and 3D

The overall dimensions of the foot are 245 mm in length, 145 mm in width, and 83.5 mm in height measured from the center of the ankle joint to the bottom of the foot plate is presented

in Figure A.4. In contrast to the adjustable components associated with the thigh, shank, and lower torso of the LLE, the foot subassembly is not an adjustable segment.

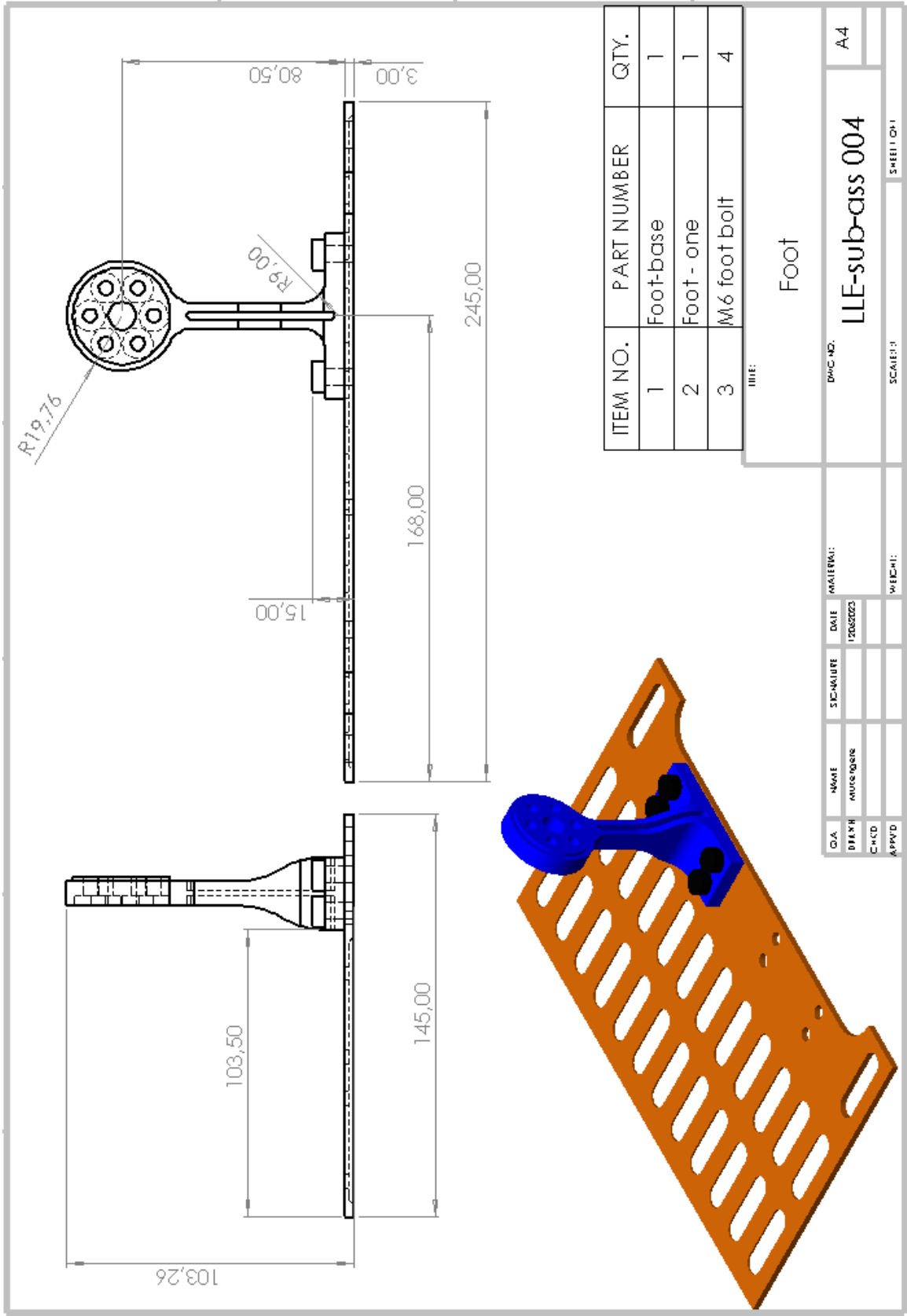


Figure A.4: Foot, front, side view, and 3D.

Finally, the overall dimensions of the LLE are provided in Figure A.5.

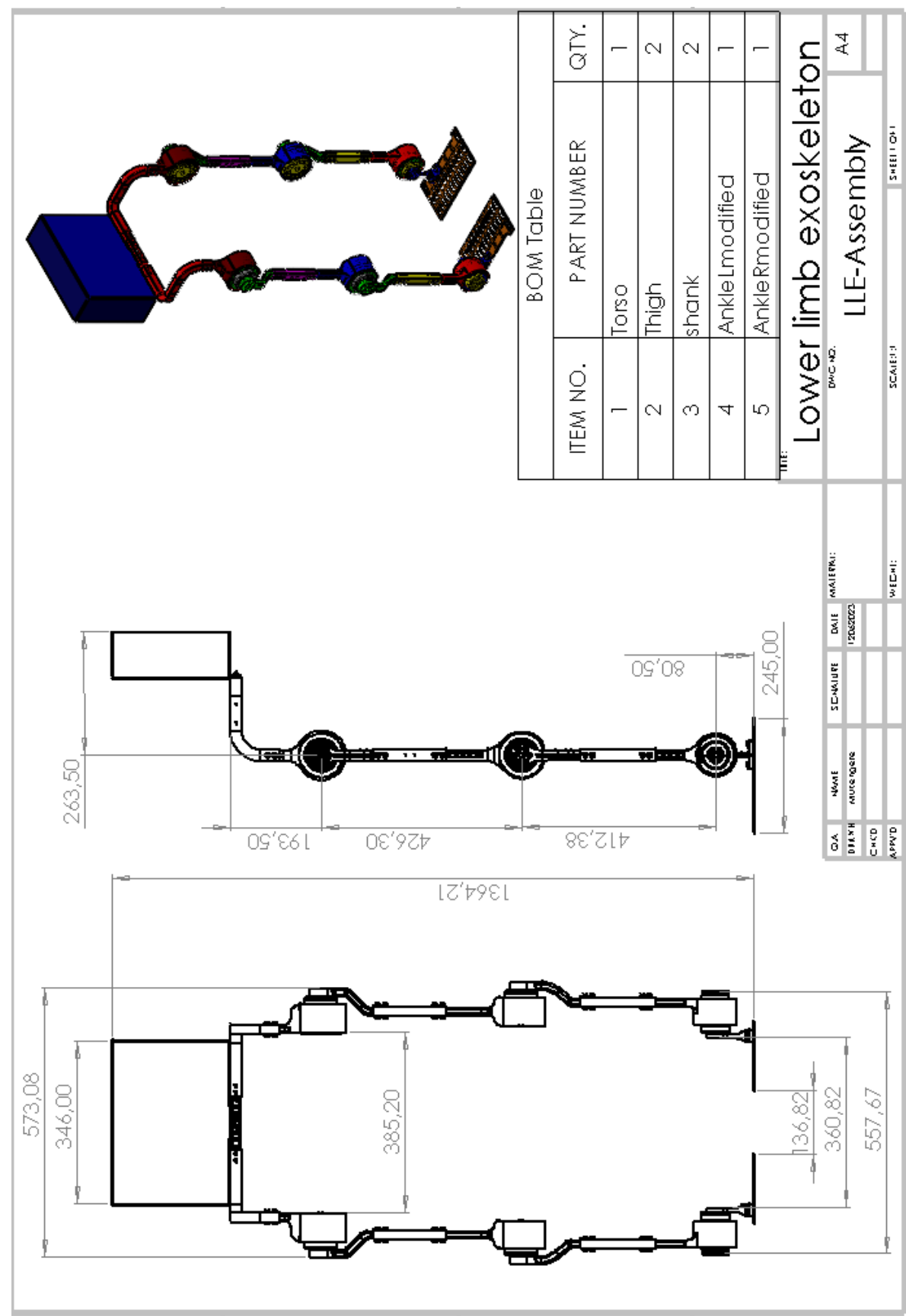


Figure A.5: Lower limb exoskeleton front and side view

Appendix B: Technical drawings of parts

From Figures B.1 to B. 12, the technical drawings of each part of the LLE are presented. These drawings are useful for replicating and manufacturing each part.

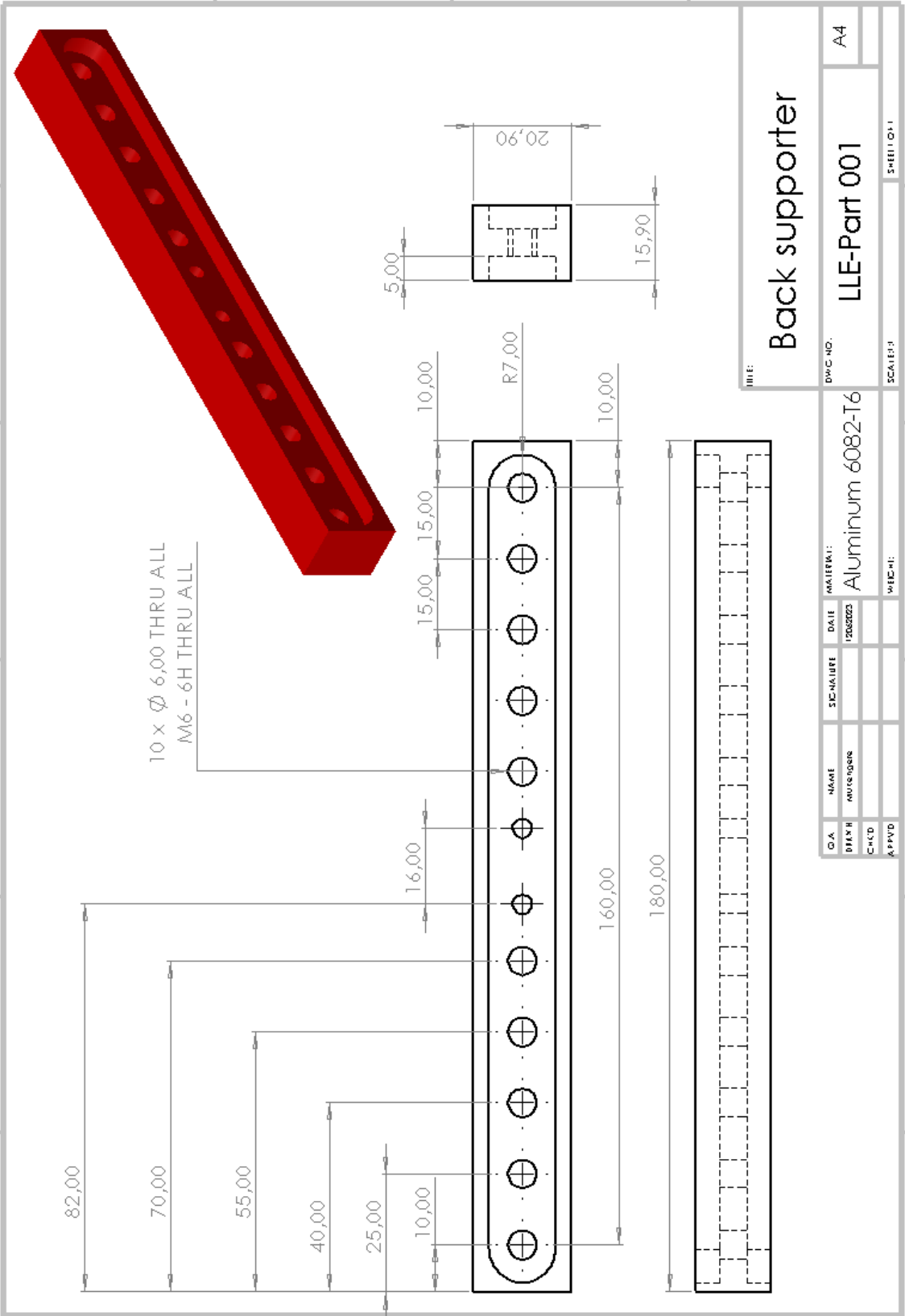


Figure B.1: Back supporter

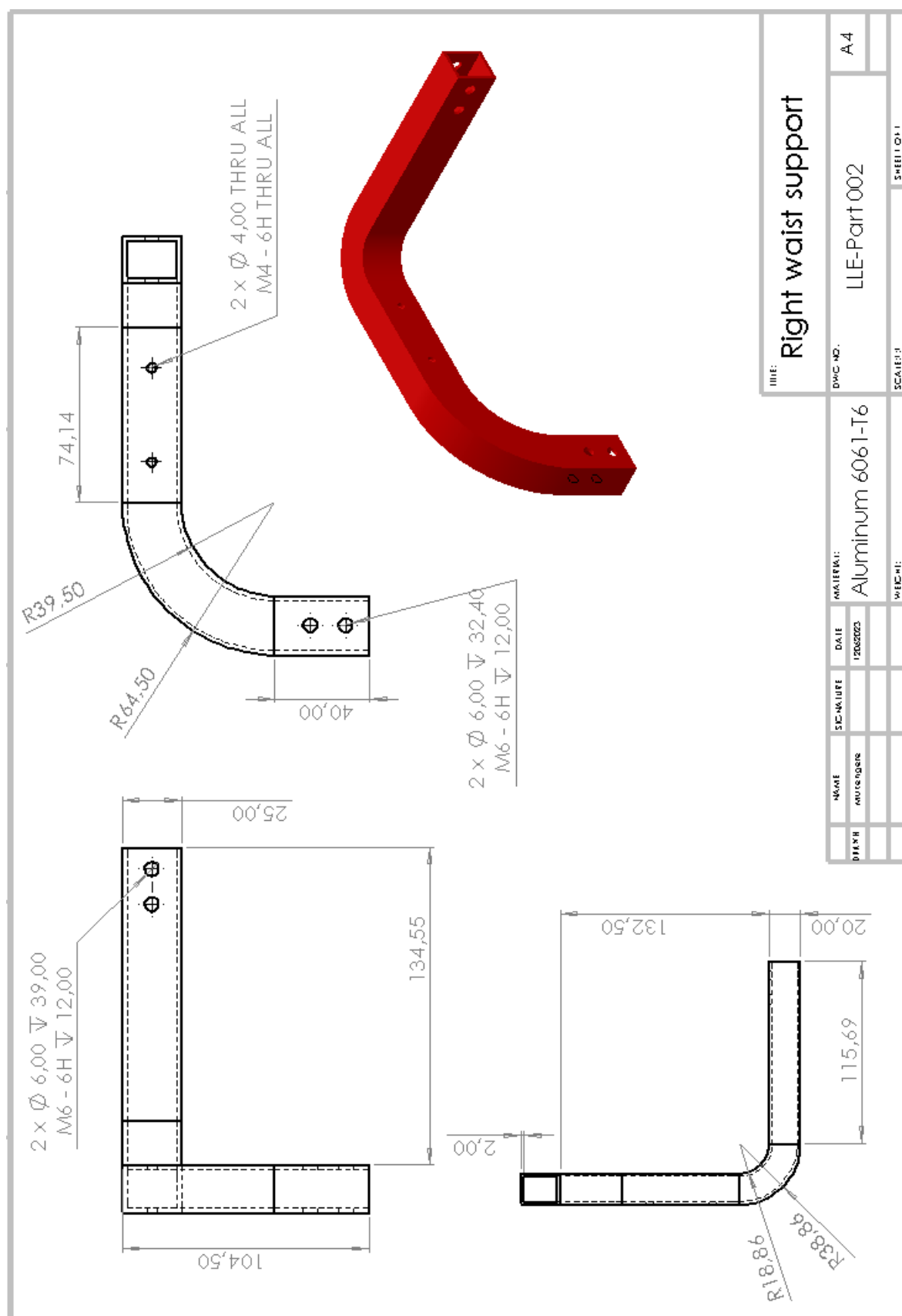


Figure B.2: Right/left waist support

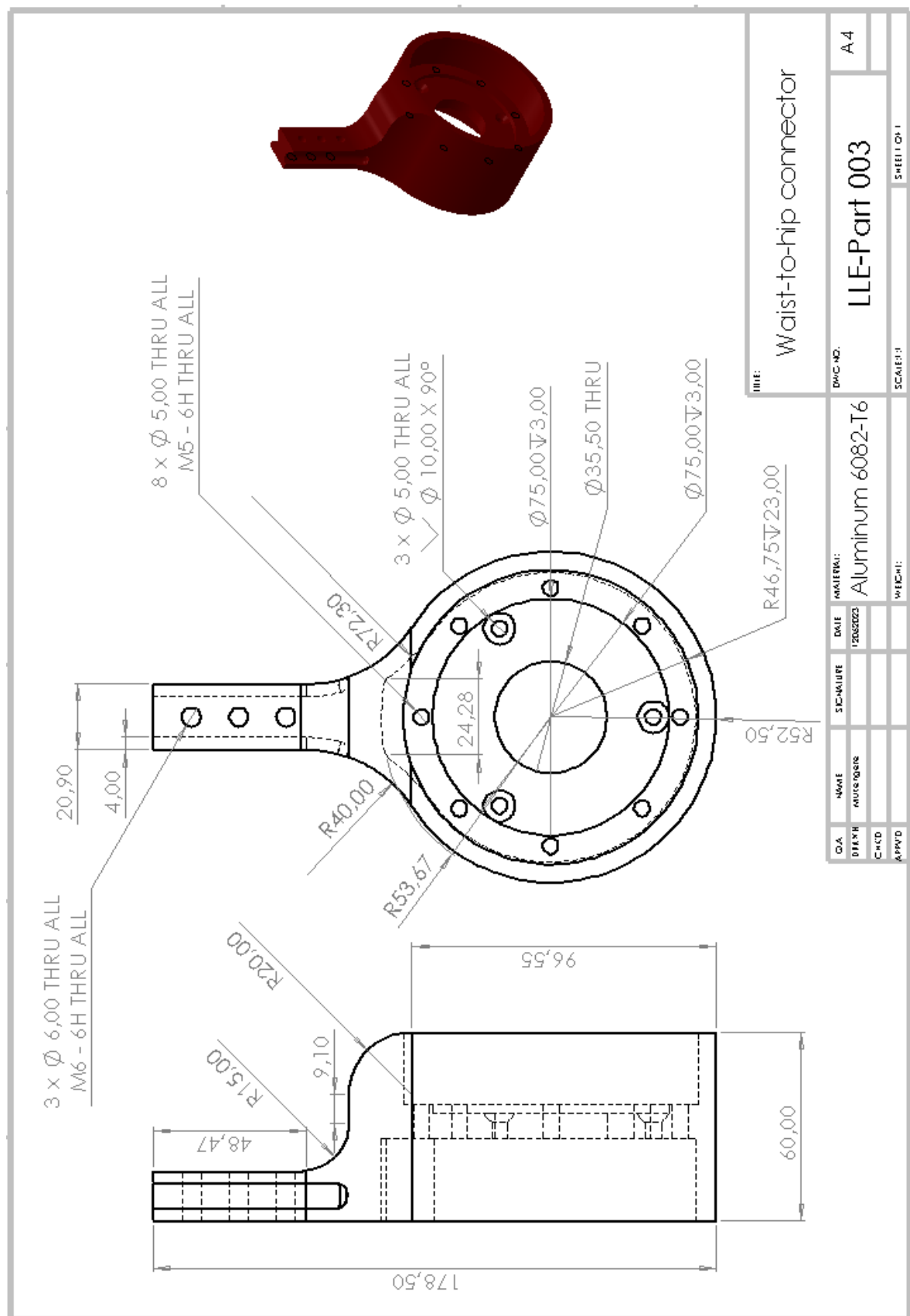
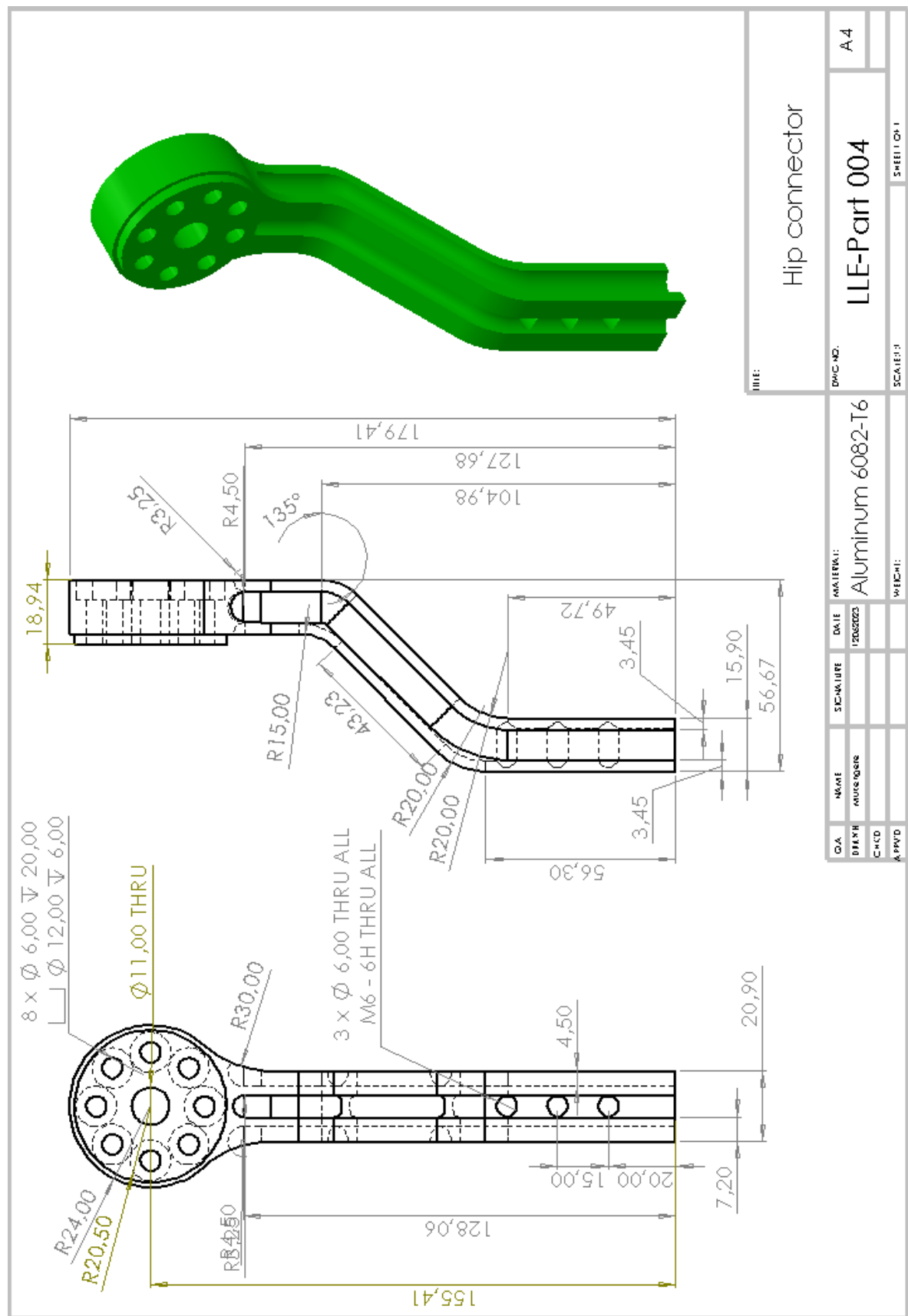


Figure B.3: Waist-to-hip



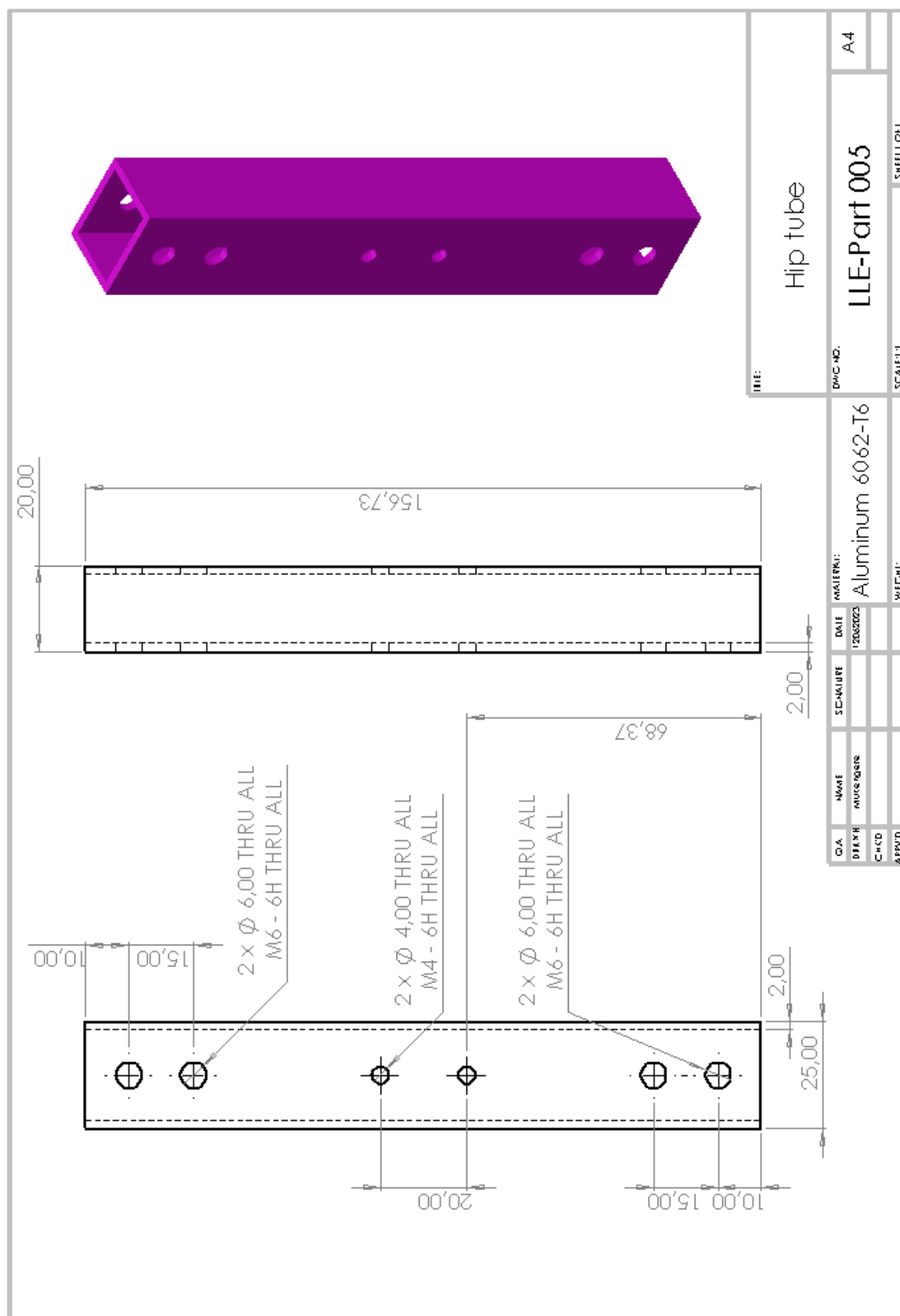
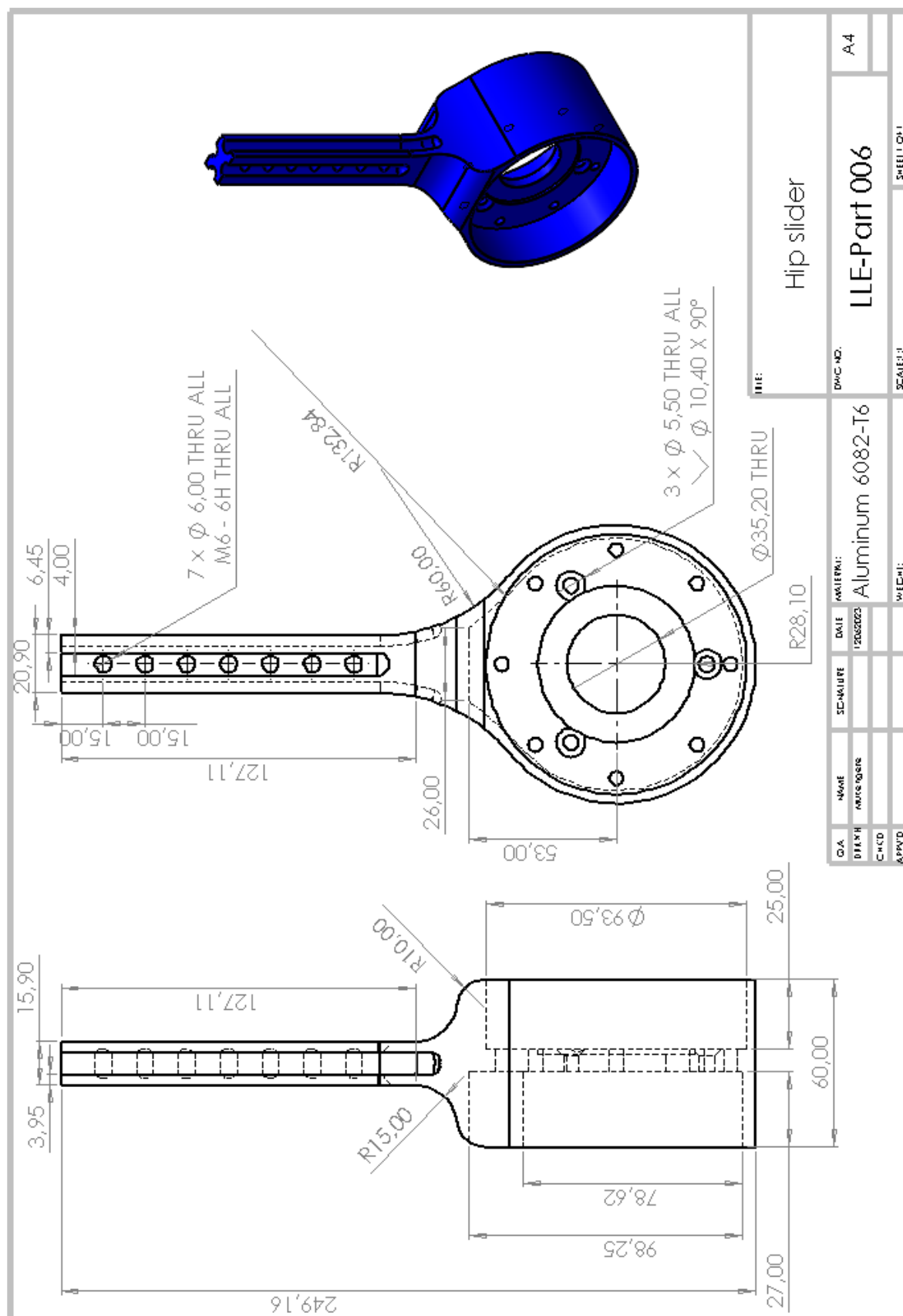
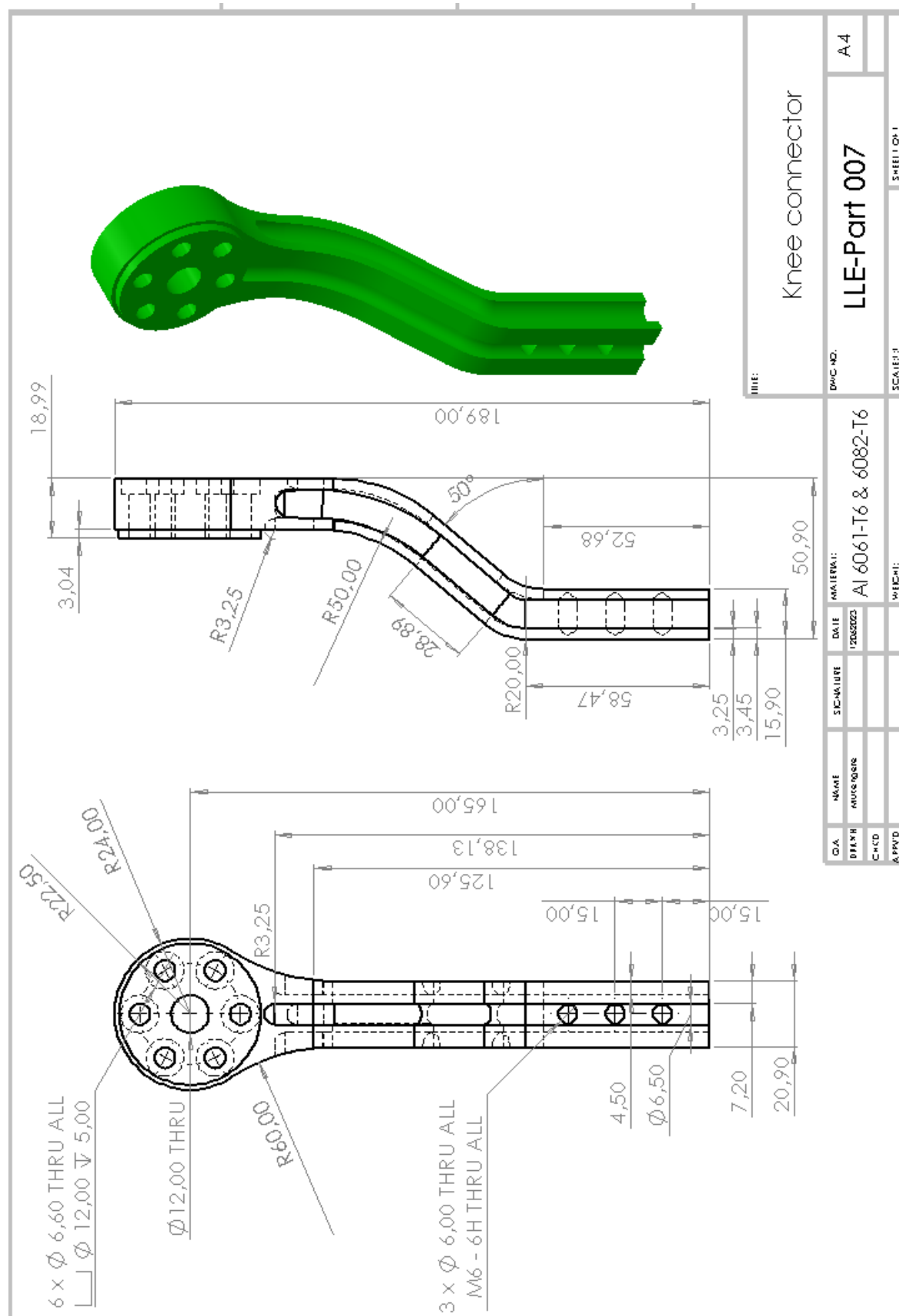
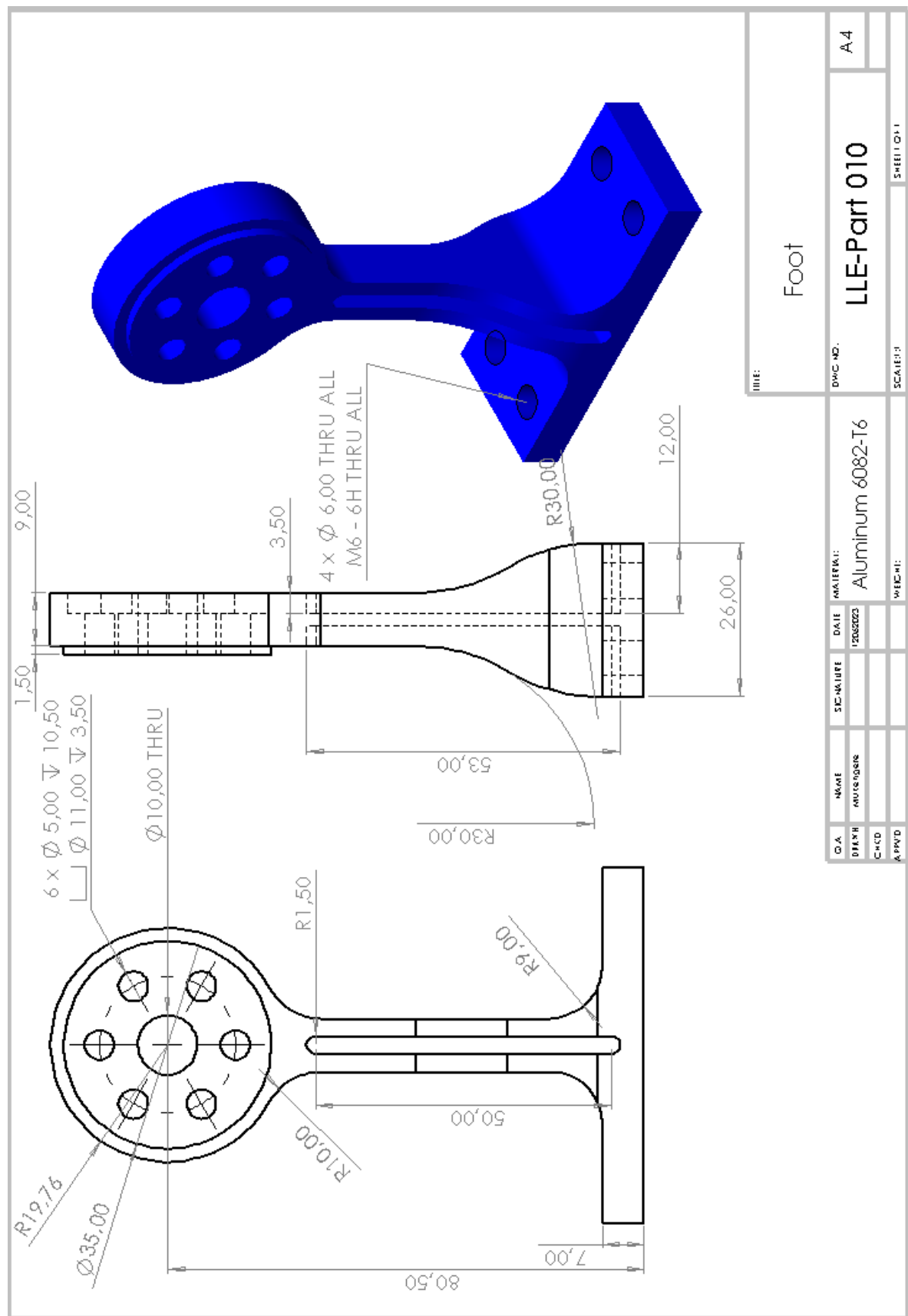


Figure B.5: Hip tube







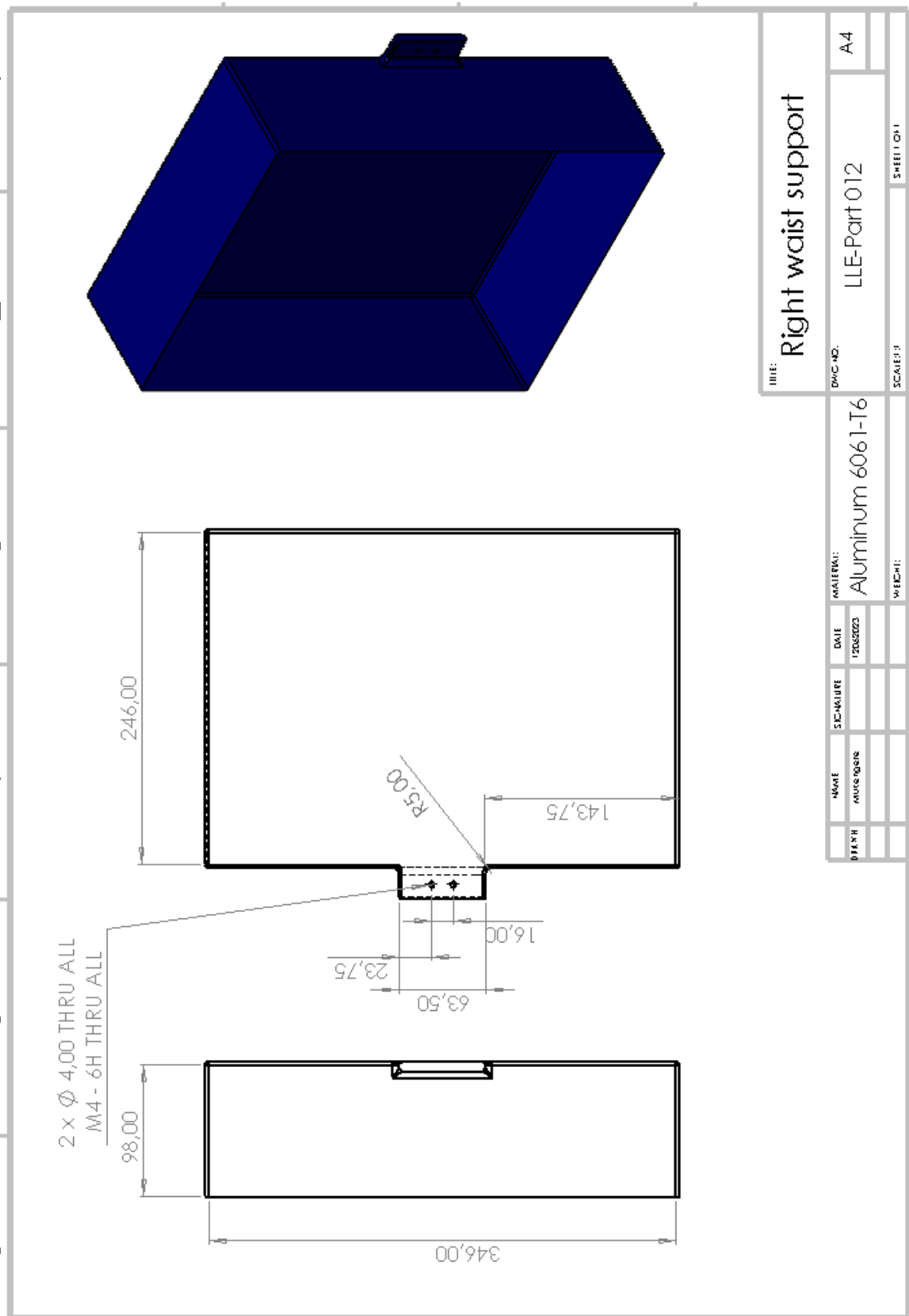


Figure B.12: Battery box

Appendix C: FEA Boundary Conditions

The boundary conditions of each segment of the LLE are presented, from Figure C.1 to C.6. In the legend, red indicates the applied force, blue indicates the fixed support, and yellow indicates standard Earth gravity.

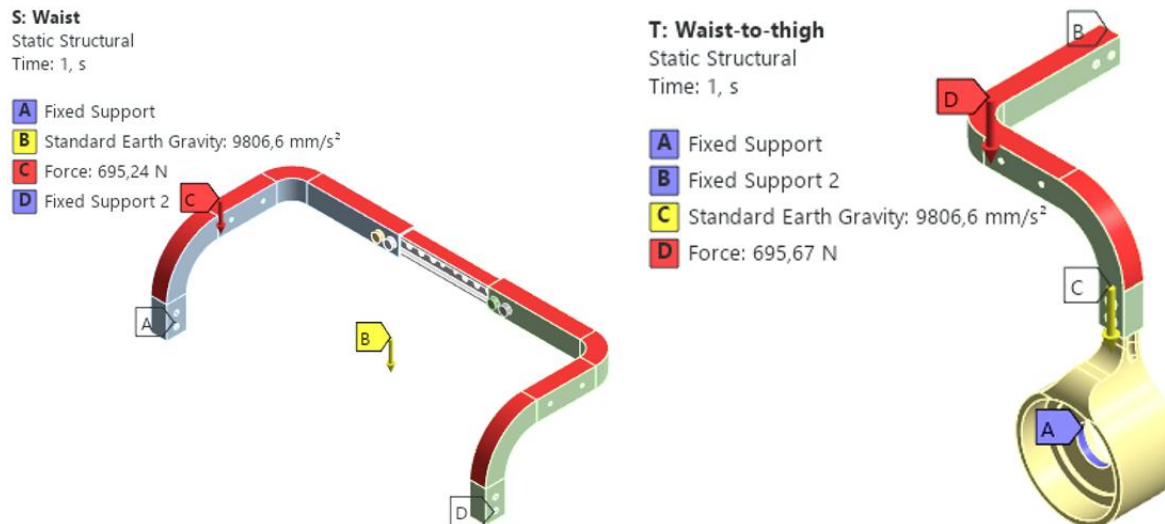


Figure C.1: Waist and waist-to-hip under axial applied load

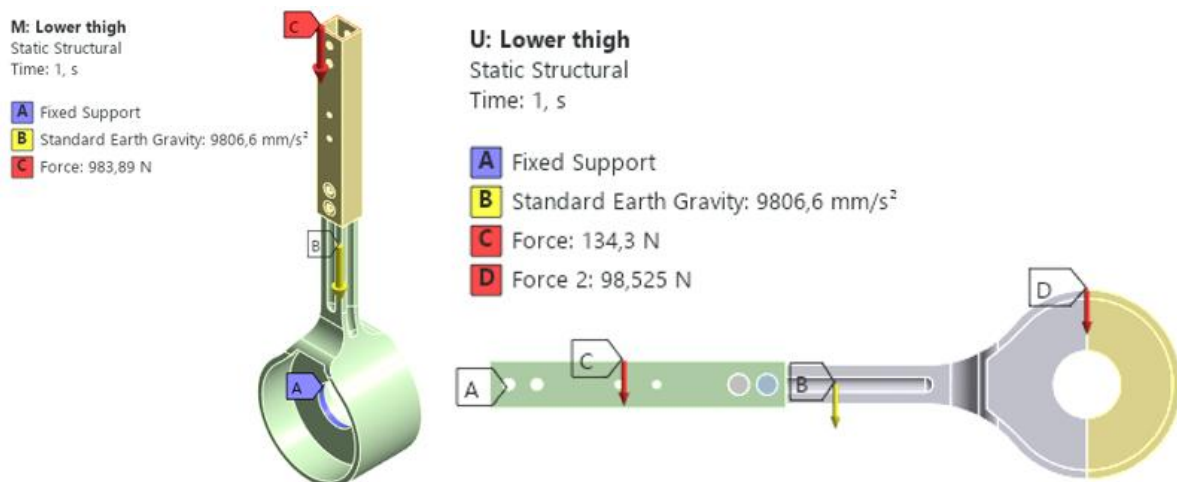


Figure C.2: Lower thigh under axial load (vertical) and under transverse load (horizontal)

E: Upper Shank
Static Structural
Time: 1, s

A Fixed Support
B Standard Earth Gravity: 9806,6 mm/s²
C Force: 1056,7 N



G: Upper Shank
Static Structural
Time: 1, s

A Fixed Support
B Standard Earth Gravity: 9806,6 mm/s²
C Force: 72,772 N
D Force 2: 25,753 N

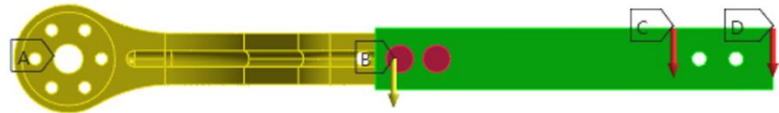
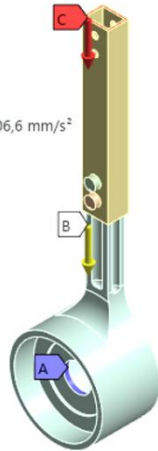


Figure C.3: Upper shank under axial load (vertical) and under transverse load (horizontal)

P: Lower shank
Static Structural
Time: 1, s

A Fixed Support
B Standard Earth Gravity: 9806,6 mm/s²
C Force: 1056,7 N



R: Lower shank
Static Structural
Time: 1, s

A Fixed Support
B Standard Earth Gravity: 9806,6 mm/s²
C Force 1: 72,819 N
D Force 2: 25,448 N

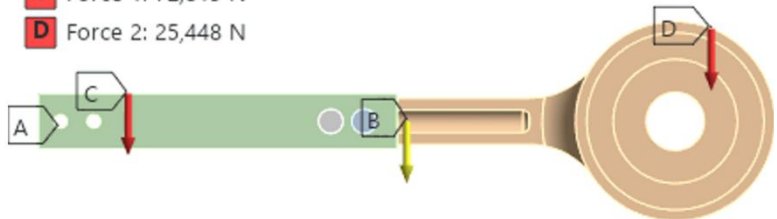
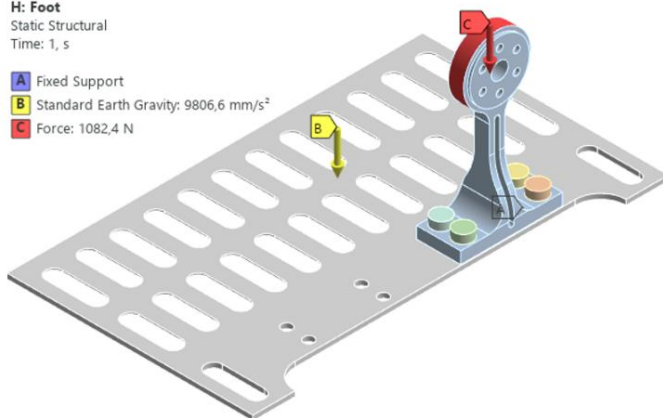


Figure C.4: Lower shank under axial load (vertical) and under transverse load (horizontal)

H: Foot
Static Structural
Time: 1, s

A Fixed Support
B Standard Earth Gravity: 9806,6 mm/s²
C Force: 1082,4 N



J: Foot
Static Structural
Time: 1, s

A Fixed Support
B Force: 25,753 N
C Standard Earth Gravity: 9806,6 mm/s²

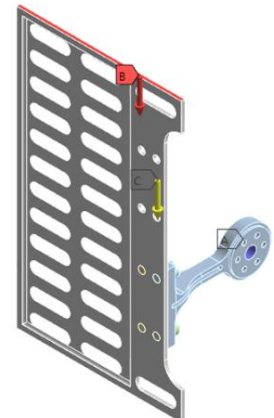


Figure C.5: Foot under axial load (vertical) and under transverse load (horizontal)

L: Foot

Static Structural

Time: 1, s

- A** Fixed Support
- B** Standard Earth Gravity: 9806,6 mm/s²
- C** Force: 332, N

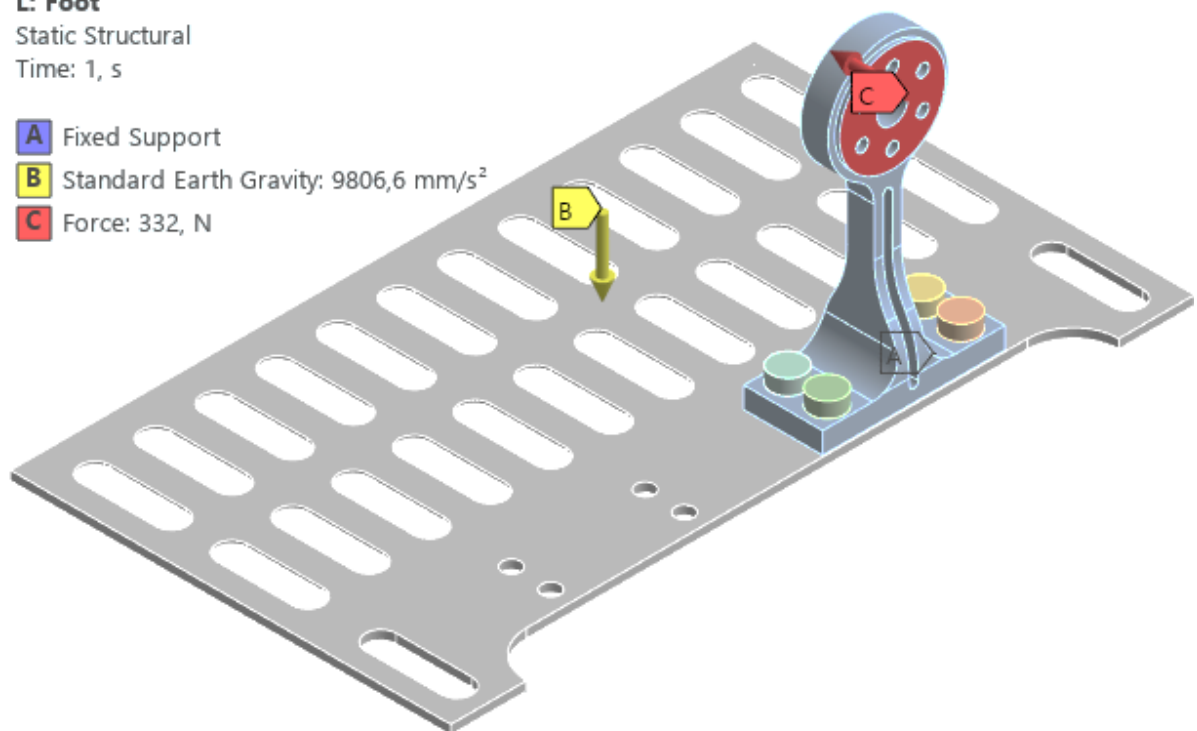


Figure C.6: Foot under transverse load

Appendix D: FEA Complementary Results

The results for the lower torso are presented in Figures D.1 to D.4, which illustrate Von Mises stress, factor of safety, and total deformation. Analysis of these figures reveals that the numerical values of the Von-Mises stress remain below the yield strength of aluminium 6061 T6, 6082-T6, and low-carbon steel. Furthermore, the safety factor is considered acceptable, with values exceeding one.

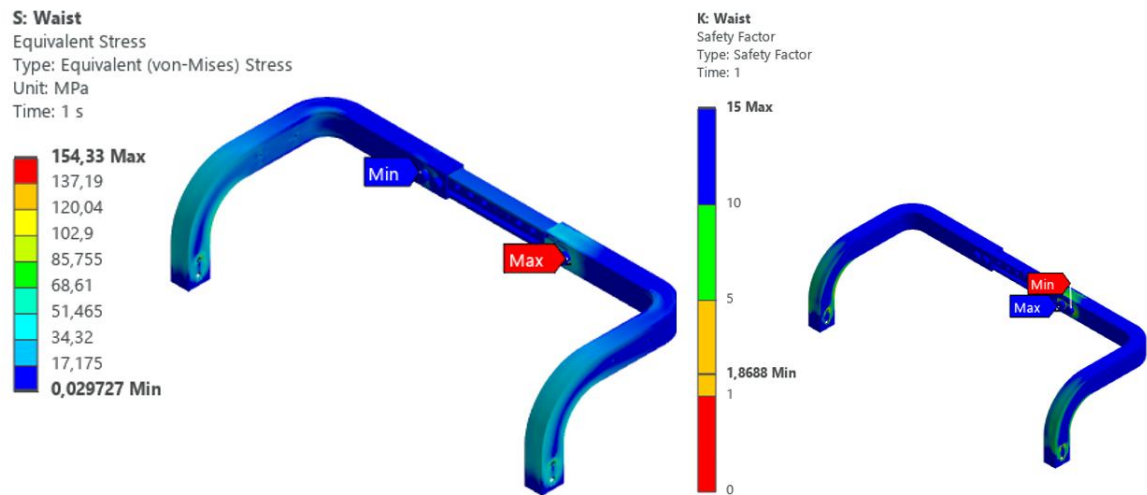


Figure D.1: Von-Mises stress and safety factor related to the waist

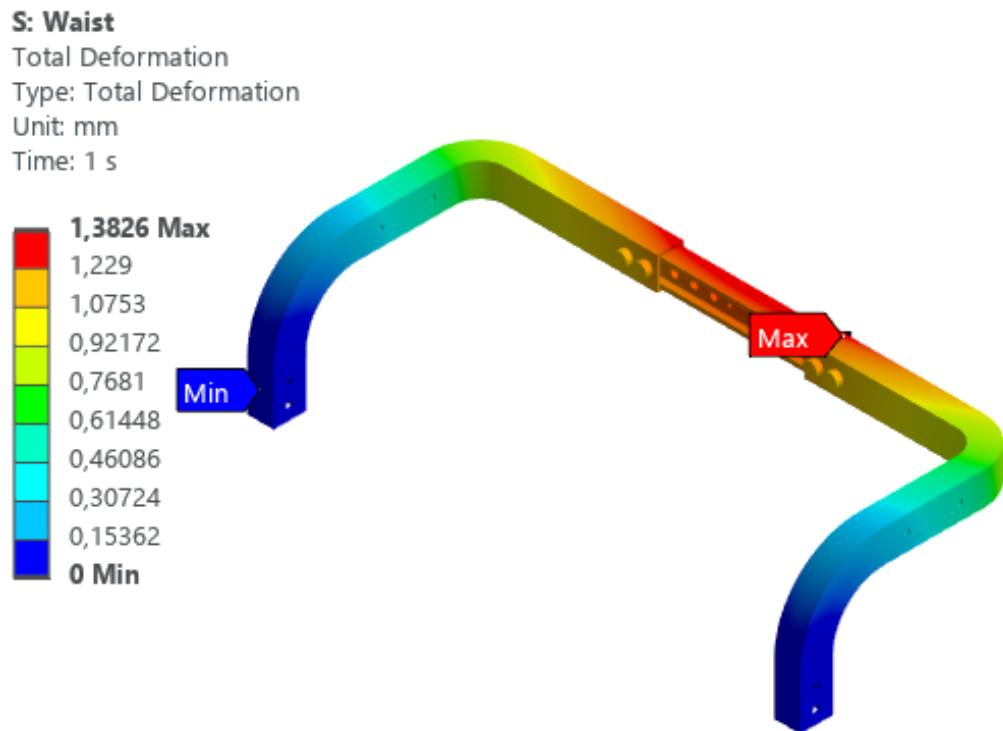


Figure D.2: Total deformation of the waist

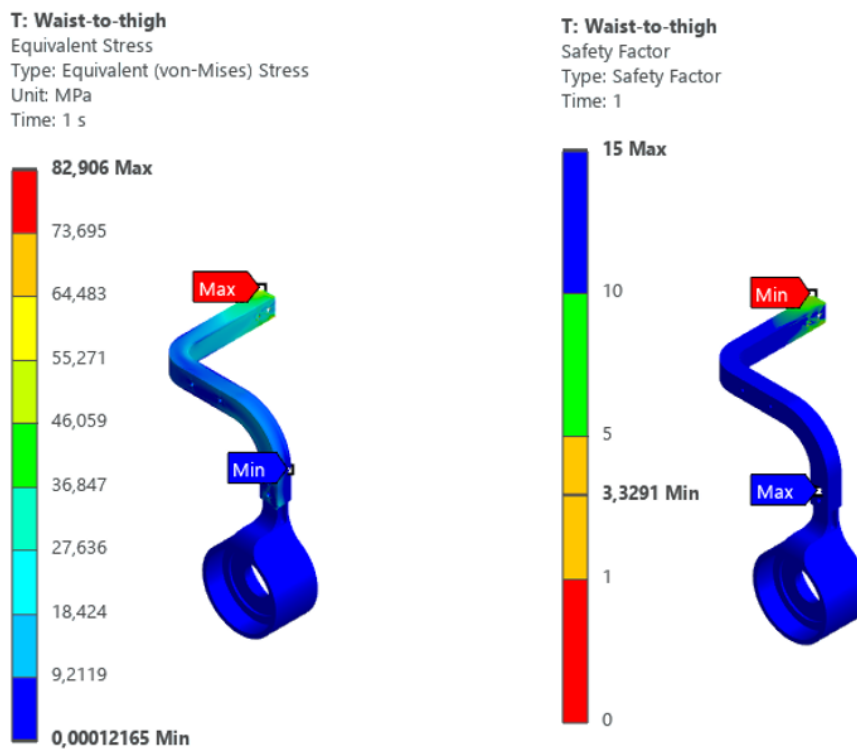


Figure D.3: Von-Mises stress and factor of safety of the waist-to-thigh

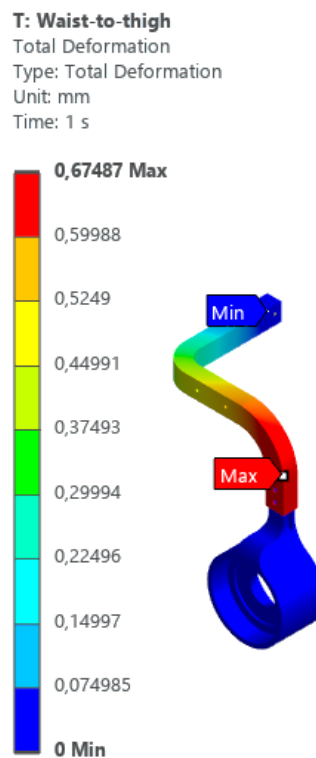


Figure D.4: Total deformation of the waist-to-hip

Appendix E: MATLAB simulation

This appendix section contains portions of code adapted to the LLE related to this research from the template developed by the MATLAB team.

Robot parameters

During simulation in MATLAB and Simulink, the LLE's parameters were defined. These parameters directly affect the behaviour and stability of the walking robot, in other words. Therefore, thorough analysis and proper calibration of the LLE parameters were fundamental steps in the design and simulation phases to ensure optimal functionality and safety. Note that the code snippets below include explanatory comments for each line.

The parameters related to the actuator control type for the foot of the LLE and ground contact, including stiffness, damping, and friction coefficients, are show in Figure E.1.

```
actuatorType = 2; % Defines the actuator control mode:
                  % 1 = motion-controlled (position input)
                  % 2 = torque-controlled (force/torque input)
                  % 3 = motor-controlled (direct motor signal input)

%% Contact and friction parameters
contact_stiffness = 981/0.001; % Contact stiffness (N/m): estimated as weight (~981 N for 100 kg)
                               % divided by the desired contact deformation (1 mm)

contact_damping = contact_stiffness/10; % Contact damping coefficient: chosen as one-tenth of stiffness
                                         % to provide a realistic energy dissipation rate

mu_s = 0.85; % Static friction coefficient: represents friction at zero velocity,
             % typical for rubber on asphalt-like surfaces

mu_k = 0.75; % Kinetic friction coefficient: friction once motion starts,
             % slightly lower than static to model slipping behavior

mu_vth = 0.1; % Velocity threshold (m/s): defines the transition speed
              % between static and kinetic friction regimes
```

Figure E.1: Actuator and contact parameter setup

To ensure the responsiveness of the joints, derivative filters were applied to the joint velocities of each joint to minimise noise; the corresponding torques for these joints were determined using the inverse dynamics of the LLE. These torques define the actuators' capabilities, ensuring the safe operation of the hip, knee, and ankle joints, as illustrated in the code snippet in Figure E.2.

```

%% === Derivative Filter Coefficients (for Noise Rejection) ===
% These coefficients are used to smooth the derivative (rate of change) signals
% from sensors or controllers to reduce noise without overly delaying response.

hip_deriv_filter_coeff = 45; % Derivative filter coefficient for hip joint:
                             % relatively high value provides strong damping
                             % to stabilize the larger, more powerful hip actuator

knee_deriv_filter_coeff = 33; % Derivative filter coefficient for knee joint:
                              % medium filtering for balanced noise rejection and response speed

ankle_deriv_filter_coeff = 22; % Derivative filter coefficient for ankle joint:
                               % lighter filtering to maintain agility and responsiveness
                               % important for quick foot-ground adjustments

%% Torque Limits
% These define the maximum and minimum allowable torques (in N·m)
% for each joint, protecting actuators and ensuring safe operation.

% Hip joint
hip_max_torque = 116.64; % Maximum torque the hip motor can safely deliver
hip_min_torque = -113.28; % Minimum (reverse direction) torque limit

% Knee joint
knee_max_torque = 114.4; % Maximum allowable torque for the knee actuator
knee_min_torque = -108.12; % Minimum torque (negative direction) limit

% Ankle joint
ankle_max_torque = 69.24; % Upper torque limit for ankle actuation
ankle_min_torque = -67.06; % Lower (negative) torque limit to prevent overloading

```

Figure E.2: Derivative Filter and Torque Limit Parameter Initialisation

To achieve a more realistic dynamic behaviour and minimise exaggerated oscillations, the joint damping and motion time constants were fine-tuned until the oscillations were significantly reduced. In addition to this configuration, the world damping parameter was used to define both translational and rotational resistance in the simulation environment, as illustrated in Figure E.3.

```

%% Robot joint parameters
% These parameters define the damping and time constants for the robot's joints
% and the surrounding simulation environment, influencing stability and motion smoothness.

joint_damping = 18.9; % Viscous damping coefficient applied to each joint
                    % (N·m·s/rad), used to suppress oscillations and improve stability

motion_time_constant = 0.0005; % Time constant (s) controlling how quickly the actuator
                                % responds to command changes; smaller value means faster response

world_damping = 14.7; % Translational damping (N·s/m) for the 6-DOF world joint,
                     % resists linear motion to reduce unwanted vibrations

world_rot_damping = 1130; % Rotational damping (N·m·s/rad) for the 6-DOF world joint,
                          % resists rotational motion, stabilizing the overall robot base

```

Figure E.3: Joint and Environmental Damping Parameter Initialisation

Figure E.4 illustrates a snippet of code that defines the fundamental parameters and initial conditions required to initiate the LLE simulation. On the one hand, it considers geometric

parameters such as Zmodel and Zrobot, while on the other hand, it specifies physical constants such as gravity and sample time. These defined parameters play a crucial role in modelling the LLE's gait.

```
%% Setup
% Set Variables and Model Initial Conditions
% These parameters define the initial geometric setup and motion conditions
% for the robot or exoskeleton model before simulation starts.
xTorso = 0.143;      % Horizontal (x-axis) distance from the torso to the foot
                    % related to SWINGFOOTPOS0 – helps define initial foot placement

g = 9.81;           % Gravitational acceleration (m/s^2)
Ts = 0.005;         % Simulation sample time (s) – determines how often the system updates
zRobot = 0.9252;     % Total height from the hip joint to the ground (m) for the actual robot
zModel = 0.87;       % Model's hip-to-ground height (m);
                    % slightly reduced (zRobot - 0.0552) for modeling consistency

swingHeight = 0.05; % Desired maximum height (m) the swing foot lifts off the ground during gait
stepLength = 0.4;   % Step length (m); defines forward distance covered per step
                    % note: y0 = stepLength / 2 in the gait initialization

dy_mid = 0.555;     % Mid-step velocity in the y-direction (m/s);
                    % represents the minimum forward velocity during walking
x0 = 0.14;          % Initial x-position (m); chosen close to xTorso for balanced stance

% Function call to compute initial motion conditions for the model
% Outputs include initial velocities and single support phase duration
[dx0, y0, dy0, tSingleSupport] = findInitialConditions(stepLength, dy_mid, x0, zModel, g, Ts);
```

Figure E.4: Simulation Initialisation and Gait Parameter Setup

The inverse dynamics of the LLE were defined using the Euler-Lagrange formulation, as illustrated in figures E.5 to E.7. The code is influenced by factors such as joint angles, velocities, accelerations, segment lengths, segment mass properties, segment rotational inertias about the COM, gravitational acceleration, and the COM itself.

```
%% =====
% Torque Evaluation for a 3R Lower-Limb Exoskeleton (Planar)
% -----
% This script computes the required joint torques at the hip,
% knee, and ankle using Euler-Lagrange dynamics formulation:
%  $\tau = M(q) \cdot \ddot{q} + C(q, \dot{q}) \cdot \dot{q} + G(q)$ 
% where:
% - M(q): Inertia (mass) matrix
% - C(q, q̇): Coriolis and centrifugal effects
% - G(q): Gravitational torque vector
% Each joint (q1, q2, q3) corresponds to:
% q1 → hip pitch
% q2 → knee pitch
% q3 → ankle pitch
% =====

clc; clear; close all;
```

Figure E.5: Description of the Euler-Lagrange dynamics formulation

```

% -----
q  = [deg2rad(35); deg2rad(305); deg2rad(25)]; % Joint angles [rad]
qd = [0.5; 0.2; 0.1]; % Angular velocities [rad/s]
qdd = [1.0; 0.5; 0.2]; % Angular accelerations [rad/s^2]
%% -----
% Model Physical Parameters
% Link geometry [m] % Thigh length (l1) % Shank length (l2) % Foot length (l3)
l1 = 0.4263; l2 = 0.41238; l3 = 0.08352;
% Center of Mass [m] % Thigh COM distance (lc1) % Shank COM distance (lc2) % Foot COM distance (lc3)
lc1 = 0.28877; lc2 = 0.27279; lc3 = 0.06847;
% Mass properties[kg] % Thigh segment mass (m1) % Shank segment mass (m2) % Foot segment mass (m3)
m1 = 13.6905; m2 = 7.41818; m3 = 2.62515;
% Segment rotational inertias about COM [kg-m^2] % Thigh inertia (I1) % Shank inertia (I2) % Foot inertia (I3)
I1 = 0.01713; I2 = 0.01485; I3 = 0.00110;
% Gravitational acceleration
g = 9.81; % [m/s^2]
% Pack parameters into a single vector
params = [l1, l2, l3, lc1, lc2, lc3, m1, m2, m3, I1, I2, I3, g];
%% -----
% Torque Evaluation
% -----
tau = torque3R(q, qd, qdd, params);
% Extract and display individual joint torques
fprintf('Hip torque = %.3f Nm\n', tau(1));
fprintf('Knee torque = %.3f Nm\n', tau(2));
fprintf('Ankle torque = %.3f Nm\n', tau(3));

```

Figure E.6: Torque Computation Setup

```

function tau = torque3R(q, qd, qdd, params)
% torque3R - Compute joint torques (hip, knee, ankle) of a 3R lower-limb exoskeleton
% using Euler-Lagrange dynamics:
% tau = M(q)*qdd + C(q,qd)*qd + G(q)
% Extract parameters
l1=params(1); l2=params(2); l3=params(3); lc1=params(4); lc2=params(5); lc3=params(6);
m1=params(7); m2=params(8); m3=params(9); I1=params(10); I2=params(11); I3=params(12); g=params(13);
q1=q(1); q2=q(2); q3=q(3); qd1=qd(1); qd2=qd(2); qd3=qd(3); qdd1=qdd(1); qdd2=qdd(2); qdd3=qdd(3);
%% Mass matrix M(q)
M11 = I1+I2+I3 + m1*lc1^2 + m2*(l1^2+lc2^2+2*l1*lc2*cos(q2)) + ...
      m3*(l1^2+l2^2+lc3^2 + 2*l1*l2*cos(q2) + 2*l1*lc3*cos(q2+q3) + 2*l2*lc3*cos(q3));
M12 = I2+I3 + m2*(lc2^2+l1*lc2*cos(q2)) + m3*(l2^2+lc3^2+l1*l2*cos(q2) + l1*lc3*cos(q2+q3)+2*l2*lc3*cos(q3));
M13 = I3 + m3*(lc3^2+l1*lc3*cos(q2+q3)+l2*lc3*cos(q3));
M22 = I2+I3 + m2*lc2^2 + m3*(l2^2+lc3^2+2*l2*lc3*cos(q3));
M23 = I3 + m3*(lc3^2+l2*lc3*cos(q3));
M33 = I3 + m3*lc3^2;

M = [M11 M12 M13; M12 M22 M23; M13 M23 M33];
%% Coriolis/Centrifugal terms (C*qd)
h1 = -m2*l1*lc2*sin(q2) - m3*l1*l2*sin(q2) - m3*l1*lc3*sin(q2+q3); h2 = -m3*l2*lc3*sin(q3);

Cqd1 = h1*(2*qd1*qd2 + 2*qd1*qd3 + qd2^2 + 2*qd2*qd3 + qd3^2) + h2*(2*qd1*qd3 + 2*qd2*qd3 + qd3^2);
Cqd2 = h1*(qd1^2) + h2*(2*qd1*qd3 + qd3^2); Cqd3 = h1*(qd1^2 + 2*qd1*qd2 + qd2^2) + h2*((qd1+qd2)^2);

Cqd = [Cqd1; Cqd2; Cqd3];
%% Gravity vector G(q)
G1 = (m1*lc1 + m2*l1 + m3*l1)*g*cos(q1) + (m2*lc2+m3*l2)*g*cos(q1+q2) + m3*lc3*g*cos(q1+q2+q3);
G2 = (m2*lc2+m3*l2)*g*cos(q1+q2) + m3*lc3*g*cos(q1+q2+q3); G3 = m3*lc3*g*cos(q1+q2+q3);
G = [G1; G2; G3];
%% Final torque
tau = M*[qdd1;qdd2;qdd3] + Cqd + G
end

```

Figure E.7: Equation of motion code related to the LLE

Appendix F: MATLAB results

The following results present the right-leg torque and angles throughout the gait cycle. The graphs displayed in Figures F1 to F3 are mirror images of those for the left leg, as discussed in Chapter 4. For illustrative clarity, Figures F4 to F9 include the complete 10-second graphs along with their zoomed-in counterparts.

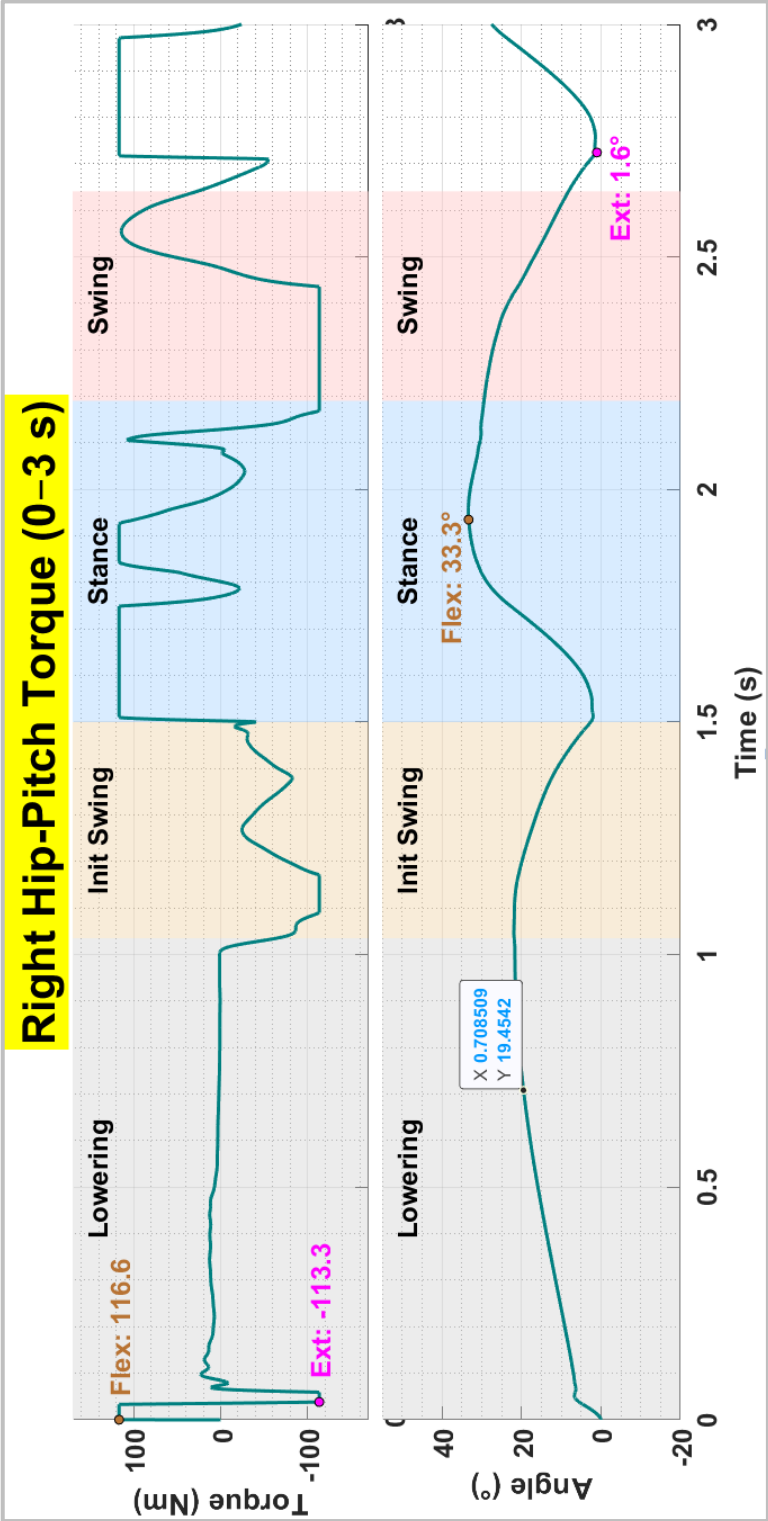


Figure F.1: Zoomed-in right hip-pitch torque and angle vs time

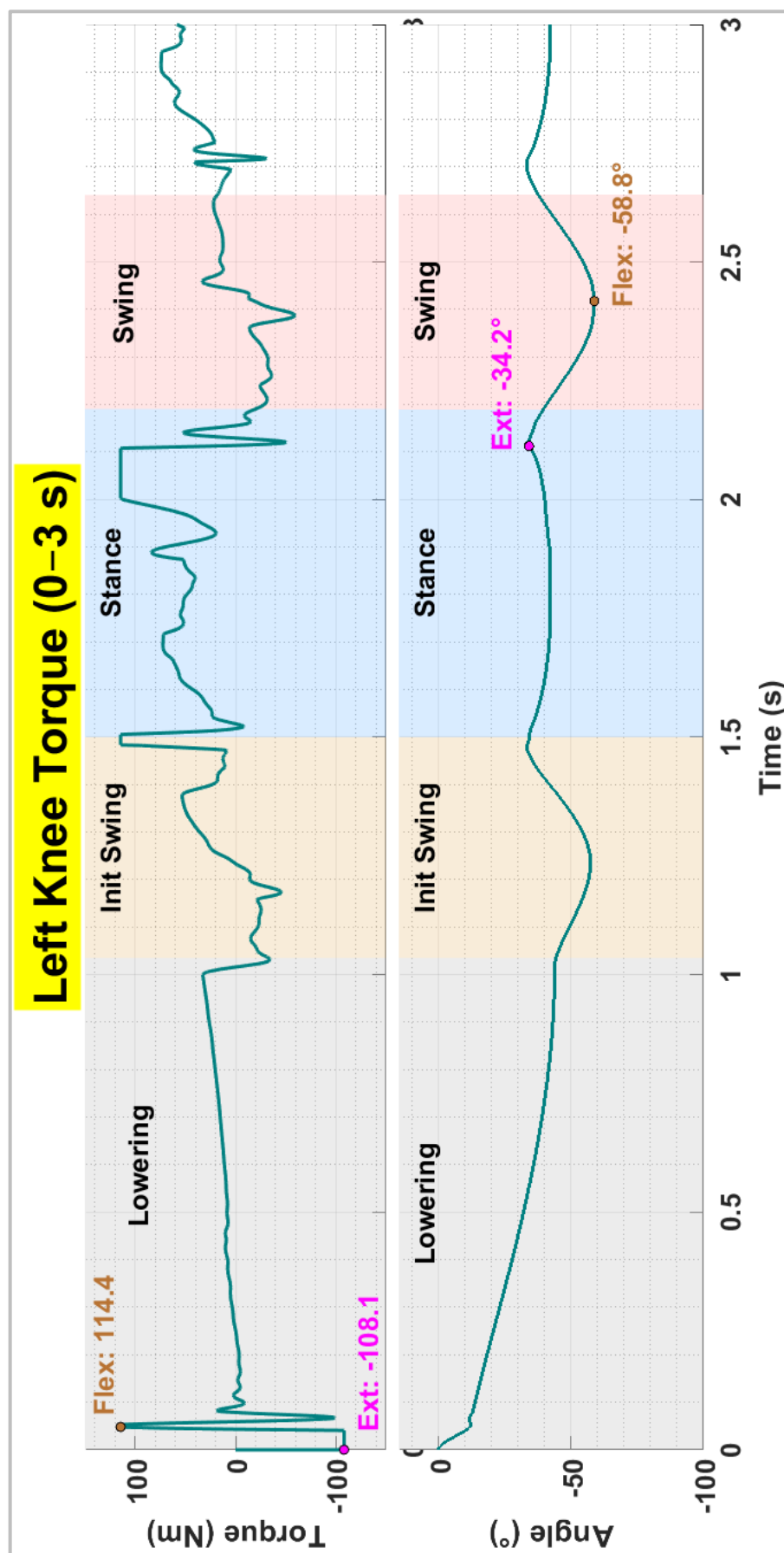


Figure F.2: Zoomed-in right knee torque and angle vs time

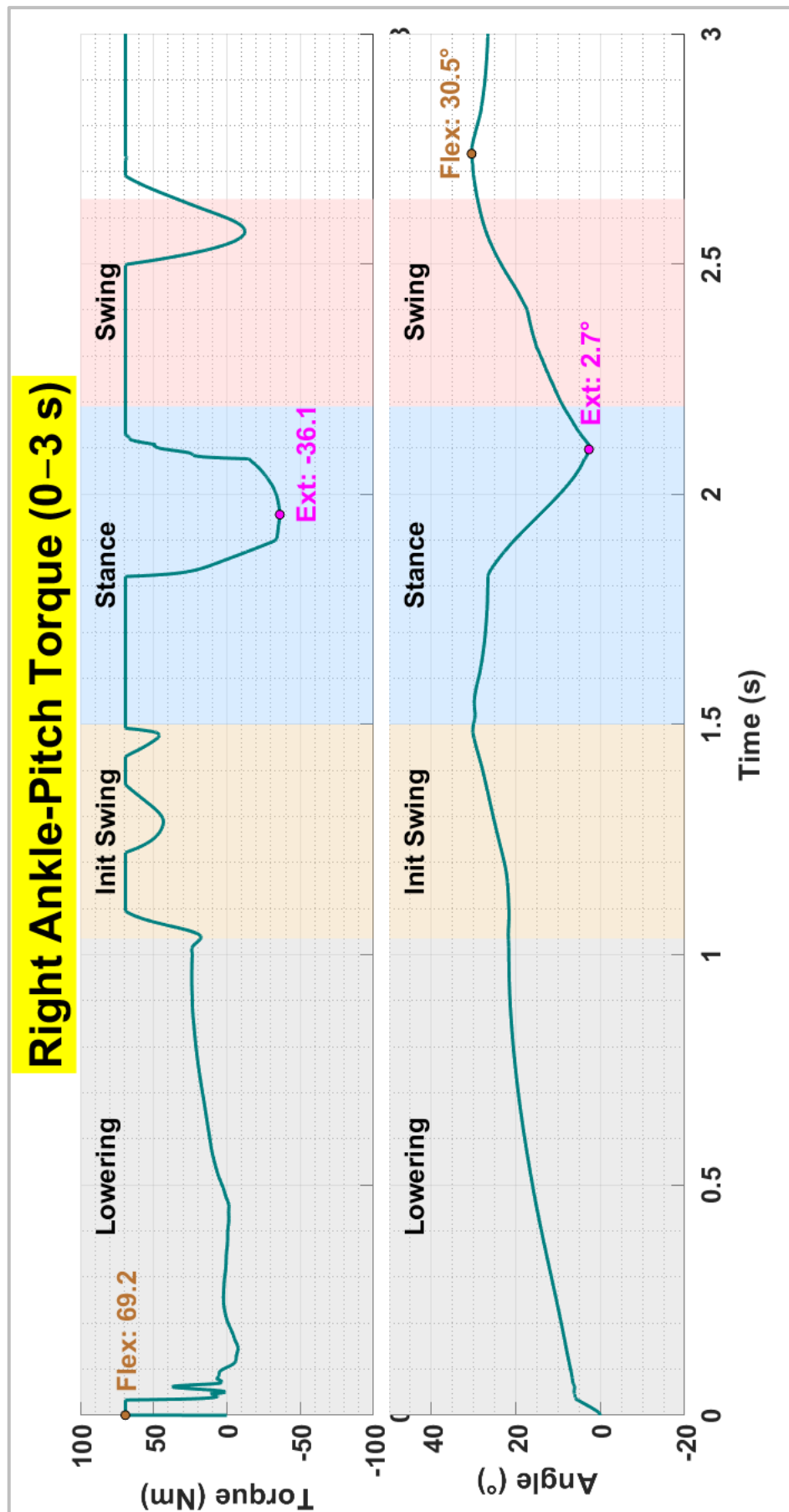


Figure F.3: Zoomed-in right ankle-pitch torque and angle vs time

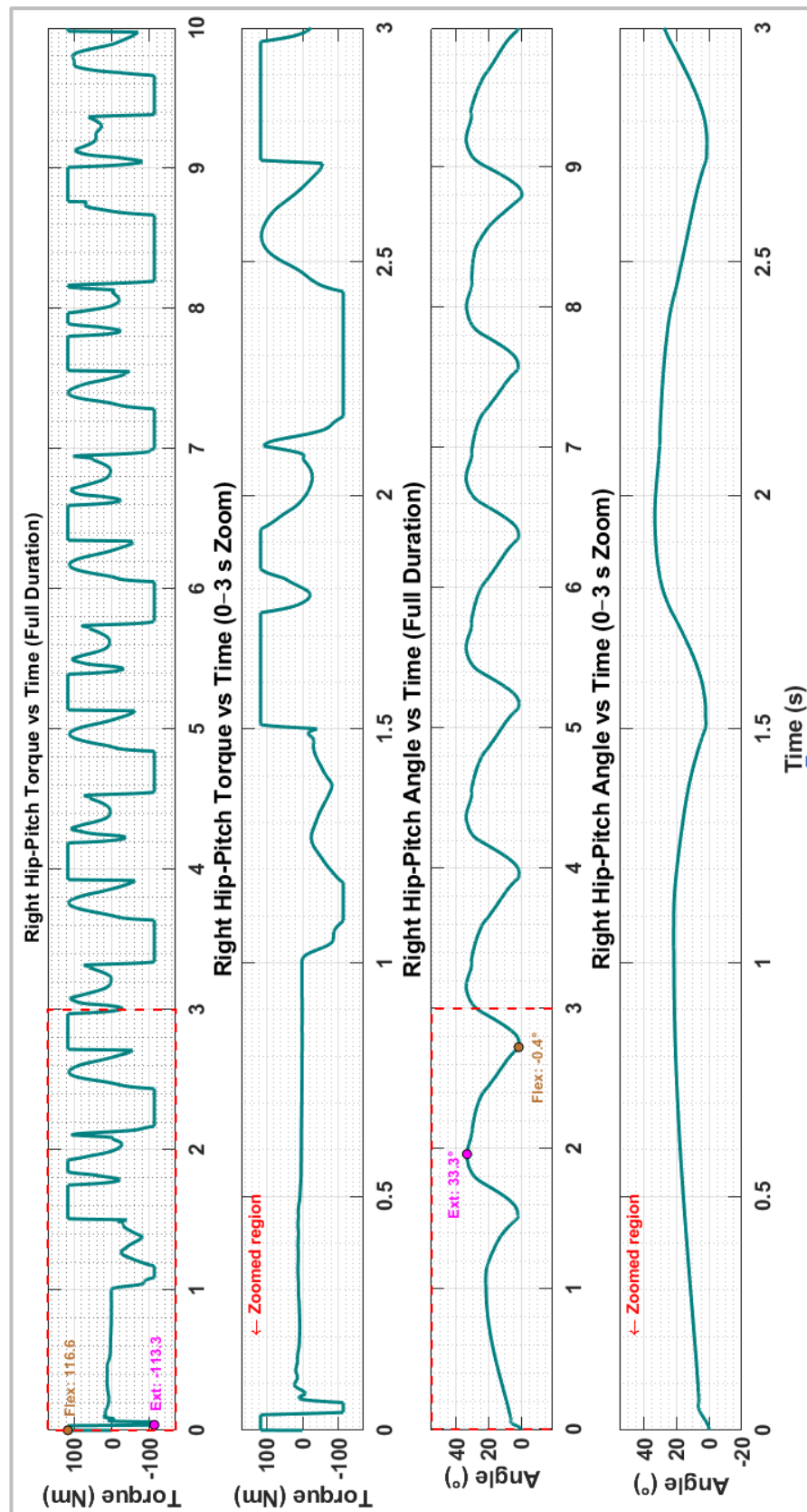


Figure F.4: Full and zoomed-in right hip-pitch torque and angle vs time

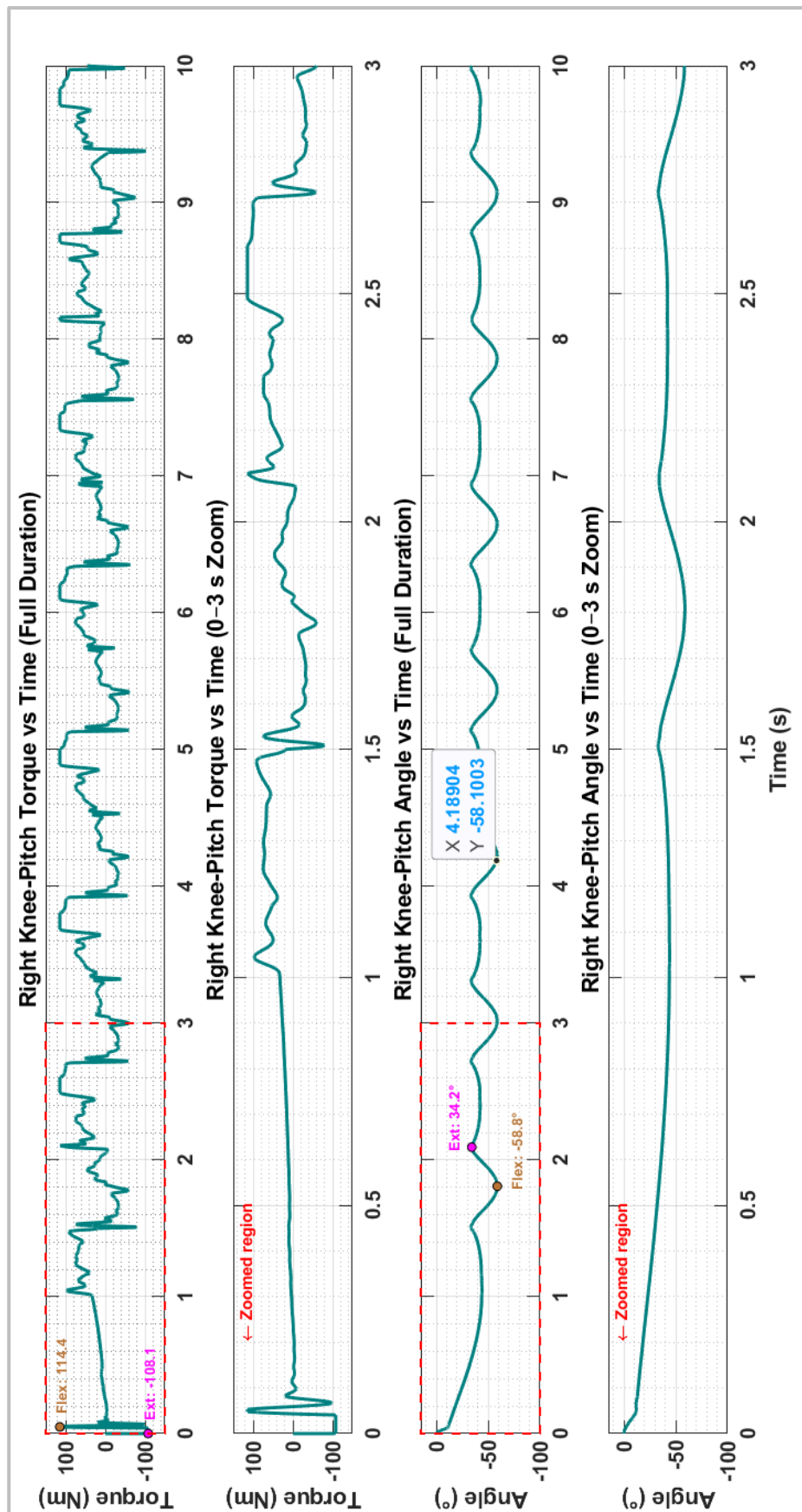


Figure F.5: Full and zoomed-in right knee torque and angle vs time

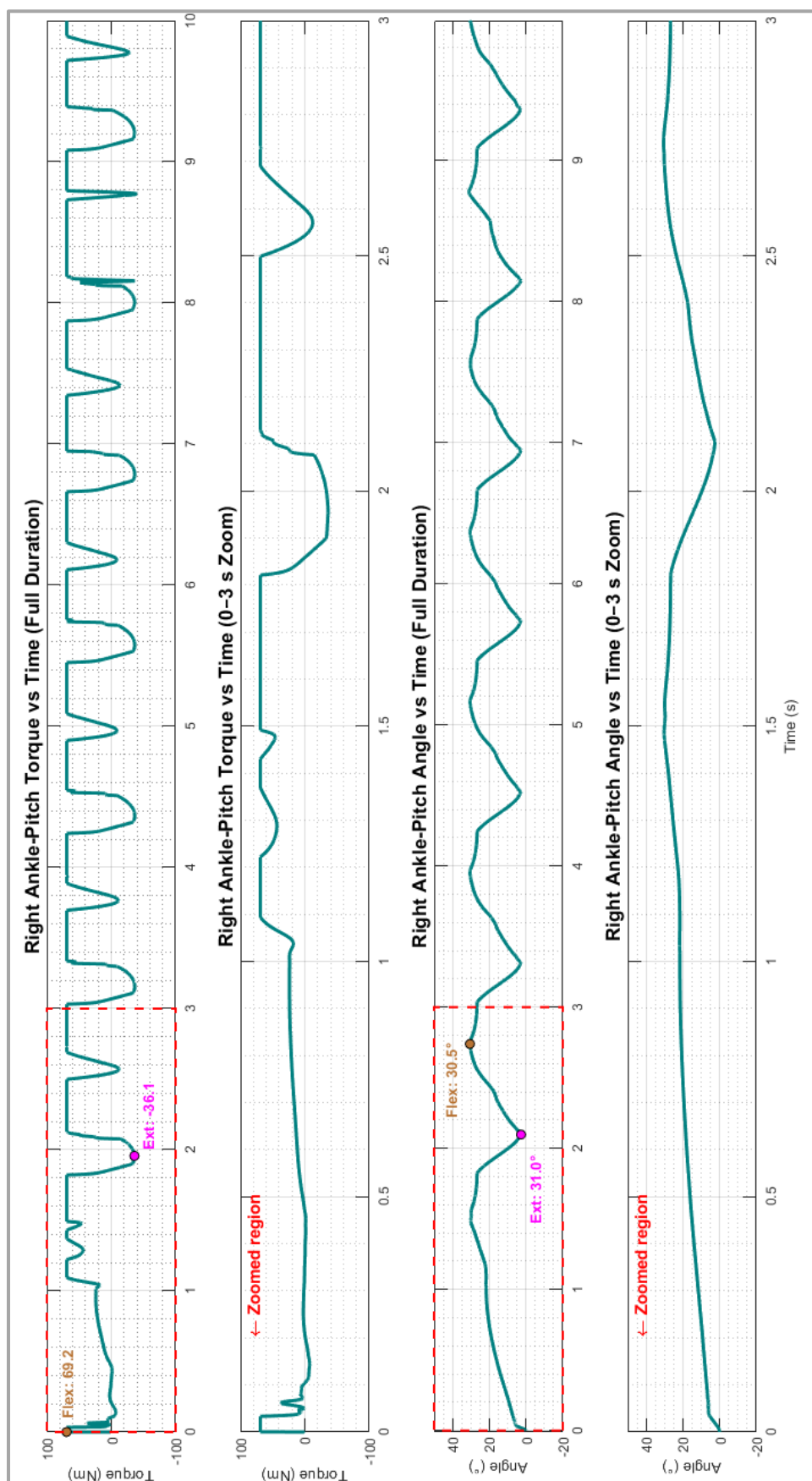


Figure F.6: Full and zoomed-in right ankle-pitch torque and angle vs time

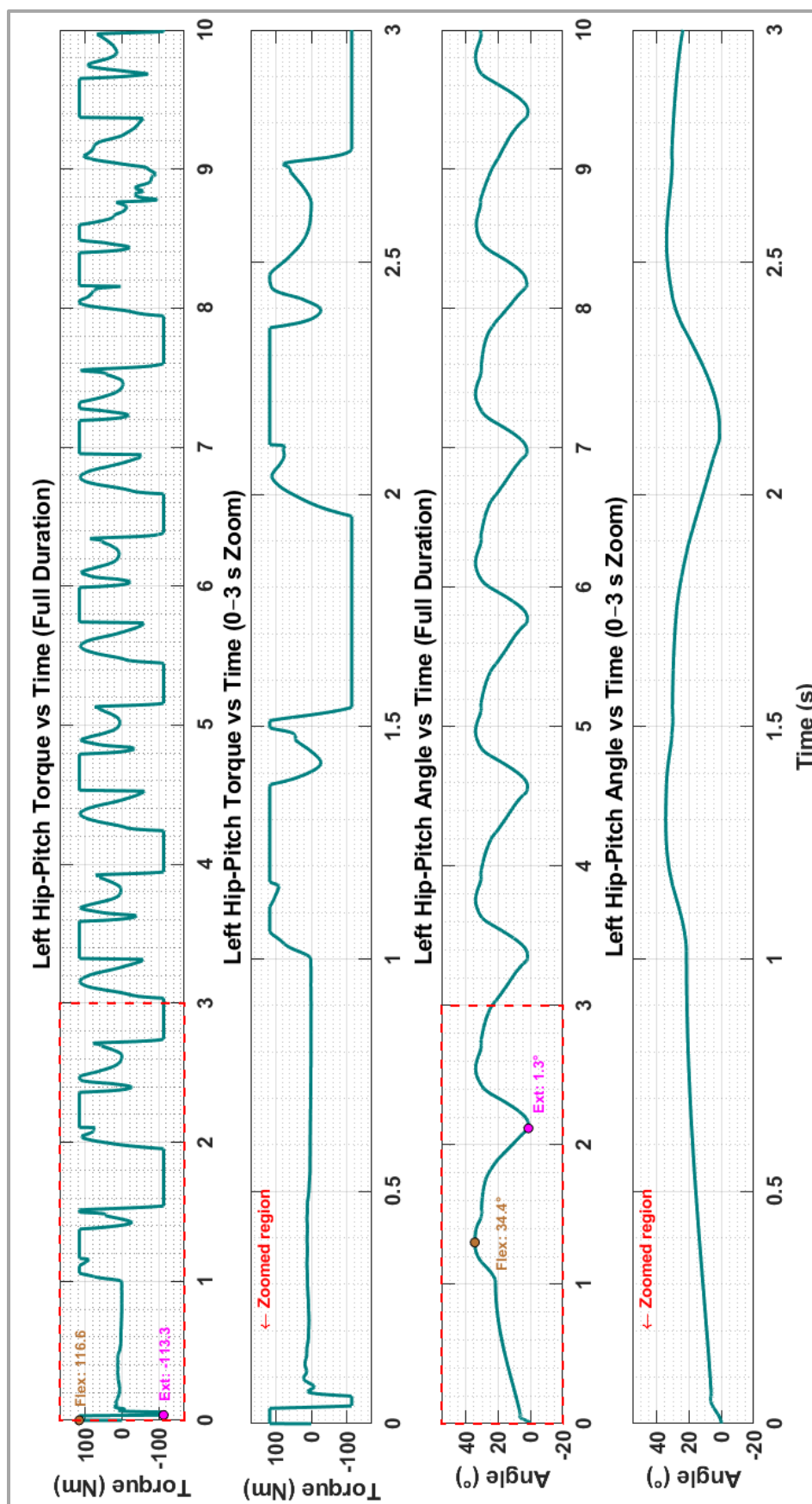


Figure F.7: Full and zoomed-in left hip-pitch torque and angle vs time

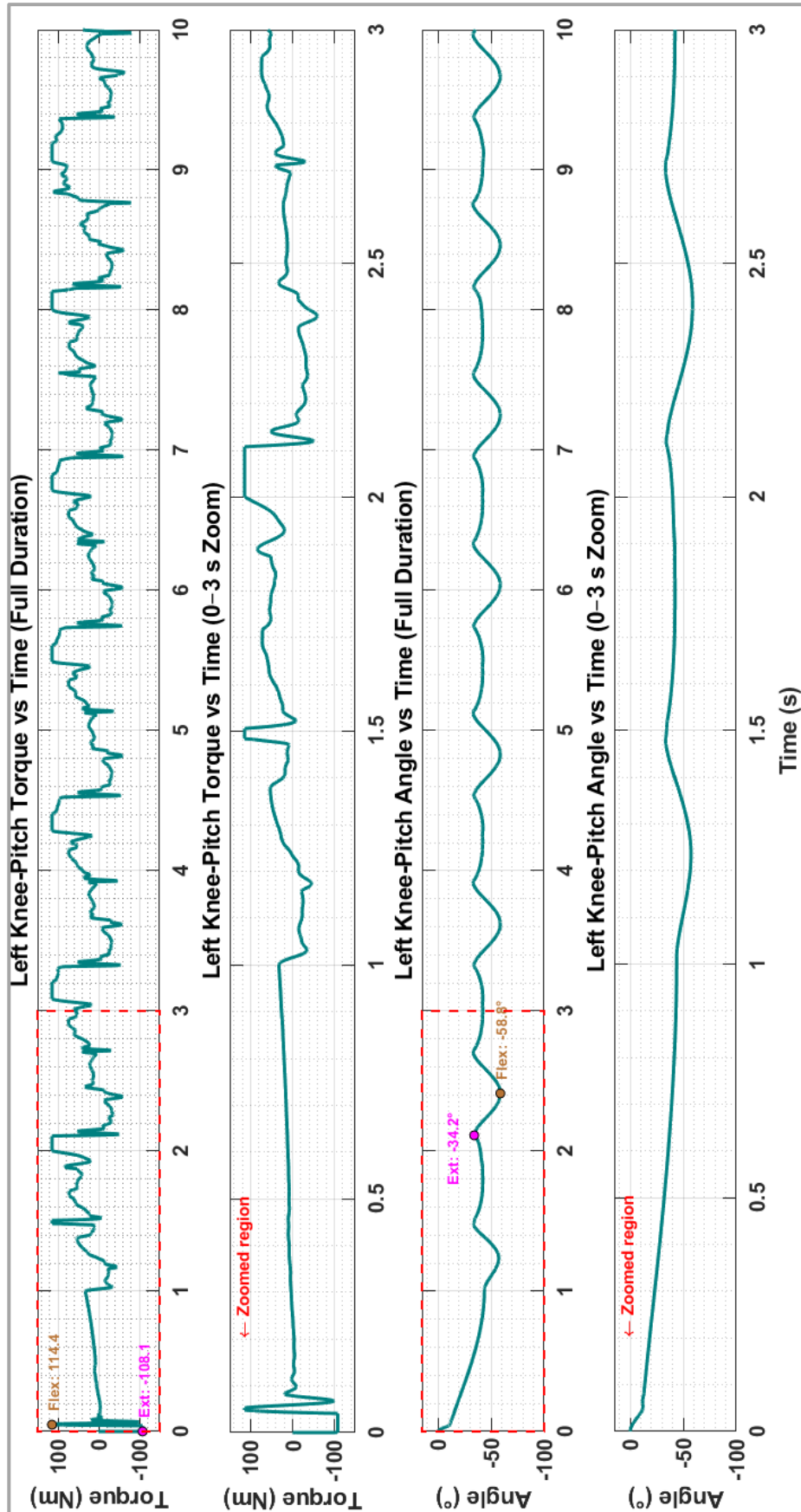


Figure F.8: Full and zoomed-in left knee torque and angle vs time

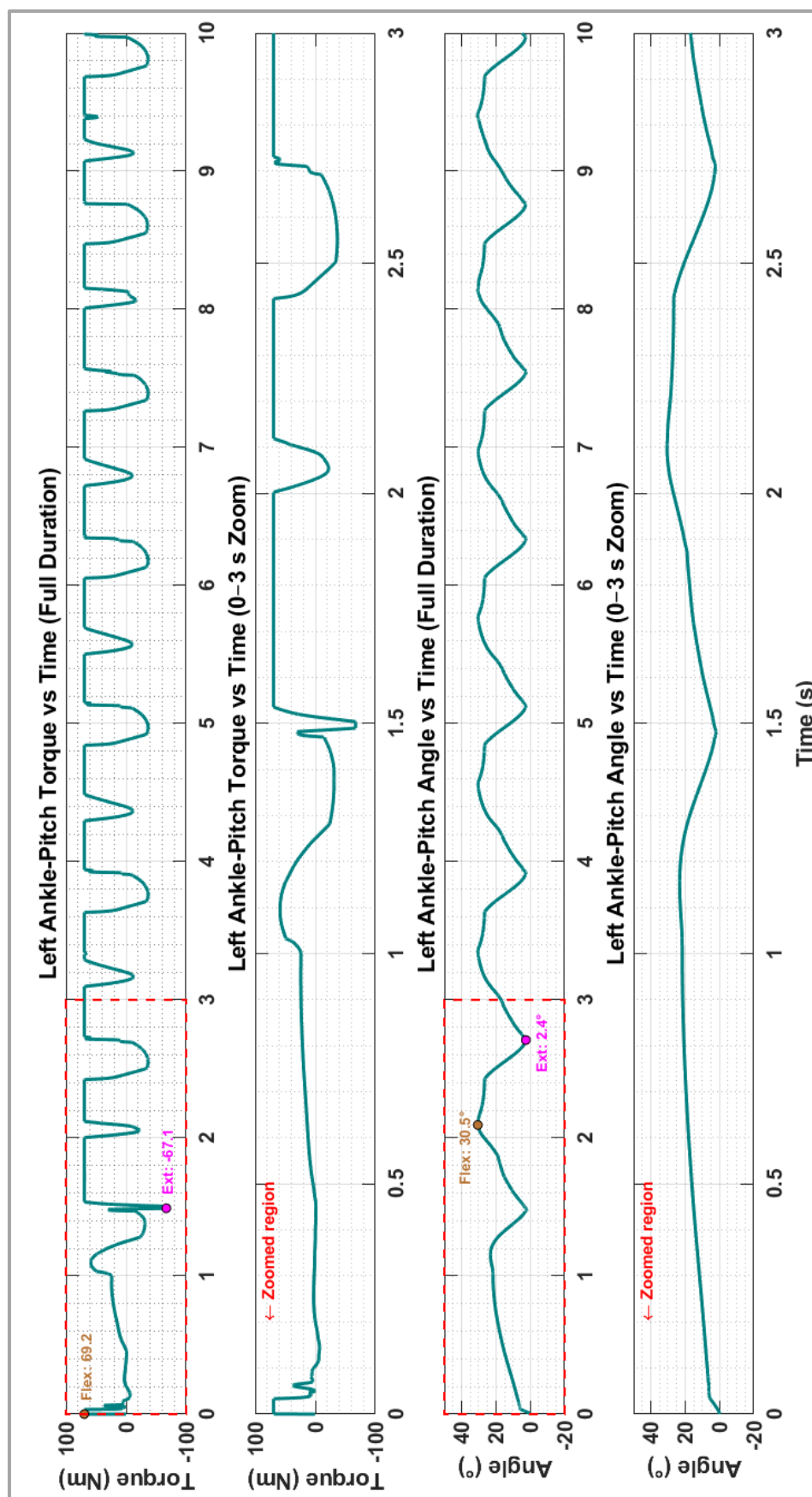


Figure F.9: Full and zoomed-in left ankle pitch torque and angle over time

Appendix G: Hardware datasheet

Figures G1 to G4 provide details on the selected motors and the harmonic drive.

EC 90 flat Ø90 mm, brushless, 220 watt
Open Rotor

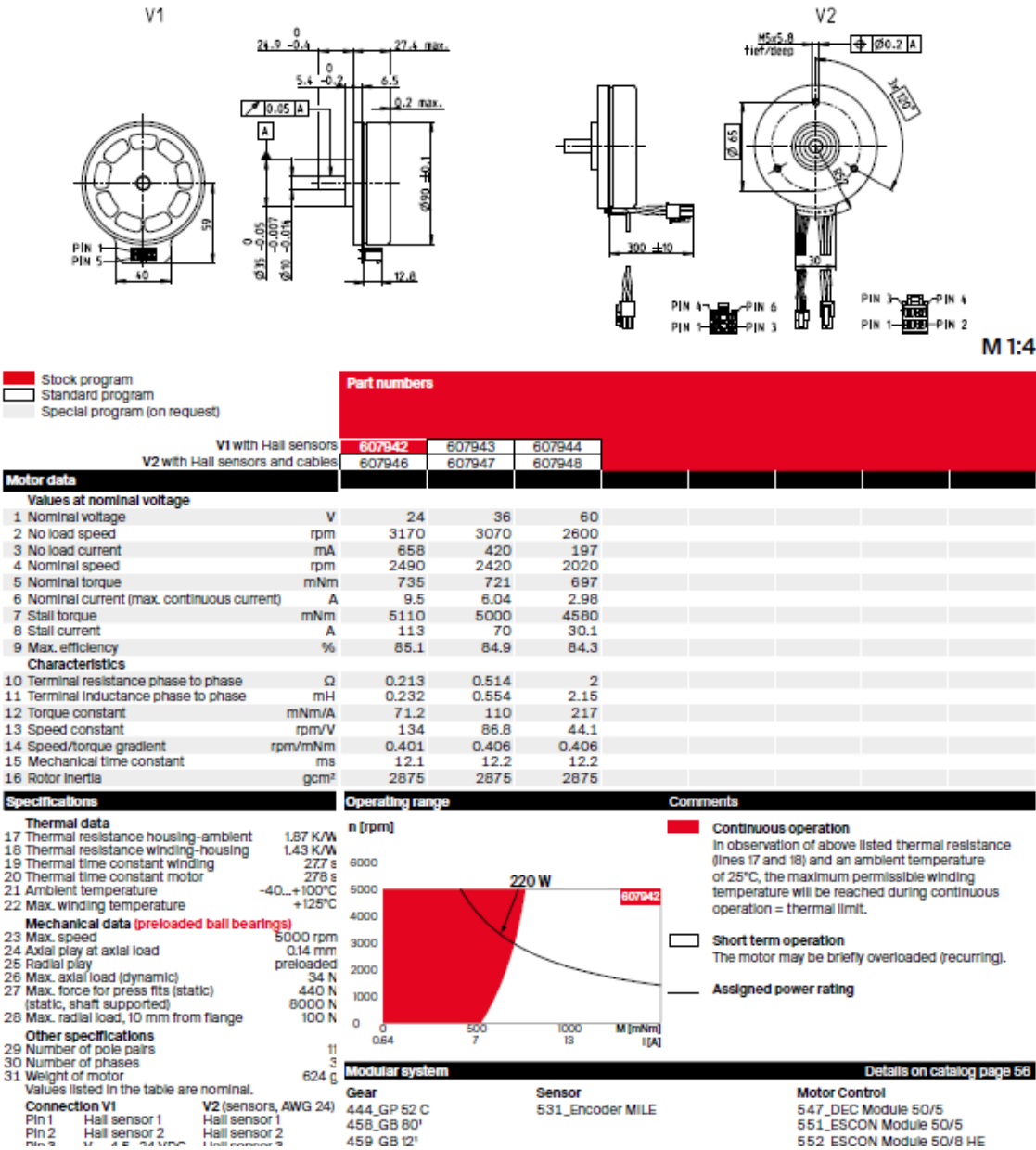
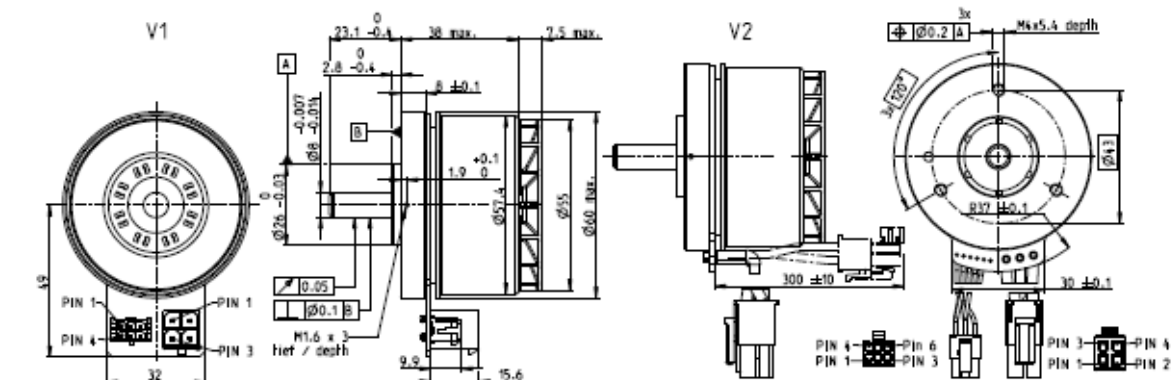


Figure G.1: EC 60 flat Ø60 mm, brushless, 150 Watt

EC 60 flat $\varnothing 60$ mm, brushless, 200 watt Ventilated



M 1:2

- Stock program
- Standard program
- Special program (on request)

Part numbers

V1 with Hall sensors	625860	614949	625861
V2 with Hall sensors and cables	647696	642221	647697

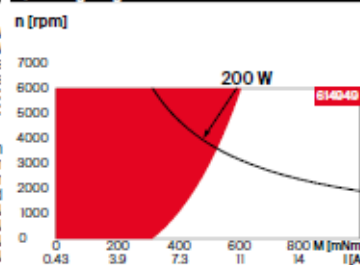
Motor data

Values at nominal voltage		12	24	48
1 Nominal voltage	V	12	24	48
2 No load speed	rpm	3760	4300	4020
3 No load current	mA	815	497	224
4 Nominal speed	rpm	2790	3240	3020
5 Nominal torque	mNm	496	540	581
6 Nominal current (max. continuous current)	A	15.2*	9.34	4.63
7 Stall torque	mNm	2210	2510	2650
8 Stall current	A	113	83.2	43.9
9 Max. efficiency	%	83.7	85.1	86.1
Characteristics				
10 Terminal resistance phase to phase	Ω	0.106	0.288	1.09
11 Terminal inductance phase to phase	mH	0.0911	0.279	1.28
12 Torque constant	mNm/A	30	52.5	113
13 Speed constant	rpm/V	318	182	84.8
14 Speed/torque gradient	rpm/mNm	1.13	0.998	0.823
15 Mechanical time constant	ms	9.8	8.69	7.17
16 Rotor inertia	gcm ²	832	832	832

Specifications

- Thermal data
 - 17 Thermal resistance housing-ambient 1.22 K/W
 - 18 Thermal resistance winding-housing 0.843 K/W
 - 19 Thermal time constant winding 9.19 s
 - 20 Thermal time constant motor 44 s
 - 21 Ambient temperature -40...+100°C
 - 22 Max. winding temperature +125°C
- Mechanical data (preloaded ball bearings)
 - 23 Max. speed 6000 rpm
 - 24 Axial play at axial load < 12.0 N 0 mm
 - 24 Axial play at axial load > 12.0 N 0.14 mm
 - 25 Radial play preloaded 12 N
 - 26 Max. axial load (dynamic) 170 N
 - 27 Max. force for press fits (static) (static, shaft supported) 8000 N
 - 28 Max. radial load, 5 mm from flange 112 N
- Other specifications
 - 29 Number of pole pairs 7
 - 30 Number of phases 3
 - 31 Weight of motor 360 g

Operating range



- Continuous operation
In observation of above listed thermal resistance (lines 17 and 18) and an ambient temperature of 25°C, the maximum permissible winding temperature will be reached during continuous operation = thermal limit.
- Short term operation
The motor may be briefly overloaded (recurring).
- Assigned power rating

Modular system

Details on catalog page 56

Figure G.2: EC 60 flat $\varnothing 60$ mm, brushless, 100 Watt

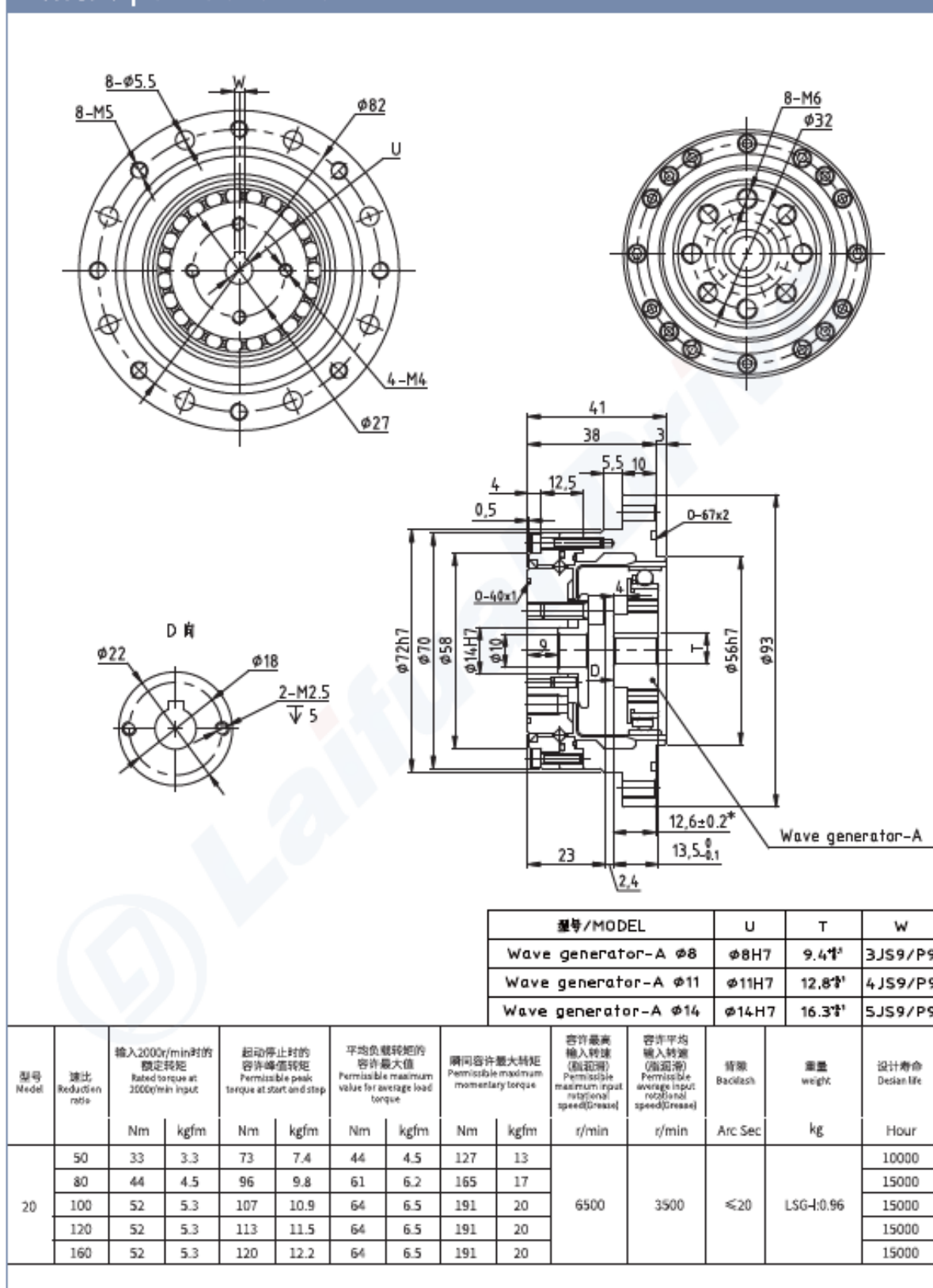
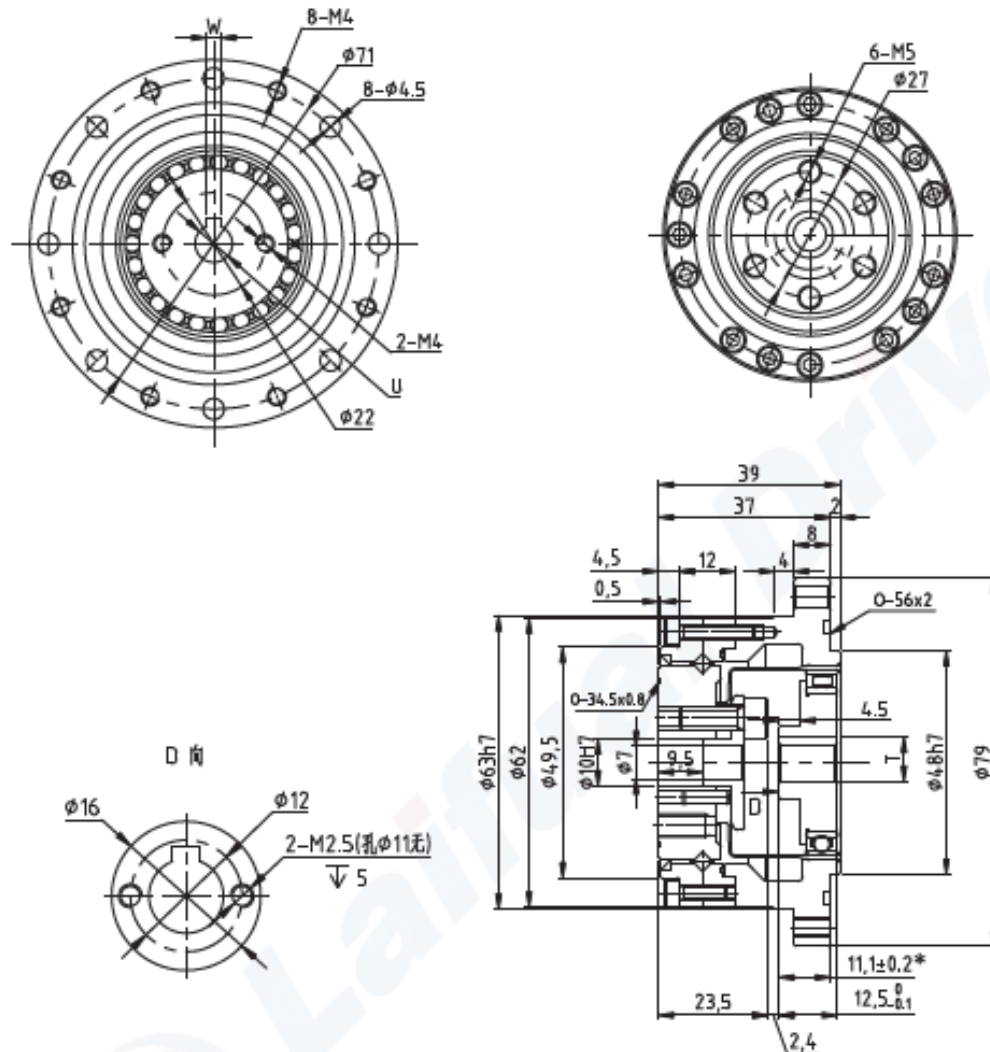


Figure G.3: Laifual drive LSG-20-160:1-U-1



型号/MODEL	U	T	W
Wave generator-A φ8	φ8H7	9.4 ⁺¹ ₋₁	3JS9/P9
Wave generator-A φ11	φ11H7	12.8 ⁺¹ ₋₁	4JS9/P9

型号 Model	速比 Reduction ratio	输入2000r/min时的 额定转矩 Rated torque at 2000r/min input		启动停止时的 容许峰值转矩 Permissible peak torque at start and stop		平均负载转矩的 容许最大值 Permissible maximum value for average load torque		瞬间容许最大转矩 Permissible maximum momentary torque		容许最大 输入转速 (Grease) Permissible maximum input rotational speed (Grease)	容许平均 输入转速 (Grease) Permissible average input rotational speed (Grease)	背隙 Backlash	重量 weight	设计寿命 Design life
		Nm	kgfm	Nm	kgfm	Nm	kgfm	Nm	kgfm	r/min	r/min	Arc Sec	kg	Hour
17	50	21	2.1	44	4.5	34	3.4	91	9	7300	3500	≤20	LSG-17-10.67	10000
	80	29	2.9	56	5.7	35	3.6	109	11					15000
	100	31	3.2	70	7.2	51	5.2	109	11					15000
	120	31	3.2	70	7.2	51	5.2	109	11					15000

Figure G.4: Laifual drive LSG-17-120:1-U-1

Brand	MaxAmps Lithium Batteries	Watt hours	1944 Wh
Capacity (milliamps - amp hours)	30000mAh - 30Ah	Energy density	250 Wh/kg
Nominal voltage	18S 64.8v	Main lead length(custom lengths available)	5.5"(140mm)
Maximum voltage	71.4v	Balance lead length(custom lengths available)	5.5"(140mm)
Minimum voltage	45v	Length	14.92"(379mm)
Recommended landing voltage(air)	51v	Width	5.82"(127mm)
Recommended cut-off voltage(ground)	48.6v	Height	2.95"(75mm)
Chemistry	Lithium-ion(Li-ion)	Weight	272.1oz(7714g)
Maximum continuous discharge	150 amps	SKU	
Maximum charge current	30 amps		

Figure G.5: MaxAmps Lithium Li-ion 30Ah 18S6P 64.8 v battery pack

Appendix H: Torque calculations

Figures H.1 to H.4 present the torque calculations for each joint of the walking using the moment method. In Figure H.1, both the use and the LLE are considered in the calculation, whereas in Figure H. 2, only the LLE frame and its actuators are considered, since the experiment does not include the user's weight. In Figures H.3 and H. 4, the calculations of the gravity compensation torque are provided.

The following variables were used in the provided Figures H1 and H2:

- τ_h .: Hip torque
- τ_k .: Knee torque
- τ_a .: Ankle torque
- m_t .: Mass of thigh
- m_{tf} .: Mass of thigh frame
- m_s .: Mass of shank
- m_{sf} .: Mass of shank frame
- m_f .: Mass of foot
- m_{ff} .: Mass of foot frame
- l_t .: Length of thigh
- l_s .: Length of shank
- l_f .: Length of foot
- h_a .: Height of the Ankle joint to the ground
- M_{wt} .: Mass of the lower body applied to the hip joint. This excludes the mass of the lower leg of the concerned leg during the stance phase, as the moment is considered about the knee joint.
- M_{ws} .: Mass of the lower body applied to the hip joint. This excludes the mass of the foot of the concerned leg during the stance phase, as the moment is taken about the ankle joint.

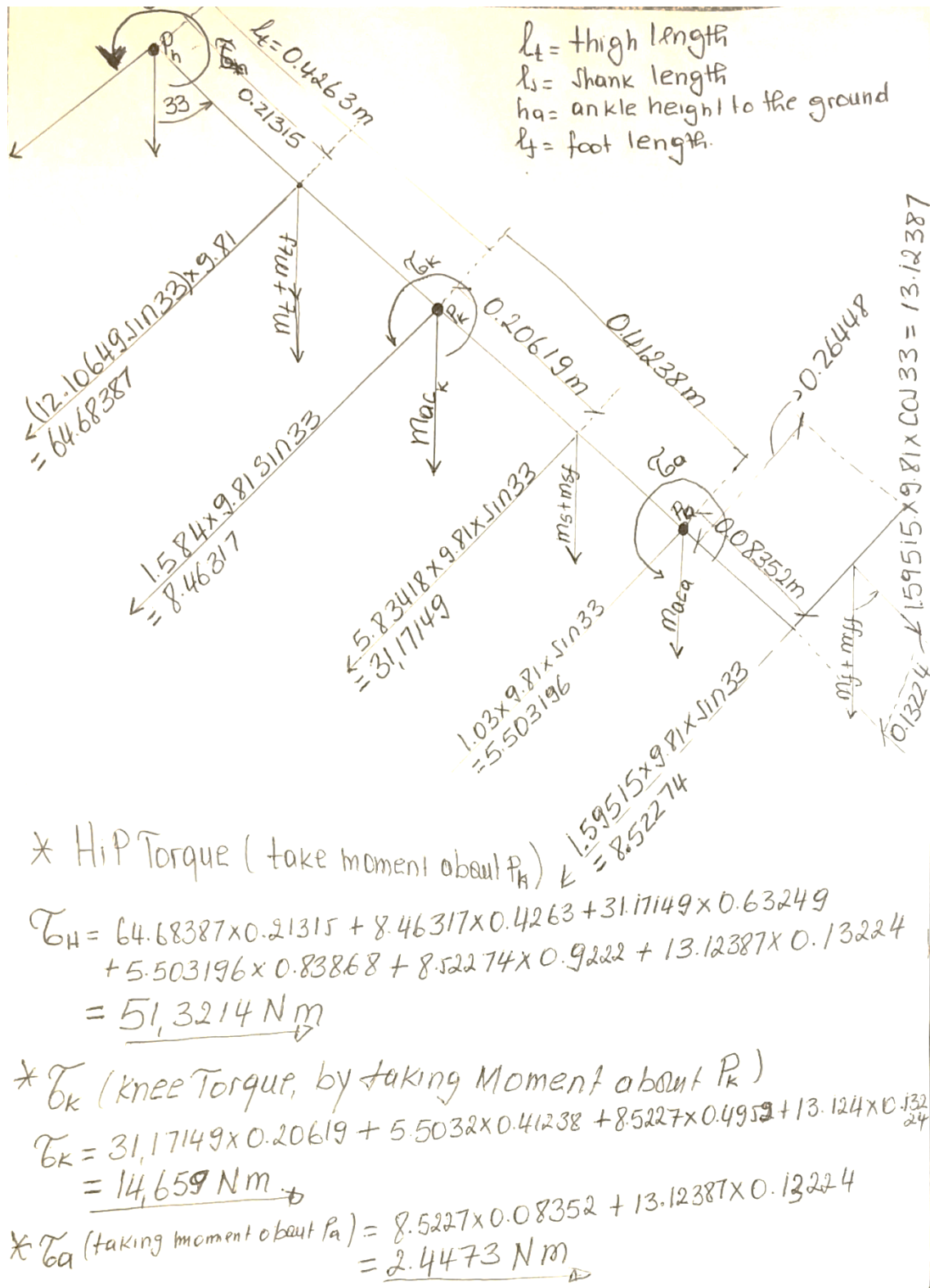
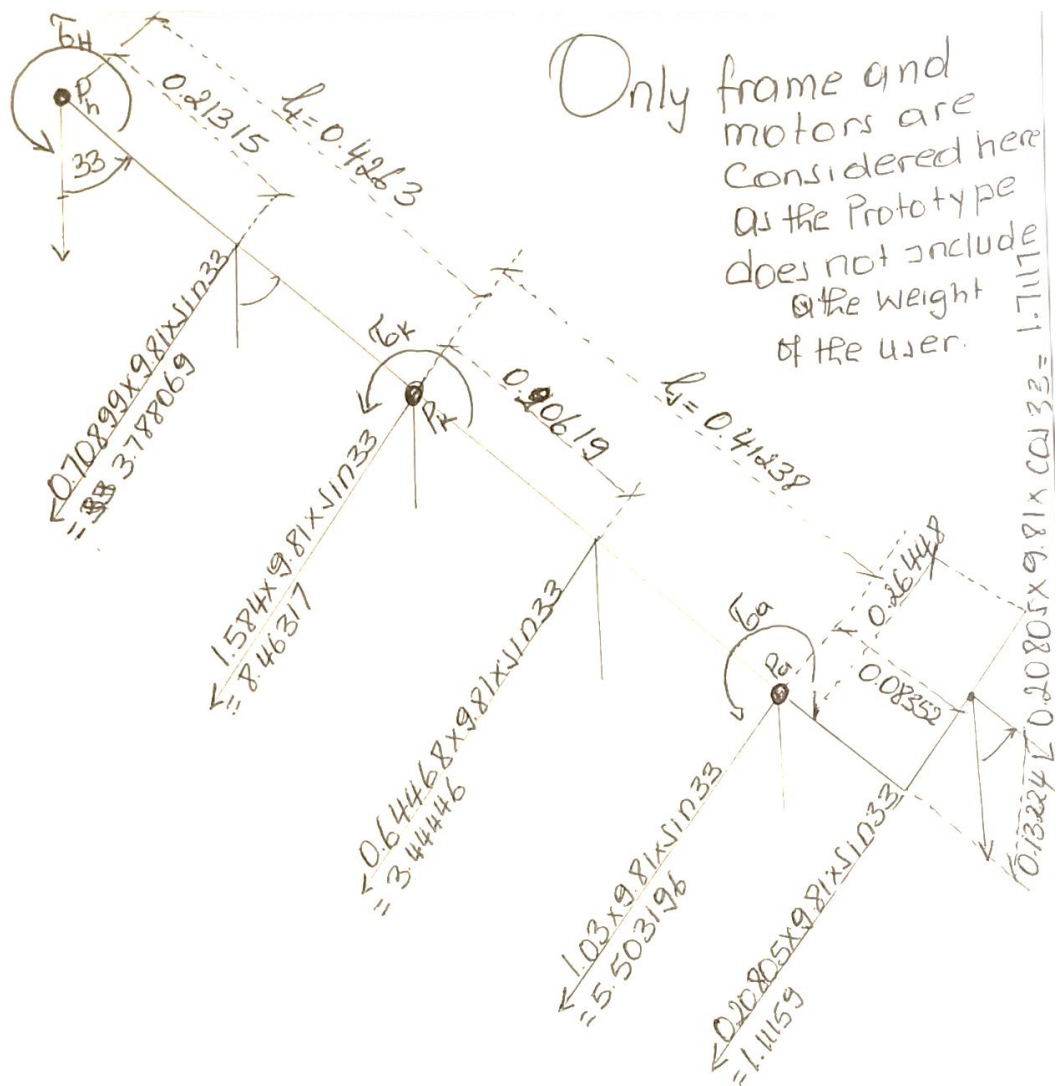


Figure H.1: Torque calculation of hip, knee, and ankle of LLE and user



* Hip torque (taking moment about P_h)

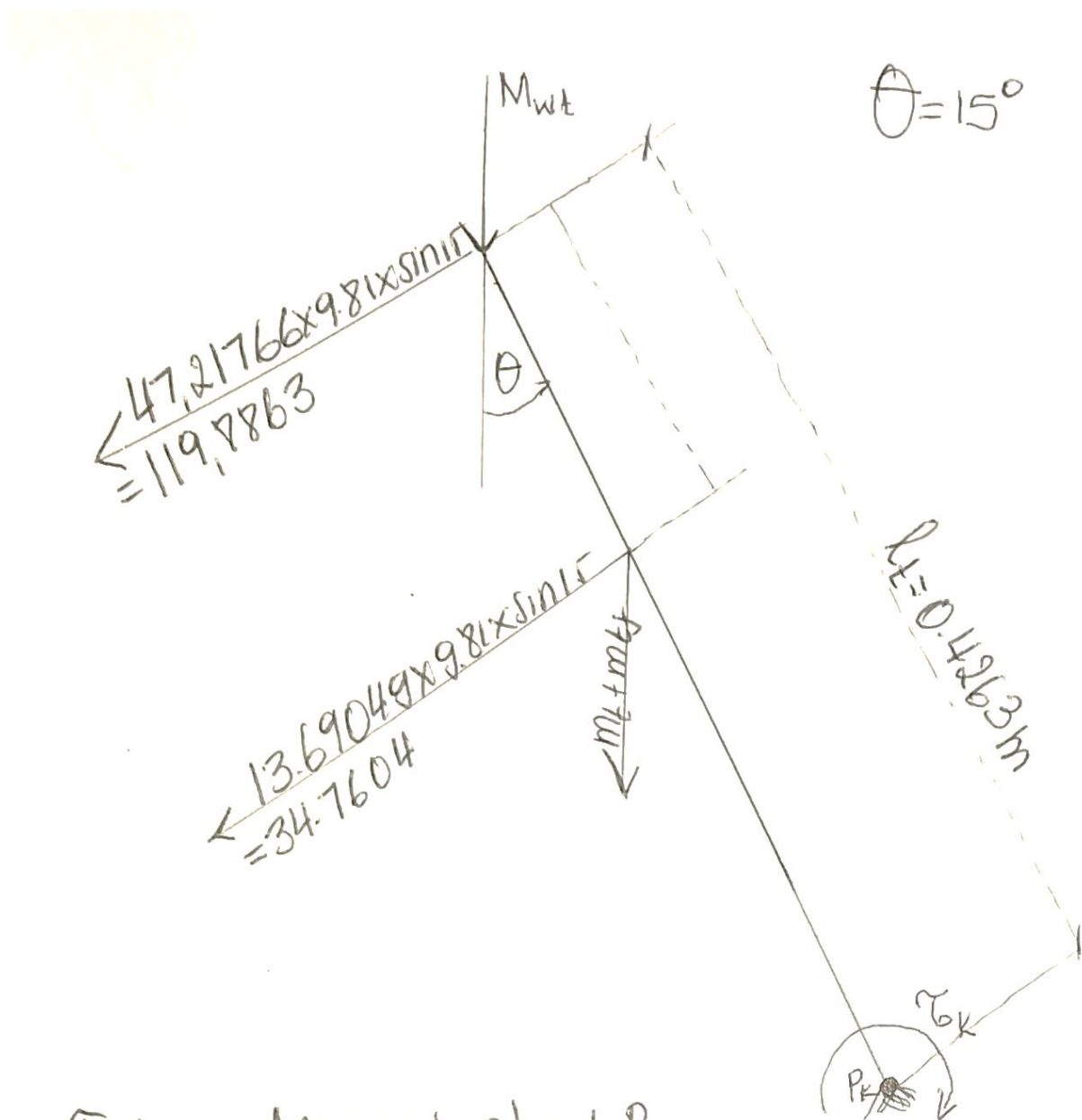
$$\tau_h = 3.78069 \times 0.21315 + 8.46317 \times 0.4263 + 3.44446 \times 0.63249 + 5.503196 \times 0.83868 + 1.1159 \times 0.9222 + 1.7117 \times 0.13224 = 12.460746 \text{ N.m}$$

* Knee torque (taking moment about P_k)

$$\tau_k = 3.44446 \times 0.20619 + 5.503196 \times 0.41238 + 1.1159 \times 0.4959 + 1.7117 \times 0.1322 = 3.7572 \text{ N.m}$$

* τ_a (Ankle Torque) $\Rightarrow 1.1159 \times 0.8352 + 1.7117 \times 0.13224 = 0.31919 \text{ N.m}$

Figure H.2: Torque calculation only LLE frame and motors

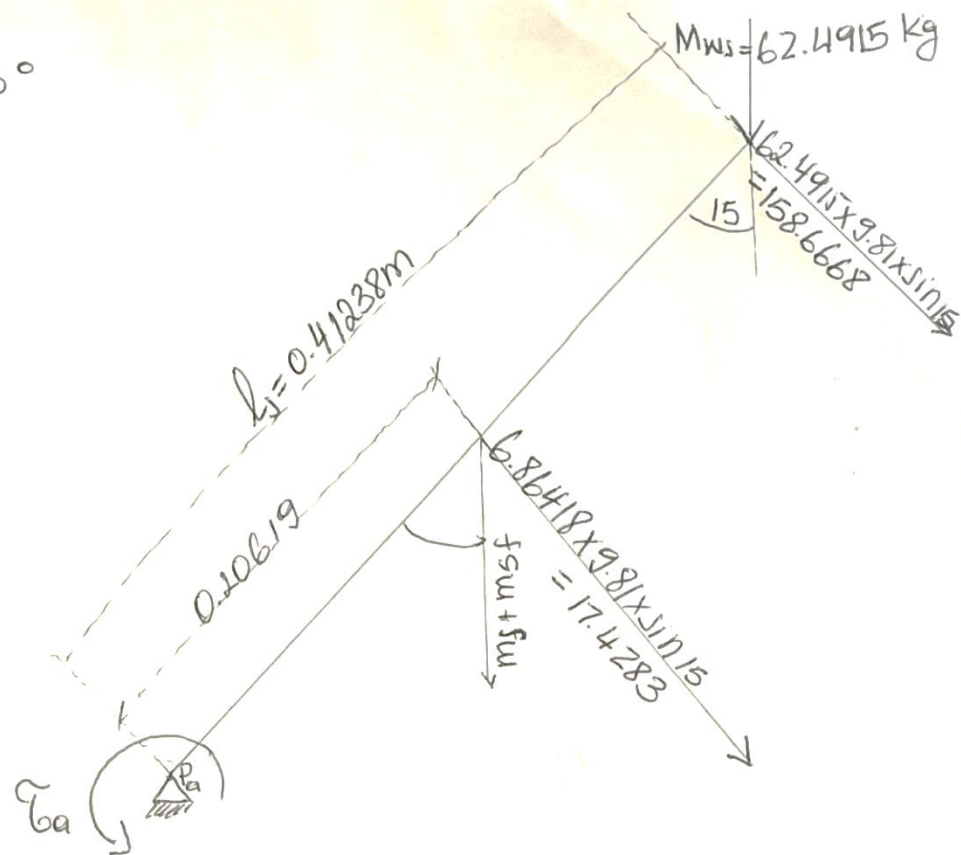


* Taking Moment about P_k

$$\begin{aligned}\tau_k &= 119.8863 \times 0.4263 + 34.7604 \times 0.21315 \\ &= 58.5167 \simeq \underline{58.517 \text{ Nm}}\end{aligned}$$

Figure H.3: Calculations of the knee Gravity compensation torque

$$\theta = 15^\circ$$



* Taking moment about P_a

$$\begin{aligned} \tau_{\text{ga}} &= 158.6668 \times 0.41238 + 17.4283 \times 0.20619 \\ &= 69.0245 \text{ Nm} \end{aligned}$$

Figure H.4: Calculations of the ankle Gravity compensation torque

Appendix I: Expansion of the Euler-Lagrange Equation for inverse dynamics

In this section the derivations of the inverse dynamic base on Euler LaGrange equations are developed, which is the continuity and expansion of the equation defined in the methodology under subsection 3.9.2.

- Thigh velocity

$$v_1^2 = \dot{x}_1^2 + \dot{y}_1^2 = l_{c1}^2 \dot{\theta}_1^2$$

- Shank velocity

$$v_2^2 = \dot{x}_2^2 + \dot{y}_2^2 = l_1^2 \dot{\theta}_1^2 + l_{c2}^2 (\dot{\theta}_1 + \dot{\theta}_2)^2 + 2l_1 l_{c2} \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2) \cos(\theta_2)$$

- Foot velocity

$$v_3^2 = \dot{x}_3^2 + \dot{y}_3^2 = l_1^2 \dot{\theta}_1^2 + l_2^2 (\dot{\theta}_1 + \dot{\theta}_2)^2 + l_{c3}^2 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3)^2 + 2l_1 l_2 \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2) \cos(\theta_2) + 2l_1 l_{c3} \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) \cos(\theta_2 + \theta_3) + 2l_2 l_{c3} (\dot{\theta}_1 + \dot{\theta}_2) (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) \cos(\theta_3)$$

- Lagrange

$$\begin{aligned} L = & \frac{1}{2} m_1 l_{c1}^2 \dot{\theta}_1^2 + \frac{1}{2} I_1 \dot{\theta}_1^2 + \frac{1}{2} m_2 \left[l_1^2 \dot{\theta}_1^2 + l_{c2}^2 (\dot{\theta}_1 + \dot{\theta}_2)^2 + 2l_1 l_{c2} \cos(\theta_2) \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2) \right] + \\ & \frac{1}{2} I_2 (\dot{\theta}_1 + \dot{\theta}_2)^2 + \frac{1}{2} m_3 \left[l_1^2 \dot{\theta}_1^2 + l_2^2 (\dot{\theta}_1 + \dot{\theta}_2)^2 + l_{c3}^2 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3)^2 + 2l_1 l_2 \cos(\theta_2) \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2) + \right. \\ & \left. 2l_1 l_{c3} \cos(\theta_2 + \theta_3) \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) + 2l_2 l_{c3} \cos(\theta_3) (\dot{\theta}_1 + \dot{\theta}_2) (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) \right] + \\ & \frac{1}{2} I_3 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3)^2 - m_1 g l_{c1} \sin(\theta_1) - m_2 g [l_1 \sin(\theta_1) + l_{c2} \sin(\theta_1 + \theta_2)] - m_3 g [l_1 \sin(\theta_1) + \\ & l_2 \sin(\theta_1 + \theta_2) + l_{c3} \sin(\theta_1 + \theta_2 + \theta_3)] \end{aligned}$$

a. Hip torque

$$\begin{aligned} \frac{\partial L}{\partial \theta_1} = & m_1 l_{c1}^2 \dot{\theta}_1 + I_1 \dot{\theta}_1 + m_2 \left[l_1^2 \dot{\theta}_1 + l_{c2}^2 (\dot{\theta}_1 + \dot{\theta}_2) + l_1 l_{c2} \cos(\theta_2) (2\dot{\theta}_1 + \dot{\theta}_2) \right] + I_2 (\dot{\theta}_1 + \dot{\theta}_2) + \\ & m_3 \left[l_1^2 \dot{\theta}_1 + l_2^2 (\dot{\theta}_1 + \dot{\theta}_2) + l_{c3}^2 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) + l_1 l_2 \cos(\theta_2) (2\dot{\theta}_1 + \dot{\theta}_2) + l_1 l_{c3} \cos(\theta_2 + \theta_3) (2\dot{\theta}_1 + \right. \\ & \left. \dot{\theta}_2 + \dot{\theta}_3) + l_2 l_{c3} \cos(\theta_3) (2\dot{\theta}_1 + 2\dot{\theta}_2 + \dot{\theta}_3) \right] + I_3 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) \end{aligned}$$

- Time derivative

$$\begin{aligned} \frac{d}{dt} \left(\frac{\partial L}{\partial \dot{\theta}_1} \right) = & \ddot{\theta}_1 [m_1 l_{c1}^2 + I_1 + m_2 (l_1^2 + l_{c2}^2 + 2l_1 l_{c2} \cos(\theta_2)) + I_2 + m_3 (l_1^2 + l_2^2 + l_{c3}^2 + \\ & 2l_1 l_2 \cos(\theta_2) + 2l_1 l_{c3} \cos(\theta_2 + \theta_3) + 2l_2 l_{c3} \cos(\theta_3)) + I_3] + \ddot{\theta}_2 [m_2 (l_{c2}^2 + l_1 l_{c2} \cos(\theta_2)) + I_2 + \\ & m_3 (l_2^2 + l_{c3}^2 + l_1 l_2 \cos(\theta_2) + l_1 l_{c3} \cos(\theta_2 + \theta_3) + 2l_2 l_{c3} \cos(\theta_3)) + I_3] + \ddot{\theta}_3 [m_3 (l_{c3}^2 + \\ & l_1 l_{c3} \cos(\theta_2 + \theta_3) + l_2 l_{c3} \cos(\theta_3)) + I_3] - m_2 l_1 l_{c2} \sin(\theta_2) (2\dot{\theta}_1 \dot{\theta}_2 + \dot{\theta}_2^2) - \\ & m_3 l_1 l_2 \sin(\theta_2) (2\dot{\theta}_1 \dot{\theta}_2 + \dot{\theta}_2^2) - m_3 l_1 l_{c3} \sin(\theta_2 + \theta_3) (2\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) (\dot{\theta}_2 + \dot{\theta}_3) - \\ & m_3 l_2 l_{c3} \sin(\theta_3) (2\dot{\theta}_1 \dot{\theta}_3 + 2\dot{\theta}_2 \dot{\theta}_3 + \dot{\theta}_3^2) \end{aligned}$$

- Partial derivative with respect to θ_1

$$\frac{\partial L}{\partial \theta_1} = -m_1 g l_{c1} \cos(\theta_1) - m_2 g [l_1 \cos(\theta_1) + l_{c2} \cos(\theta_1 + \theta_2)] - m_3 g [l_1 \cos(\theta_1) + l_2 \cos(\theta_1 + \theta_2) + l_{c3} \cos(\theta_1 + \theta_2 + \theta_3)]$$

- Final τ_1

$$\begin{aligned} \tau_1 = & \ddot{\theta}_1 [m_1 l_{c1}^2 + I_1 + m_2 (l_1^2 + l_{c2}^2 + 2l_1 l_{c2} \cos(\theta_2)) + I_2 + m_3 (l_1^2 + l_2^2 + l_{c3}^2 + 2l_1 l_2 \cos(\theta_2) + \\ & 2l_1 l_{c3} \cos(\theta_2 + \theta_3) + 2l_2 l_{c3} \cos(\theta_3)) + I_3] + \ddot{\theta}_2 [m_2 (l_{c2}^2 + l_1 l_{c2} \cos(\theta_2)) + I_2 + m_3 (l_2^2 + l_{c3}^2 + \\ & l_1 l_2 \cos(\theta_2) + l_1 l_{c3} \cos(\theta_2 + \theta_3) + 2l_2 l_{c3} \cos(\theta_3)) + I_3] + \ddot{\theta}_3 [m_3 (l_{c3}^2 + l_1 l_{c3} \cos(\theta_2 + \theta_3) + \\ & l_2 l_{c3} \cos(\theta_3)) + I_3] - m_2 l_1 l_{c2} \sin(\theta_2) (2\dot{\theta}_1 \dot{\theta}_2 + \dot{\theta}_2^2) - m_3 l_1 l_2 \sin(\theta_2) (2\dot{\theta}_1 \dot{\theta}_2 + \dot{\theta}_2^2) - \\ & m_3 l_1 l_{c3} \sin(\theta_2 + \theta_3) (2\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) (\dot{\theta}_2 + \dot{\theta}_3) - m_3 l_2 l_{c3} \sin(\theta_3) (2\dot{\theta}_1 \dot{\theta}_3 + 2\dot{\theta}_2 \dot{\theta}_3 + \dot{\theta}_3^2) + \\ & m_1 g l_{c1} \cos(\theta_1) + m_2 g [l_1 \cos(\theta_1) + l_{c2} \cos(\theta_1 + \theta_2)] + m_3 g [l_1 \cos(\theta_1) + l_2 \cos(\theta_1 + \theta_2) + \\ & l_{c3} \cos(\theta_1 + \theta_2 + \theta_3)] \end{aligned}$$

b. Knee torque

- Partial derivative with respect to θ_2

$$\frac{\partial L}{\partial \theta_2} = m_2 [l_{c2}^2 (\dot{\theta}_1 + \dot{\theta}_2) + l_1 l_{c2} \cos(\theta_2) \dot{\theta}_1] + I_2 (\dot{\theta}_1 + \dot{\theta}_2) + m_3 [l_2^2 (\dot{\theta}_1 + \dot{\theta}_2) + l_{c3}^2 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) + l_1 l_2 \cos(\theta_2) \dot{\theta}_1 + l_1 l_{c3} \cos(\theta_2 + \theta_3) \dot{\theta}_1 + l_2 l_{c3} \cos(\theta_3) (2\dot{\theta}_1 + 2\dot{\theta}_2 + \dot{\theta}_3)] + I_3 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3)$$

- Time derivative

$$\begin{aligned} \frac{d}{dt} \left(\frac{\partial L}{\partial \theta_2} \right) = & \ddot{\theta}_1 [m_2 (l_{c2}^2 + l_1 l_{c2} \cos(\theta_2)) + I_2 + m_3 (l_2^2 + l_{c3}^2 + l_1 l_2 \cos(\theta_2) + l_1 l_{c3} \cos(\theta_2 + \theta_3) + \\ & 2l_2 l_{c3} \cos(\theta_3)) + I_3] + \ddot{\theta}_2 [m_2 l_{c2}^2 + I_2 + m_3 (l_2^2 + l_{c3}^2 + 2l_2 l_{c3} \cos(\theta_3)) + I_3] + \ddot{\theta}_3 [m_3 (l_{c3}^2 + \\ & l_2 l_{c3} \cos(\theta_3)) + I_3] - m_2 l_1 l_{c2} \sin(\theta_2) \dot{\theta}_1 \dot{\theta}_2 - m_3 l_1 l_2 \sin(\theta_2) \dot{\theta}_1 \dot{\theta}_2 - m_3 l_1 l_{c3} \sin(\theta_2 + \theta_3) \dot{\theta}_1 (\dot{\theta}_2 + \\ & \dot{\theta}_3) - m_3 l_2 l_{c3} \sin(\theta_3) (2\dot{\theta}_1 + 2\dot{\theta}_2 + \dot{\theta}_3) \dot{\theta}_3 + m_2 l_1 l_{c2} \sin(\theta_2) \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2) + \\ & m_3 l_1 l_2 \sin(\theta_2) \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2) + m_3 l_1 l_{c3} \sin(\theta_2 + \theta_3) \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) \end{aligned}$$

- Partial derivative with respect to θ_2

$$\begin{aligned} \frac{\partial L}{\partial \theta_2} = & -m_2 l_1 l_{c2} \sin(\theta_2) \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2) - m_2 g l_{c2} \cos(\theta_1 + \theta_2) - m_3 l_1 l_2 \sin(\theta_2) \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2) - \\ & m_3 l_1 l_{c3} \sin(\theta_2 + \theta_3) \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) - m_3 l_2 l_{c3} \sin(\theta_3) (\dot{\theta}_1 + \dot{\theta}_2) (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) - \\ & m_3 g [l_2 \cos(\theta_1 + \theta_2) + l_{c3} \cos(\theta_1 + \theta_2 + \theta_3)] \end{aligned}$$

- Final τ_2

$$\begin{aligned} \tau_2 = & \ddot{\theta}_1 [m_2 (l_{c2}^2 + l_1 l_{c2} \cos(\theta_2)) + I_2 + m_3 (l_2^2 + l_{c3}^2 + l_1 l_2 \cos(\theta_2) + l_1 l_{c3} \cos(\theta_2 + \theta_3) + \\ & 2l_2 l_{c3} \cos(\theta_3)) + I_3] + \ddot{\theta}_2 [m_2 l_{c2}^2 + I_2 + m_3 (l_2^2 + l_{c3}^2 + 2l_2 l_{c3} \cos(\theta_3)) + I_3] + \ddot{\theta}_3 [m_3 (l_{c3}^2 + \\ & l_2 l_{c3} \cos(\theta_3)) + I_3] + m_2 l_1 l_{c2} \sin(\theta_2) \dot{\theta}_1^2 + m_3 l_1 l_2 \sin(\theta_2) \dot{\theta}_1^2 + m_3 l_1 l_{c3} \sin(\theta_2 + \theta_3) \dot{\theta}_1^2 - \end{aligned}$$

$$m_3 l_2 l_{c3} \sin(\theta_3) (2\dot{\theta}_1 + 2\dot{\theta}_2 + \dot{\theta}_3) \dot{\theta}_3 + m_2 g l_{c2} \cos(\theta_1 + \theta_2) + m_3 g [l_2 \cos(\theta_1 + \theta_2) + l_{c3} \cos(\theta_1 + \theta_2 + \theta_3)]$$

c. Ankle torque

- Partial derivative with respect to θ_3

$$\frac{\partial L}{\partial \dot{\theta}_3} = m_3 [l_{c3}^2 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) + l_1 l_{c3} \cos(\theta_2 + \theta_3) \dot{\theta}_1 + l_2 l_{c3} \cos(\theta_3) (\dot{\theta}_1 + \dot{\theta}_2)] + I_3 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3)$$

- Time derivative

$$\frac{d}{dt} \left(\frac{\partial L}{\partial \dot{\theta}_3} \right) = \ddot{\theta}_1 [m_3 (l_{c3}^2 + l_1 l_{c3} \cos(\theta_2 + \theta_3) + l_2 l_{c3} \cos(\theta_3)) + I_3] + \ddot{\theta}_2 [m_3 (l_{c3}^2 + l_2 l_{c3} \cos(\theta_3)) + I_3] + \ddot{\theta}_3 [m_3 l_{c3}^2 + I_3] - m_3 l_1 l_{c3} \sin(\theta_2 + \theta_3) \dot{\theta}_1 (\dot{\theta}_2 + \dot{\theta}_3) - m_3 l_2 l_{c3} \sin(\theta_3) (\dot{\theta}_1 + \dot{\theta}_2) \dot{\theta}_3 + m_3 l_1 l_{c3} \sin(\theta_2 + \theta_3) \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) + m_3 l_2 l_{c3} \sin(\theta_3) (\dot{\theta}_1 + \dot{\theta}_2) (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3)$$

- Partial derivative with respect to θ_3

$$\frac{\partial L}{\partial \theta_3} = -m_3 l_1 l_{c3} \sin(\theta_2 + \theta_3) \dot{\theta}_1 (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) - m_3 l_2 l_{c3} \sin(\theta_3) (\dot{\theta}_1 + \dot{\theta}_2) (\dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3) - m_3 g l_{c3} \cos(\theta_1 + \theta_2 + \theta_3)$$

- Final τ_3

$$\tau_3 = \ddot{\theta}_1 [m_3 (l_{c3}^2 + l_1 l_{c3} \cos(\theta_2 + \theta_3) + l_2 l_{c3} \cos(\theta_3)) + I_3] + \ddot{\theta}_2 [m_3 (l_{c3}^2 + l_2 l_{c3} \cos(\theta_3)) + I_3] + \ddot{\theta}_3 [m_3 l_{c3}^2 + I_3] + m_3 l_1 l_{c3} \sin(\theta_2 + \theta_3) \dot{\theta}_1^2 + m_3 l_2 l_{c3} \sin(\theta_3) (\dot{\theta}_1 + \dot{\theta}_2)^2 + m_3 g l_{c3} \cos(\theta_1 + \theta_2 + \theta_3)$$

Appendix J: Impact of torque and velocity variation on displacements

The figures in this section pertain to the relationship between the walking robot's displacement and variations in torque and velocity. Figures J.1 through J.4 focus on a torque variation of $\pm 10\%$, while Figures J.5 to J.8 concentrate on the influence of velocity variation (± 0.100 m/s) on the displacement.

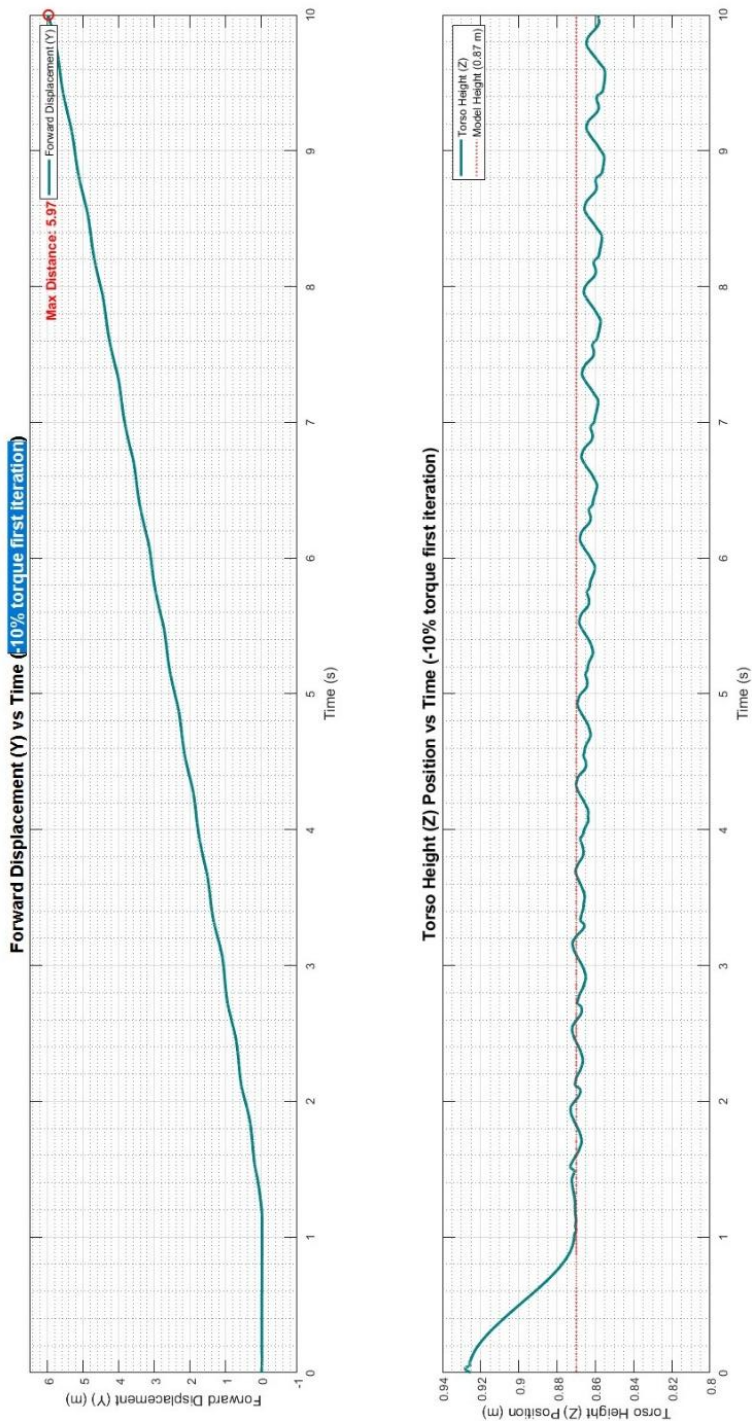


Figure J.1: Displacement with respect to torque variation (-10% 1st iteration)

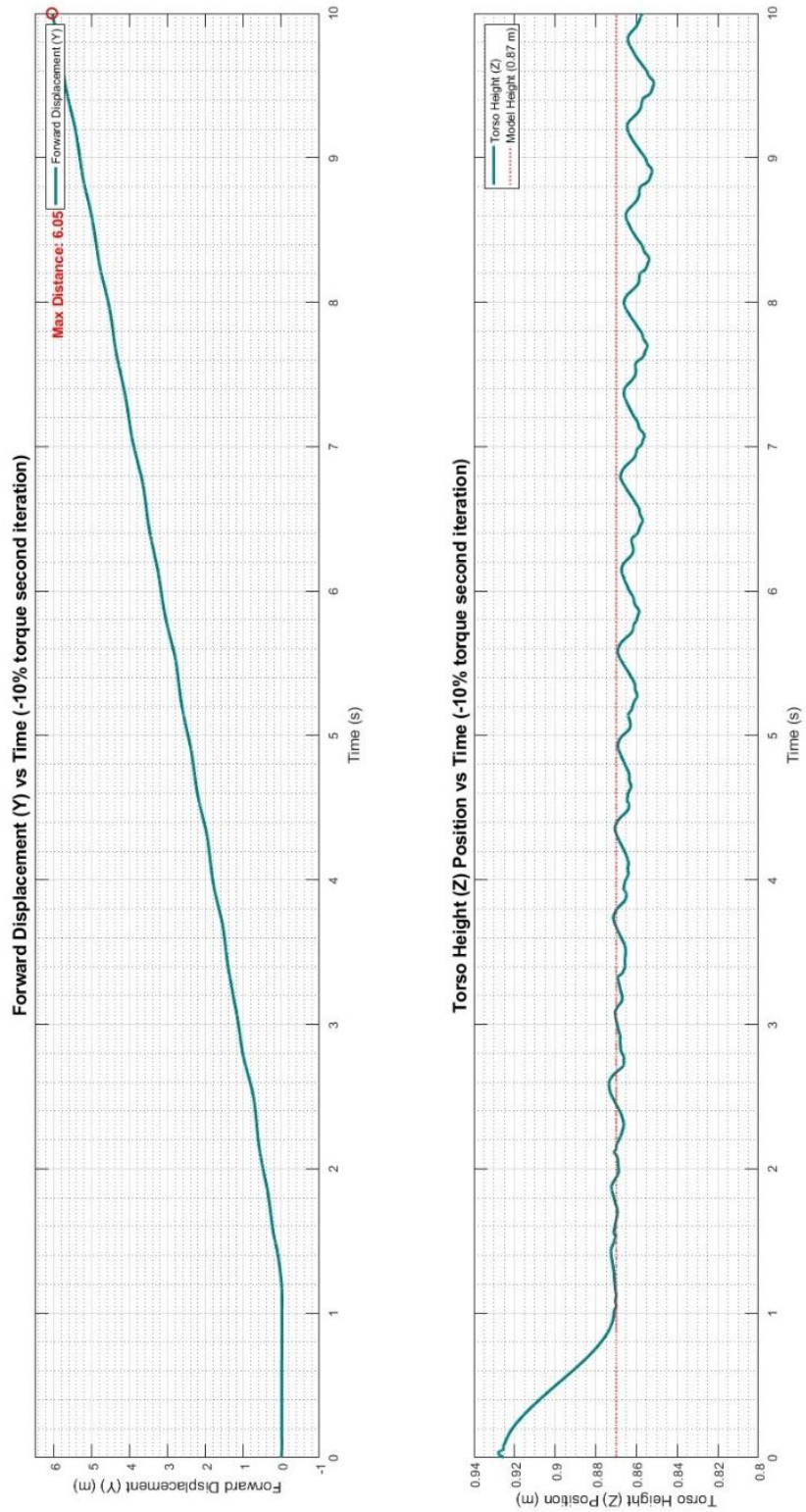


Figure J.2: Displacement with respect to torque variation (-10% 2nd iteration)

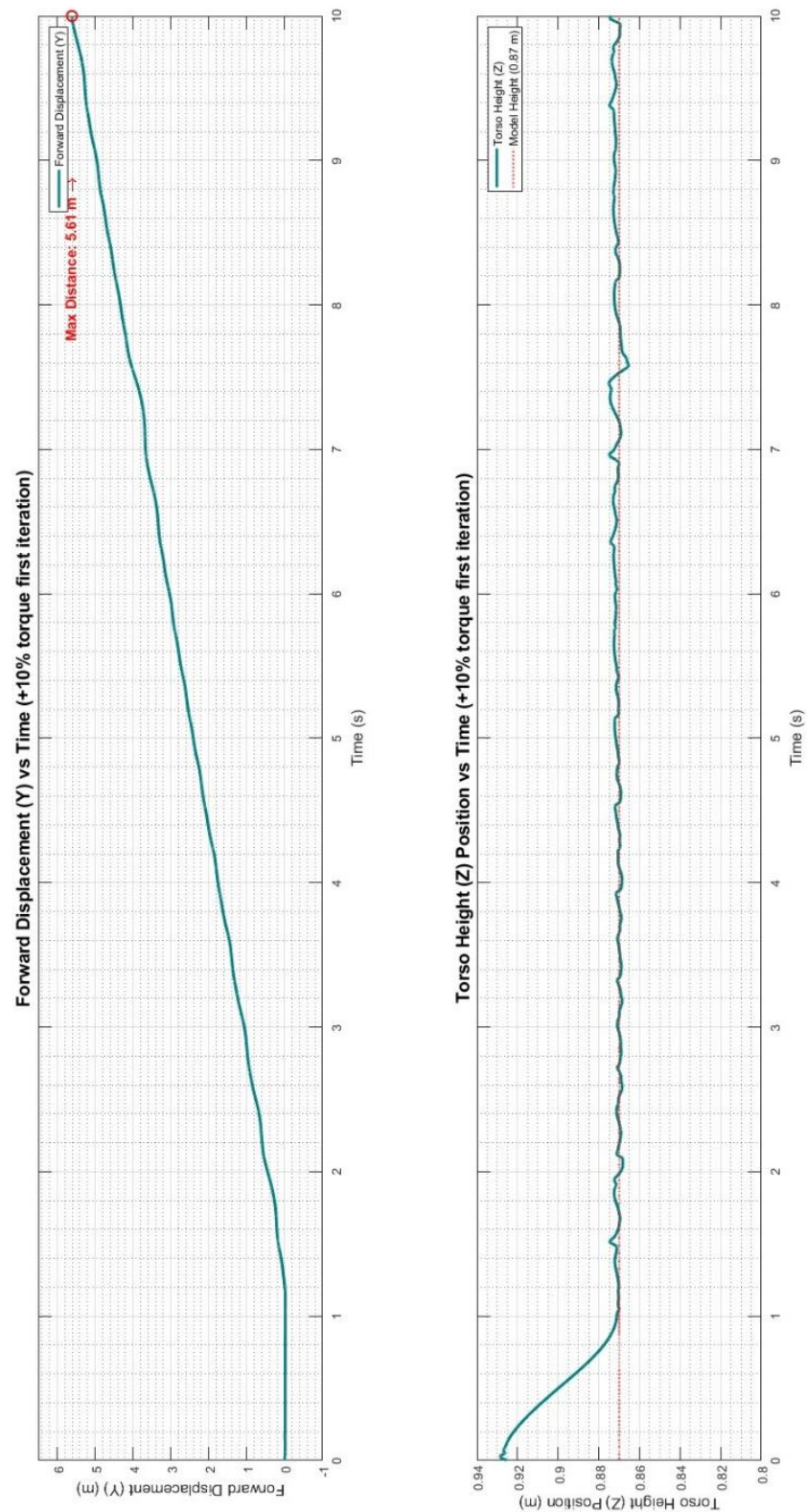


Figure J.3: Displacement with respect to torque variation (+10% 1st iteration)

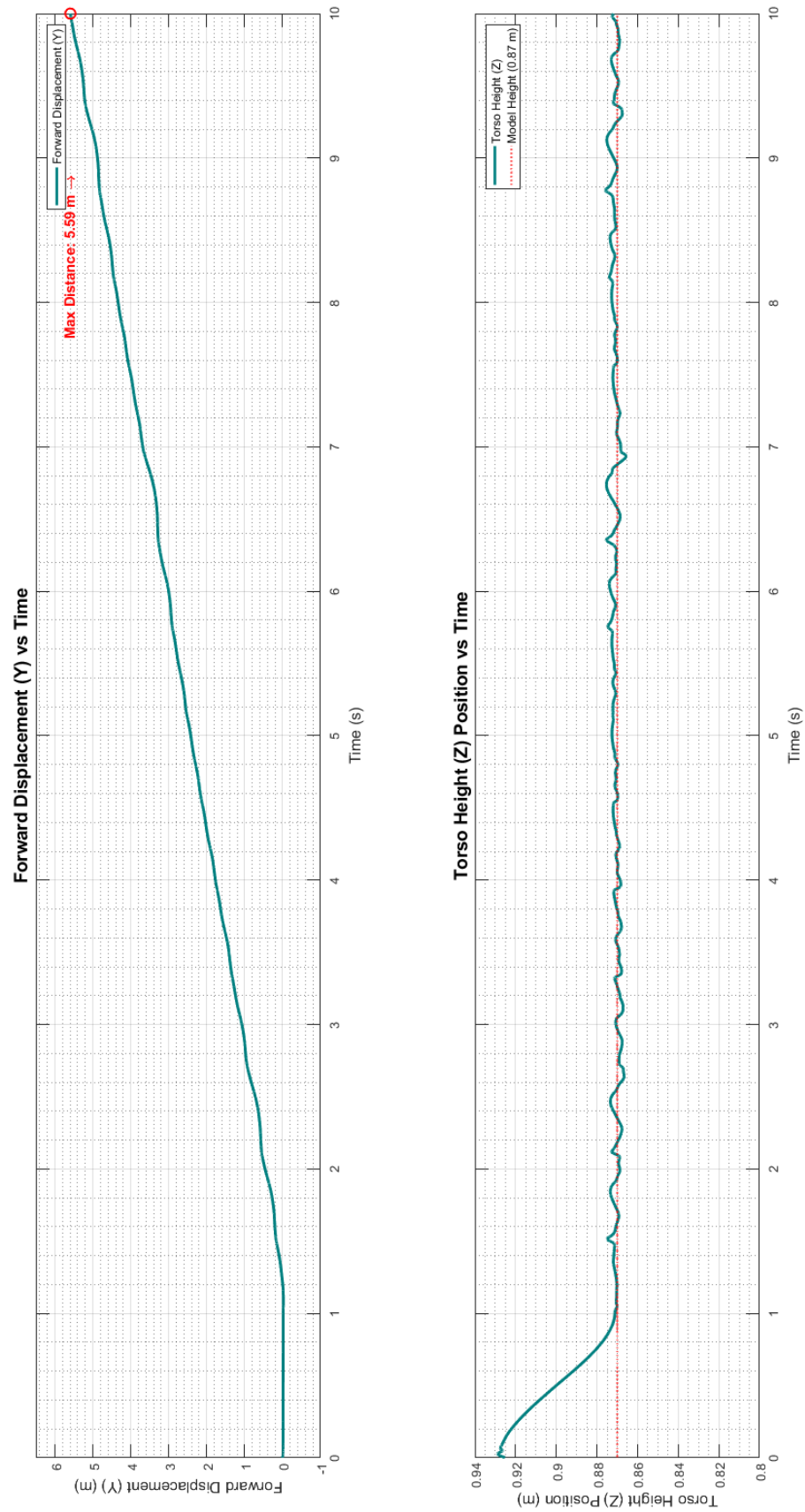


Figure J.4: Displacement with respect to torque variation (+10% 2nd iteration)

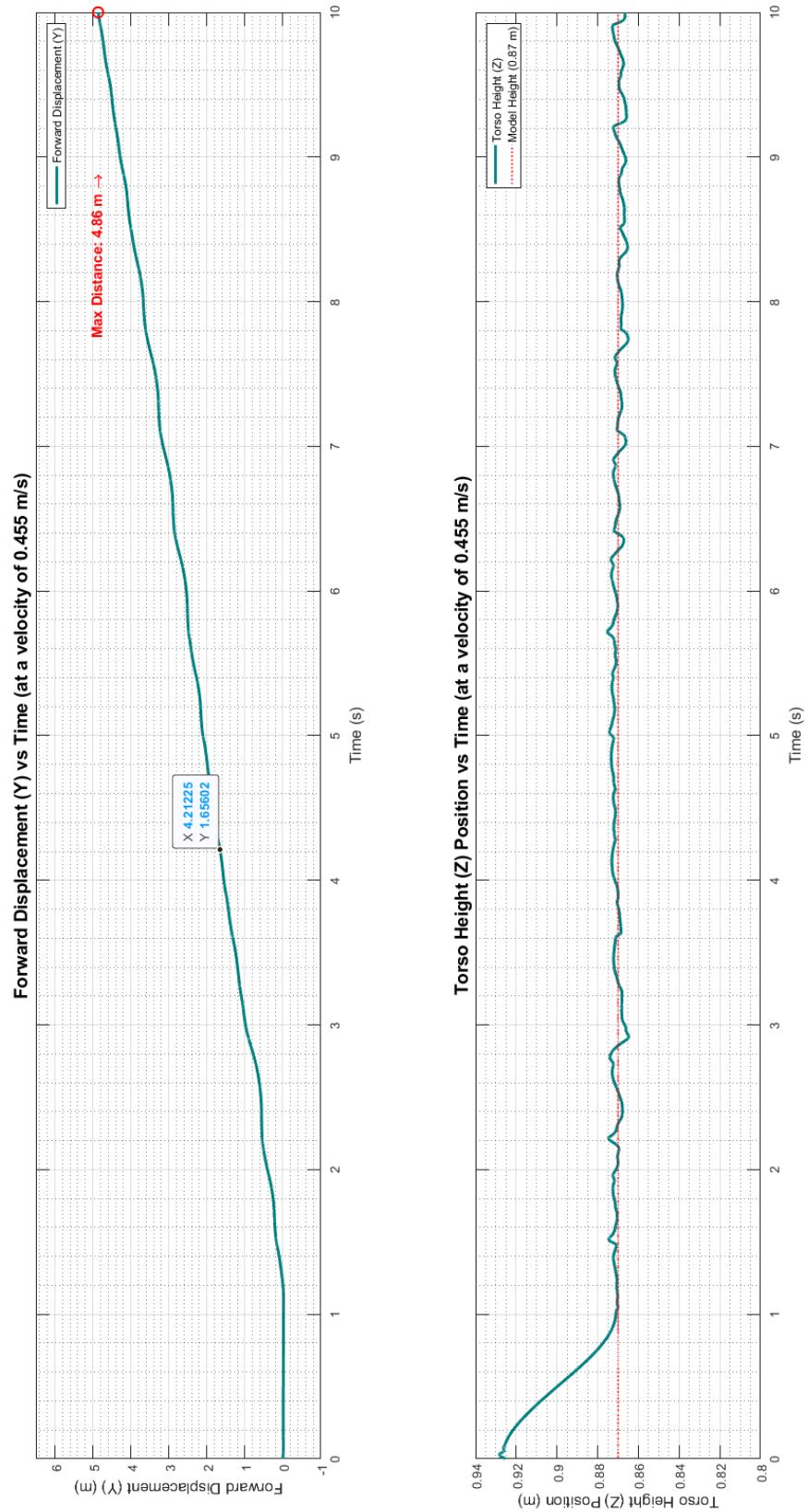


Figure J.5: Displacement with respect to velocity variation (-0.100 m/s 1st iteration)

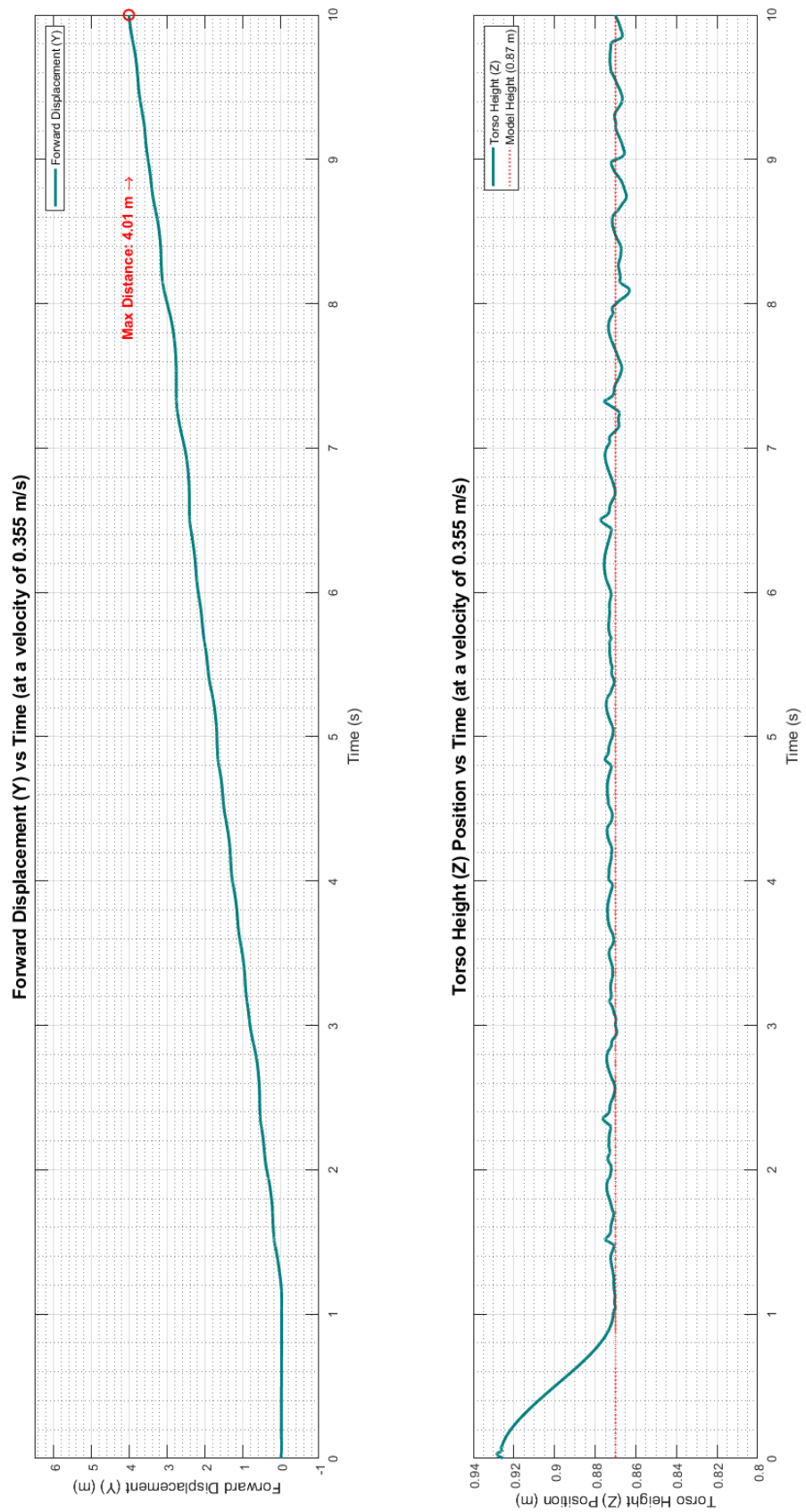


Figure J.6: Displacement with respect to velocity variation (-0.100 m/s 2nd iteration)

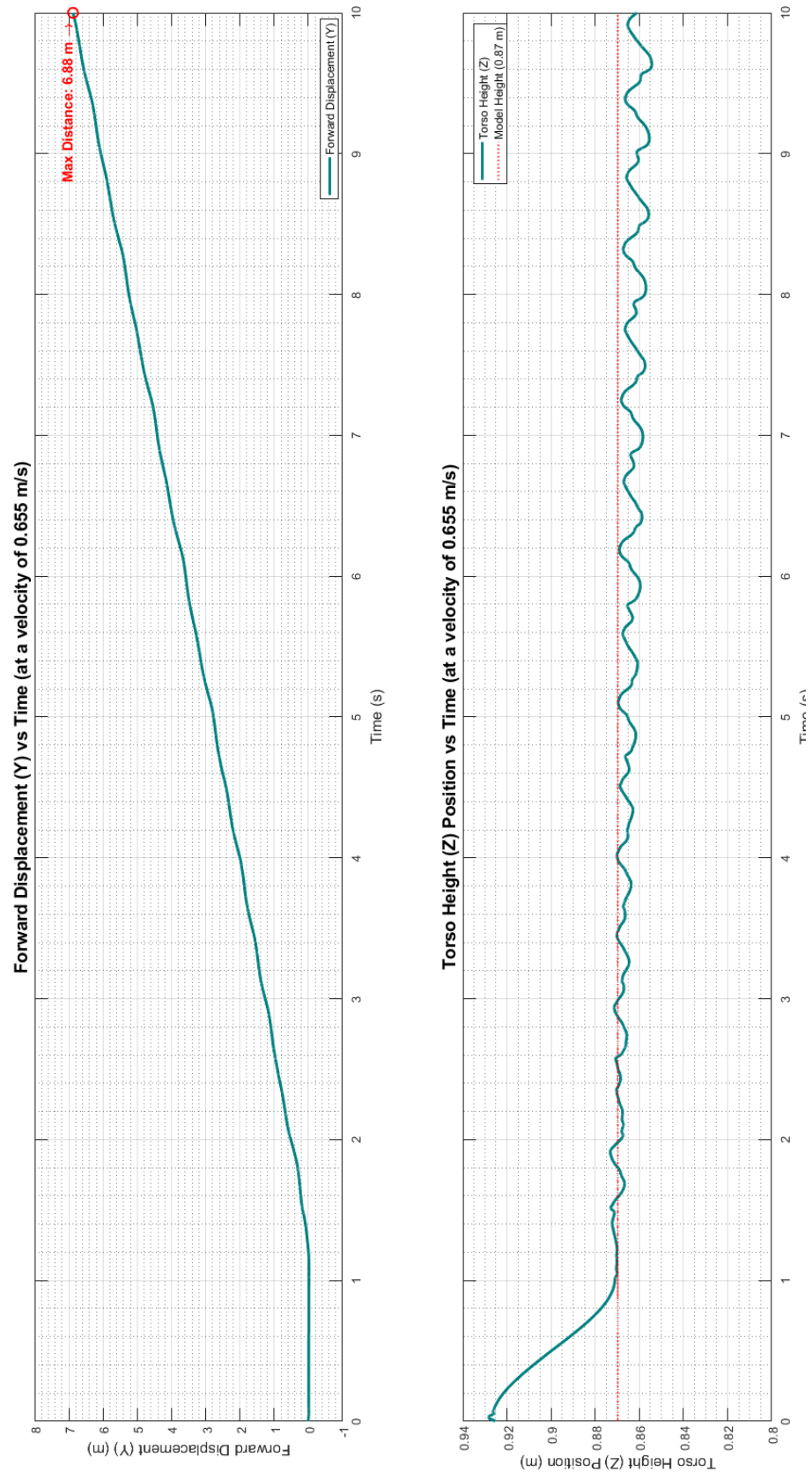


Figure J.7: Displacement with respect to velocity variation (+0.100 m/s 1st iteration)

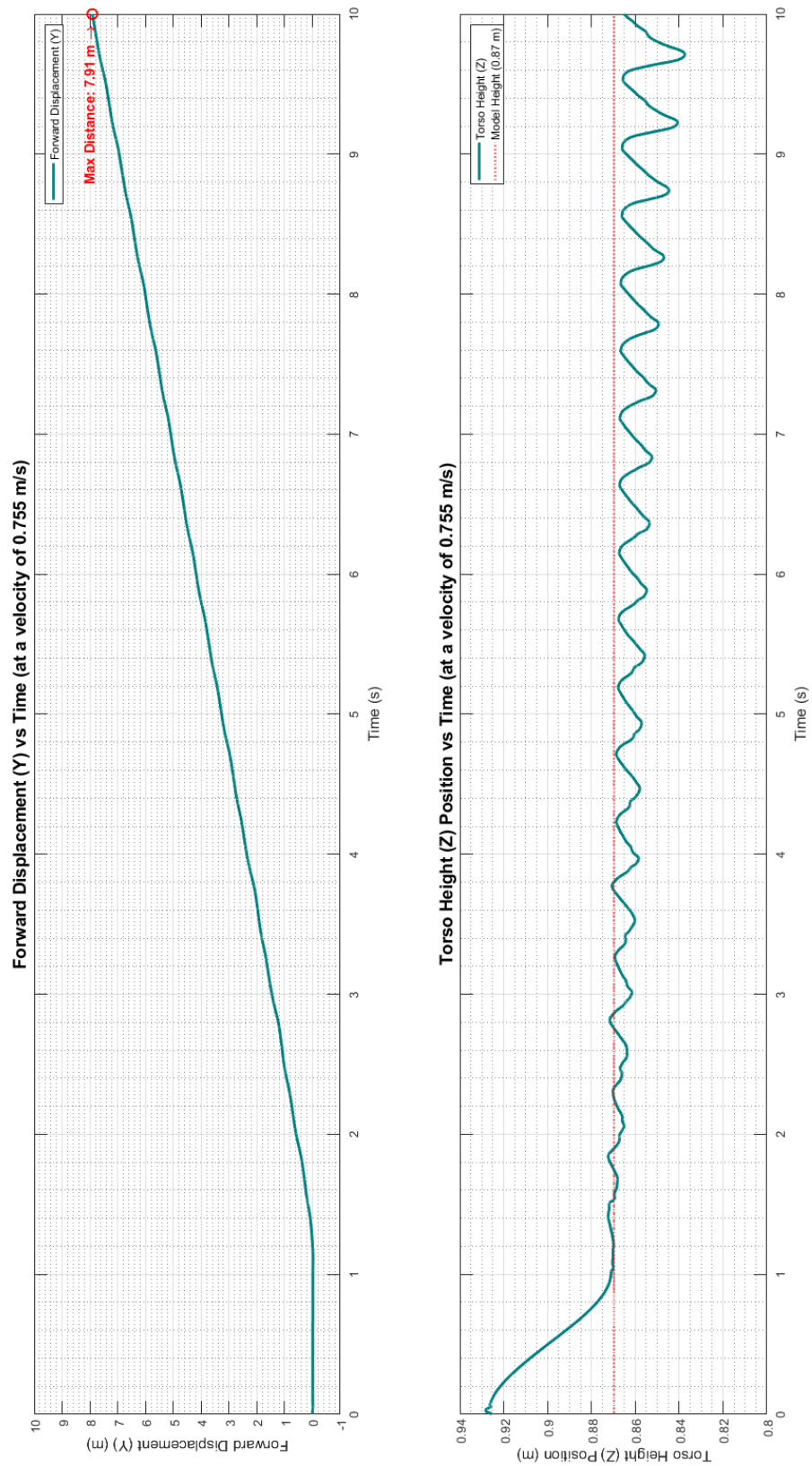


Figure J.8: Displacement with respect to velocity variation (+0.100 m/s 2nd iteration)