# Pedalling Optimisation and Balance Challenges in Leisure Cycling with Unilateral Transtibial Prostheses

エロイザ セラチュキ フロレス



出版情報：Kyushu University, 2025, 博士（工学）, 課程博士
バージョン：
権利関係：

Kyushu University
**Graduate School of Design**

# Pedalling Optimisation and Balance Challenges in Leisure Cycling with Unilateral Transtibial Prostheses

Heloísa Seratiuk Flores
セラチュキ・フロレス・エロイザ

June, 2025

Supervised by
Satoshi Muraki

# Abstract

Leisure cycling is a widely practiced activity among individuals who wear unilateral transtibial prostheses and offers various physical, psychological, and social benefits. As a low-impact alternative to walking, cycling increases physical activity levels and promotes social engagement. Despite its benefits, challenges such as movement and power asymmetries can arise owing to differences between intact and amputated limbs, affecting multiple aspects of cycling. This thesis investigated equipment adjustments and situational factors that can enhance leisure cycling for individuals with unilateral transtibial amputations.

The first study focused on saddle height settings. While higher saddle heights are generally recommended to optimise joint engagement, lower saddle heights may offer advantages for leisure cycling, particularly in mitigating the effects of stiff ankle joints in prostheses. This study assessed the impact of both high and low saddle heights on reducing asymmetries. The participants wore orthoses to simulate the prosthetic conditions. The results showed that a lower saddle height, achieving a knee angle of 37–45° at maximum extension, led to joint movements more akin to intact cycling. This adjustment also resulted in improved power delivery symmetry and muscle activity, similar to those of intact cycling.

The second study explored prostheses, specifically the ankle joints. This study evaluated how the addition of ankle movement through compression springs may bring the biomechanics of the affected limb closer to those of an intact limb, improving comfort and diminishing asymmetry. Therefore, this study aimed to investigate the use of springs of varying stiffness to allow prosthetic movements at the ankle. A prototype prosthesis was tested with two participants: one with a traumatic amputation and the other with a

congenital amputation. Results showed that stiffer springs improved comfort, muscle activity regularity, and power symmetry in participants with traumatic amputation. However, no significant improvements were observed in participants with congenital amputation, highlighting the need for personalised prosthetic solutions.

The final section addresses the situational challenges of balancing bicycles. Fear of falling, often present in individuals with lower limb amputations, can pose a barrier to the adoption of leisure cycling. It is generally understood that cycling speed affects cycling balance. Consequently, this study employed speed perturbations to evaluate the balancing challenges in simulated prosthetic conditions by comparing the results with those of regular, intact cycling. Prosthetic conditions caused slight instability and specific lean angles, but balance was largely unaffected within the established cycling motion. This study also found that the anticipation of speed changes possibly influenced balance and control.

In conclusion, this thesis highlights key considerations for leisure cycling among individuals with unilateral transtibial amputations. Lowering the saddle height was found to enhance symmetry between the affected and unaffected limbs while also improving manoeuvrability, as evidenced by the maintained balance. Furthermore, the observed balance improvements and heightened symmetry suggest that incorporating prosthetic ankle mobility can further increase user comfort. Overall, these findings suggest that refining prostheses and bicycles to promote natural limb symmetry and control could improve cycling accessibility, broadening their adoption as beneficial activities.

# Acknowledgements

First and foremost, I wish to express my deepest gratitude to my supervisor, Professor Satoshi Muraki, whose guidance, trust, and encouragement were instrumental in the successful completion of this thesis. This work, and my academic journey, would not have been possible without him. I would also like to extend my sincere thanks to Professor Kosuke Morinaga for his practical assistance—his contributions were essential to the completion of this thesis. For supervising this thesis, I am grateful to Professor Tomoaki Fuji and Professor Ping Yeap Loh, who provided valuable insights. My gratitude is extended to Professor Ping Yeap Loh as an examiner and to Professor Shigekazu Ishihara. Additionally, I am grateful to Professor Wen Liang Yeoh for his technical expertise and insightful feedback.

I would like to extend my thanks to Naoki Chijiwa and Takahito Oda from Arizono Orthopedic Supplies Co., Ltd., Kitakyushu, Japan, and Teppei Morimoto, CPO and CEO of JST Fit Inc., for their significant technical support, participant recruitment, and generous allocation of time and resources to aid in the experiments conducted for this research. I am also very grateful for all the participants who took part in the presented studies.

My heartfelt thanks go to everyone in the Ergonomics for All Ages and Abilities Lab, whose friendship and support over the past six years have made them like family to me.

I would also like to express my deepest gratitude to my mother and Christian, whose constant support have been invaluable, and to Ana. This work is dedicated to the memory of my grandparents and that of my first supervisor, Professor Alberto Ireneu Puppi.

Finally, I am sincerely thankful to Japan's Ministry of Education, Culture, Sports, Science and Technology for the financial support that made it possible for me to pursue both my Master's and Doctorate degrees at Kyushu University.

# Contents

# List of Figures

# List of Tables

# List of Publications

1. **Biomechanical Effects of Saddle Height Changes in Leisure Cycling with Unilateral Transtibial Prostheses: A Simulated Study**
   https://journals.plos.org/plosone/article?id=10.1371/journal.pone.0317121
   Heloísa Seratiuk Flores; Yeoh Wen Liang; Ping Yeap Loh; Kosuke Morinaga; Satoshi Muraki

2. **Biomechanical Analysis of Recreational Cycling with Unilateral Transtibial Prostheses**
   https://www.mdpi.com/2673-1592/5/3/52
   Heloísa Seratiuk Flores; Yeoh Wen Liang; Ping Yeap Loh; Kosuke Morinaga; Satoshi Muraki

# Chapter 1
# General Introduction

## 1.1. Background

Prosthetic devices are engineered to substitute impaired or missing body parts, potentially augmenting their functionality. This dissertation focuses on lower limb prostheses, which are pivotal in restoring the abilities lost due to amputation and are essential for facilitating mobility.

The International Organization for Standardization defines lower limb amputation as "the removal of a leg at any level below the pelvis" (*ISO 8548-2*, n.d.). Such amputations can occur at various levels (Fig. 1.1), and they include hemipelvectomy; hip disarticulation; knee and ankle disarticulation; and transfemoral, partial foot, and toe amputation, each necessitating specific rehabilitation strategies. Although all amputation levels can technically receive a prosthesis, the actual prescription of these devices depends on the patient's overall health and specific needs. In the United States, amputations due to vascular diseases predominantly involve the removal of a toe or toes (33.2%), followed by transtibial amputations (28.2%) (Renzi et al., 2006). Other amputation levels, such as hip disarticulations and hemipelvectomies, represent less than 1.5% of all cases. Accordingly, the present research will concentrate on unilateral transtibial amputations, given their prevalence and relevance to prosthetic application.

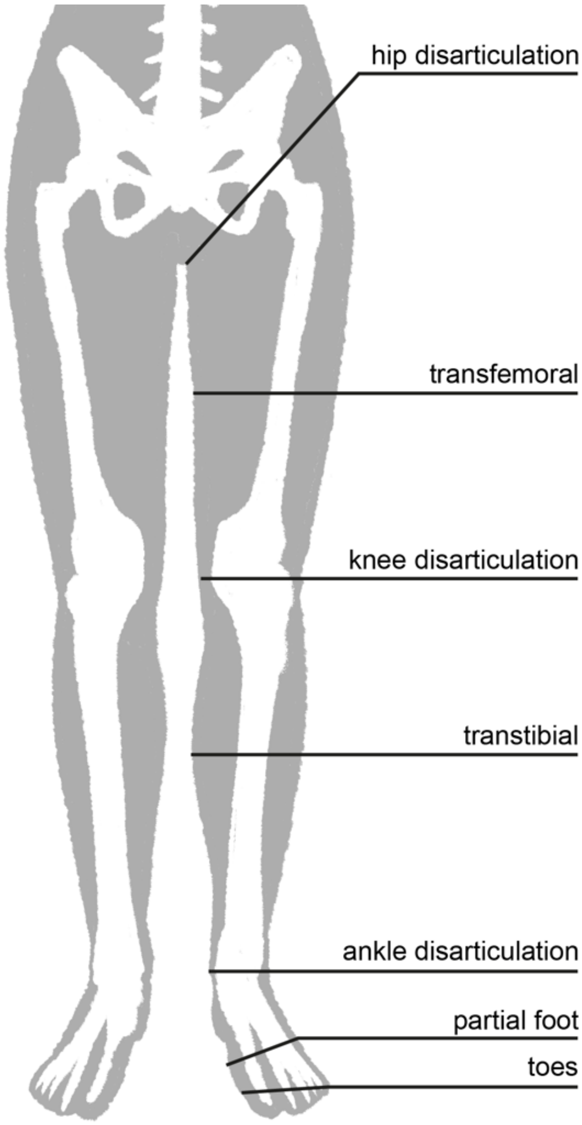


**Figure 0.1.** Levels of amputation.

An understanding of the significance of lower limb prostheses necessitates an awareness of the widespread occurrence of amputations. Data from the World Health Organization indicate that the incidence of amputations is influenced by geographical location and socioeconomic factors. In more developed nations such as the United States, the prevalence of vascular disease-related amputations is particularly high, affecting an estimated 5.6 million individuals (Caruso & Harrington, n.d.). Projections suggest a substantial increase in this figure by 2050, driven by the growing prevalence of diabetes and vascular diseases

(Ziegler-Graham et al., 2008). Conversely, in Japan, the prevalence is significantly lower, with around 150,000 individuals affected by amputation, according to the National Rehabilitation Centre for Persons with Disabilities. Globally, the majority of lower limb amputations are attributable to vascular conditions or diabetes (*Lower Limb Amputations*, 2017). However, in less developed regions, amputations are frequently caused by conflict and limited access to healthcare (Barth et al., 2021).

The loss of a lower limb not only diminishes physical capability but also incites psychological challenges related to bodily function and self-image, exacerbating existing health conditions or triggering new ones. Hence, leg prostheses play a crucial role in rehabilitation, profoundly affecting the user's daily life, sense of identity, and autonomy (Rybarczyk et al., 2004; Schaffalitzky et al., 2011).

## 1.2. Sports practice with lower limb prostheses

As previously established, lower limb amputations profoundly affect the patient's mobility and overall quality of life. Engaging in physical exercise can significantly bolster the rehabilitation process for individuals who have endured such amputations, as shown by previous research (Ülger et al., 2018). Notable benefits include reduced energy expenditure, alleviated circulation issues, and enhanced control and stability of prostheses. Such improvements can also positively influence primary health concerns commonly faced by amputees, including conditions such as vascular complications, which may have been the primary drive for the amputation.

The spectrum of sports accessible to individuals with lower limb amputations is varied and has grown with technological advances in prosthetics and increasing availability of adaptive sports. A systematic review by Bragaru et al. (2011) highlights swimming, athletics, and cycling as popular sports among amputees. The selection of a particular sport typically depends on the level of amputation, the individual's health status, and personal preferences. It is noteworthy that besides athletics, which enhances walking capabilities, swimming and cycling are favourable as they reduce or support the body weight independently of the lower limbs. This characteristic makes these activities particularly suitable for rehabilitating various motor impairments (Gloc et al., 2021; Marinho-Buzelli et al., 2019; Yum et al., 2021).

Cycling, in particular, is highly beneficial during the early stages of rehabilitation following amputations. The motion involved in cycling is generally simpler and less demanding than walking, making it an accessible and low-impact exercise that can be enjoyed at leisure by most individuals equipped with lower limb prostheses.

### *1.2.1. Cycling practice with lower limb prostheses*

A study by Poonsiri et al. (2021) examined the cycling habits of individuals with lower limb amputations in the Netherlands. Among the 207 participants, 141 (68%) engaged in cycling post-amputation, with 80% cycling for leisure and 74% for physical fitness. Notably, while most participants used their everyday prostheses (33%, with 4% using specialized cycling prostheses), 19.1% reported experiencing pain or discomfort during cycling, and 6.4% faced specific challenges using their regular prostheses for cycling. These data underscore the prevalent use of cycling as a form of exercise among lower limb amputees and also point to potential issues with the suitability of standard prostheses for this activity. Additionally, half of the study participants used unilateral transtibial prostheses, the same type being evaluated in this research.

Previous research has indicated that cycling can aid in regaining walking abilities, serving as a beneficial rehabilitation tool following surgeries and strokes (Yang et al., 2014), and it is recommended for patients with certain orthopaedic conditions (Yum et al., 2021). The limb movements required for cycling are fundamentally less complex than those required for walking, yet they demand a degree of dexterity, thus enabling individuals who are unable to walk to still engage in cycling (Childers et al., 2009). During cycling, the body weight is supported independently of leg movement, simplifying the motion of the lower limbs. Moreover, when cycling on an ergometer or stationary bike, maintaining balance is also more manageable. However, the application of cycling as a rehabilitation method for lower limb amputees has been scantily explored, and the adoption of cycling-specific prostheses for post-amputation rehabilitation remains relatively novel.

## 1.3. Biomechanics of cycling

To comprehend the requirements and limitations of cycling with a unilateral transtibial prosthesis, first, the biomechanics of regular cycling need to be understood. The cycling movement involves the lower limbs, utilising a bicycle to transform mechanical effort applied to the crank into kinetic energy (Turpin & Watier, 2020). This dynamic requires the coordination of mainly eight muscle groups—the tibialis anterior, soleus, gastrocnemius, vastus medialis, vastus lateralis, rectus femoris, hamstrings, and gluteus maximus—and three main joints: the hip, knee, and ankle (So et al., 2005), all of which are shown in figure 1.2. These muscles and joints interact with the bicycle, generating crank power via the pedals, with additional support from the saddle and handlebars.

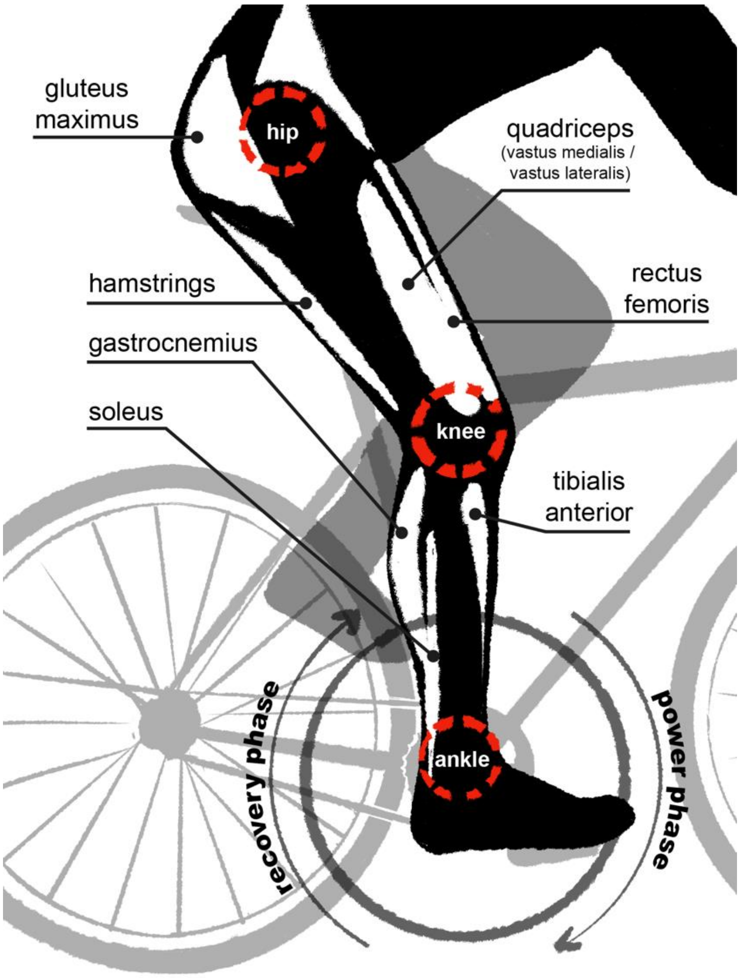


**Figure 0.2.** Muscles and joints involved in the cycling movement. The phases of the pedal stroke are also shown.

Modifications in crank length (So et al., 2005) and saddle height (Bini, 2020) significantly influence joint angles, kinematics, and metabolic costs. These adjustments also alter muscle activation patterns and affect which muscle groups are predominantly engaged during cycling. Research indicates that the hip and knee joints are responsible for approximately 40% each of the torque produced at the crank, with the ankle contributing around 15% and the upper body joints, only 5% (Elmer et al., 2011). The power generation process in cycling is distributed throughout the crank's rotation phases; the distal muscles such as the tibialis anterior, soleus, and gastrocnemius primarily provide stabilization and provide dorsiflexion and plantarflexion at the ankle. The proximal muscles, including the quadriceps, rectus femoris, hamstrings, and gluteus maximus, deliver the majority of the power during the pedal stroke. Notably, the quadriceps and gluteus maximus are the main power producers, whereas the iliopsoas plays a critical role during the recovery phase, after the pedal passes its lowest position (Childers et al., 2009).

The forces applied to the pedals vary depending on the cycling cadence and the resistance or power output. High resistance on the crank necessitates faster cadences (greater than 100 rpm) to maintain lower metabolic costs, optimizing power output at these rates. Conversely, recreational cyclists often choose lower cadences (40–80 rpm), which, while demanding less energy, require greater pedal forces to maintain consistent power output (Abbiss et al., 2009). Force application during cycling creates angled force vectors that are most intense between the 90° and 180° marks—the critical power phase of the pedal stroke (Turpin & Watier, 2020)—and are both normal and tangential in relation to the pedal.

### *1.3.1. Biomechanics of cycling with a prosthesis*

Unilateral transtibial amputations result in the loss of the ankle joint on one side of the body, affecting the associated muscle groups in the distal portion of the leg. This includes a reduction in 15% of the torque generated by the ankle, which functions as a biological spring, as well as in the muscles responsible for stabilization, namely, the tibialis anterior, soleus, and gastrocnemius. Due to the absence of ankle movement, the prosthesis used in such amputations maintains a fixed effective cycling length, leading to kinetic asymmetries that pose significant challenges for unilateral amputee cyclists (Childers et al., 2009).

Comparative studies on cycling between individuals with unilateral transtibial prostheses and those with intact limbs have identified several key factors contributing to these asymmetries. Notably, the range of motion (ROM) in the knee and hip joints is altered, exhibiting more extension at the bottom of the cycle, compared to intact cycling, due to the inability of the prosthetic foot to plantarflex (Childers et al., 2014). There is also an increased hip sway, which is the side-to-side sliding movement on the saddle, observed in unilateral amputee cyclists (Childers et al., 2009).

Further research into professional-level cycling with prostheses has explored the impact of these asymmetric kinematics on the power exerted during cycling. Findings suggest that despite the kinematic disparities, the kinetics of amputee cycling can approach levels seen in intact cycling when the kinematic asymmetries are more pronounced (Childers & Kogler, 2014). Additionally, differences in muscle activation patterns are noted in the intact leg of unilateral amputees, with a tendency to delay peak muscle activity, achieving maximum activation later in the cycle and at different angles compared to intact cyclists. This nuanced understanding highlights the complex interplay between prosthetic design and biomechanical adaptations in amputee cyclists.

Another key factor contributing to asymmetry in cycling among unilateral amputees is the reduced ability to generate and deliver power through the prosthetic limb. In addition to the loss of distal muscle groups, the capacity to produce power in the prosthetic limb is significantly diminished. Studies have demonstrated that this results in a pronounced work asymmetry, defined as the percentage difference in power contribution between the prosthetic and intact limbs over one complete pedal stroke. Childers et al. (2014) found that unilateral amputees exhibit a substantial work asymmetry, with the prosthetic limb contributing approximately 24.5% ± 10.0% less to the total power output compared to the intact limb. This disparity in power generation is a major factor contributing to the biomechanical challenges faced by amputee cyclists and emphasizes the need for more effective prosthetic designs tailored to athletic activities such as cycling.

## 1.4. The need for prosthesis for leisure cycling

As discussed, the current body of research on cycling with transtibial prostheses primarily focuses on the needs of professional athletes, leaving a notable gap concerning leisure cyclists. The lack of guidance for leisure cyclists, who often lack the close supervision of healthcare or sports professionals, can result in improper adjustments of either the bicycle or prosthesis. Such conduct may lead to both short- and long-term injuries. Furthermore, poorly adjusted equipment can create psychological barriers, causing leisure cyclists to believe they are incapable of cycling when, in fact, the issue lies in the configuration of the bicycle or prosthesis.

The available literature on professional cycling prostheses does not fully address the needs of leisure cyclists, at most, making indications of possible differences (Childers et al., 2009). Professional cycling typically involves higher resistance and load, with cyclists using higher saddle heights to maximise the use of effective muscle lengths and fully engage the ankle joint (Bini & Priego-Quesada, 2022). Leisure cyclists, on the other hand, may prioritise lower saddle heights for improved manoeuvrability and a lower centre of gravity (Burke, 2003). This lower position also makes it easier for cyclists to stop and start—especially important in parks or residential areas—by allowing them to place their feet on the ground without dismounting the saddle.

In professional cycling, the foot is typically placed on the pedal in a tip-of-the-foot position, facilitating greater dorsiflexion and plantarflexion (Bini et al., 2013). However, at the lower resistance levels common in recreational cycling, with lower saddle heights, leisure cyclists often position the middle of the foot on the pedal, still employing the ankle joint to a degree, utilising it to provide clearance the top of the cycle and to reach further down when the pedal is at the bottom position. Leisure cyclists also commonly use platform pedals, which provide a less secure connection between the foot and pedal compared to the clipless pedals used in professional cycling. However, the platform pedals better enable the act of stopping and starting the cycling movement and can improve the response speed when preventing falls, which are important factors for leisure cycling.

While there are unilateral transtibial prostheses designed for cycling, these are generally tailored for professional use, focusing on power transmission (Childers et al., 2014; Childers & Kogler, 2014; Dyer & Disley, 2020; Koutny et al., 2013). They are often stiff and connect directly to the pedal using attachment systems that maximise the power output of the intact leg. These prostheses are not suitable for walking and are custom-built to meet the specific needs of the user, typically using non-standard parts. Additionally, these prostheses may require an adapted bicycle, often featuring modifications such as crank shortening on the affected side (Koutny et al., 2013). As a result, these prostheses may not be ideal for leisure cycling, where comfort and practicality are the main priorities. Figure 1.3 outlines some possible requirements for a recreational prosthesis, which would prioritise user comfort, adaptability, and ease of use over maximum performance. These requirements would ideally be added to a regular walking prosthesis, enabling a “cycling mode” to prioritise convenience.

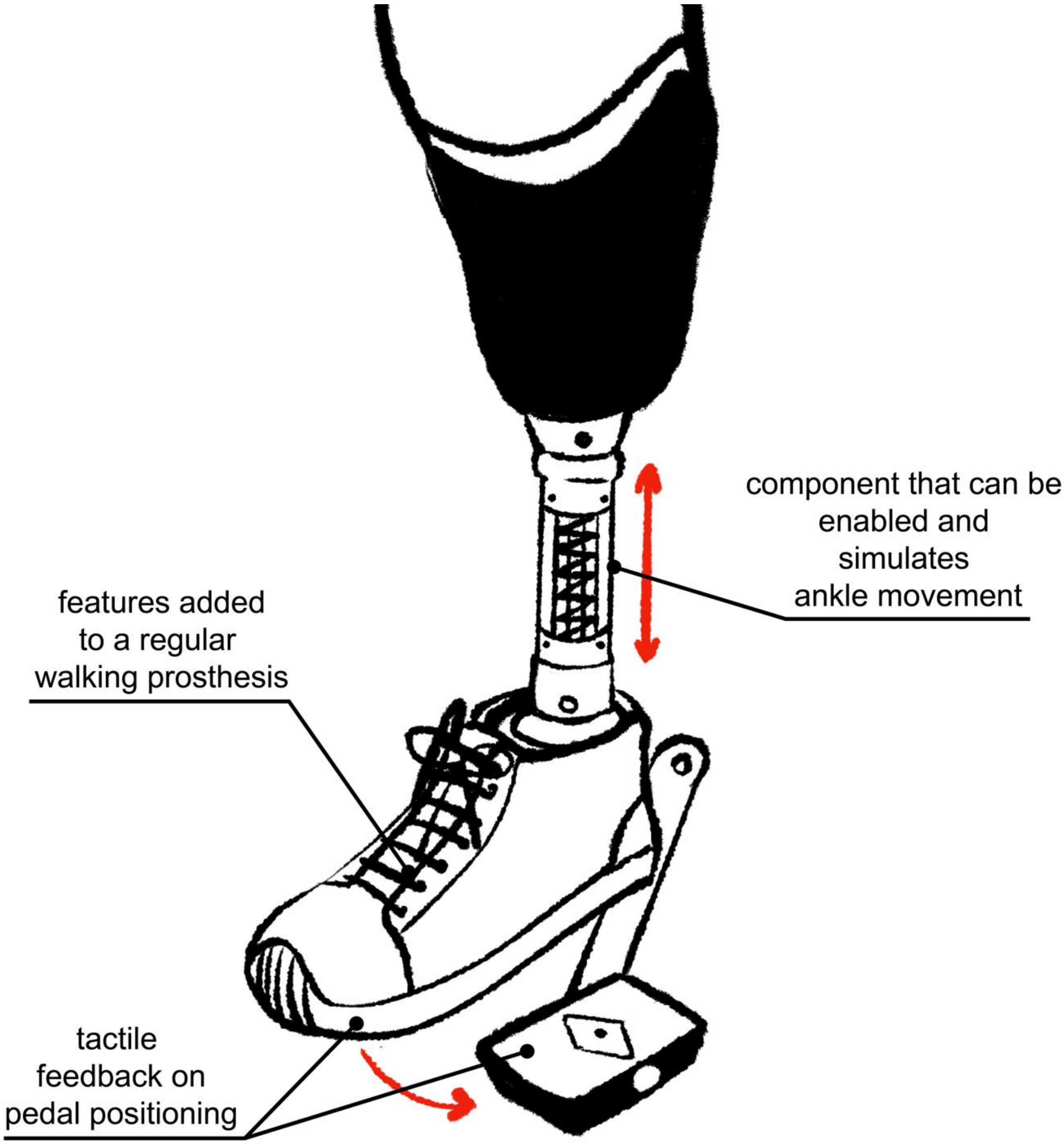


**Figure 0.3.** Prosthetic requirements for recreational use.

## 1.5. Path to the practical application of leisure cycling prostheses and its challenges

Given the differing demands between leisure and professional cycling, the design requirements for unilateral transtibial prostheses in a leisure context must be tailored accordingly. Existing research, including my own, has highlighted a few factors. For instance, my previous findings suggest that positioning the pedal in a middle-of-the-foot position is beneficial for leisure cyclists with transtibial prostheses (Seratiuk Flores et al., 2023). Additionally, research suggests that incorporating ankle movement into prostheses can enhance comfort, as it more closely mimics the natural movement of a biological limb (Childers et al., 2009; Tiele et al., 2020). Since leisure cycling prostheses do not need to prioritise full power transmission from the residual limb to the pedal, some energy could be used to deform a spring or another energy-storing mechanism to simulate ankle movement, benefitting the cycling experience.

Other adaptive equipment is already commercially available, such as low-entry bicycles designed for individuals with reduced mobility. These bicycles make mounting and dismounting easier, especially when balance is a concern. Magnetic pedals are another useful innovation (*MagLOCK Bike Pedal - Magnetic Bike Pedals*, n.d.), offering feedback to prosthesis users regarding the position of their foot on the pedal, which enhances control and reduces the risk of accidents.

Nevertheless, leisure cycling with prostheses remains underexamined in several aspects. The interplay among bicycle design, rider characteristics, prosthetic components, and environmental conditions can impose substantial barriers to cycling for individuals with amputations. Figure 1.4 outlines potential areas of concern, highlighting the specific topics addressed in this thesis.

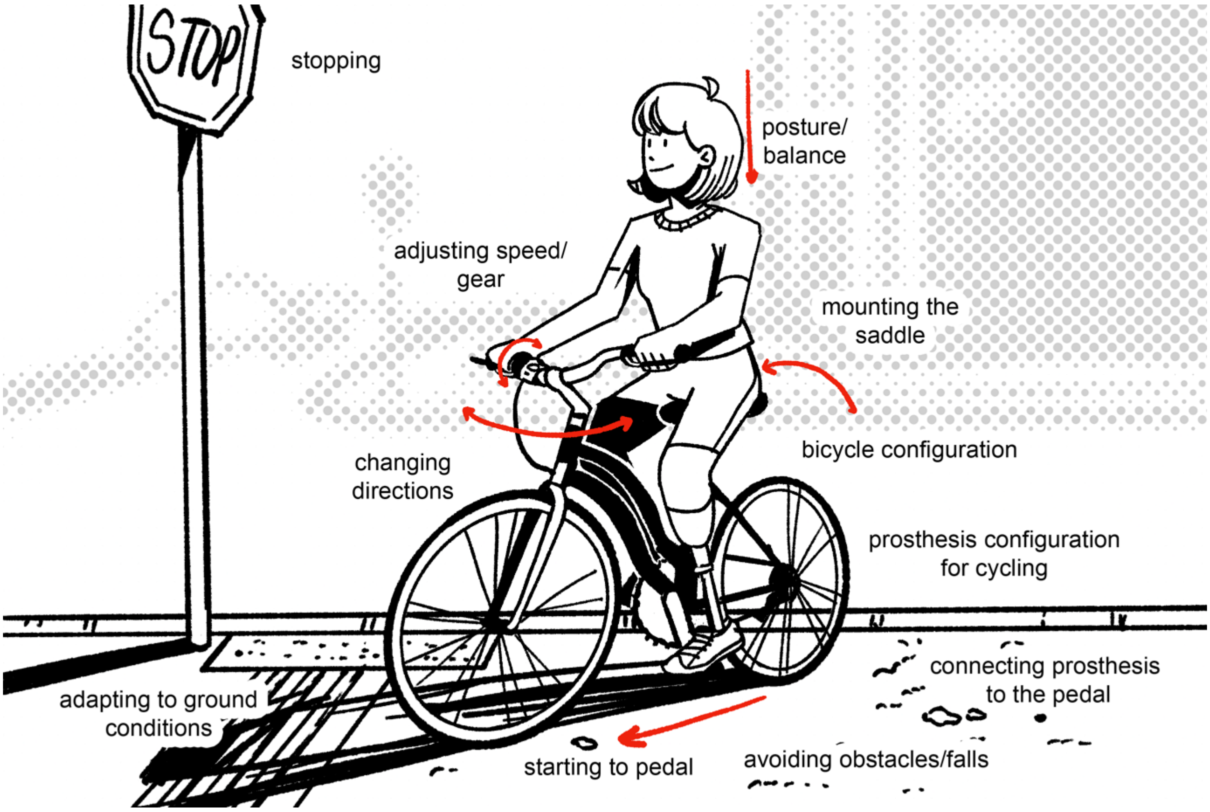


**Figure 0.4.** Potential research areas in leisure cycle prostheses. Highlighted topics will be explored in this thesis.

Pertaining to the highlighted bicycle configuration, saddle height adjustment is a component that could be explored, as it directly affects the movement of the hip and knee, both of which are influenced by the use of a prosthesis. Improper saddle height can lead to balance difficulties and potential injuries. In professional cycling, unilateral crank shortening is often employed, and adjusting the saddle height in leisure cycling could simulate similar biomechanical effects, helping users achieve a more comfortable and efficient ride.

Another underexamined area is the issue of balance when using a prosthesis while cycling. Interviews conducted with amputees frequently reveal anxiety about maintaining balance on a bicycle. One study (Poonsiri et al., 2019) found that 14.7% of amputees who did not cycle cited fear of injury as their primary barrier to taking up the activity, marking it as the most significant concern in the study. Similarly, another study (Poonsiri et al., 2021) showed that 32% of participants with prostheses refrained from cycling because they considered their bicycle or prosthesis unsuitable for the activity. These findings highlight the need for further research and development in this area, ensuring that leisure cyclists with prostheses have access to equipment that meets their physical and psychological needs, while also addressing the barriers to participation in cycling. The already explored areas and those that will be addressed in this thesis are shown in figure 1.4.

## 1.6. Aim

Considering the apprehensions users have regarding adapted equipment and the fear of injuries, this research aims to address two critical aspects that could significantly impact user experience: saddle height adjustments and balancing challenges. Additionally, to further the scope of existing studies, this research also explores the viability of incorporating ankle movement into prosthetic designs.

To standardise results and facilitate direct comparisons between individuals with and without amputations, the studies described in Chapters 2 and 4 employed participants without amputations using orthoses to simulate the conditions of prosthetic use. This method ensures a uniform level of simulated amputation across participants, allowing for a controlled assessment of prosthetic functionality versus results in able-bodied individuals.

### *1.6.1. Thesis outline*

***Chapter 2 – Biomechanical Effects of Saddle Height Changes in Leisure Cycling with Unilateral Transtibial Prostheses***

One of the simplest alterations that can be made to cycling equipment is to the saddle height of the bicycle. This study evaluated biomechanical variations under five different saddle heights. I utilised joint movement, EMG, and instrumented pedal data to compare simulated prosthetic conditions, while also comparing results to intact cycling.

*Chapter 3 – Development of a Transtibial Prosthesis Employing Compression Springs in the Ankle Component*

In this study, I compared the biomechanical effects of a prototype prosthesis featuring different levels of ankle mobility. The study involved two participants with traumatic and congenital amputations, assessing the use of three compression springs with varying spring constants in an ankle component installed in a prototype prosthesis. Results obtained with the prototype were compared with those from participants using their standard walking prostheses.

*Chapter 4 – Evaluation of Balancing Strategies while Cycling with Unilateral Transtibial Prostheses*

This investigation focused on the strategies developed by participants to maintain balance while cycling on rollers using a cross bicycle. By comparing regular cycling results to those obtained under simulated prosthetic conditions, in this study I analysed the movement of the points of contact between the wheel and roller, changes in the trunk and bicycle angle, and accelerometer results.

# Chapter 2
# Biomechanical Effects of Saddle Height Changes in Leisure Cycling with Unilateral Transtibial Prostheses: A Simulated Study

## 2.1. Introduction

For bicycle configurations, the guidelines for non-amputees may be applicable to lower-limb amputees without comorbidities (Childers et al., 2009). However, cycling movement is altered in individuals with a prosthesis primarily because of the absence of ankle joint movement, which is typically rigid in walking prostheses and absent in professional-level cycling prostheses. This difference can cause asymmetries, making cycling uncomfortable (Childers et al., 2009). When using a unilateral prosthesis, the leg with the prosthesis must extend further to the bottom pedal position to compensate for the lack of ankle movement. Meanwhile, in the highest pedal position, the knee and hip joints may flex more compared with an intact leg because of the absence of ankle dorsiflexion (Pierson-Carey et al., 1997). These exaggerated joint movements lead to asymmetry between the intact and prosthetic limbs, potentially causing other musculoskeletal disorders (Devan et al., 2014; Morgenroth et al., 2012).

Previous studies have explored mitigating this asymmetry by altering the crank length on the affected side (Koutny et al., 2013), which reportedly increases the comfort of prosthesis-wearing cyclists unfamiliar with cycling (Childers et al., 2009). However, bicycle cranks are only commercially available at specific lengths, which are not highly customizable and may not fit individual anthropometric dimensions. One way to emulate crank shortening without special parts is to alter the saddle height, allowing for more precise adjustments. In settings such as elite cycling, athletes often wear cycling-specific prostheses, and use a higher saddle height, compatible with able-bodied elite cycling, while possibly utilizing the shorter crank on the affected side (Koutny et al., 2013). This higher saddle height enables more effective use of muscle lengths and joint power (Bini et al., 2014; Robergs et al., 2005); therefore, the lower limbs reach an ideal setting for power production, at the cost of kinematic symmetry, inferred by the unilateral shorter crank and rigid cycling prostheses. In contrast, leisure cycling often employs suboptimal heights. Leisure cycling involves shorter activity periods, lower resistance, and slower cadences, thus minimizing the risk of acute and overuse injuries (Bini & Priego-Quesada, 2022). Additionally, leisure cyclists tend to prefer lower saddle heights, which can improve manoeuvrability and stability (Burke, 2003).

In the context of unilateral prostheses, the lower saddle heights can minimize the use of the ankle joint on the affected side; thus, making its kinematics more similar to that at the prosthetic side, which often has a rigid ankle joint. This would come at the cost of reduced power production ability; however, leisure cycling does not reach the same levels of power requirements as elite cycling. Potentially, the lower saddle height could lead to increased comfort stemming from the improved movement symmetry. Therefore, it can be theorized that leisure cycling with unilateral transtibial prostheses would benefit from either the higher saddle height employed in elite cycling, or lower saddle height already commonly used in leisure cycling.

This study aimed to evaluate the biomechanical effects of leisure cycling at different saddle heights using a simulated unilateral transtibial prosthesis. To achieve this, biomechanical data were collected between affected and unaffected legs, pertaining kinetic and kinematic aspects. Parameters involved the muscle activity of the gastrocnemius medialis, vastus medialis, and semitendinosus, joint movement of hip and knee, symmetry of force applied to the pedals, as well as torque effectiveness and pedal smoothness, and subjective feedback on perceived exertion and comfort. To elucidate the effects of the different saddle heights, results were compared between simulated prosthetic cycling conditions and a control, intact cycling condition.

## 2.2. Methods

### *2.2.1. Participants*

Ten female (n = 6; age: 27.5 ± 2.7 years; height: 157.6 ± 1.9 cm; weight: 59.0 ± 10.8 kg) and male (n = 4; age: 28.3 ± 3.6 years; height: 167.7 ± 2.7 cm; weight: 67.0 ± 8.4 kg) able-bodied participants were recruited. The inclusion criteria consisted of being between the ages of 25 and 35 years and practicing leisure or commuting cycling with at least monthly regularity. Using cycling as physical training or sports practice and presenting any condition that could affect the practice of cycling or could worsen during the experiment were exclusion criteria. In addition, the participant height was limited to 172 cm because of experimental setting limitations on saddle height variance.

Participants answered a health and bicycle-riding habit questionnaire that enforced the exclusion criteria, and provided written informed consent to participate in the study. Leg dominance was assessed through asking the question "if you were to kick a ball, which leg would you use" (van Melick et al., 2017), with the majority of participants reporting right-leg dominance (n = 9). Anthropometric data was then collected, including the length of the right leg inseam (75.1 ± 2.4 cm) and foot length (23.7 ± 1.1 cm).

This study was approved by the Ethics Committee of the Faculty of Design, Kyushu University (approval number 532).

### 2.2.2. Simulated prosthesis condition

To collect data comparable to intact cycling and standardize the results, participants without amputations were recruited for the experiment. The participants wore custom orthoses (Arizono Orthopedic Supplies Co., Ltd., Kitatkyushu, Japan) simulating prosthetic conditions, which prevented the biological foot from contacting the pedal and ankle from exerting force throughout the crank cycle. The inability to contact the pedal also simulated the lack of tactile feedback, experienced by prosthesis users. This simulated prosthetic condition allows for the isolation of ankle-movement factors and minimizes possible influence of amputation and disability levels. Furthermore, it also allows a better evaluation of the adaptation to the prosthesis. The orthoses were worn on the right side and attached below the knee, designating the right leg as the affected leg (AL) and left leg as the unaffected leg (UL). The orthoses featured an aluminium strut connected directly to the pedal via a cleat, aligning the foot in a mid-foot position. Previous research suggests that this position can help mitigate the asymmetry between intact and amputated limbs (Seratiuk Flores et al., 2023). After the donning of the orthosis, participants were given up to 5 min to practice cycling with it.

### 2.2.3. Saddle height setting

Based on previous cycling research, the ideal saddle height is achieved when the knee is at 25–35° extension with the crank at the 180° position (bottom of the cycle) (de Vey Mestdagh, 1998; Robergs et al., 2005). This setting optimizes power output and muscle performance and is suitable for physical training or competitive cycling. However, for leisure cycling, where the power output is lower, cyclists may prefer a lower saddle height. In this experiment, the standard saddle height was set to achieve a 45° knee angle at the bottom of the cycle, measured statically with a goniometer and corresponding to 46° ± 3° when assessed dynamically.

#### 2.2.3.1. Experimental conditions

The different heights of the saddle to be evaluated were measured from the top of the saddle to the crank axis and set as variants of 3.5% (2.31 ± 0.07 cm) of the standard height, based on previous research (Bini, 2020). Therefore, the conditions for the experiment were defined as (Fig. 2.1):

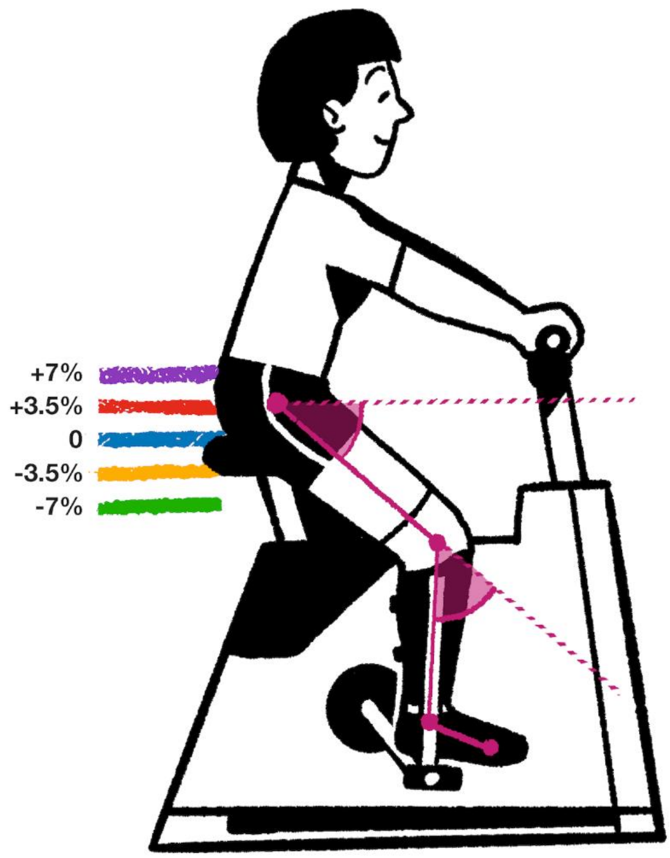


**Figure 2.1.** Graphical representation of the saddle heights used during the experiment (+7%, +3.5%, 0, -3.5%, and -7%), tracking points, and calculated angles.

**Control:** Regular cycling without orthosis.

**0:** Employing the orthosis, participants cycled at the same saddle height as in the control condition.

**-7%, -3.5, +3.5, and +7%:** Saddle height variations calculated over the control/0 height, performed while wearing unilateral orthosis.

### *2.2.4. Experimental setup*

Experiments were conducted in an air-conditioned room kept at approximately 22°C. The participants cycled on an AeroBike 75XL ergometer (Konami Sports Co., Ltd., Kanagawa, Japan) with a crank length of 170 mm (Fig. 2.2). The ergometer was modified to include Assioma Duo power-meter pedals (Favero Electronics Srl., Arcade, Italy). The handlebar height was set to 100 mm above the saddle height in all the conditions.

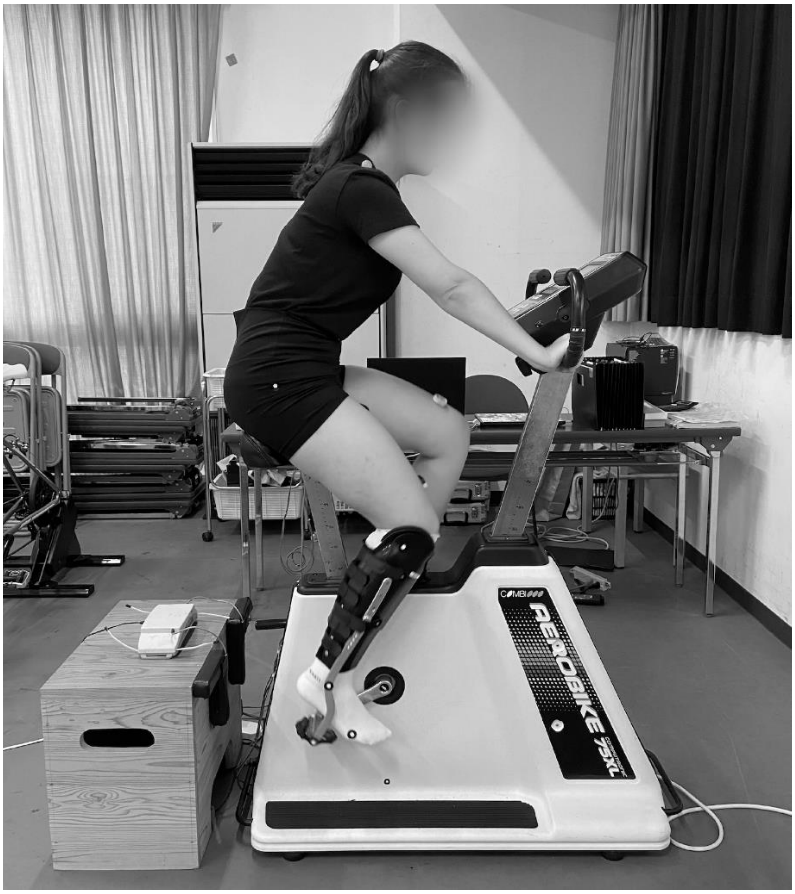


**Figure 2.2.** Experimental environment.

Clipless pedals were equipped with three-dimensional (3D)-printed attachments to function as platform pedals, complemented by thermoplastic fastening to limit the forward foot position. This limiter prevents upward pedal pulling, which is possible with clipless pedals. The 3D-printed attachments were used on both sides in the control condition and only on the unaffected side in subsequent trials. The orthosis was directly connected to the pedal via a cleat.

### *2.2.5. Experimental protocol*

For the experiment, the participants wore tight-fitting clothing (t-shirt and shorts) and ankle-covering socks. Laceless MW100 trainers (New Balance Athletics Inc., Boston, MA, USA) of appropriate sizes were provided. These trainers were used during the control condition and only on the unaffected side during the orthosis conditions. Cycling was performed with 40 W resistance at 60 rpm, simulating everyday leisure cycling. A metronome ensured consistent cadence.

Each experimental trial lasted 2 min, with three replicates for each of the six conditions. The conditions were randomized except for the control condition, which was always performed first. Breaks of 30 s were enforced between trials, and of 3 min between conditions. The first trial of each condition served as practice, with data being collected during the latter two trials. The first and last 30 s of each trial were discarded, and data were analysed for 10 consecutive crank revolutions.

### *2.2.6. Measurements*

#### 2.2.6.1. Joint angles

Two-dimensional motion analysis was conducted using a Panasonic HC-300M camera (Panasonic Co., Osaka, Japan) which recorded at 30 frames per second and 1080p resolution.

The camera was positioned 3.8 m from the ergometer on the participant's affected (right) side. A secondary camera (HDR-CX560V; Sony Co., Tokyo, Japan) with identical capture settings was placed 1 m from the unaffected side for data verification. Four colour-contrasting self-adhesive markers were symmetrically placed on anatomical landmarks: the greater trochanter, lateral femoral epicondyle, lateral malleolus, and fifth metatarsal head, corresponding to the hip, knee, ankle, and foot, respectively. Ankle joint movement was collected for monitoring reasons and was not used in the study. An additional marker was placed on the pedal to segment the data for each complete cycle.

Video footage was processed using Adobe Premiere Pro 15.4 (Adobe Inc., San Jose, CA, USA). Motion analysis was performed using motion analysis software (Yeoh & Seratiuk Flores, 2023/2024) that employed a Discriminative Correlation Filter Tracker with Channel and Spatial Reliability (Lukežič et al., 2018) algorithm for automatic motion tracking. Each frame was verified manually. The angles were calculated using software, as illustrated in figure 2.1.

#### 2.2.6.2. Muscle activity

Muscle activity was recorded using a telemeter surface electromyography (EMG) system (WEB7000; NIHON Kohden Co., Tokyo, Japan) at 1 kHz. The signal was internally bandpass-filtered (15–500 Hz) and rectified. Following the Surface Electromyography for the Non-Invasive Assessment of Muscles guidelines (Hermens et al., 2000), four electrodes were placed bilaterally on the proximal portion of the legs to monitor the vastus medialis and semitendinosus muscles. An additional electrode was placed on the gastrocnemius medialis muscle of the UL. These muscles were specifically chosen based on previous research, with the vastus medialis, semitendinosus, and gastrocnemius showing stronger changes in activity under different unilateral prosthesis cycling conditions (Childers et al., 2009; Childers et al., 2011; Koutny et al., 2013; Seratiuk Flores et al., 2023). A direct-current signal sampled wirelessly at 1 kHz was correlated with a Hall sensor designating the crank orientation. After collection, an 8 Hz low-pass linear envelope filter was applied to smooth the rectified EMG signal (Burden et al., 2003), reflecting the magnitude of muscle activity. The data were normalized to the control condition and segmented into cycles using Hall sensor data.

**2.2.6.3. Instrumented pedals**

Assioma Duo instrumented pedals were installed bilaterally on the ergometer to assess the force applied to the pedals (Bini & Hume, 2014). Data were recorded on a computer using an ANT+ receptor at 1 Hz using GoldenCheetah ver. 3.6 software. The parameters measured included cadence and power, which were used to monitor the participants' adherence to target power settings and cadence. Left and right balance assessed the contribution of each leg to a full revolution of the crank, with values closer to 0 indicating a higher contribution from the right leg and those closer to 100 indicating that the left leg was more prominently used. The torque effectiveness measures the influence of the force applied to the pedals on the resulting force vectors, with values close to 100 indicating the greatest effectiveness. Pedal smoothness was calculated as the constant application of power to the pedals throughout the cycle, with constant and uniform power delivery achieved at 100. Torque effectiveness and pedal smoothness were measured independently on each side.

**2.2.6.4. Subjective evaluation**

Thirty seconds prior to the conclusion of each trial, the participants were asked to assess their perceived exertion using Borg's 6-20 RPE scale (Borg, 1970), which was visibly positioned on a wall in front of the ergometer. During the practice trial for each condition, the participants were instructed to evaluate their comfort level in comparison to that during the immediately preceding condition and indicate the height at which they were most comfortable at that point. Additionally, during the control trial, participants who owned a bicycle ($n = 7$) were questioned whether the standard saddle height resembled that of their regular bicycle, with four participants stating that it was the same, two indicating that it was lower, and one reporting that it was higher.

### *2.2.7. Statistical analysis*

Thirty seconds prior to the conclusion of each trial, the participants were asked to assess their perceived exertion using Borg's 6-20 RPE scale (Borg, 1970), which was visibly positioned on a wall in front of the ergometer. During the practice trial for each condition, the participants were instructed to evaluate their comfort level in comparison to that during the immediately preceding condition and indicate the height at which they were most comfortable at that point. Additionally, during the control trial, participants who owned a bicycle ($n = 7$) were questioned whether the standard saddle height resembled that of their regular bicycle, with four participants stating that it was the same, two indicating that it was lower, and one reporting that it was higher.

## 2.3. Results

### *2.3.1. Joint angles*

#### 2.3.1.1. Knee

The knee joint angles are shown in figure 2.3. The -7% saddle position exhibited joint movement closest to the control. ANOVA revealed a significant effect of saddle height on the knee joint angles throughout the cycle (0° = $F_{[4,36]} = 231.92$, $p < 0.001$; 90° = $F_{[4,36]} = 233.51$, $p < 0.001$; 180° = $F_{[1.88,16.90]} = 314.88$, $p < 0.001$; 270° = $F_{[4,36]} = 408.86$, $p < 0.001$). Pairwise comparisons indicated significant differences between all saddle heights throughout the crank rotation ($p < 0.05$). The +7% height differed the most from the control and exhibited the highest extension at the 180° crank position, and the -7% height showed the most flexion throughout, comparable to the intact cycling levels. Dunnett's test revealed significant differences between the control condition and the -3.5%, 0, +3.5%, and +7% conditions throughout the cycle, all of which showed more extension than that of the control, with only the -7% condition showing no significant difference from the intact cycling levels (Appendix 1).

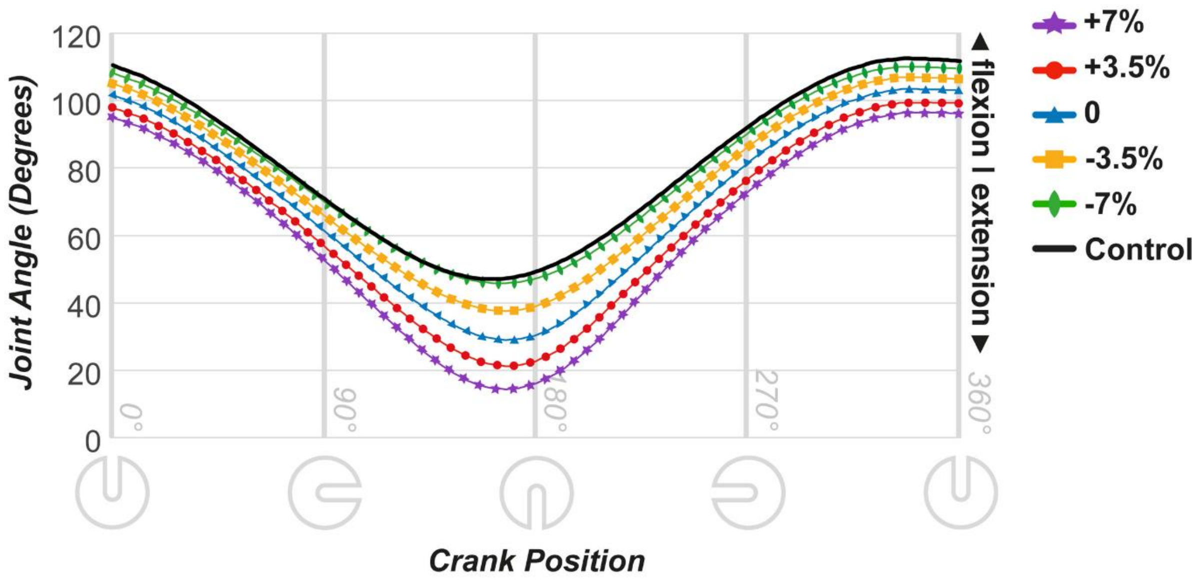


**Figure 2.3.** Joint movement data throughout the cycle in the affected leg for the knee joint.

#### 2.3.1.2. Hip

Figure 2.4 shows the hip joint movement. In the 0° crank position, hip flexion was closer to the control values at +7% saddle height. When the crank was at the bottom dead centre position (180°), -3.5% height movement became the closest to control. Similar to the knee joint mean angles, the ANOVA results indicated a main effect of saddle height across all measured crank angles (0° = $F[4,36] = 337.25$, $p < 0.001$; 90° = $F[4,36] = 410.98$, $p < 0.001$; 180° = $F[1.46,13.18] = 302.17$, $p < 0.001$; 270° = $F[4,36] = 259.95$, $p < 0.001$). Post-hoc analyses revealed that for the whole cycle, differences between the saddle height means were significant ($p < 0.01$), with the +7% height exhibiting the most extension throughout, peaking at the 180° crank position, and the -7% height showing the most flexion. Dunnett's test results (Appendix 1) revealed that at the start of the cycle, saddle heights of -7%, -3.5%, 0, and +3.5% significantly differed from the control, with +7% showing flexion levels similar to those of intact cycling. At 180° (lowest pedal position), -7%, 0, +3.5%, and +7% heights differed significantly, whereas -3.5% closely matched control extension levels. At 90° and 270°, which are flexion-extension transition points, all heights significantly deviated from the control. Overall, all saddle heights exhibited significant differences at specific cycle points, with the +7% height being closest to the control in flexion at the start and end, and -3.5% height aligning at mid-cycle during peak extension.

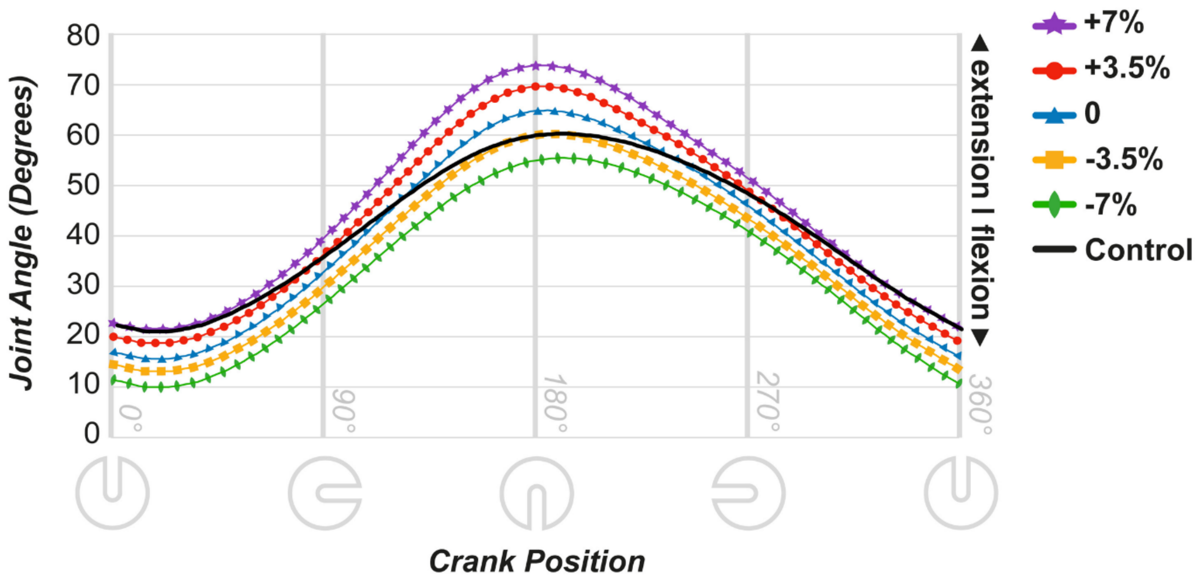


**Figure 2.4.** Joint movement data throughout the cycle in the affected leg for the hip joint.

### *2.3.2. Muscle activity*

EMG data were calculated as % variation over the control condition and are shown as means in figure 2.5. These data are plotted in a circular graph, as shown in figure 2.6. The dotted line in the centre of the graph represents the mean muscle activity in the control condition. Lines correlating with the conditions represent a percentage increase or decrease over the control. The degrees of circumference in the graph correspond to the crank position, and muscle activity refers to the represented positions. ANOVA was conducted with the overall mean muscle activity calculated over each cycle.

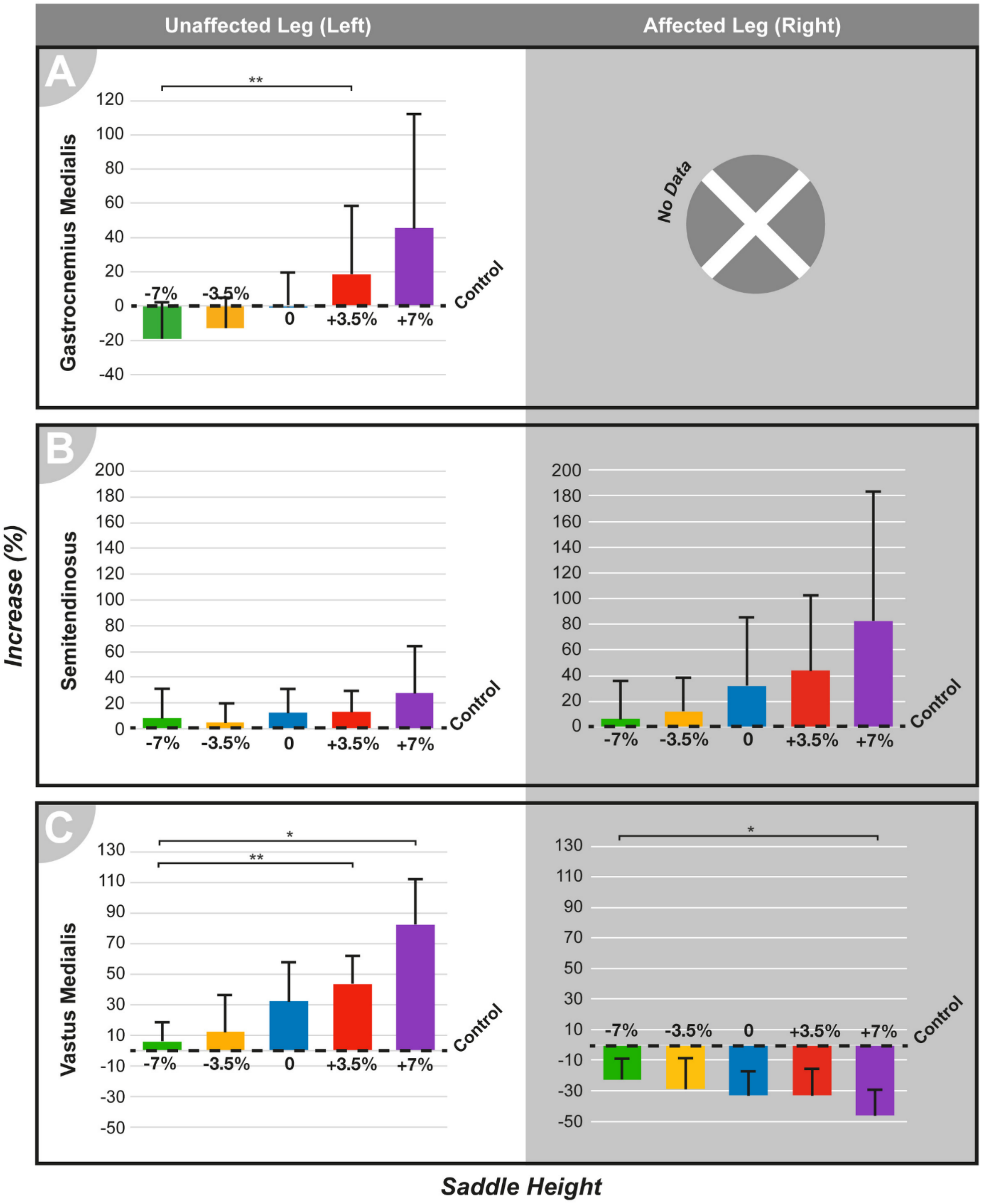


**Figure 2.5.** Mean muscle activity for all seat heights in both the unaffected and affected legs. Control condition is set as baseline (0) and is indicated. (A) gastrocnemius medialis; (B) semitendinosus; (C) vastus medialis. $^{**}p < .01$, $^{*}p < .05$.

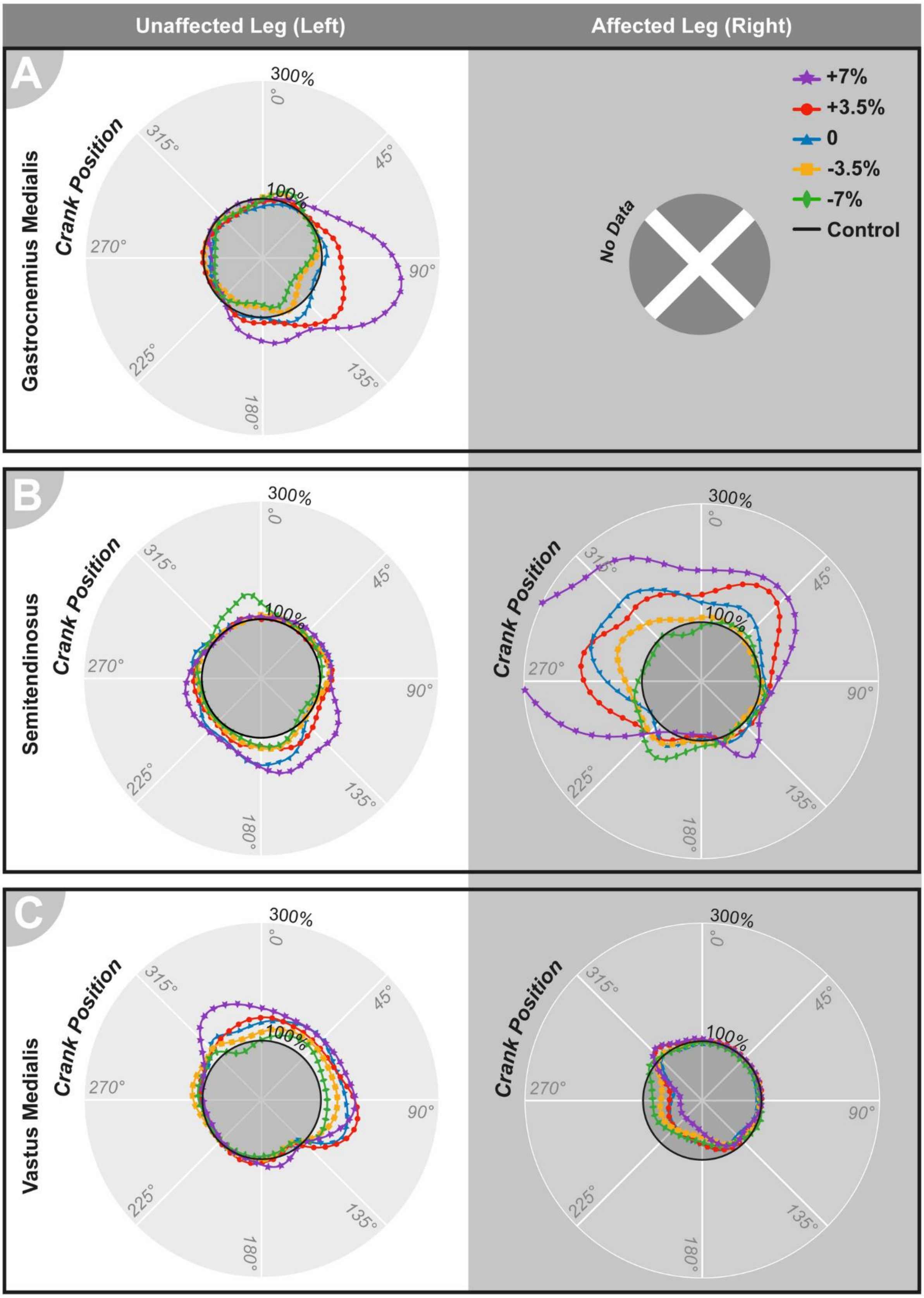


**Figure 2.6.** Mean muscle activity in the unaffected and affected legs in relation to the crank angle. (A) gastrocnemius medialis; (B) semitendinosus; (C) vastus medialis.

#### 2.3.2.1. Gastrocnemius medialis

Mean results for the gastrocnemius medialis (Fig. 2.6A) in the UL indicated that altering the saddle height from 180° to 360° had minimal impact on muscle activity. However, between 90° and 135°, a significant increase in muscle activity was observed for higher saddle heights of +3.5% and +7%, with average increases of 19% and 46%, respectively, throughout the cycle. Visual inspection showed that the 0% and -3.5% heights were closer to those of the control. ANOVA for the overall means (Fig. 2.5A) indicated a significant effect of saddle height ($F [4,36] = 7.26$, $p < 0.001$). The +7% height showed the greatest overall increase, with the +3.5% height significantly differing from -7% ($p < 0.01$, 95% CI = 0.08, 0.38).

#### 2.3.2.2. Semitendinosus

In the UL, muscle activity levels for the semitendinosus (Fig. 2.6B) showed no significant alteration under different saddle heights, as corroborated by ANOVA (Fig. 2.5B): no main effect of saddle height was found. The results were different in the AL, with most saddle heights leading to increased muscle activity (6–83%), especially from 225° to 90°. The +7% saddle height showed the greatest increase, with +3.5% and 0% presenting similar levels of muscle activity above the control condition. Heights -7% and -3.5% showed levels closer to the control. Similar to the UL, the ANOVA results indicated main effect of saddle height ($F [2.12,19.09] = 4.36$, $p < 0.01$), but the Bonferroni-corrected pairwise comparisons revealed no significant differences between the height conditions.

#### 2.3.2.3. Vastus medialis

Muscle activity levels in the UL were increased in the vastus medialis (Fig. 2.6C) in all saddle height conditions (9–40%), with the highest levels of increase observed in the +7% condition. The most reduced saddle height, -7%, presented results closest to control. The main effect of the saddle height (Fig. 2.5C) was found through ANOVA ($F [4,36] = 7.26$, $p <$

0.001), and pairwise comparisons revealed a significant difference between the heights -7%, which showed levels closest to those of the control, and both +7% ($p < 0.05$, 95% CI = -0.62, -0.01) and +3.5% ($p < 0.01$, 95% CI = -0.38, -0.08), which presented the strongest increase. The AL showed an overall decrease in the mean muscle activity (-46–-23%), more notably occurring between 135 and 315°. The +7% height results deviated the most from those of the control group, showing the greatest decrease. Statistical analysis showed a significant effect of saddle height ($F [4,36] = 7.87$, $p < 0.001$), with a significant difference between the -7% height, which was the closest to the control, and the +7% height, which showed the most reduced levels of muscle activity ($p < 0.05$, 95% CI = 0.04, 0.44).

### *2.3.3. Instrumented pedals*

Figure 2.7 illustrates the mean percentages of the left and right balance. Figure 2.8 shows the mean percentages of torque effectiveness, and pedal smoothness. For left and right balance (Fig. 2.7), ANOVA revealed a significant effect of saddle height ($F [4,36] = 16.13$, $p < 0.001$). The +7% height exhibited the most asymmetry, significantly differing from all other saddle heights (-7% = $p < 0.01$, 95% CI = 10.24, 32.01; -3.5% = $p < 0.01$, 95% CI = 4.63, 28.50; 0 = $p < 0.01$, 95% CI = 3.50, 17.06, and +3.5% = $p < 0.05$, 95% CI = 0.39, 19.65), which were more symmetrical. However, Dunnett's test results (Appendix 1) showed that all heights were significantly different from the control, which exhibited the most symmetry.

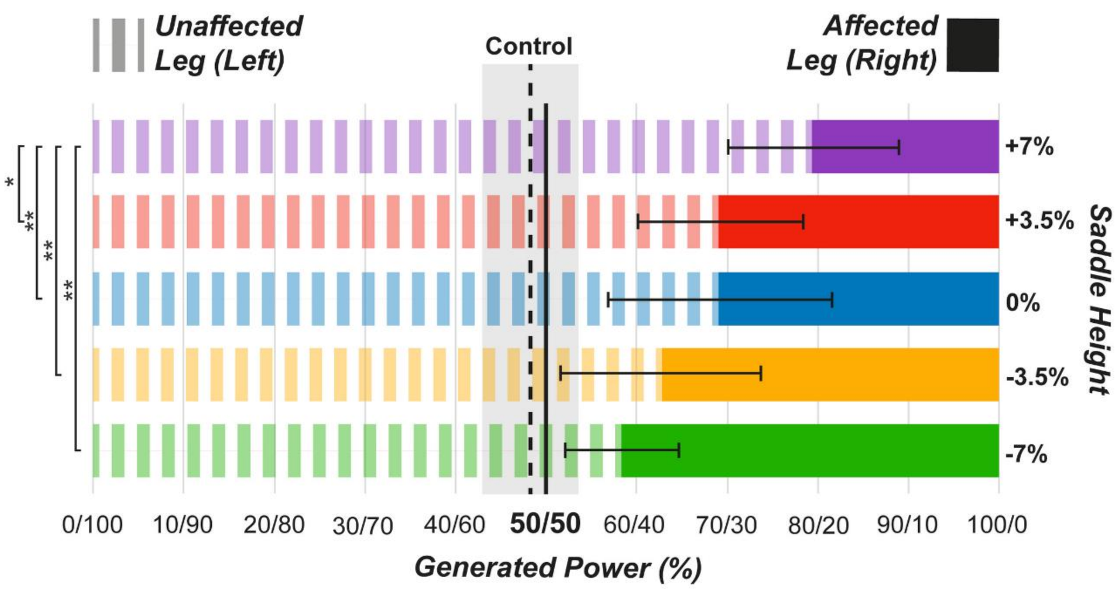


**Figure 2.7.** Generated power percentage for left (unaffected leg) and right (affected leg) balance. ***p < .001, **p < .01, *p < .05.

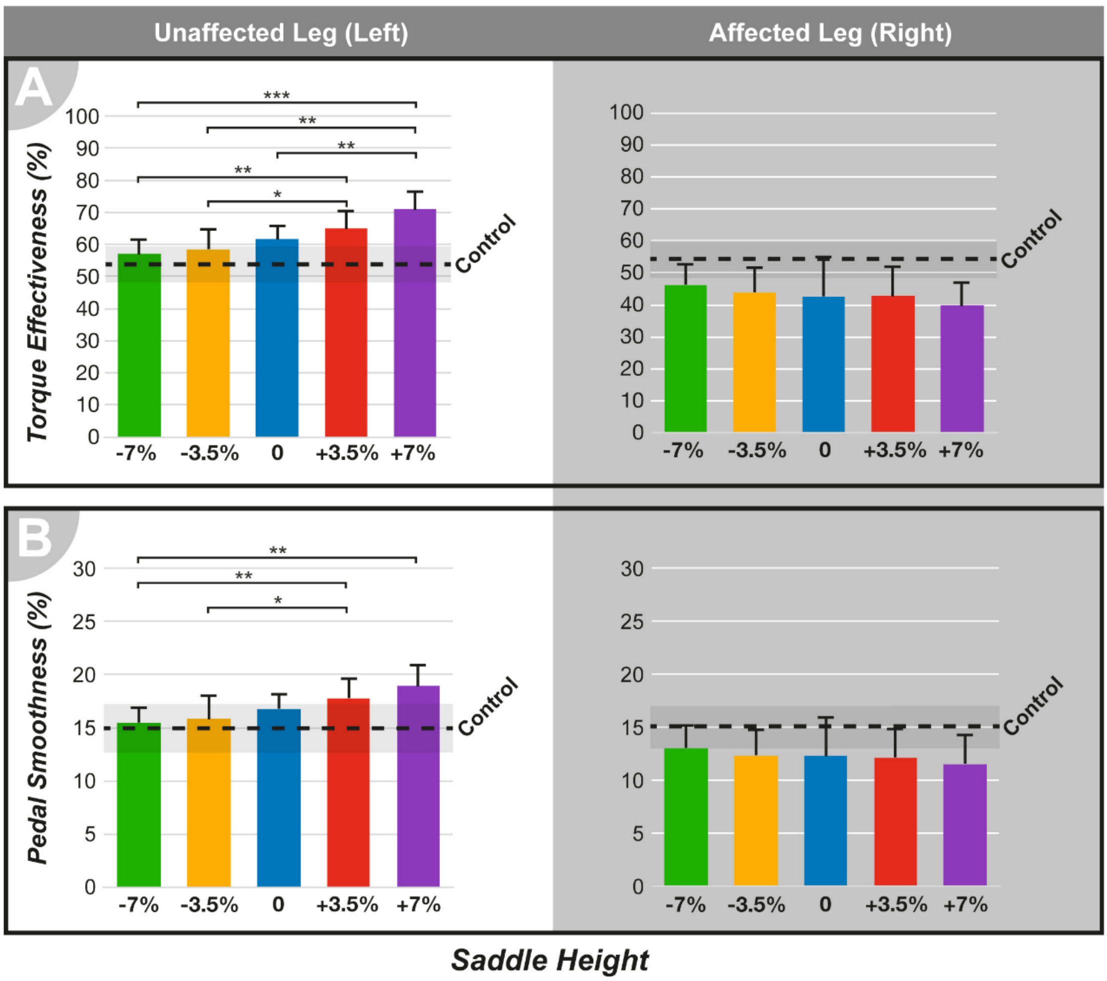


**Figure 2.8.** Instrumented pedal data means for the unaffected and affected legs. (A) torque effectiveness, (B) pedal smoothness. ***p < .001, **p < .01, *p < .05.

In the UL, the mean torque effectiveness (Fig. 2.8A) was increased with saddle height. Statistical analysis indicated significant differences between the groups ($F [4,36] = 23.74$, $p < 0.001$). While all heights showed increased torque effectiveness, the +7% condition deviated the most from the control, presenting the highest increase and significantly differing from -7% ($p < 0.001$, 95% CI = 7.20, 20.44), -3.5% ($p < 0.01$, 95% CI = 4.23, 20.57), and 0 ($p < 0.01$, 95% CI = 3.85, 14.67). This trend extended to the +3.5% condition, which also significantly differed from -7% ($p < 0.01$, 95% CI = 2.94, 12.70) and -3.5% ($p < 0.05$, 95% CI = 0.74, 12.04). The AL showed no main effect of saddle height. In the UL, Dunnett's test (Appendix 1) showed no significant differences between the -7% saddle height and control. In the AL, all heights showed results that were significantly different and lower than those of the control.

For pedal smoothness (Fig. 2.8B), ANOVA indicated a main effect of saddle height on the UL side ($F [4,36] = 12.78$, $p < 0.001$). The +7% height, similar to the torque effectiveness, showed the strongest increase in pedal smoothness and was significantly different from the reduced saddle height -7% ($p < 0.01$, 95% CI = 1.47, 5.39), which had percentages closer to those of the control. The +3.5% height also showed significant differences between the -7% ($p < 0.01$, 95% CI = 0.83, 3.69) and -3.5% ($p < 0.05$, 95% CI = 0.08, 3.74) heights, with the -7% height showing overall lower levels of pedal smoothness. In the AL, no main effect of saddle height was found. Dunnett's test (Appendix 1) showed no significant differences between the -7% and -3.5% saddle heights and control in the UL. For the AL, results for all heights significantly differed from the control, showing lower levels of pedal smoothness.

### *2.3.4. Subjective evaluation*

When asked to subjectively choose the most comfortable saddle height, one participant chose the -7.5% height, and another three chose the standard 0 height. One participant chose a +3% increase in height. Most participants (n = 5) elected the -3.5% height as the most comfortable.

Perceived exertion was also evaluated using a 6–20 points Borg scale, the means of which are shown in figure 2.9. ANOVA showed the main effect of the saddle height (F [4,36] = 10.71, $p < 0.001$), with post-hoc analyses showing significant differences between the height that was perceived as the most strenuous, +7%, and the heights -3.5% ($p < 0.05$, 95% CI = 0.33, 4.40) and 0 ($p < 0.05$, 95% CI = 0.40, 4.20). Additionally, Dunnett's test (Appendix 1) showed significant differences between all experimental conditions and the control condition, all of which presented higher scores than control.

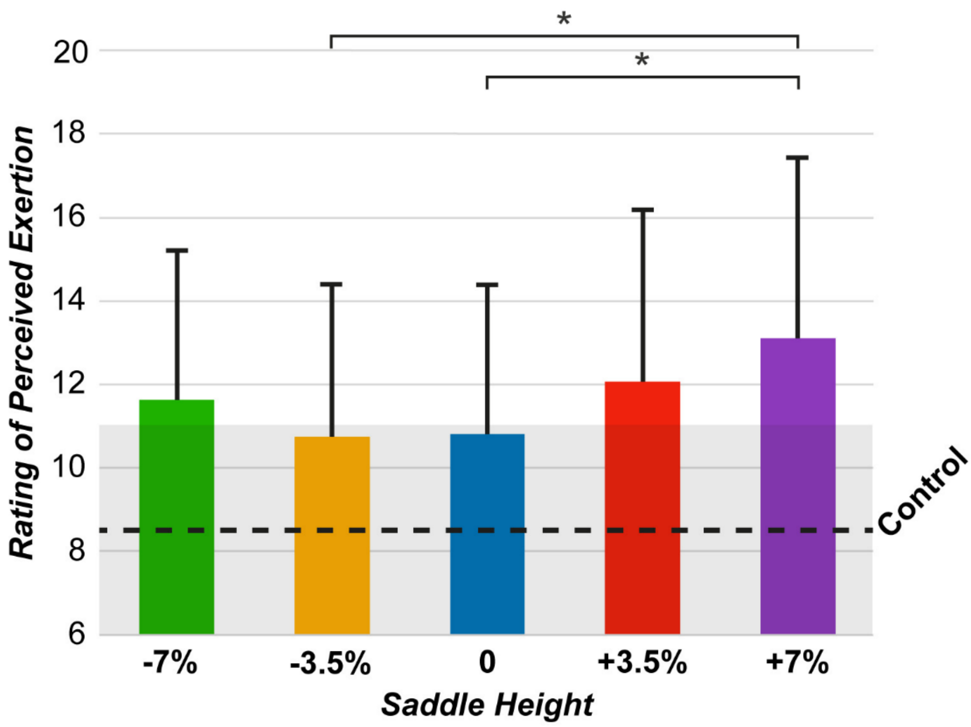


**Figure 2.9.** Mean rating of perceived exertion. *p < .05.

## 2.4. Discussion

This study assessed the effect of varying saddle heights on movement and power asymmetries during cycling under simulated unilateral transtibial prosthesis conditions. This novel evaluation in the context of leisure cycling offers valuable insights into optimising bicycle fit in individuals with unilateral amputations. A previous study with this population employed the LeMond method for saddle height adjustment (LeMond & Gordis, 1987; Seratiuk Flores et al., 2023), resulting in heights similar to the +3.5% and +7% conditions used in this study. Other studies on professional cycling with amputations (Childers et al., 2011, 2014; Koutny et al., 2013) either utilised methods designed for intact cyclists or allowed cyclists to set saddle heights at their own discretion. Thus, while the effects of saddle height changes in intact cycling are known, an evaluation of the impact of these configuration changes in cycling with prostheses has not been conducted thus far, either in elite or leisure cycling contexts. Furthermore, it can have great influence on the biomechanics of cycling, as well as perceived comfort.

### *2.4.1. Effects of raising the saddle height*

According to previous research on professional cycling, the recommended saddle height that provides optimal power output would result in a knee angle of 25–35° at the bottom of the cycle, or at its maximum extension (Bini, 2020; de Vey Mestdagh, 1998). This saddle height is commonly applied in road cycling at both professional and amateur levels. Other methods of selecting the saddle height (Bini & Priego-Quesada, 2022; LeMond & Gordis, 1987) also result in near-full knee extension at the bottom of the cycle. This higher saddle heights offers numerous benefits such as reduced knee overuse injuries and improved cycling performance (Bini et al., 2011; Priego Quesada et al., 2017).

In the AL, the 25–35° maximum knee angle joint range was achieved at the 0% saddle height (29°) because of the lack of ankle joint movement in the simulated prosthesis condition. Increasing the saddle height led to a greater deviation of the kinematic parameters from the control condition. As shown in figure 2.3, at the peak extension, the knee joint reached 14° at

+7% height and 21° at +3.5% height. Similarly, a higher extension was observed in the hip angle (Fig. 2.4). Peak extension at the 180° pedal position was higher than that in the control under the +3.5% and +7% conditions, which was corroborated by Dunnett's test results.

The kinetic changes due to the increased saddle height may be linked to a difference in the pedalling technique, as evidenced by the EMG results. Starting at the beginning of the cycling movement, according to the pedal position of the AL, changes in the muscle activity in both legs were as follows. During the first and second quarters, muscle activity remained similar to that of the control. In the third quarter, the vastus medialis (Fig. 2.5C and 2.6C) showed increased activity on the UL side and reduced activity on AL side. Both the increase and reduction were exacerbated under the +7% and +3.5% conditions, suggesting that, with increased saddle height, the UL becomes the primary driver of knee extension, whereas AL relies on the increased activity of the UL to complete the cycling movement, thus increasing the asymmetry in muscle activity. This pattern was also noted in a previous study (Seratiuk Flores et al., 2023) but contrasts with findings in high-resistance unilateral amputee cycling, which described increased muscle activity in the vastus medialis of the AL (Childers et al., 2014; Dyer, 2016; Koutny et al., 2013).

During the fourth cycling quarter of the AL, the gastrocnemius medialis muscle (Fig. 2.5A and 2.6A) became more active in the UL, peaking at a 250% increase over the control condition at +7% height. As mentioned previously, the higher saddle heights provide many advantages to professional cyclists and also enable better use of the involved muscles' effective length (Connick & Li, 2013). Previous studies (Ericson et al., 1985; Verma et al., 2016) on intact cycling have shown that higher saddle heights increase gastrocnemius muscle activity, similar to the effects observed here. This suggests that regardless of the simulated prosthesis conditions, higher saddle heights allow the UL to reach an optimal position within the maximum knee extension range of 25–35°. Consequently, gastrocnemius activity increases as plantar flexion is used more extensively at the bottom of the cycle, in contrast to that at lower saddle heights. Additionally, in the +7% condition, the knee and hip joints in the AL achieved the highest extension, deviating the most from those of the control. This extension (14° for the knee joint) was significantly beyond normal levels and may have hindered the participants' ability to use the AL effectively. Therefore, the participants may have preferred using the UL, which was in an optimal position within the 25–35° range.

Additionally, in the fourth quarter on the AL side and at +7% height, the semitendinosus muscle (Fig. 2.5B and 2.6B) activity in the AL was increased by over 300%. A moderate increase in semitendinosus activity under similar conditions has been reported before (Seratiuk Flores et al., 2023) and is compatible with the knee flexor patterns. However, the movement analysis showed that the +7% saddle height had the lowest maximum knee flexion (Fig. 2.3). This peak semitendinosus activity might be related to the muscle performing at an alternate force-length setting (da Silva et al., 2016). Another possibility is the use of a clipless pedal attachment for the AL during the experiment, leading the participants to employ more muscle force during knee flexion at higher saddle heights to pull the pedal up at the end of the cycle. Higher hamstring activity during the last quarter of the cycle was documented when pulling clipless pedals (Mornieux et al., 2008). In this context, it might be used to compensate for reduced muscle activity in the knee extensor muscles of the AL such as the vastus medialis, shifting peak power production from pushing down to pulling up during an upstroke.

The effects described for the muscle activity and joint movement were directly reflected in the data collected by the instrumented pedals. At +7% height, the UL was responsible for 79% of the power production (Fig. 2.7). This height also resulted in increased torque effectiveness (Fig. 2.8A) and power smoothness (Fig. 2.8B) in the UL, whereas both parameters were decreased in the AL. The increased power asymmetry, difference in pedalling technique, and higher extension of the AL affected the perceived exertion (Fig. 2.9), with higher saddle heights being rated as the most strenuous. Therefore, it can be concluded that at increased saddle heights, the pedalling technique mainly employs the UL. The sound side becomes the main driver of the cycling movement, thereby increasing the asymmetry between the AL and UL sides.

### *2.4.2. Effects of lowering the saddle height*

Although lower saddle heights are associated with knee pain and other negative factors in competitive cycling (Bini et al., 2011), cyclists may choose lower the saddle for commuting and leisure activities. This choice provides a lower centre of gravity, which enhances balance and facilitates starting and stopping, which is often necessary in urban environments (Burke, 2003). Consequently, the control saddle height in this study targeted a maximum knee extension angle of 45°. Under the simulated prosthesis conditions, lower saddle heights brought the kinematic and kinetic parameters of the AL closer to those of the control condition.

The -7% condition showed knee angles throughout the cycle that matched those in the control condition (Fig. 2.3). The simulated prosthesis condition prevents ankle movement, which typically provides extra reach and force at the lowest pedal position. Therefore, the 0% saddle height shows increased extension, which decreases with lower saddle heights, closely matching the control at -7%. However, for hip angles (Fig. 2.4), the -7% height deviated the most from the control at the beginning and end of the cycle. With the pedal in its highest position, the ankle cannot dorsiflex for clearance, resulting in a higher hip flexion. However, the reduced distance between the saddle and crank spindle at lower saddle heights diminished hip overextension. Consequently, at -3.5%, the hip angle matched the maximum hip extension of the control condition. This aligns with the findings of a previous study on joint movement in triathletes and cyclists at different saddle heights, in which the knee and hip joints showed increased flexion at lower saddle heights (Bini et al., 2014).

Both vastus medialis and semitendinosus muscles (Fig. 2.5B, C and Fig. 2.6B, C) showed results closer to those in the control condition at lower saddle heights throughout the cycle for both UL and AL. Because these muscles contribute to knee motion as knee extensors and flexors, respectively, the observed muscle activity levels can be related to knee angles being closest to the control at -7% height. However, lower saddle heights resulted in activation levels below the control during the second quarter of the cycle in the gastrocnemius medialis muscle (Fig. 2.5A; Fig. 2.6A) in the UL. The ankle joint has a limited range of dorsiflexion (10–20°), with most of its motion in plantarflexion (40–55°) (Brockett & Chapman, 2016). With a lower saddle height, the ankle joint in the UL may enter a range where only dorsiflexion is required and plantarflexion becomes restricted (Moreno-Pérez et al., 2021). Therefore, the ankle power production shifts to the knee and hip joints, making the UL resemble a prosthetic condition in which the ankle joint is absent (Childers et al., 2009; Childers & Kogler, 2014).

Lower saddle heights resulted in knee and ankle movements and muscle activity in the proximal portion of the legs that closely matched the control conditions, suggesting better power symmetry. At -7% saddle height, the UL produced 58% of power (Fig. 2.7), indicating near-symmetric power production. Similar findings in intact cycling (Bini et al., 2011) showed greater pedalling power but reduced force effectiveness at lower saddle heights. In this study, the torque effectiveness (Fig. 2.8A) was increased in the AL at lower saddle heights, similar to the pedal smoothness results (Fig. 2.8B). Despite better power symmetry at -7%, this was neither the least strenuous condition (Fig. 2.9) nor the most comfortable (n = 1). The -3.5%

condition had lower perceived exertion and was deemed most comfortable (n = 5). This may be related to ankle joint movement; increased dorsiflexion and restricted plantarflexion may cause discomfort. A previous study (Bini, 2020) found that comfort was decreased at lower saddle positions among recreational cyclists, with no significant difference between preferred and higher heights. Consistent with the results of the present study, saddle height did not significantly affect perceived exertion.

### *2.4.3. Implications*

Musculoskeletal injuries are prevalent among cyclists and may be exacerbated by lower saddle positions. However, this study suggests that for leisure-level cycling with unilateral prostheses, lower saddle heights are recommended. Since leisure-level cycling does not require the increased power output resultant of higher saddle positions, this population can focus on comfort and the enhanced symmetry in power application and reduced strain achieved by the lower saddle heights. Thus, it can be concluded that the optimal saddle height in this population should maintain the dynamically measured maximum knee extension of the affected side within the 37–45° range, corresponding to the -3.5% and -7% saddle height conditions used in this study.

Following these findings, healthcare professionals can prescribe lower saddle heights for patients willing to start cycling. Furthermore, when having a first contact with cycling practice, users with amputations can potentially be discouraged from cycling if using incorrected settings, thus making the activity uncomfortable or impossible to carry out. This may lead to the belief that they are unable to practice leisure cycling. This is also reflected in research results (Poonsiri et al., 2021) which show that one of the main barriers for persons with amputations in taking up cycling is the inappropriate bicycle or prosthesis configuration.

These findings can also inform the design of cycling equipment specifically directed at individuals with amputations.

### *2.4.4. Limitations*

The use of actual lower-limb prostheses requires motor adaptation (Childers et al., 2014) and adjustment to limit muscle function after the procedure. However, these factors were not evaluated in this study. Additionally, the age range of the participants in this study was notably lower than that of the population with transtibial amputations that partake in cycling (62.0 ± 13.0 years) (Poonsiri et al., 2021); this was due to the introduction of the age-related factors (decline in muscle strength, endurance and balance, etc.), injury risk inferred by the task and, the possibility of exhaustion during the required physical activity. Future studies involving persons with lower-limb amputations will include middle-aged and older participants. A key element of the saddle height selection in the leisure cycling context is the actual daily use of bicycles and the manoeuvrability provided by the saddle height. Since this short study was conducted with a stationary bicycle, these adaptations and the development of a cycling technique could not be considered. Furthermore, the affected side was evaluated using the kinematic data. With the aim of collating data, a full evaluation of the unaffected side could provide further insights into the asymmetries.

## 2.5. Conclusion

Lower saddle heights resulted in joint movement and muscle activity levels more closely resembling those in control conditions and improved power symmetry between the AL and UL. Thus, adopting lower saddle heights in leisure cycling is a feasible and straightforward modification of the cycling equipment for individuals with unilateral transtibial prostheses.

# Chapter 3
# Development of a Transtibial Prosthesis for Cycling Employing Compression Springs in the Ankle Component

## 3.1. Introduction

While modifications can be made to various components of the bicycle to optimise cycling for individuals with unilateral transtibial prostheses, adjusting parts of the prosthesis itself can significantly enhance the cycling experience (Childers et al., 2009, 2014; Dyer & Disley, 2020; Koutny et al., 2013). Standard prostheses are primarily designed for walking, a movement that utilises the same joints involved in cycling, but in a distinct manner—particularly at the ankle joint. Walking prostheses often incorporate energy-storing components, such as carbon fibre blades, which deform at the toe under the user's weight to simulate the push-off phase of walking (Childers & Takahashi, 2018; Prost et al., 2022). However, cycling places different demands on the ankle joint, requiring more controlled movement rather than the passive reliance on weight distribution used in walking (Ericson et al., 1988). While carbon fibre blades effectively store and release energy during walking, they are less suited for cycling, where the movement of the ankle should occur within the joint itself, different from the blade deformation that happens at the toe.

Therefore, the current study seeks to explore the potential for enhancing standard prosthetic designs with a component that offers added functionality: ankle movement. The feature under investigation is a component attached to the pylon, which can be activated or deactivated by the user as needed in daily life. This possibility of adjustment could provide significant benefits in terms of adaptability and user control, allowing persons with amputation to tailor the functionality of the prosthesis according to their daily needs, whether for general activities or specific recreational pursuits like cycling.

During cycling with intact limbs, muscles such as the tibialis anterior, gastrocnemius, and soleus play crucial roles in plantarflexing and dorsiflexing the foot, contributing up to 15% of the joint forces required for cycling, at the ankle, and stabilising the foot throughout the pedal stroke (Elmer et al., 2011; Fonda & Sarabon, 2010). However, in the case of transtibial amputation, these muscles are not present, and standard walking prostheses typically do not simulate ankle movement, leading to two main biomechanical challenges.

First, the inability of the prosthetic foot to plantarflex at the bottom of the pedal stroke means that it cannot generate additional power to drive the pedal, nor can it provide the extra reach usually facilitated by the natural plantarflexion of a biological ankle. This limitation requires greater extension from the knee and hip than would be required with an intact limb. Additionally, the lack of dorsiflexion at the top of the pedal stroke leads the knee and hip to compensate by flexing more (Childers et al., 2009, 2014). Second, the loss of the distal leg muscles results in reduced stabilisation of the prosthetic limb (So et al., 2005), complicating the placement of the foot on the pedal. This absence also eliminates the tactile feedback that a biological limb provides, which helps leisure cyclists to utilise regular platform pedals to ascertain whether the foot is in proper contact with the pedal and its specific positioning. These biomechanical deficits lead to kinetic and kinematic asymmetries between the intact and amputated limbs, potentially causing detrimental effects in both the short and long term during cycling practice (Childers et al., 2009; Poonsiri et al., 2021).

To address these issues, this study hypothesises that incorporating ankle movement through passive compression springs into the prosthesis design could alleviate some of the biomechanical challenges posed by cycling with a standard prosthesis. The springs may help replicate the natural flexion and extension of the ankle joint during cycling. This in turn reduces the compensatory strain on the knee and hip, improves pedal power distribution, and ultimately approximates the prosthesis movement to that of a biological limb, enhancing the overall cycling experience for the user. Previous research has explored the use of a transtibial prosthesis prototype that incorporated ankle joint movement (Tiele et al., 2020). This study examined kinematic heel misalignment and pedal pressure, finding that the addition of ankle movement through an elastic neoprene band on the prosthetic heel improved bilateral heel alignment and reduced unwanted pedal pressure during the latter half of the cycling motion.

It is important to note that this solution may not be suitable for competitive cycling. Professional cycling prostheses typically do not feature ankle movement; instead, they are designed to provide a rigid connection between the crank and the residual limb, maximising the power delivery from the intact portion of the limb (Dyer & Disley, 2020; Koutny et al., 2013). To accommodate biomechanical differences in joint motion, modifications such as crank shortening (Koutny et al., 2013) on the affected side are often made to the bicycle in professional settings. Recreational cycling, on the other hand, places more emphasis on comfort and stability, making the inclusion of movement in the ankle component advantageous in this context. The advantages of the proposed solution and its differences from professional cycling can be seen in figure 3.1.

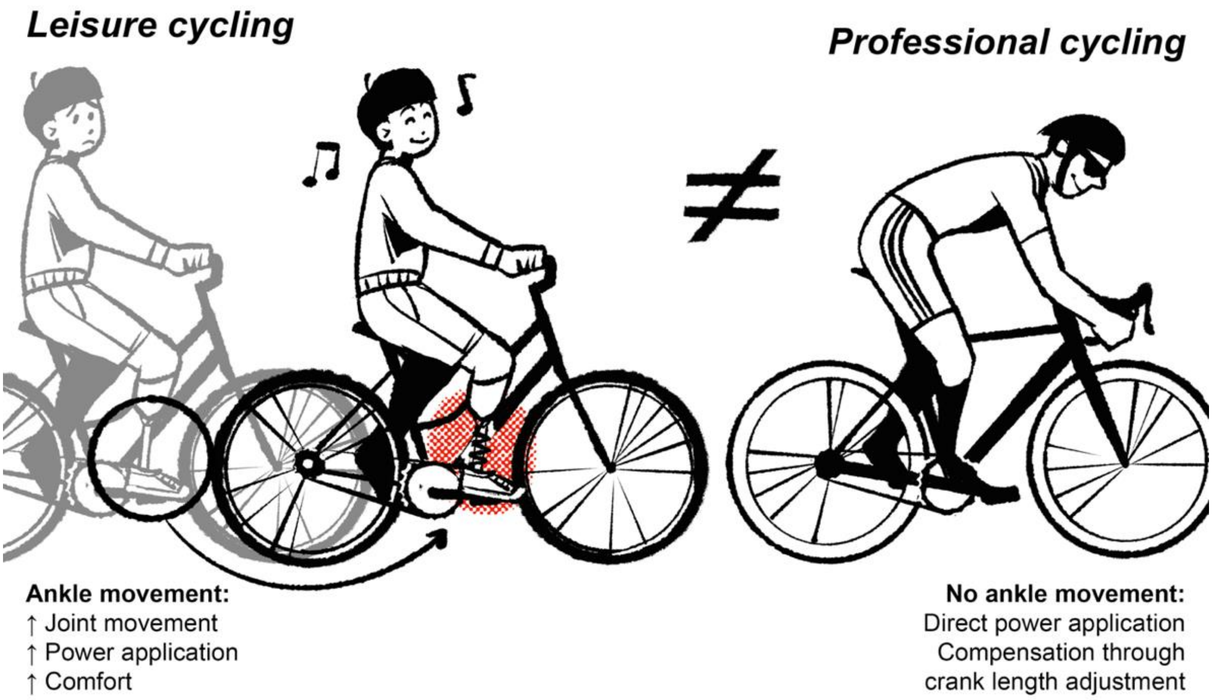


**Figure 3.1.** Differences in prosthetic needs between leisure and professional cycling.

Hence, the current study aimed to assess the biomechanical effects—including muscle activity, joint movement, force application to the pedals, and subjective feedback on perceived exertion—of using a prototype prosthesis equipped with three compression springs varying in spring constants, placed in the pylon component. This prototype also integrated findings from other previous studies that I conducted regarding the optimal positioning of the foot on the pedal (Seratiuk Flores et al., 2023), as well as the findings from Chapter 2. Two participants with transtibial amputations were recruited for this study, one with a traumatic amputation and the other with a congenital amputation. The results allowed an exploration of the biomechanical impacts of adding ankle movement in these two amputation contexts.

## 3.2. Methods

### *3.2.1. Participants*

Two male participants wearing transtibial prostheses were recruited for this experiment. The primary distinction between the participants lies in the cause of their amputations: one participant experienced a traumatic amputation, whereas the other had a congenital amputation. Both participants completed a health questionnaire, providing information about the nature of their amputation and their cycling experience. Written consent was obtained from both participants prior to the commencement of the experiment. Leg dominance was determined by asking each participant, "If you were to kick a ball, which leg would you use?" (van Melick et al., 2017). Additionally, measurements of their unaffected leg (UL) and affected leg (AL) were taken, and participants self-reported their height and weight. The full demographic and physical details of the participants are presented in table 3.1.

**Table 3.1.** Participants' characteristics.

| **Participant Information** | **Participant with Traumatic Amputation (TA)** | **Participant with Congenital Amputation (CA)** |
|---|---|---|
| Height | 186 cm | 157 cm |
| Weight | 98 kg | 54 kg |
| Age | 43 years old | 42 years old |
| Inseam | 81.5 cm | 63.1 cm |
| Foot Size | 28.6 cm | 22.7 cm |
| Prosthetic Length | 35.2 cm | 28.6 cm |
| Ankle distance to the pedal attachment | 7.2 cm | 5.6 cm |
| Dominant leg | Right | Right |
| Affected leg | Left | Right |
| Time since amputation procedure | 10 years | From childhood |
| Time to start using a prosthesis after the amputation procedure | Within one year | Since childhood |
| Experience with bicycle riding after the amputation | No experience | Cycles once a week |

The experiment was approved by the Ethics Committee of the Faculty of Design, Kyushu University (approval number 532).

### *3.2.2. Prototype prosthesis*

The primary feature of the prosthesis prototype used in this study was its ability to accommodate interchangeable spring components. The specifications of the prototype are shown in figure 3.1A. Beginning at the top, the design incorporated an Ottobock tube adaptor (2R50; Ottobock SE & Co. KGaA, Duderstadt, Germany) to attach to most prosthetic sockets, which typically use a pyramid nut as the connecting interface between the socket and prosthesis. A tube clamp was employed to connect the standard ⌀30 mm prosthetic pylon to the housing for the piston of the prototype, allowing for adjustments to different prosthetic lengths (Fig. 3.2B).

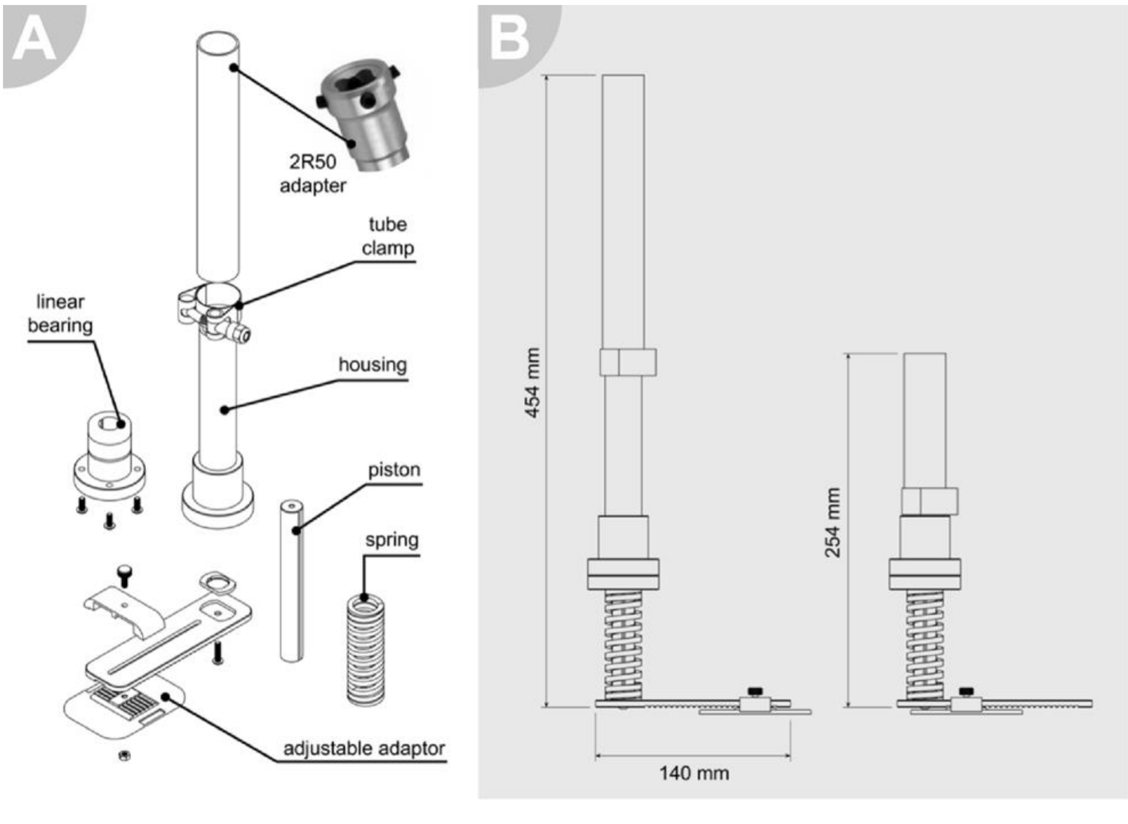


**Figure 3.2.** Prototype specifications. (A) Exploded view with components, (B) maximum and minimum height settings of the prototype.

The piston itself was placed in a housing machined from A7075 aluminium, which contained a linear bearing (E-BSHM16-132-M4-N4, Misumi Group Inc., Tokyo, Japan). This bearing enabled the piston to move with minimum resistance within the housing. The piston was attached to an aluminium plate, also machined from A7075 aluminium, through a screw, allowing for the spring surrounding the piston to be easily swapped. The plate was designed to connect to a 3D-printed polylactic acid (PLA) adaptor, which served as the interface between the ergometer and the prosthesis. The adaptor could be positioned at a specified length relative to the piston, ensuring precise adjustments for the prosthetic setup.

### *3.2.3. Experimental setup*

The experiments were conducted at two locations, both of which maintained air-conditioned environments set to a consistent temperature of 22°C. The cycling tasks were performed using a mechanically powered, constant-load FITBOX Lite ergometer (AINEXT Inc., Tokyo, Japan), with a crank length of 170 mm. The original pedals of the ergometer were replaced with Assioma DUO (Favero Electronics Srl., Arcade, Italy) clipless power meter pedals. The saddle height for each participant was adjusted following the findings detailed in Chapter 2. For the CA participant, the saddle height was set to achieve a knee angle of 45° at full extension, measured statically with a goniometer. This height was compatible with the previously set minimum height, and was chosen due to the ankle movement employed in the prototype, which would further extend the knee at the bottom of the cycle. In contrast, the saddle height for the TA participant was set to provide a maximum knee angle of 55°, due to the participant's height. The handlebar was positioned at a height of 100 mm above the saddle height. The full setup is presented in figure 3.3.


**Figure 3.3.** Cycling environment.

The clipless pedals were outfitted with 3D-printed PLA platform attachments (Figure 3.4), designed to accommodate plates affixed to both the shoes and the prototype prosthesis used in the experiment. These plates were positioned at a distance equivalent to one-quarter of the participant's biological foot length from the ankle joint or the piston attachment point in the prototype. This configuration was implemented to standardise a middle-of-the-foot position on the pedal, a setup identified in my previous research as reducing asymmetries between intact and affected limbs in transtibial leisure cyclists (Seratiuk Flores et al., 2023).

**Figure 3.4.** Plates used to control the positioning of the foot on the pedal. A diamond shape present in the pedal cleat fitted into a plate on the prosthesis. This plate was also used in the unaffected leg.

## *3.2.4. Spring settings*

Previous research has indicated significant variability in the normal forces exerted on the pedals, influenced by different cycling conditions, methods of calculation, and resistance levels during pedalling. Tiele et al. (2020) documented force of up to 1000 N on the pedals at a resistance setting of 100 W. Conversely, research by Bini et al. (2013) identified an effective force of approximately 470 N at a 90° pedal position in more professional cycling contexts, with a pedal resistance of 350 W. Specifically, the earlier study by Tiele et al. (2020) investigating transtibial prostheses with ankle movement utilized an extension neoprene band with a spring constant of 3.54 N/mm, with the cycling task being conducted at a 100 W resistance setting.

### 3.2.4.1. Experimental conditions

Given the objective of this study to assess leisure cycling, two resistance settings were selected for the stationary bicycle: 50 W at 60 rpm, simulating an everyday commute or leisure cycling on a level path, and 60 W at 50 rpm, mimicking a slight incline. In light of these lower resistance settings, three commercially available 80-mm compression springs with varying spring constants were chosen to be coupled to the prototype prosthesis (Fig. 3.5):

**Figure 3.5.** Prototype prosthesis fitted with the soft spring.

**Soft (S):** 4.27 N/mm

**Medium (M):** 6.17 N/mm

**Hard (H):** 13.08 N/mm

In addition to using the springs with different constants, a fourth condition was also employed: **regular prosthesis** (RP), where participants used their regular walking prosthesis.

These specific spring constants were chosen based on a pilot study, which concluded that a spring constant of less than 3 N/mm would lead to a full collapse of the spring during the pedal stroke when cycling at 50 W resistance. The hard setting was also chosen based on pilot results, which indicated that the 13.08 N/mm constant led to only slight ankle movement during cycling practice at the aforementioned resistance settings.

### 3.2.5. Experimental protocol

Participants were instructed to wear tight-fitting clothing, consisting of t-shirts and shorts, as well as ankle-covering socks. Trainers with 2.3 cm sole thickness at their middle section were worn on both feet during the regular prosthesis (RP) conditions and only on the UL during the other experimental conditions. The cycling task was performed across three trials, each lasting 1 minute, with a 30-second rest period between trials and a longer 3-minute rest between each condition.

The RP condition was conducted first, after which participants were fitted with the prototype prosthesis, a process supervised by a certified prosthetist. The prosthesis was adjusted to match the height of the unaffected leg while wearing the trainer, ensuring consistency in leg length. Following the RP condition, the order of the remaining conditions was randomised. The first trial in each condition was considered practice, and data collection occurred during the subsequent two trials. The initial and final 15 seconds of each trial were discarded, and analysis was conducted over 10 consecutive crank revolutions within the remaining time. In total, the experiment was carried out using eight conditions.

### 3.2.6. Measurements

#### 3.2.6.1. Joint angles

Two Panasonic HC-300M cameras (Panasonic, Osaka, Japan) recording at 60 frames per second (fps), with 1080-pixel resolution, were placed 2 metres away from the participant at both right and left sides. The remaining settings, including data collection and processing and marker placement, were identical in procedure to what is presented in section 2.2.6.1.

#### 3.2.6.2. Muscle activity

The procedures for data collection and electrode placement were the same as previously described in section 2.2.6.2. However, the gastrocnemius medialis muscle was not targeted in this study.

#### 3.2.6.3. Instrumented pedals

Pedal data were collected and processed exactly as described in section 2.2.6.3, also applying the same parameters: left and right balance (in this study, adjusted to represent affected and unaffected balance), torque effectiveness, and pedal smoothness.

#### 3.2.6.4. Subjective evaluation

The use of the Borg 6-20 RPE scale (Borg, 1970) was conducted exactly as previously described in section 2.2.6.4. The scale was placed in front of the participant on a support integrated to the ergometer handle.

## 3.3. Results

Results for pedal data and overall muscle activity are presented as average ± standard deviation. The muscle activity circular graphs and joint movement are shown as averages.

### *3.3.1. Joint movement*

#### 3.3.1.1. Ankle

Figure 3.6 illustrates distinct ankle movement patterns in both participants. In the TA participant's unaffected limb (Fig. 3.6A), the S spring showed more acute angles, whereas the RP setting produced more obtuse angles. Peak acute angles occurred near the 90° pedal position, transitioning to maximal obtuse angles at 180°. In the affected limb, the M spring most closely reproduced this pattern, although the S spring shifted peak obtuse angles to about 270° pedal position, followed by a sharp transition into acute angles.

For the CA participant (Fig. 3.6B), the unaffected limb exhibited more pronounced acute and obtuse angles between 90° and 180° pedal positions. Similar patterns emerged in the affected limb under the M spring, with comparable trends for the S setting. Both participants showed limited ankle movement under the H spring, although it remained more variable than in the RP condition.

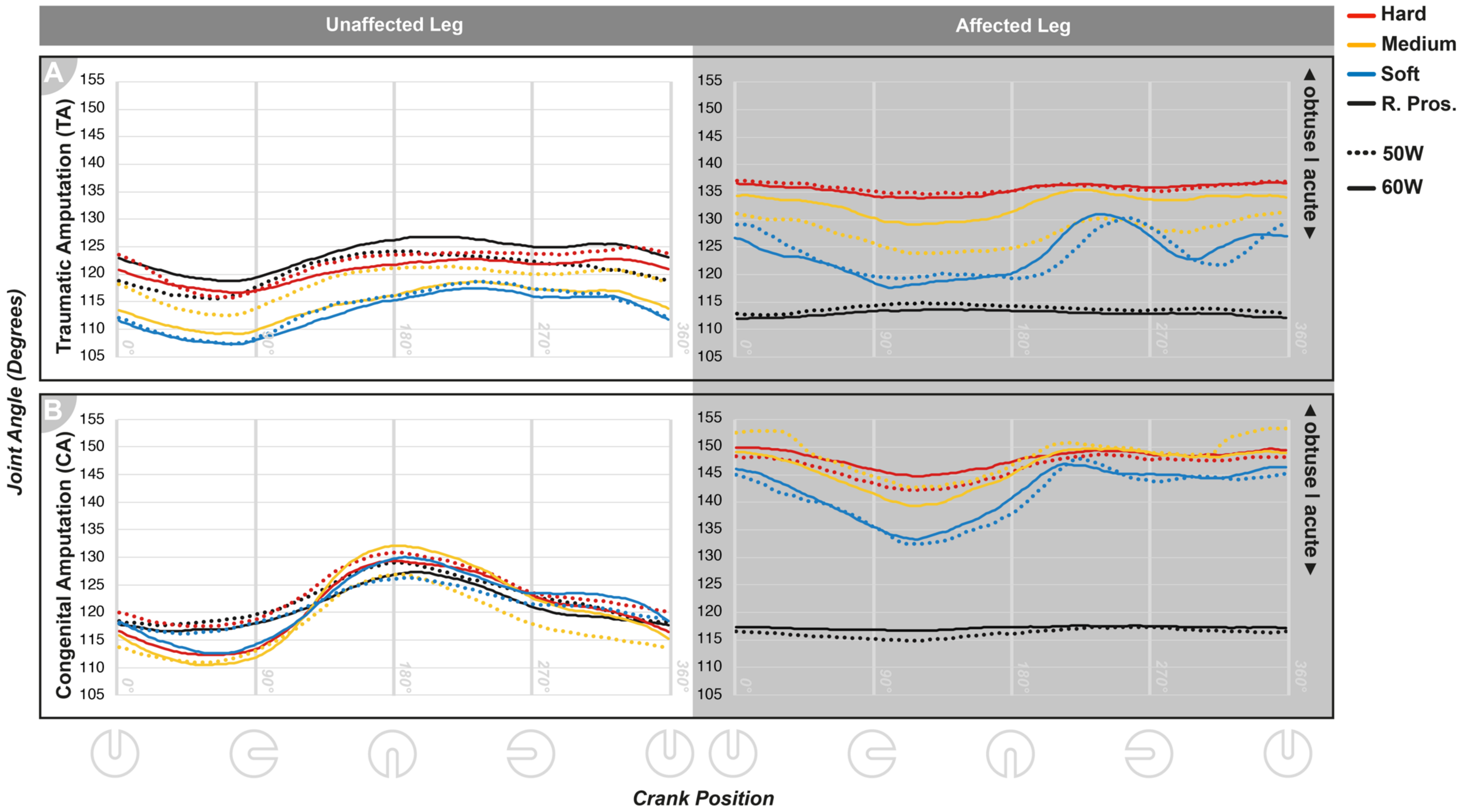


**Figure 3.6.** Ankle joint movement throughout the cycle. (A) traumatic amputation participant, (B) congenital amputation participant. R. Pros.: regular prosthesis.

#### 3.3.1.2. Knee

Figure 3.7 illustrates knee movement under different prosthesis settings. In the TA participant (Fig. 3.7A), the UL showed minimal changes, with only a slight flexion increase at 60 W under the S and H spring settings. By contrast, the AL exhibited more extension overall, peaking in the S setting at 60 W. Flexion levels in that same condition remained consistent with the UL.

For the CA participant (Fig. 3.7B), the UL displayed greater variability. The M setting at 60 W produced the most extension, whereas the 50 W M spring setting yielded the most flexion. In the AL, the RP at 60 W offered limited movement but improved at 50 W, converging with UL behaviour. Extension increased further under the M and S settings at 60 W. In contrast, the H setting promoted more symmetry, aligning knee motion in both limbs.

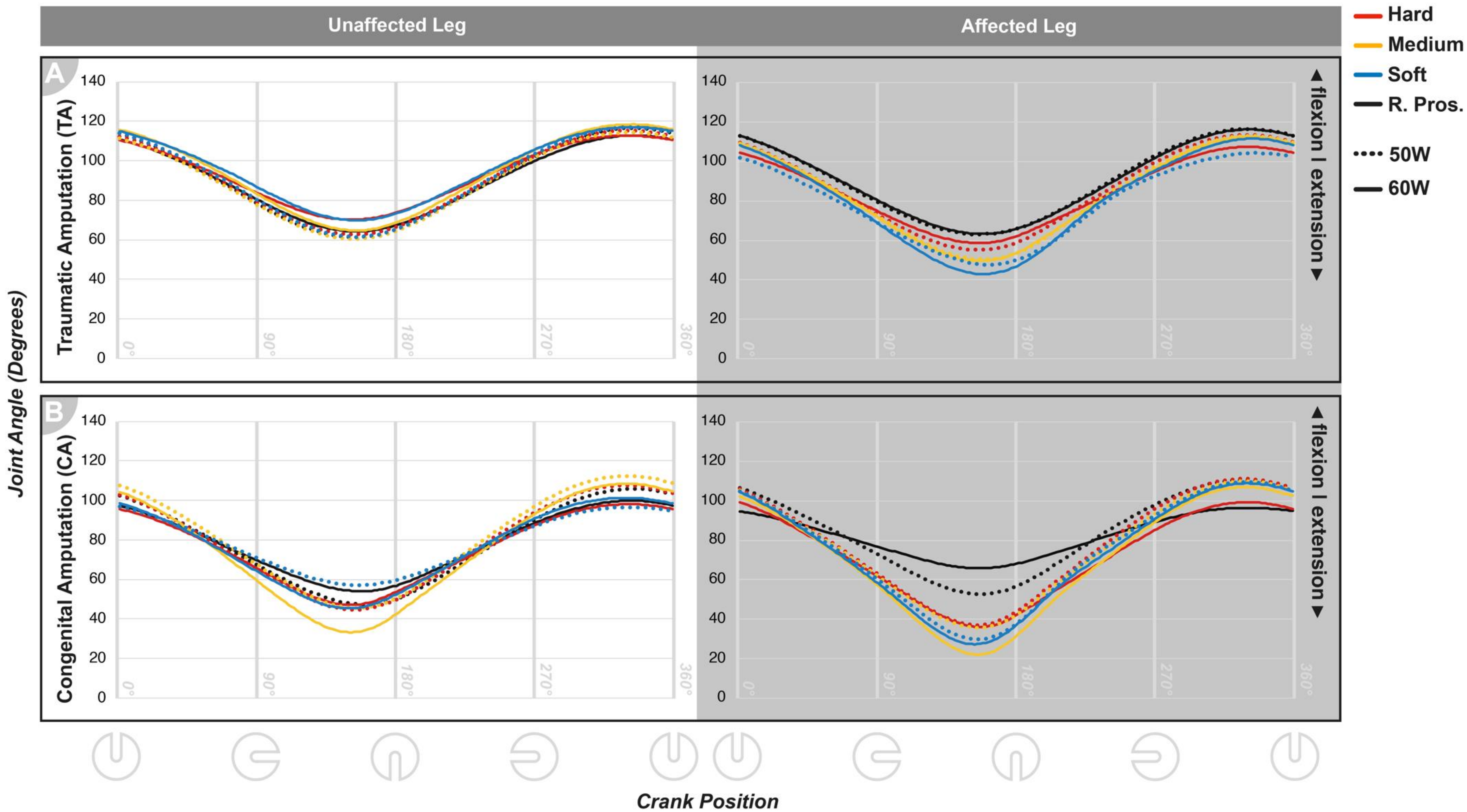


**Figure 3.7.** Knee joint movement throughout the cycle. (A) traumatic amputation participant, (B) congenital amputation participant. R. Prost.: regular prosthesis.

#### 3.3.1.3. Hip

Figure 3.8 shows hip joint movement patterns similar to those at the knee for the TA participant (Fig. 3.8A). In the UL, hip motion remained largely unchanged, although the S and H springs at 60 W resistance induced slightly increased flexion. By contrast, the AL displayed greater extension as spring constants decreased, peaking with the S spring at 60 W. This condition also revealed a pronounced extension peak, including a minor secondary curve near the 270° pedal position.

In the CA participant (Fig. 3.8B), prosthetic conditions led to more substantial changes in the UL. The M spring at 60 W produced the highest extension peak, and other settings likewise deviated markedly from the RP at 60 W. A secondary peak around 270° also appeared in the UL. Overall, the prosthetic settings narrowed differences between 50 W and 60 W in both limbs, with the M spring at 60 W yielding the greatest extension and flexion in the AL. Variations between 50 W and 60 W were less pronounced in the AL than in the UL.

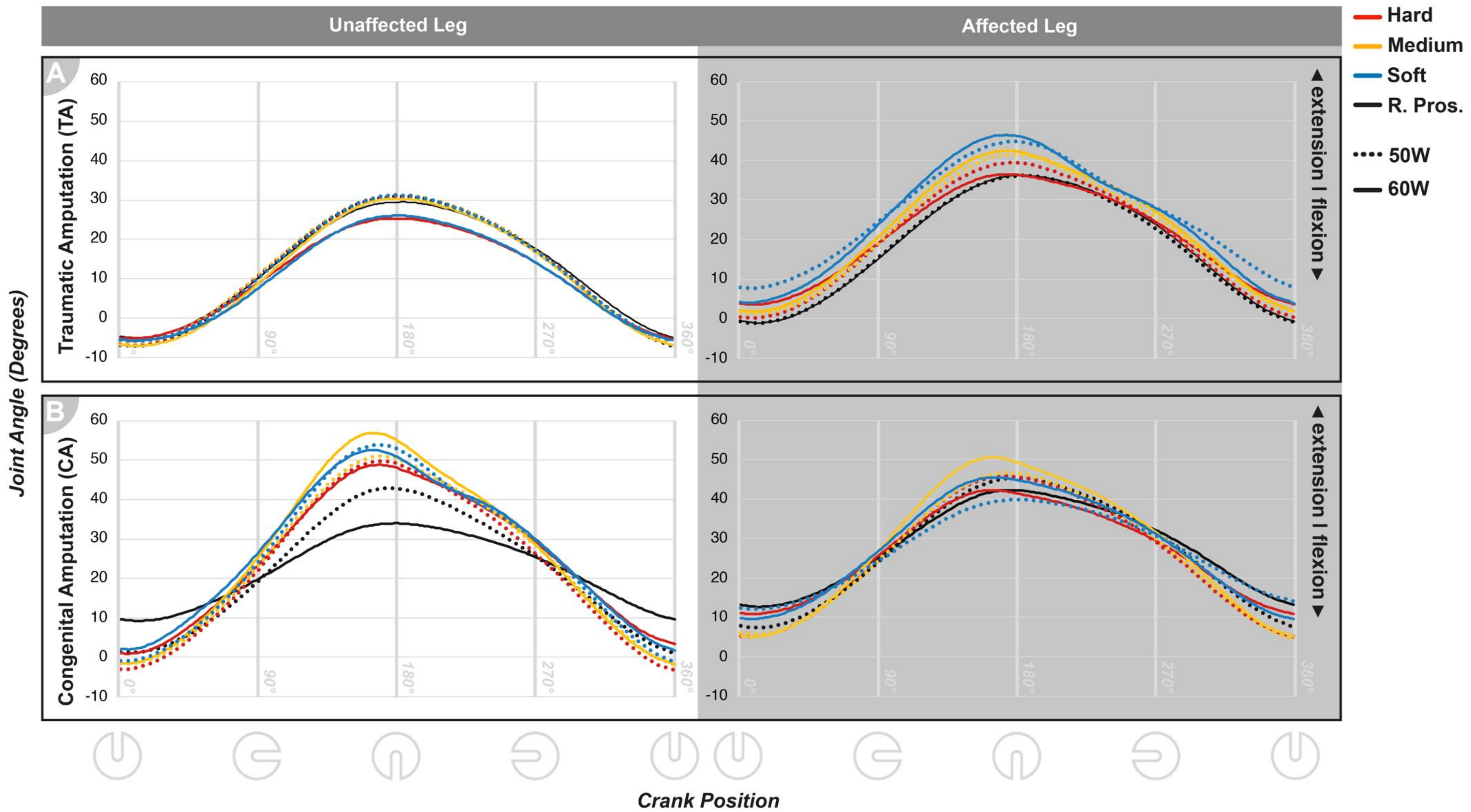


**Figure 3.8.** Hip joint movement throughout the cycle. (A) traumatic amputation participant, (B) congenital amputation participant. R. Pros.: regular prosthesis.

### *3.3.2 Muscle activity*

Muscle activity results were normalised by the RP condition. Figures 3.9 (TA participant) and 3.10 (CA participant) show the overall mean muscle activity results for each condition. In figures 3.10 and 3.11, the EMG results are shown as percentage increase or decrease over the RP condition, which corresponds to the dark circle in the centre of the graph. These results are presented according to the pedal position.

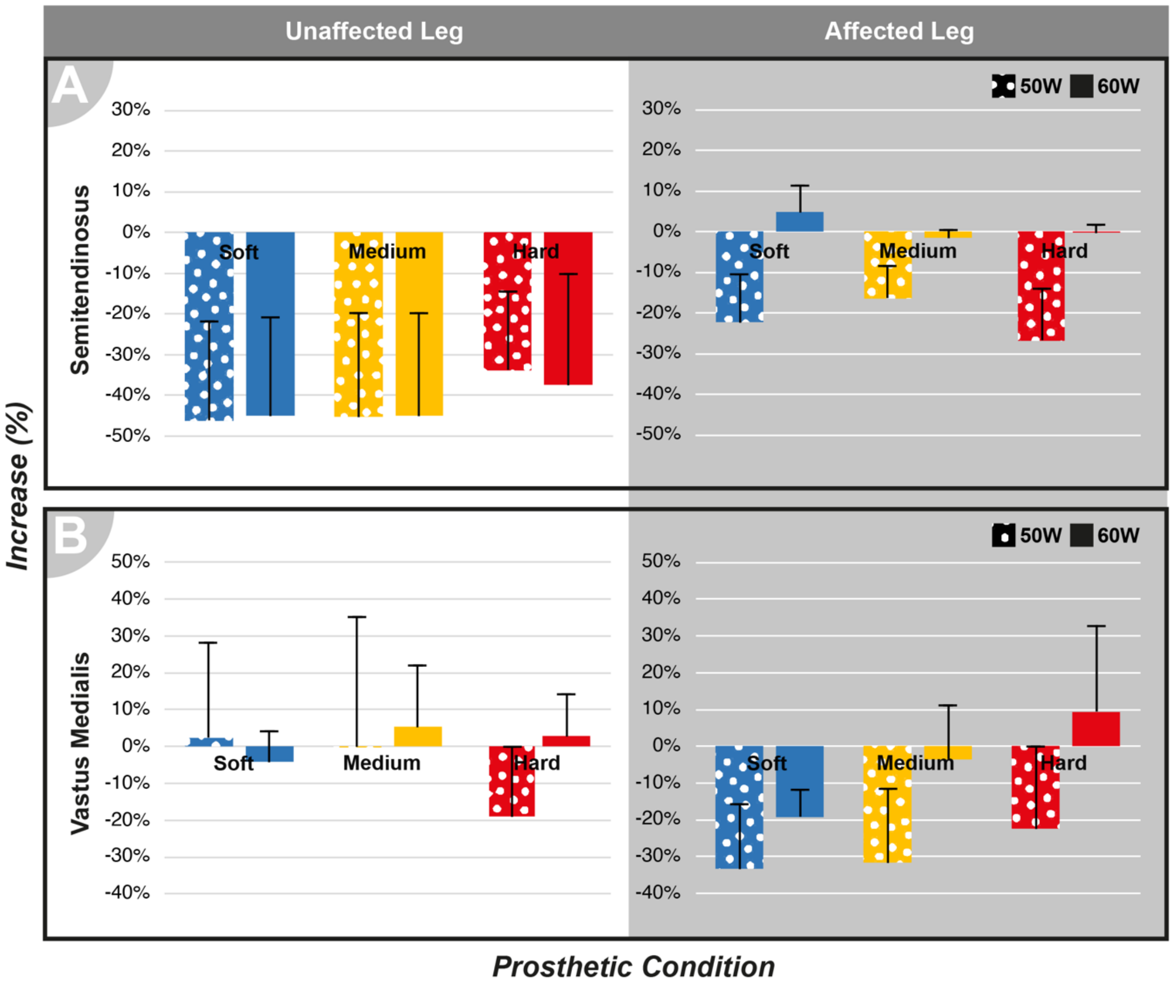


**Figure 3.9.** Mean muscle activity of the (A) semitendinosus and (B) vastus medialis muscles for the TA participant.

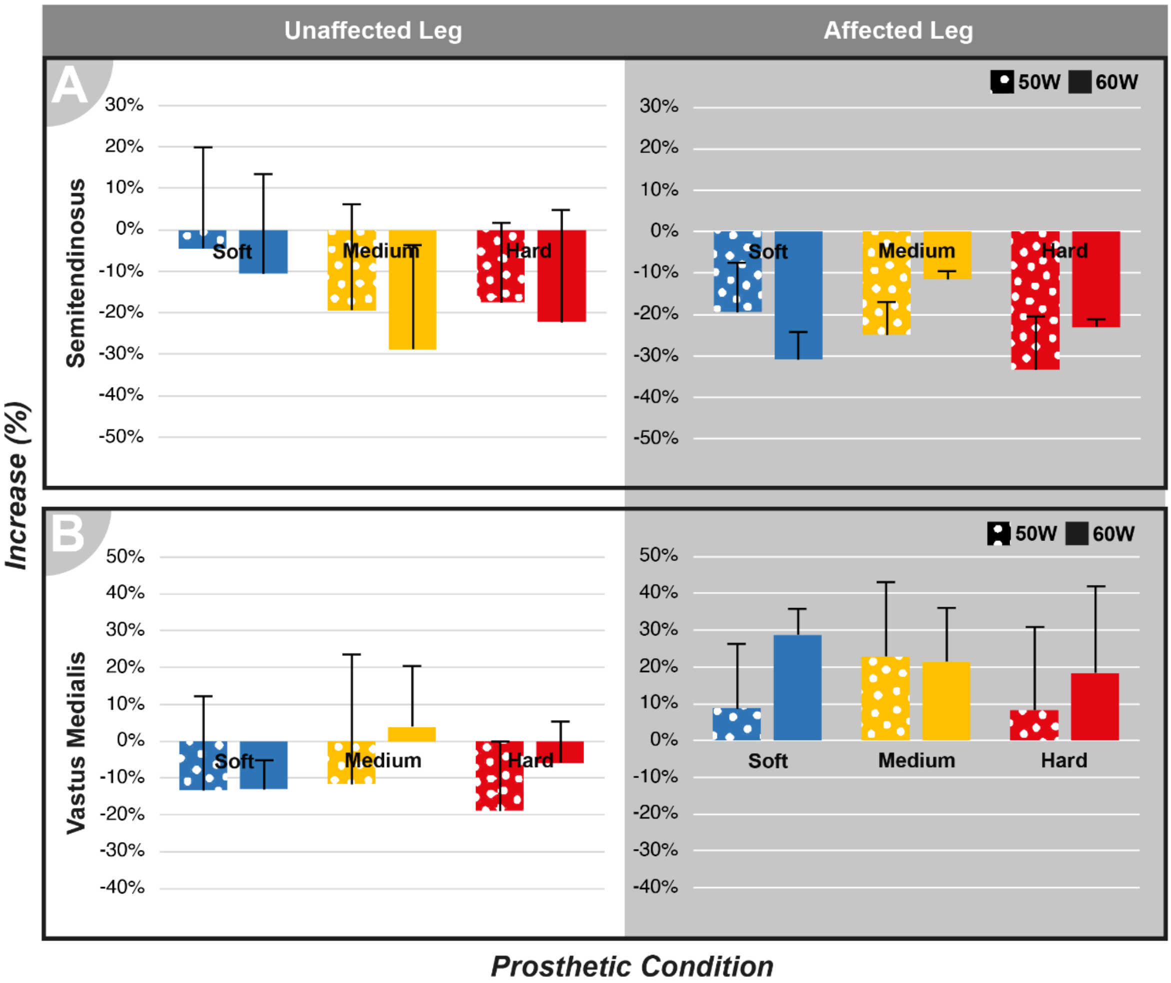


**Figure 3.10.** Mean muscle activity of the (A) semitendinosus and (B) vastus medialis muscles for the CA participant.

### 3.3.2.1. Semitendinosus

Figure 3.11A indicates that in the TA participant's UL, the semitendinosus activity declined during the second and final quarters of the pedalling cycle at 50 W. At 60 W, a more pronounced decrease appeared in the second half, although a slight increase emerged in the second quarter under the H spring. In the AL, 60 W conditions were broadly similar to the RP, whereas 50 W showed lower activity in the fourth and first quarters.

For the CA participant (Fig. 3.12A), the UL revealed a drop in semitendinosus activation in the second quarter at both 50 W and 60 W, with a notable first-quarter peak at 60 W under the S spring. In the AL, most prototype settings reduced activity compared to the RP, although a peak appeared in the third quarter for the S spring at 50 W and in the fourth quarter for the M spring at 60 W.

Overall mean values (Figures 3.9A, 3.10A) point to decreased activity across all conditions. In the TA participant's UL, the H spring best approximated RP levels, whereas the S spring was closer to the RP in the CA participant. For the AL, the TA participant's 60 W conditions broadly paralleled RP outcomes, whereas the CA participant's results diverged more notably from the RP at every setting.

**3.3.2.2. Vastus medialis**

Vastus medialis activity varied considerably between participants. In the TA participant (Fig. 3.11B), the UL's activity peaked during the second quarter at 60 W and during the fourth quarter at 50 W under the S and M spring conditions. In the AL, a strong first-quarter peak was present at 50 W under the H spring setting, with a second-quarter peak at 60 W under the same setting. Most conditions showed decreased activity near the 270° pedal position.

For the CA participant (Fig. 3.12B), the UL largely mirrored the RP, yet exhibited increased fourth-quarter activity across all conditions, especially with the M spring at 60 W. In the AL, activity consistently rose from the late fourth quarter through the second quarter, particularly at 60 W, with a pronounced dip between 135° and 315° pedal positions for all settings.

Mean results (Figs. 3.9B, 3.10B) indicate that UL activity typically matched RP values, although the 50 W H spring yielded up to 18% lower activity. In the AL, the TA participant recorded an overall decrease except under the 50 W H spring, whereas the CA participant showed a general increase in muscle activation.

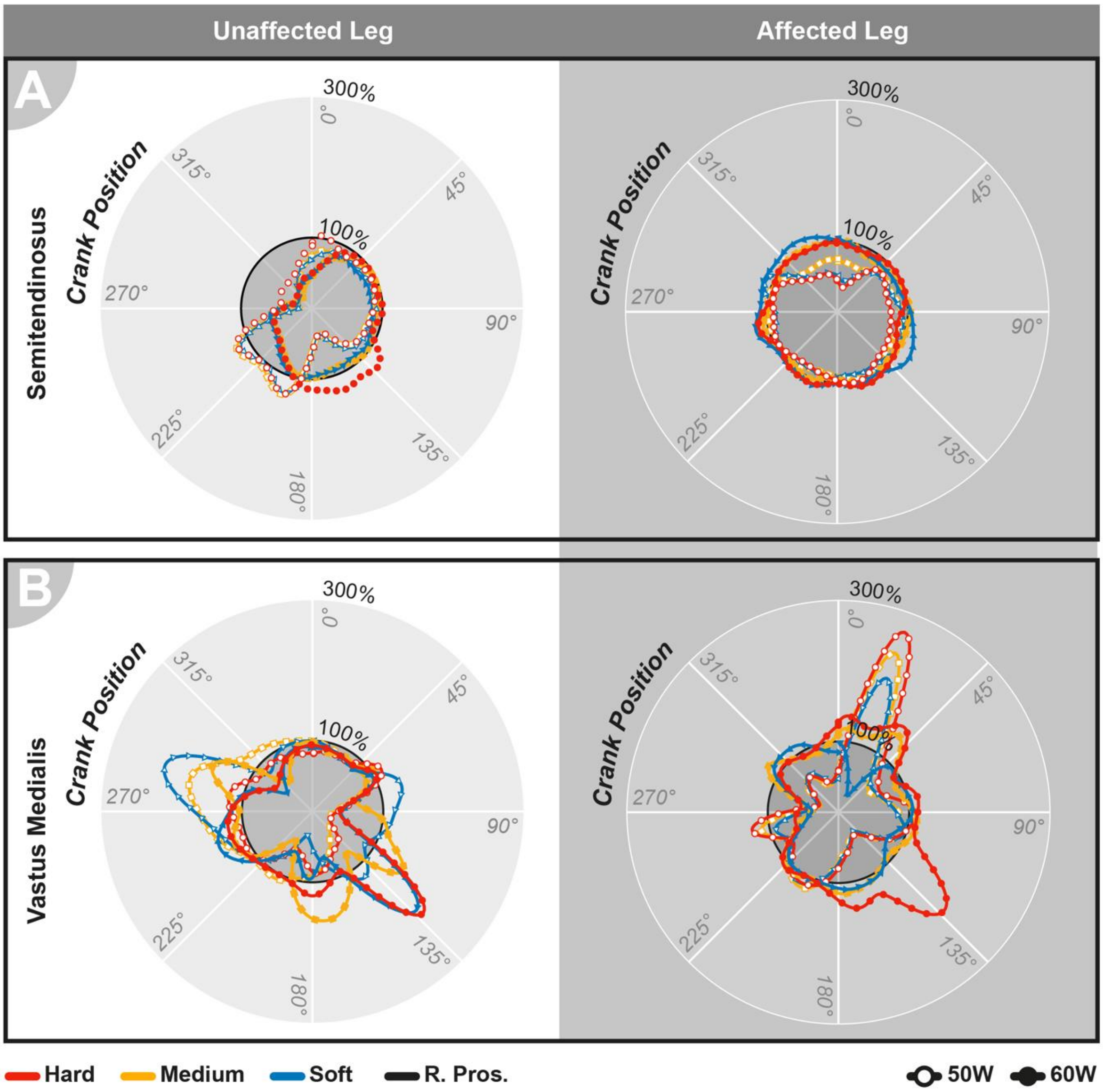


**Figure 3.11**. Mean muscle activity according to pedal position for the TA participant. (A) semitendinosus and (B) vastus medialis. R. Pros.: regular prosthesis.

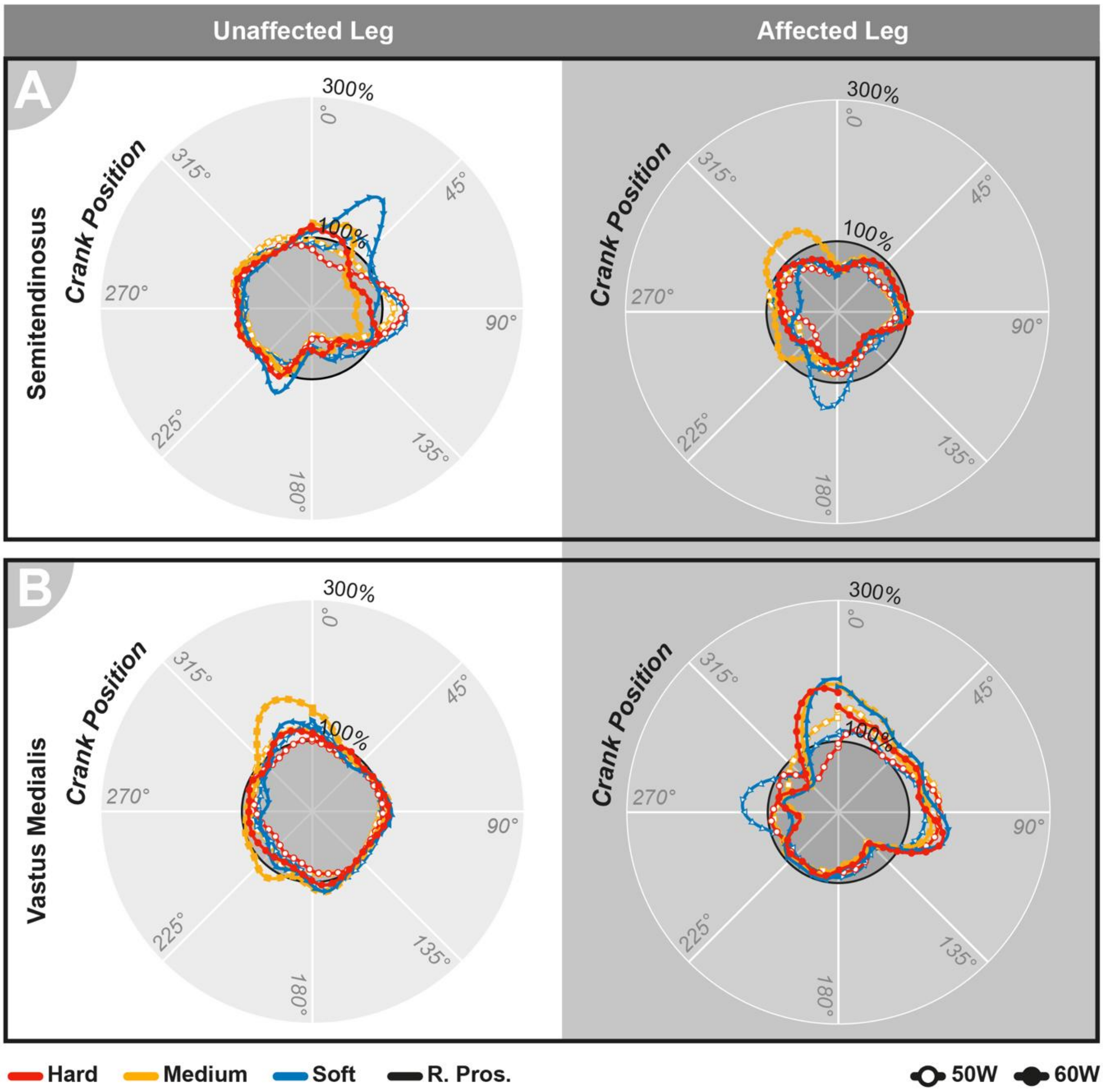


**Figure 3.12.** Mean muscle activity according to pedal position for the CA participant. (A) semitendinosus and (B) vastus medialis. R. Pros.: regular prosthesis.

### *3.3.3. Instrumented pedals*

#### 3.3.3.1. Balance between affected and unaffected legs

Figures 3.13 (TA) and 3.14 (CA) illustrate the power distribution between the UL and AL. The TA participant achieved a more balanced power distribution under the H spring condition for both the 50 W and 60 W resistance settings. In contrast, the CA participant exhibited a more symmetrical balance between the UL and AL in the RP condition.

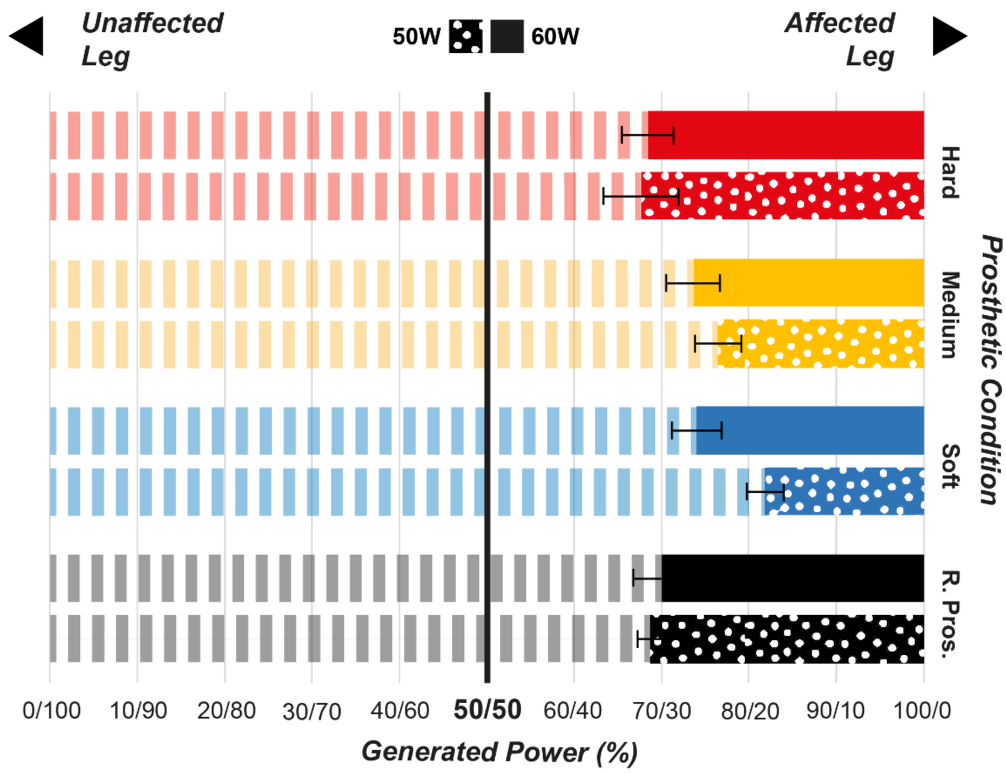


**Figure 3.13.** Unaffected leg and affected leg power balance means for the TA participant. R. Pros.: regular prosthesis.

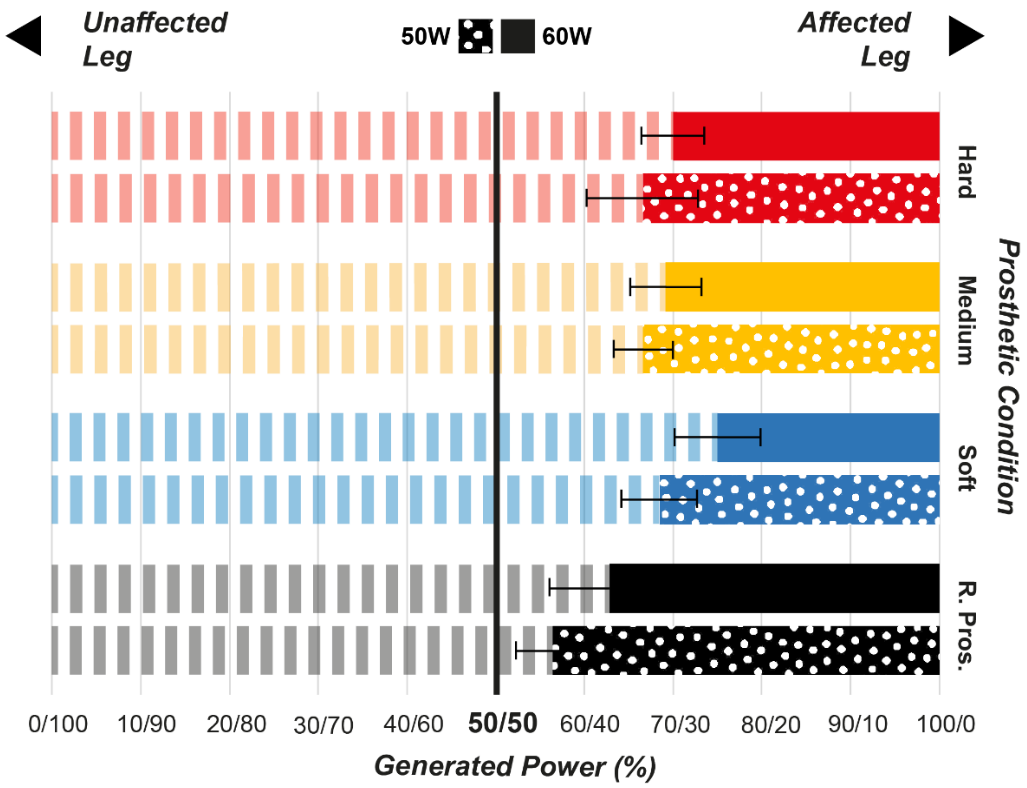


**Figure 3.14.** Unaffected leg and affected leg power balance means for the CA participant. R. Pros.: regular prosthesis.

### 3.3.3.2. Torque effectiveness and pedal smoothness

Figure 3.15 (TA) and Figure 3.16 (CA) illustrate torque effectiveness and pedal smoothness for both the UL and AL. In the TA participant, UL torque effectiveness (Fig. 3.15A) generally fell below that of the RP, except under the S spring at 50 W. By contrast, the AL showed higher torque effectiveness at all 60 W settings and under the H spring at 50 W. Pedal smoothness (Fig. 3.15B) in the UL declined at 50 W but rose at 60 W, whereas in the AL, it surpassed the RP only when using H springs at both loads.

For the CA participant, torque effectiveness (Fig. 3.16A) increased in the UL but decreased in the AL across both loads, mirroring trends in pedal smoothness (Fig. 3.16B). The sole exception was the S spring at 50 W, where AL torque effectiveness exceeded the RP.

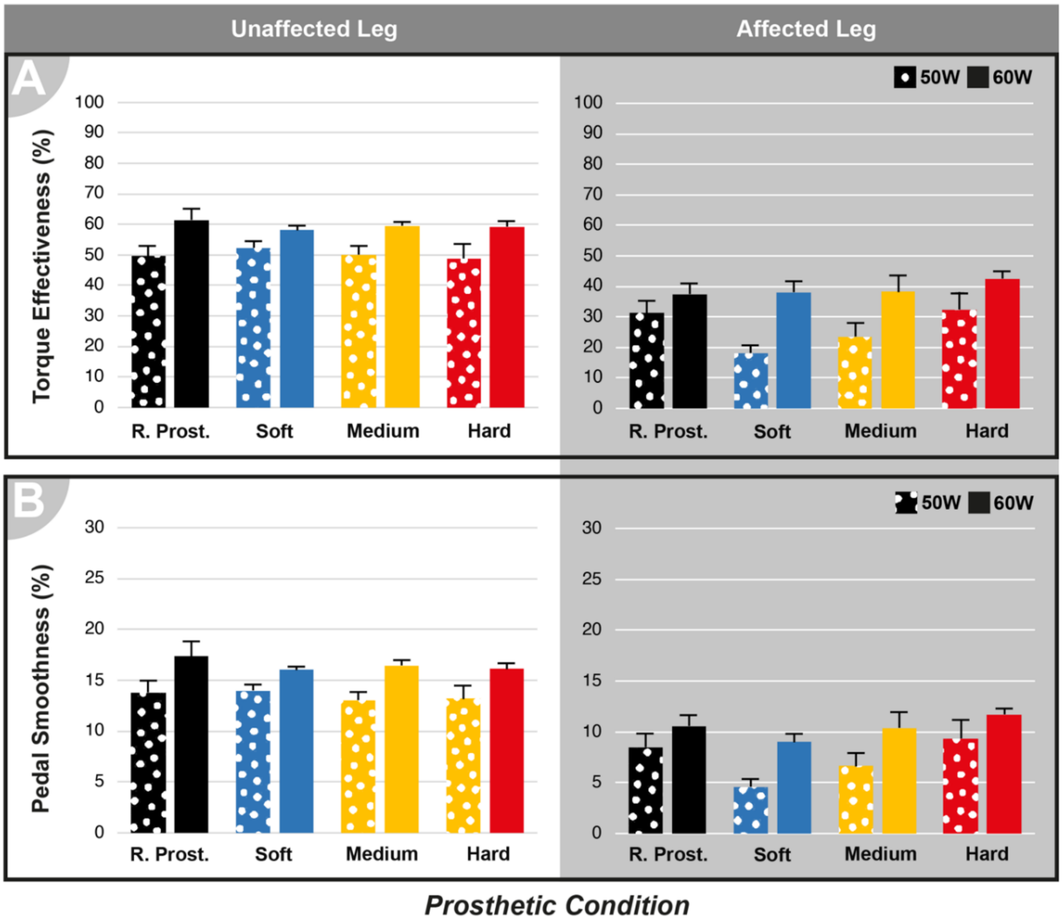


**Figure 3.15.** Unaffected leg and affected leg (A) torque effectiveness and (B) pedal smoothness means for the TA participant. R. Pros.: regular prosthesis.

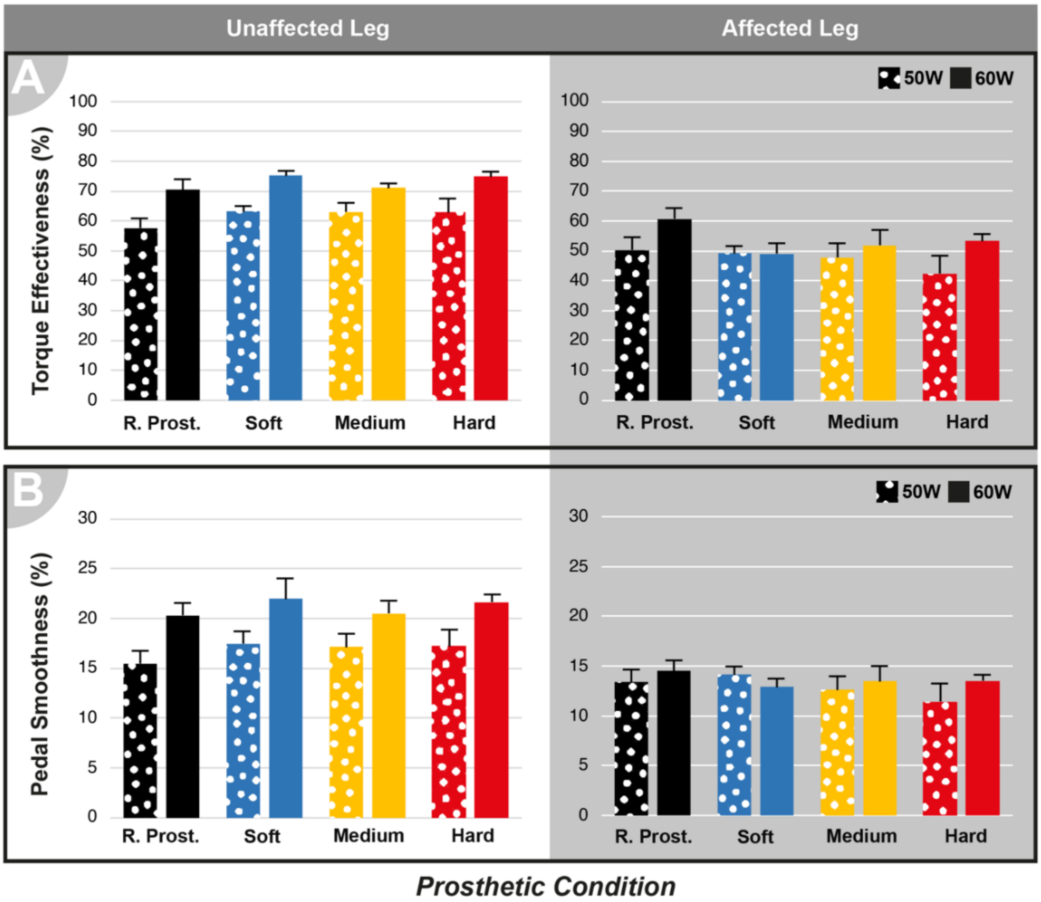


**Figure 3.16.** Unaffected leg and affected leg (A) torque effectiveness and (B) pedal smoothness means for the CA participant. R. Pros.: regular prosthesis.

### *3.3.4. Perceived exertion*

Figures 3.17 (TA) and 3.18 (CA) present the rating of perceived exertion. The TA participant demonstrated consistent levels of perceived exertion across all 50 W conditions, but a more pronounced increase was observed under the S spring condition at 60 W. In contrast, the CA participant reported lower perceived exertion in the S and M spring conditions at 50 W compared to the RP condition, with the H condition at 50 W remaining the same as that of the RP. For the 60 W conditions, the CA participant experienced an increase in perceived exertion under the S and M spring settings compared to that of the RP, whereas the H condition showed a slight decrease in perceived exertion.

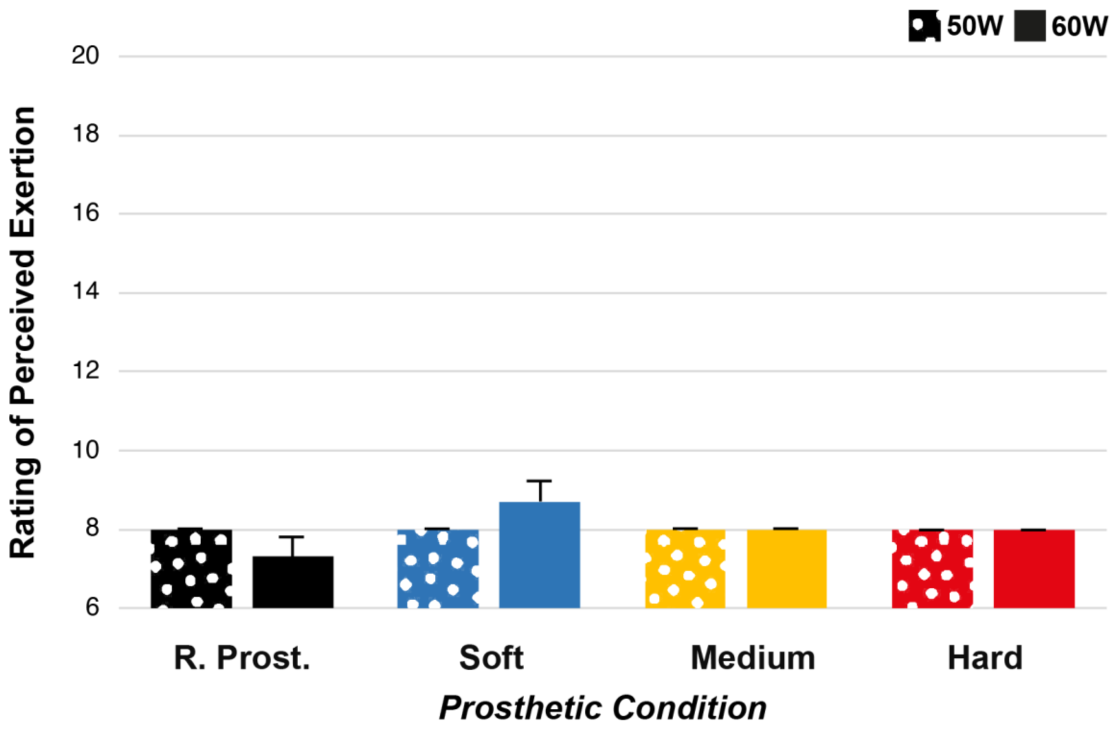


**Figure 3.17.** Rating of perceived exertion means for the TA participant. R. Pros.: regular prosthesis.

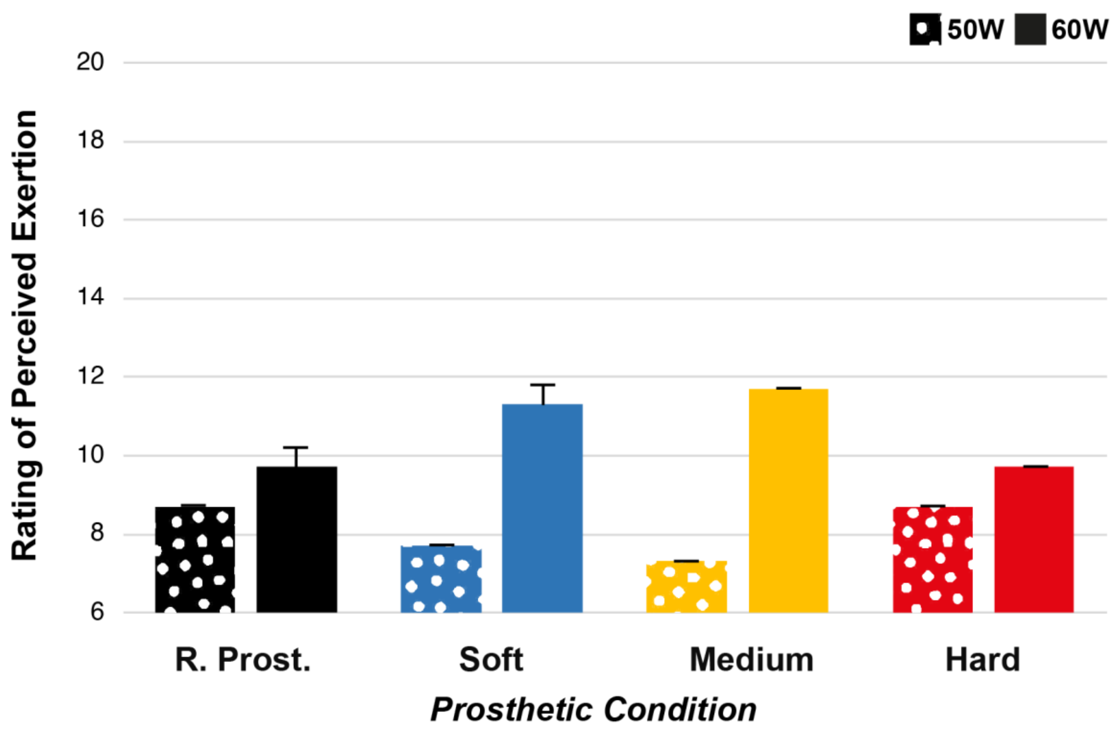


**Figure 3.18.** Rating of perceived exertion means for the CA participant. R. Pros.: regular prosthesis.

## 3.4. Discussion

This study investigated the potential benefits of incorporating an energy-storage component—specifically, a compression spring—into the ankle section of a unilateral transtibial prosthesis for leisure cycling. Unlike previous approaches that often required complete foot redesign (Childers et al., 2011; Tiele et al., 2020), the proposed method only necessitates sufficient clearance in the pylon, thus rendering it compatible with a wide range of conventional prostheses. By introducing an engageable feature, users have the flexibility to activate or deactivate the added ankle movement as needed, potentially improving pedalling efficiency and overall comfort. Such ankle articulation may be crucial for a more natural cycling experience, as it could lead to more symmetric biomechanics between intact and amputated limbs and increased comfort. However, while some similarities were present, the impact of the use of the spring component was dissimilar between the TA and CA participants of this study.

### *3.4.1. Comparison with previous studies*

#### 3.4.1.1. By the author

Both participants exhibited increased hip (Fig. 3.8) and knee (Fig. 3.7) extension at the 180° pedal position in the affected limb when using the prototype prosthesis, consistent with earlier observations (Chapter 2 and Seratiuk Flores et al., 2023) of amplified joint extension at the bottom of the pedal cycle in the absence of ankle movement. In the prototype, this increased extension could additionally be caused by the need to compress the spring at the bottom of the cycle.

In both participants, semitendinosus activity (Figs. 3.9A and 3.10A) decreased in both limbs under prototype conditions, diverging from previous studies (Chapter 2 and Seratiuk

Flores et al., 2023), where semitendinosus activation was higher. This discrepancy may be attributed to the 3D-printed pedal fittings, which prevent the upward pull allowed by clipless pedals. Specifically, in the TA participant's UL (Fig. 3.11A), this phenomenon occurred during the final quarter of the cycle. This decrease suggests a more normalised muscle response, given that it occurs at the same cycle phase where results in Chapter 2 pointed to elevated semitendinosus activity in simulated prosthesis comparisons. Moreover, energy stored through spring compression at the bottom of the cycle may be released during the upstroke, thereby reducing reliance on knee flexion.

#### 3.4.1.2. Other studies in cycling

In the CA participant, the decrease in semitendinosus activity happened with concomitant increases in vastus medialis activation during the initial quarter of the pedalling cycle. This pattern implies a shift in cycling strategy, where the knee joint becomes the principal driver of pedal force, whereas the hip, despite achieving greater extension, remains relatively passive. Such a trend aligns with conventional cycling studies showing reduced hip extensor activity and increased knee extensor/flexor activity at higher loads (Bing et al., 2024; O'Bryan et al., 2014). In the UL, the semitendinosus exhibited reduced activity in only the first quarter of the cycle under all prototype conditions—a finding consistent with research on cyclists with amputations (Watanabe et al., 2020) but contrasting with results for intact cyclists, who typically show proportional increases in semitendinosus activity with load (Sarabon et al., 2012; Silva et al., 2018). These observations may signify unique neuromuscular adaptations in amputee cyclists.

The pedal data for the CA participant indicated that the participant relied more on the UL under higher loads yet achieved superior overall power balance (Fig. 3.14) with the regular prosthesis. Torque effectiveness (Fig. 3.16A) and pedal smoothness (Fig. 3.16B) improved in the UL yet declined in the AL for all prototype conditions, suggesting that the participant compensated with the UL rather than distributing effort evenly when using the prototype. Increased reliance on the sound limb when cycling is common among individuals with unilateral amputations (Childers et al., 2011; Dyer, 2016; Koutny et al., 2013).

### *3.4.2. Effectiveness of introducing the spring and effects of spring constant*

#### 3.4.2.1. Similarities between participants

Despite individual variations, participants shared a unilateral amputation level, leading to specific consistent outcomes; furthermore, the levels of daily physical activity with the prosthesis for both participants were also similar. The prototype prosthesis enabled ankle movement (Figure 3.6), bringing it closer to a biological ankle's motion. Specifically, the spring flexion angle became more acute at 90° of pedal rotation and more obtuse during the third quarter of the cycle. Although the extent of this improvement varied by participant and spring setting, it was consistently apparent. These results align with previous designs requiring a complete foot and ankle overhaul (Childers et al., 2011; Tiele et al., 2020). By contrast, the current prototype replicates aspects of natural ankle kinematics with only minimal modifications.

Regarding muscle activity, vastus medialis activation in the AL increased during the first quarter of the pedalling cycle (Figs. 3.11B and 3.12B), reflecting patterns observed in professional-level cycling (Childers et al., 2009, 2011; Koutny et al., 2013). However, it can be theorised that unlike elite cycling—where elevated vastus medialis activation arises from compensating for absent ankle movement and an increased saddle height—the added spring led to this effect by requiring additional force for deformation.

Instrumented pedal data revealed few similarities between participants. In terms of power balance (Figs. 3.13 and 3.14), the H spring setting produced values akin to those of a regular prosthesis. By contrast, the S spring setting increased power asymmetry between the UL and AL, likely because greater spring movement diverted energy into spring deformation rather than pedal propulsion. Consequently, the stored energy was released neither in an effective nor timely manner, reducing overall efficiency. Elevated perceived exertion under the S spring (Figs. 3.17 and 3.18) further suggests that while increased movement may enhance kinematics, it also imposes higher physical demands on the user.

Although both participants responded differently to the various spring resistances, the findings were broadly consistent across the 50 W and 60 W conditions. This indicates that the mechanical changes introduced by the springs did not produce substantial biomechanical differences at these moderate resistance levels.

**3.4.2.2. Differences between participants**

Individuals with traumatic or congenital amputations may respond differently to prostheses that introduce ankle motion. Traumatic amputation involves sudden limb loss, demanding neuromuscular adaptation to both limb absence and subsequent prosthesis use (Claret et al., 2019). Incorporating ankle motion via a spring mechanism can more closely replicate natural limb function, as evidenced by improvements observed in the TA participant. In contrast, persons with congenital amputation may develop distinct biomechanics, potentially shaped by prolonged prosthesis use from early childhood (Boccia et al., 2020; Eshraghi et al., 2018; Koenig et al., n.d.).

For the TA participant, the prototype improved cycling kinematics compared to a standard prosthesis. Knee (Fig. 3.7A) and hip (Fig. 3.8A) angles remained largely unchanged across conditions. In the ankle (Fig. 3.6A), the M spring setting most closely matched the movement of the UL. Power balance data (Fig. 3.13) indicated that the H spring setting moved power output closer to the symmetry observed with a regular prosthesis, whereas torque effectiveness and pedal smoothness (Figs. 3.15A and 3.15B) demonstrated greater engagement of the amputated limb (AL). This finding is in line with the study of Tiele et al. (2020), which showed that utilising an energy-storing component in a cycling prosthesis led to improved power application with the AL. However, these results and the results of the present study contradict the findings in a study that employed foot-blade deformation for professional-level cycling (Childers et al., 2011).

Therefore, for the TA participant, the softer springs (S and M) resulted in increased prosthesis ankle-level motion at the expense of higher hip and knee extension, ultimately reducing power balance. Hence, the H spring offered a trade-off, with minimal energy loss and reduced perceived exertion (Fig. 3.17). Additionally, the TA participant reported feeling more comfortable using the M and H springs than the standard prosthesis.

In contrast, for the CA participant, adding ankle motion did not enhance the cycling experience. Although the M and S springs induced ankle motion more closely matching that of the UL, they also led to greater knee motion (Fig. 3.7B) in the AL. Additionally, this participant's hip joint was particularly affected by the prototype (Fig. 3.8B). Perceived exertion (Fig. 3.18) showed no improvement over the regular prosthesis.

Two factors may explain these results. First, the CA participant regularly cycles with a rigid prosthesis; hence, introducing ankle movement could disrupt an established pedalling technique. Second, the nature of the congenital amputation, along with prolonged prosthesis use from early childhood, may have led to a distinct acquired body schema (Mayer et al., 2008; Reilly & Sirigu, 2011; Takeuchi et al., 2016) or sensorimotor adaptation (Bramati et al., 2019). Because the participant had never experienced movement in the affected limb, introducing it may have felt unnatural, as mentioned by the participant during the experimental trials.

Another factor potentially affecting these outcomes and the observed differences between participants concerns familiarity with cycling itself. Although the TA participant had cycled before the amputation procedure, he had not resumed cycling post-amputation. By contrast, the CA participant regularly rode with a prosthesis, suggesting a motor adaptation more akin to that of professional cyclists, who rely on proximal leg segments to deliver power (Childers et al., 2011; Koutny et al., 2013). This resemblance is further evidenced by the CA participant's muscle activity patterns, which included an increase in vastus medialis (Fig. 3.13) activity under the prototype, which is characteristic of professional-level cycling. Meanwhile, the TA participant lacked comparable post-amputation cycling experience and may have benefited more from the prototype's added ankle mobility, which demanded an activation pattern closer to that of intact cycling.

Overall, while the TA participant benefited from increased kinematic symmetry, stable levels of perceived exertion, and increased comfort while using the H spring, the CA participant found limited advantage in the prototype.

### *3.4.4. Implications*

Beginning cycling practice can be daunting for individuals with amputations, and creating more comfortable, accessible prosthetic solutions may significantly enhance their experience. The findings presented here indicate that a relatively simple alteration: the addition of an energy-storing spring to the ankle section of a prosthesis, may offer benefits for specific users with amputations. One of the participants showed enhanced comfort and more balanced power distribution when using a stiffer (H) spring, suggesting that this feature could be advantageous for leisure cycling and could potentially have more impactful benefits with its prolonged use.

Nevertheless, results also showed that specific approaches might be needed when incorporating novel prosthetic components into devices that users are already accustomed to employing in daily life. By addressing these considerations, prosthetists and researchers can work towards designing more adaptable prostheses that support both everyday activities and recreational pursuits such as cycling.

### *3.4.5. Limitations*

As a case study, the most significant limitation of the present findings lies in the small participant cohort. In the two specific cases presented here, the generalisability of findings may be influenced by several variables, including the participants' body dimensions, body weight in relation to spring stiffness, and overall cycling experience. Nonetheless, the clear influence of the spring on participants' performance highlights the need for continued investigation in this area. Future investigations should incorporate larger sample sizes to enhance the generalisability of the results. In addition, data were gathered in a controlled setting, whereas leisure cycling ordinarily takes place outdoors on a moving bicycle that requires balance—a factor not accounted for in this study.

Although the prototype successfully evaluated the viability of introducing an additional prosthetic feature, it was employed as an entirely new prosthesis. Outcomes more aligned with the proposed application could be obtained by incorporating the passive spring mechanism into participants' regular prostheses, thus capturing everyday usage conditions. Moreover, learning a novel cycling technique to accommodate the added ankle movement may require more time than was allocated here, indicating that future research should explore long-term adaptation to this component.

## 3.5. Conclusion

Incorporating the spring component led to potentially beneficial outcomes for the TA participant, enhancing power delivery symmetry and partially normalising muscle activity without substantially increasing perceived exertion. These advantages were observed under the stiffest (H) spring setting. Conversely, the softer springs offered no observable benefits, and the CA participant showed no improvements under any spring setting.

# Chapter 4
# Evaluation of Balancing Strategies while Leisure Cycling with Unilateral Transtibial Prostheses

## 4.1. Introduction

One of the main challenges in recovering ambulatory function after a lower-limb amputation and prosthetic fitting is regaining balance. For individuals with unilateral lower-limb prostheses, factors such as postural instability, proprioceptive deficits, cognitive burden, and gait asymmetry—arising from prosthetic adaptations or the prosthesis's inherent characteristics (e.g. movement, weight)—can hinder rehabilitation. Moreover, these users face a higher risk of falls, with incidence rates as high as 52% in community settings (Miller et al., 2001). Such concerns, coupled with diminished balance confidence (Kaufman et al., 2014; Wong et al., 2014), may also impede participation in cycling, an activity heavily dependent on balance. Nonetheless, research indicates that engaging in cycling following disorders that may have affected motor function of the lower limbs can substantially improve overall balance (Baughn et al., 2022; Rissel et al., 2013; Thevarajah et al., 2023). This enhancement, in turn, can play a pivotal role in the broader rehabilitation process, complementing the physiological and functional gains associated with post-amputation recovery.

With the ability to balance as the basis for adopting cycling as a possible leisure activity, having this skill is necessary for engaging in this activity. In a study by Poonsiri et al. (2021), 27% of respondents with unilateral lower limb amputations who did not cycle reported a negative attitude towards cycling, possibly stemming from a fear of falling. Similarly, Carey et al. (2023) found that 94.1% of participants with motor impairments believed they needed assistance with balance and stability before starting leisure cycling. Research targeting the general population has likewise identified motor impairments as a barrier to leisure cycling adoption (Chowdhury & Costello, 2016; Nikitas, 2018; Schneider & Hu, 2015; van Bekkum et al., 2011), underscoring the significance of effective balance strategies in promoting widespread participation.

To my knowledge, no research thus far has directly assessed whether individuals with unilateral transtibial amputations experience worsened balance while cycling. In principle, bicycles self-stabilise as long as their wheels remain aligned (Carvallo, 1901; Lowell & McKell, 1982; Whipple, 1899) and sufficient speed is maintained, yet stability diminishes at lower speeds or following speed disturbances. Riders typically compensate by steering the front wheel or leaning to manipulate their centre of mass (Cain et al., 2016). By this definition, wearing a prosthesis should not lead to worsened balance, because factors such as the weight difference between intact and amputated limbs can be remediated by changes in posture, altering the centre of mass. However, due to impaired balance confidence and proprioceptive deficit, it can be theorised that the cycling balancing task carried out by persons wearing a unilateral prosthesis might be affected.

Therefore, this study investigated whether individuals with unilateral transtibial prostheses exhibit impaired balance during cycling practice. To address this question, participants without amputations were equipped with orthoses simulating prosthetic conditions, utilising the same method as described in Chapter 2. They then cycled on rollers using a cross bicycle, both with and without the orthoses, while performing a perturbation task (gear shifting) that changed wheel speed into two conditions: faster and slower. Measured parameters included the point of contact (POC) between both wheels and the rollers, trunk and bicycle angle, instrumented pedal data on the application of power, and accelerometers placed on the participant's body and on the bicycle, providing insight into any balance changes introduced by simulated prosthetic usage.

## 4.2. Methods

### *4.2.1. Participants*

Ten participants without amputations—five women (age: 27.6 ± 2.2 years; height: 164.9 ± 8.9 cm; weight: 63.1 ± 12.8 kg) and five men (age: 25.2 ± 4.4 years; height: 171.8 ± 3.3 cm; weight: 67.8 ± 12.5 kg)—were recruited. Balancing on rollers required a certain degree of cycling proficiency; therefore, participants were included if they regularly engaged in leisure or commuting cycling (at least once per month). Additional criteria specified an age range of 20–35 years and the absence of any physical conditions that might impair balance or be exacerbated by cycling.

All participants provided written informed consent and completed a health and bicycle-riding questionnaire to confirm eligibility. Leg dominance was determined by asking the question "If you were to kick a ball, which leg would you use?" (van Melick et al., 2017), with eight participants reporting right-leg dominance. Anthropometric data were recorded, including each participant's foot size (25.5 ± 1.4 cm).

This study received ethical approval from the Faculty of Design at Kyushu University (approval number 534).

### *4.2.2. Simulated prosthesis condition*

The same orthoses used in Chapter 2 were used for this study, fitted on the participants' right legs, at a level below the knee. Therefore, the right leg was designated as the affected leg (AL) and the left leg as the unaffected leg (UL) during trials with the prostheses. For further details, consult section 2.2.2.

### *4.2.3. Experimental setup*

Participants cycled on a standard Giant Gravier 2022 cross bicycle (Giant Co., Kanagawa, Japan) equipped with a 24-speed Shimano Atlus EF500 (Shimano Inc., Sakai, Japan) gear-shift system and Shimano EF500 (Shimano Inc., Sakai, Japan) rim brakes on both front and rear wheels. The bicycle was modified to include Assioma Duo power-meter pedals (Favero Electronics Srl., Arcade, Italy), each fitted with a 3D-printed platform enabling foot placement without clipless attachment. For the orthosis conditions, a hook-and-loop fastener was secured the orthosis to the pedal, allowing rapid release if balance was lost.

The bicycle was ridden on rollers positioned 150 mm from a wall on the right side, with a barrier of 69.5 cm in height placed on the left. The right-side wall served as the body support when initiating movement and in the event of imbalance, thus deeming it a safe side, and was used as a reference for results. Achieving stability necessitated first accelerating to maintain momentum, requiring the participant to lean against the wall and keep both hands on the handlebars. Should participants lose balance, they could lean against the wall for full-body support; however, an unexpected tilt could also lead to a left side imbalance. In this situation, the participant could use the barrier to prevent a fall.

To enable motion tracking, a white stripe with a 2-mm red line was marked along the rollers' centre, enabling precise measurement of the point of contact (POC) deviation. Additionally, 8-mm colour-contrasting stripes were painted on the bicycle's tyres for improved visual detection.

Saddle height was determined in accordance with Chapter 2, ensuring that maximum knee extension on the affected side remained within 40°–45°. This criterion resulted in an average saddle height of 61.7 ± 4.2 cm, measured from the saddle's upper surface to the centre of the crank axis. The experiment was conducted in a temperature-controlled environment maintained at 22°C.

### *4.2.4. Experimental protocol*

During the experiment, participants wore shorts and loose-fitting t-shirts, as well as trainers (MW100; New Balance Athletics Inc., Boston, MA, USA) provided in their foot size. Participants were also instructed to wear a helmet. Before data collection commenced, they practised riding the bicycle on rollers for up to 20 minutes, enabling them to gain sufficient confidence and familiarity with this setup. Before mounting the bicycle, participants received specific guidance on roller cycling, which was as follows.

Participants initially leaned against a wall on the right side, maintaining contact until reaching sufficient speed to stabilise the bicycle. During each trial, they were instructed to maintain contact with the wall during the acceleration period, only losing contact once the trial started. They were further instructed to lean on the wall if balance was compromised, designating the right side as a safe side. In the event of more pronounced instability, they could also place a hand on a barrier on the left side. During the trials, participants were instructed to align the front wheel with the central mark on the front roller for as long as possible, thereby centring the bicycle on the rollers.

Each trial lasted 1 minute 30 seconds, with a 30-second initial period dedicated to acceleration. Three trials were conducted for each condition, and data were recorded during the final two trials. Cycling cadence was standardised at 60 rpm using an audio file (Fig. 4.1), which contained a metronome track and announced the trial start, gear change, and trial end, allowing a 10-second countdown for each event. Utilising the formula below, the bicycle speed during trials was determined:

$$speed\ (m/s) = \frac{60 \times \frac{28}{ct} \times 2.164}{60}$$

In this equation, 60 denotes the cadence, 2.164 the wheel circumference, 28 the number of teeth on the chainring, and *ct* the variable related to the number of teeth in the rear wheel cog. The experiment was conducted at a standard speed, utilising the cog with 24 teeth, which was switched to a speed-up condition (21), making balancing easier, or speed-down (28), making balancing more difficult. Hence, all trials began at an initial speed of 2.53 m/s. After 45 seconds of steady cycling, a perturbation was introduced: a randomly assigned gear change either increased (2.87 m/s) or decreased (2.16 m/s) the bicycle's speed, while participants maintained 60 rpm. This standard speed was chosen based on previous research (Cain et al., 2016), where less-experienced participants cycling at 2.58 m/s were still able to maintain balance, whereas at the following lowest speed (1.29 m/s), some participants were unable to finish trials. Participants could briefly deviate by up to ±15% of 60 rpm to maintain balance. Participants were informed of which gear change would need to be performed before starting trials. Data were collected during the stable cycling periods.

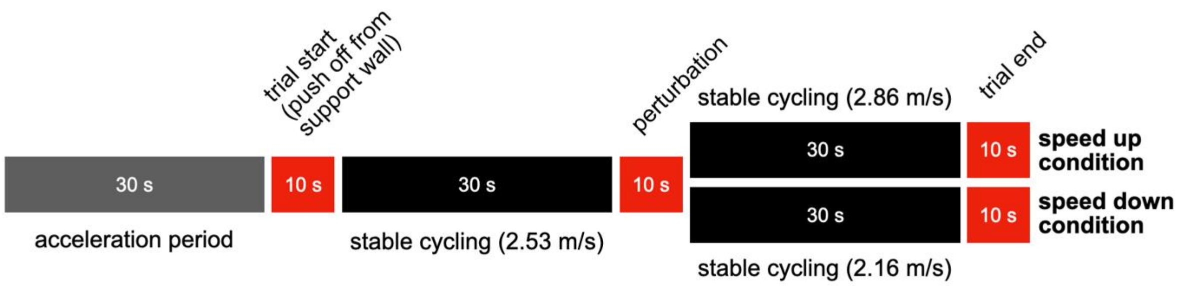


**Figure 4.1.** Trial conditions. Solid black periods denote data collection.

Four conditions in total were carried out, as defined below. While speed shift was randomised, all no-orthosis (NO) conditions were conducted before orthosis (O) conditions.

1. No orthosis, speed-up
2. No orthosis, speed-down
3. Orthosis, speed-up
4. Orthosis, speed-down

### *4.2.5. Measurements*

#### 4.2.5.1. Point of contact

The POC between the centre of each wheel and the centre of the rollers was calculated using two Panasonic HC-300M cameras (Panasonic Co., Osaka, Japan), positioned 2.2 m in front of and 3.9 m behind the rollers. The front camera was at a 17 cm height, and the rear camera was at a 70 cm height. Video footage was recorded at 1080 pixels and 60 fps and then trimmed to the relevant intervals using Adobe Premiere Pro 15.4 (Adobe Inc., San Jose, CA, USA) software. The rollers measured 250 mm in total width, a dimension used to determine the POC location within a motion analysis software tool (Yeoh & Seratiuk Flores, 2023/2024). This software employed a discriminative correlation filter tracker with channel and spatial reliability (Lukežič et al., 2018) to automatically track the POC. Each frame underwent manual verification after the automatic tracking procedure. Figure 4.2 illustrates the front wheel's tracking and reference points.

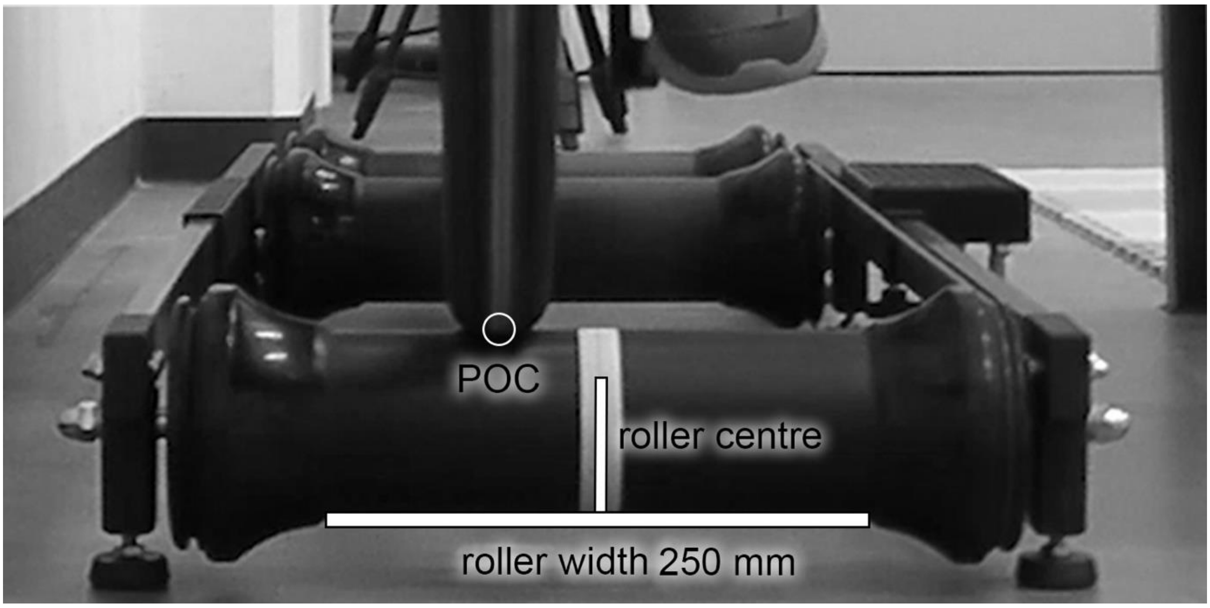


**Figure 4.2.** Tracking points and reference measurements for the front wheel. Same tracking points and references were used for the back wheel.

#### 4.2.5.2. Bicycle and trunk angles

Bicycle and trunk angles were analysed using footage from the rear camera, following the same software and methodology described in Section 4.2.5.1. Adhesive, colour-contrasting markers were placed on the participant's seventh cervical vertebra and on the back of the bicycle saddle, and angles were then measured as illustrated in Figure 4.3.


**Figure 4.3.** Trunk and bicycle angles.

#### 4.2.5.3. Accelerometer

Two wireless accelerometer units (MuscleBAN BLE, Biosignalsplux, PLUX Wireless Biosignals S.A., Lisbon, Portugal) were used for data collection. One was positioned on the centre of the breastbone, affixed directly on the participant's skin, with care taken to avoid contact with the t-shirt worn during trials; this was named the body accelerometer. The second accelerometer was fixed on the top of the bicycle's top tube, ensuring it was not inadvertently touched during cycle practice; this was deemed the bicycle accelerometer. Data were recorded at 200 Hz using OpenSignals version 2.2.5 software. Raw data were converted to g-forces within the software. With the goal of understanding the effects of stabilisation on the rollers, data were only collected for the x-axis (lateral variation).

#### 4.2.5.4. Instrumented pedals

Pedal data were collected as described in section 2.2.6.3 but only utilising one parameter: left and right balance.

### *4.2.6. Statistical analysis*

The Shapiro–Wilk test was conducted to assess data normality. The results indicated deviations from normality for different parameters under the simulated prosthesis conditions: the mean POC of the front wheel and rear wheel, and the standard deviation of accelerometer data from both the body and the bicycle. Consequently, a Wilcoxon signed-rank test (SPSS Statistics, version 21.0; IBM Co., Armonk, NY, USA) was employed to determine whether significant differences existed between orthosis and non-orthosis conditions. The analysed variables included the means for front and rear wheel POC, bicycle and trunk angle, and power applied to the pedals during the stable cycling phases. For accelerometer data, mean standard deviation was used to investigate the effects of data variability during the cycling practice. Statistical significance was set at $\alpha = 0.05$.

## 4.3. Results

### *4.3.1. Point of contact*

#### 4.3.1.1. Front wheel

Figure 4.4 presents the POC for the front wheel. Although results remained broadly consistent across all conditions—with no significant differences identified by the Wilcoxon test—visual inspection shows that the O conditions displayed slightly higher variability. This is evident in the larger standard deviations (20.02–23.06 mm) compared to those under the NO conditions (16.80–20.84 mm). Prior to the speed perturbation, the O conditions also exhibited mean POC values slightly closer to the roller's centre.

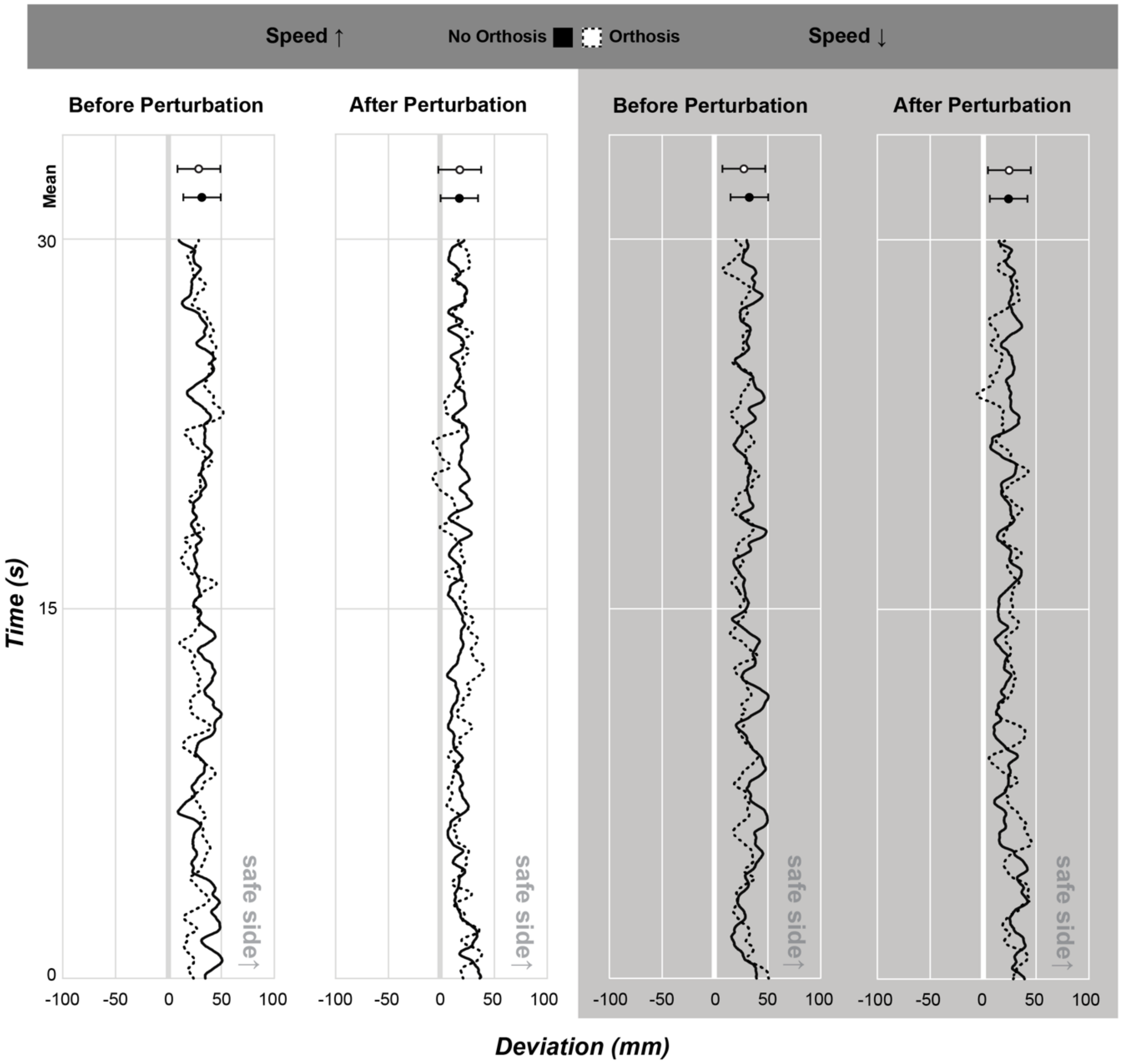


**Figure 4.4.** Point of contact between the front wheel and the roller throughout the trial and mean values.

#### 4.3.1.2. Rear wheel

Rear wheel POC means (Fig. 4.5) were similar to the front wheel counterpart but exhibited overall lower variability, and no significant differences were found between O and NO means in the Wilcoxon test. Visual inspection shows that the O conditions achieved higher lateral movement variability (19.52–21.64 mm standard deviations), whereas the NO conditions demonstrated less variability (15.01–18.92 mm). Before the speed perturbation, the O conditions maintained a POC slightly closer to the roller's centre. After the perturbation, both NO and O conditions converged, displaying similar distances from the roller.

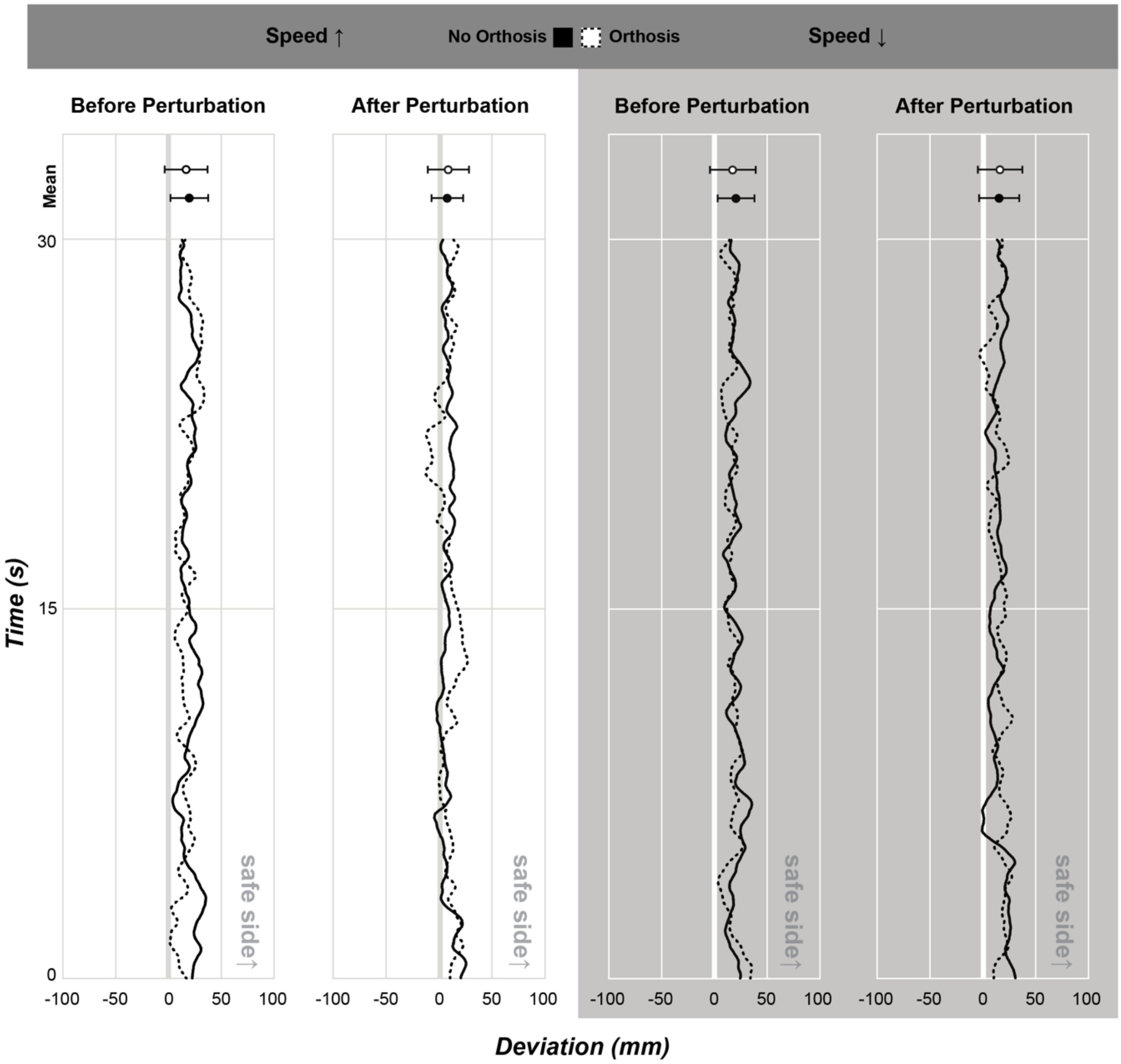


**Figure 4.5.** Point of contact between the back wheel and the roller throughout the trial and mean values.

## *4.3.2. Angles*

### 4.3.2.1. Bicycle angles

Figure 4.6 illustrates bicycle angles under different conditions. O conditions showed a more acute angle overall, with the bicycle tilting more toward the safe side. The Wilcoxon test showed a significant difference between NO and O conditions before the perturbation in the speed-up condition ($W = 1$, $z = -2.70$, $p < 0.01$). Moreover, O conditions displayed reduced variability, with standard deviations ranging from 0.49° to 0.62°, whereas NO conditions exhibited larger fluctuations (0.60°–0.83°).

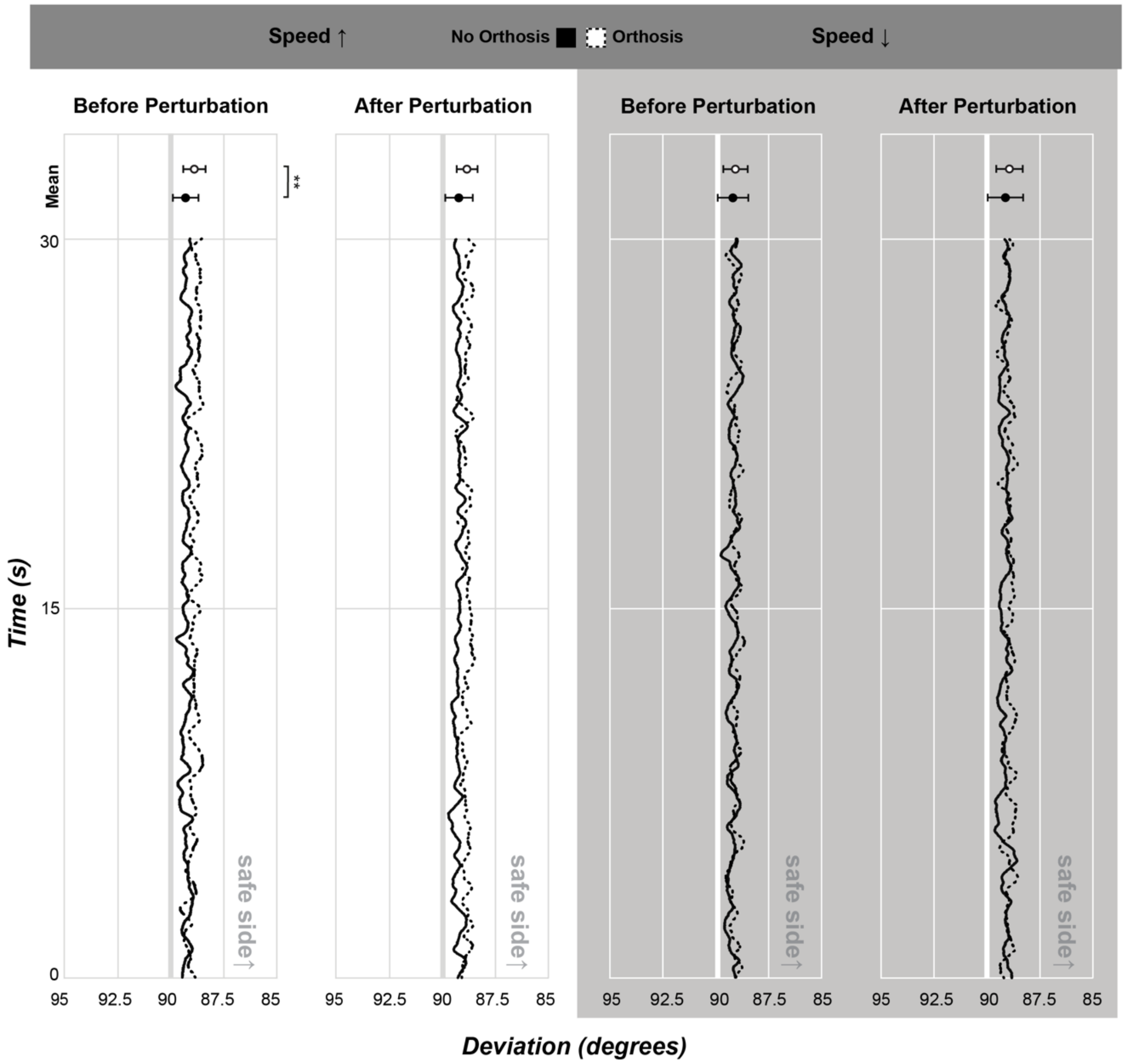


**Figure 4.6.** Bicycle angles throughout the trial and mean values. **p < .01.

### 4.3.2.2. Trunk angles

Unlike the bicycle angles, trunk angles (Fig. 4.7) exhibited higher variability and tended to remain oriented away from the safe side. The Wilcoxon test revealed no significant differences between O and NO conditions. However, visual inspection indicates greater standard deviation in the speed-up scenario under O conditions, whereas the speed-down scenario showed lower standard deviations but more pronounced peaks over time.

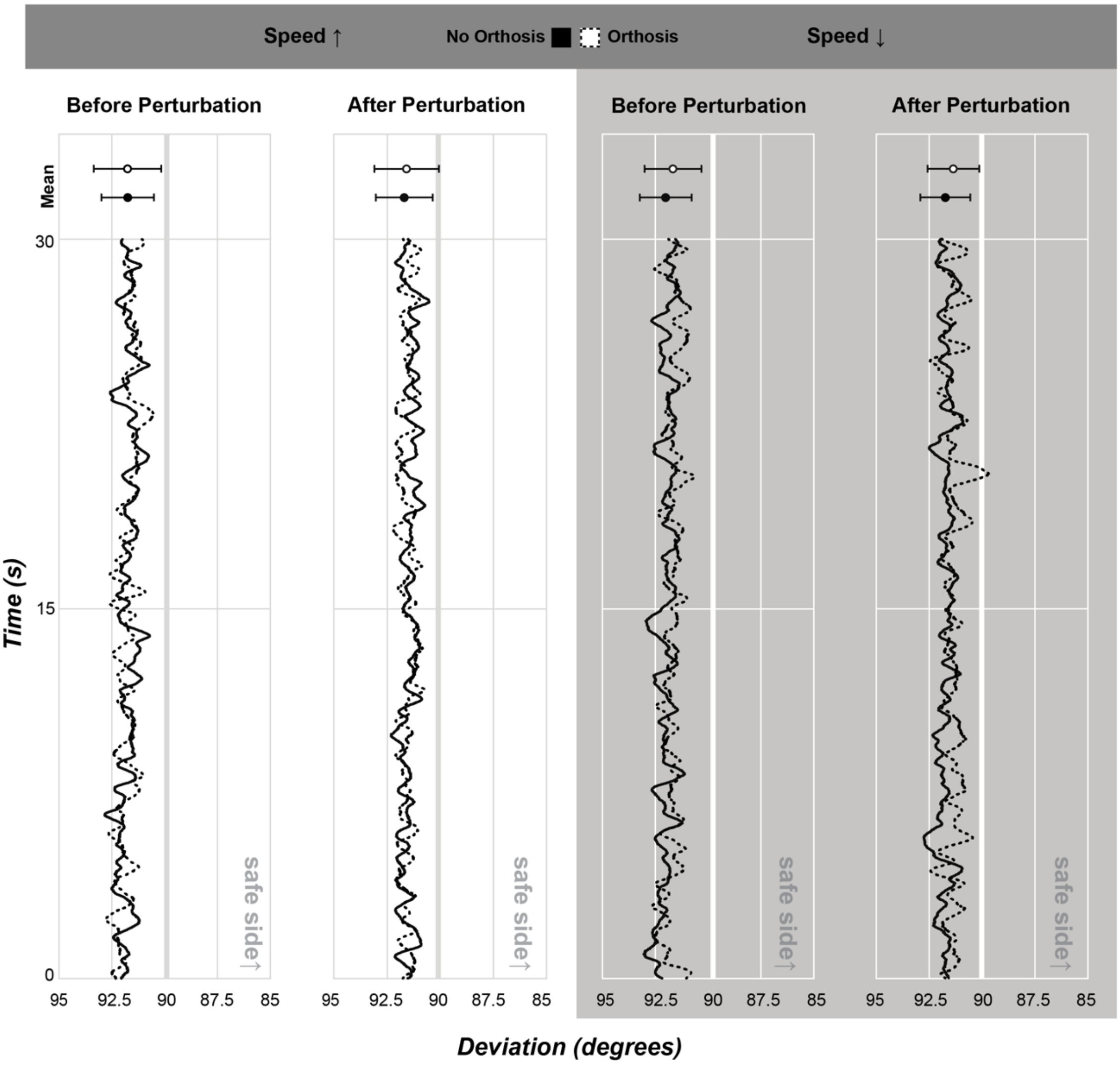


**Figure 4.7.** Trunk angles throughout the trial and mean values.

### *4.3.3. Accelerometer*

#### 4.3.3.1. Body

Figure 4.8 displays box plots of the mean standard deviation recorded by the body accelerometer. Overall, no significant differences were shown between O and NO conditions in the Wilcoxon test, and results were broadly consistent across both speed-up and speed-down scenarios. Visual inspection indicates a slight elevation in the distribution of mean values following the perturbation, yet the overall trends remained comparable between conditions.

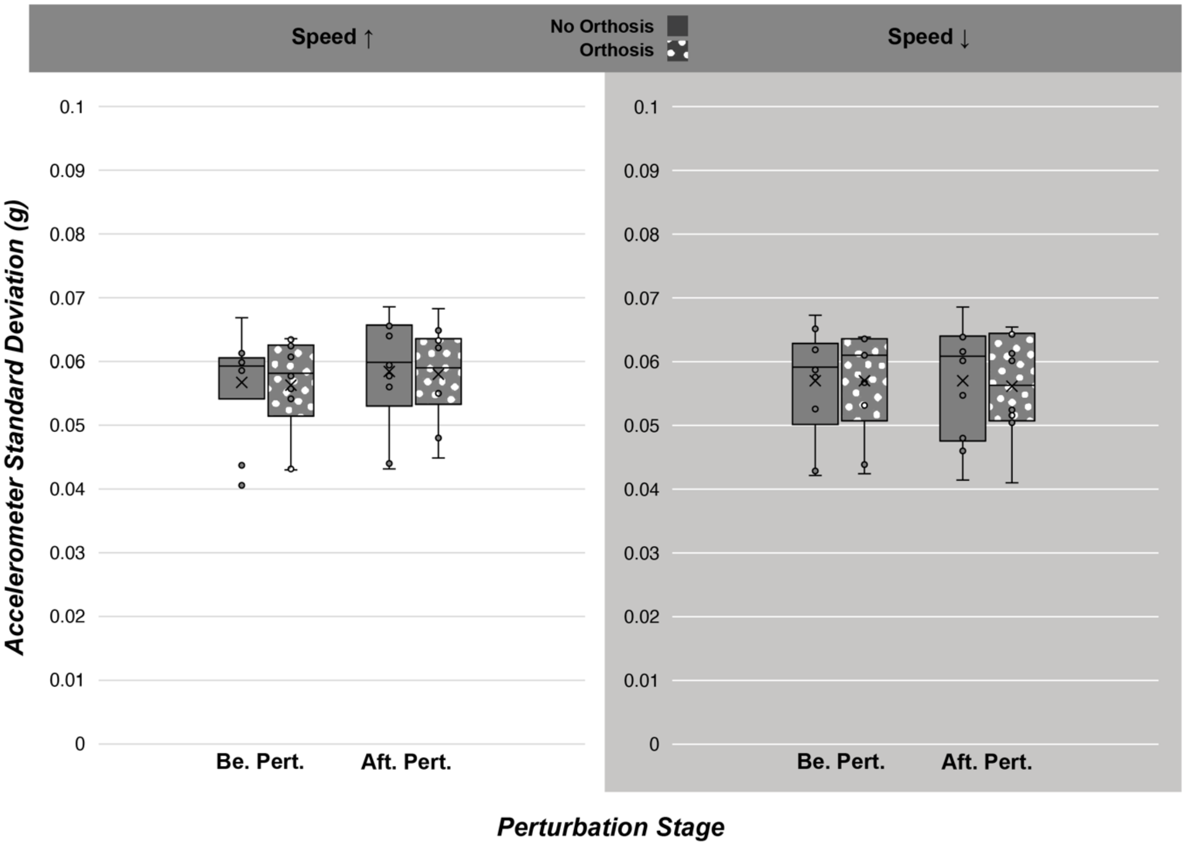


**Figure 4.8.** Mean standard deviations for the body accelerometer. Be. Pert.: before perturbation, Aft. Pert.: after perturbation.

### 4.3.3.2. Bicycle

Figure 4.9 presents box plots of mean standard deviation for bicycle accelerometer data. The Wilcoxon test detected no significant differences between the O and NO conditions. Nevertheless, visual inspection indicates that, while the results were broadly comparable before the perturbation, the speed-down scenario exhibited reduced variability. Overall, O conditions in the speed-down trials yielded the highest distribution of means.

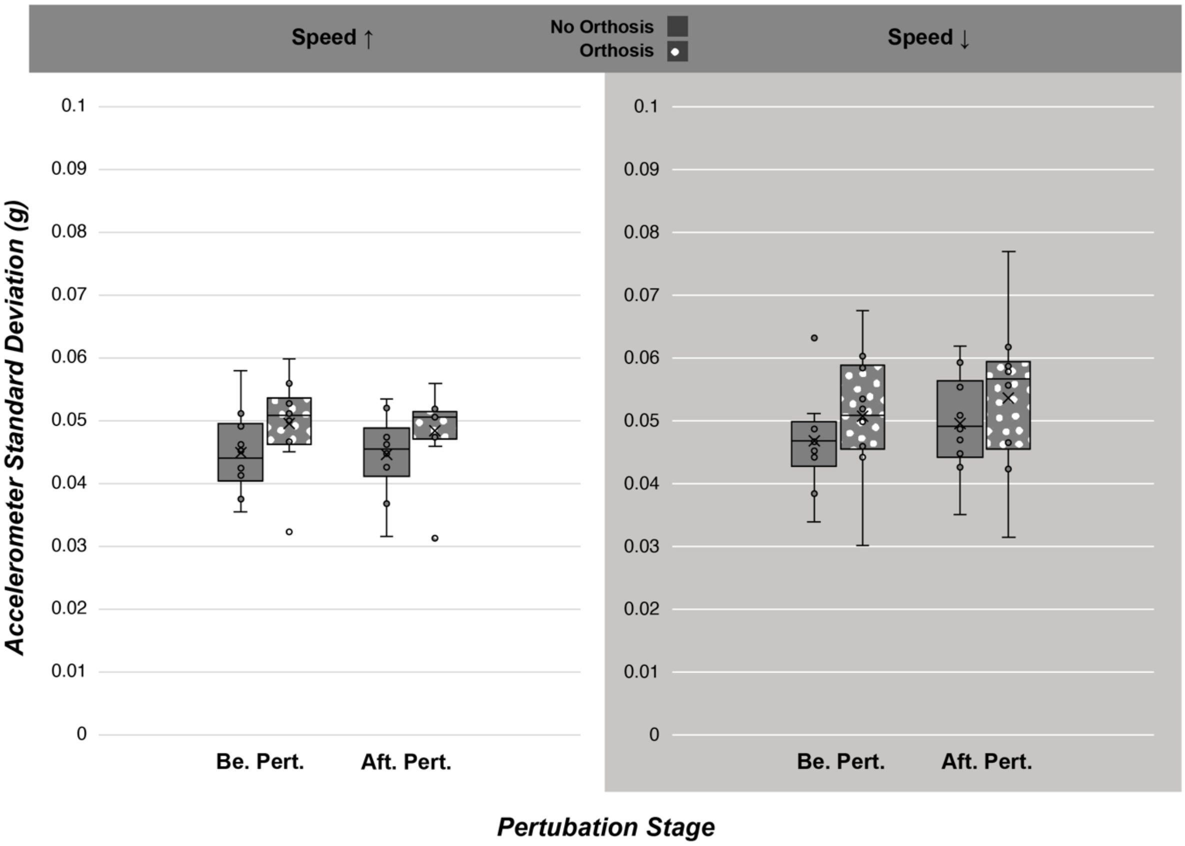


**Figure 4.9.** Mean standard deviations for the bicycle accelerometer. Be. Pert.: before perturbation, Aft. Pert.: after perturbation.

### *4.3.4. Instrumented pedals*

Figure 4.10 presents the power balance for the left (unaffected) and right (affected) legs. The Wilcoxon test revealed significant differences between orthosis O and NO conditions in all comparisons (speed-up, before perturbation: $W = 0$, $z = -2.81$, $p < 0.01$; speed-up, after perturbation: $W = 0$, $z = -2.80$, $p < 0.01$; speed-down, before perturbation: $W = 1$, $z = -2.70$, $p < 0.01$; speed-down, after perturbation: $W = 3$, $z = -2.50$, $p < 0.05$). Visual inspection indicates broadly similar power balance across speed-up and speed-down trials. However, in the O condition, the speed-down scenario showed the greatest imbalance, with most of the power produced by the unaffected leg.

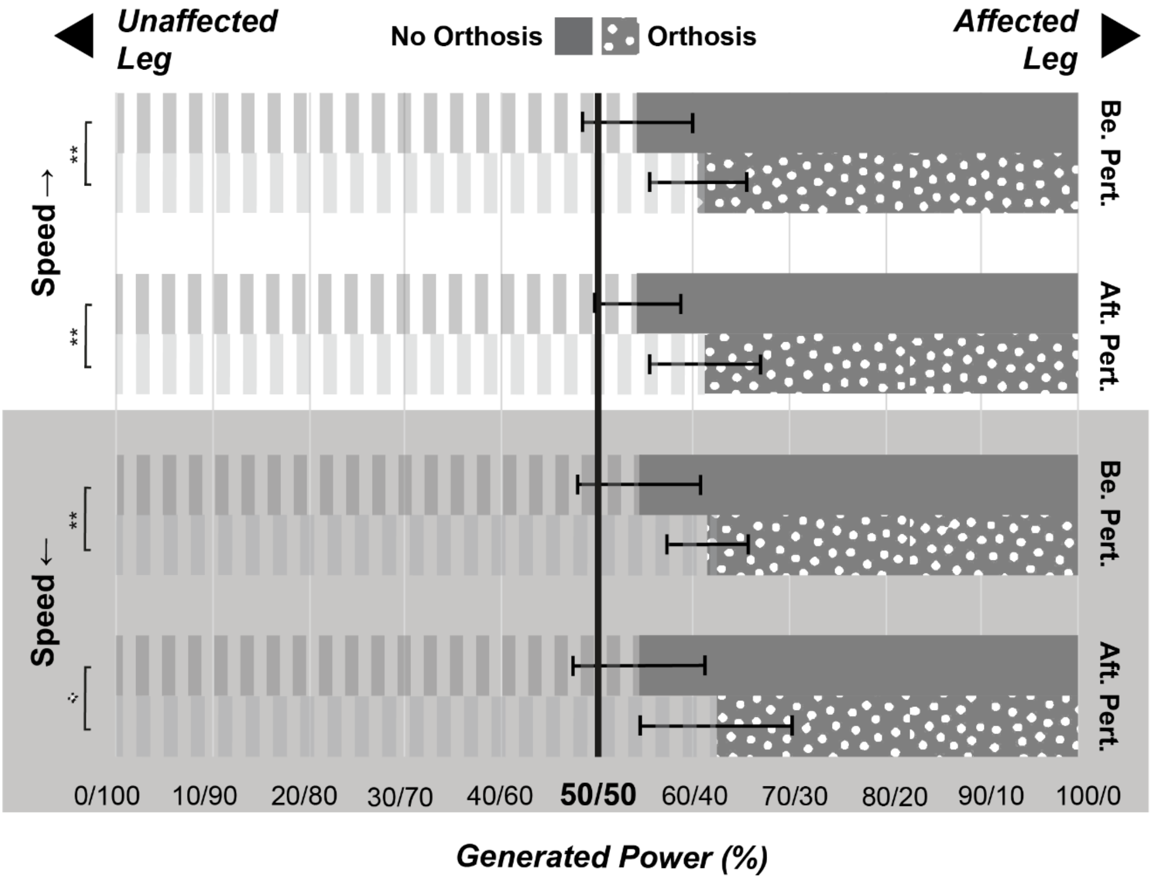


**Figure 4.10.** Generated power percentage for the left (unaffected) and right (affected) legs. Be. Pert.: before perturbation, Aft. Pert.: after perturbation.

## 4.4. Discussion

This study examined whether simulated unilateral transtibial prostheses affect balance while cycling. Although no research has directly assessed balance in prosthetic cycling, individuals with motor impairments often cite balance as a barrier to participation (Carey et al., 2023; Chowdhury & Costello, 2016; Nikitas, 2018; Schneider & Hu, 2015). Balance issues are also common among prosthesis users (Kaufman et al., 2014; Wong et al., 2014), making it both a practical concern and a psychological barrier to recreational cycling. Due to the lack of studies on balance disturbances during cycling, these findings will be compared to gait and standing balance research.

### *4.4.1. Effects of speed perturbation in intact cycling*

The speed-up and speed-down perturbation conditions led to some specific coincidental effects due to their nature, reflecting how alterations in riding speed drove participants to modify their balance strategies. While maintaining a cadence of 60 rpm, reducing speed made balancing more challenging, whereas increasing speed facilitated stability. Although no formal statistical comparisons were performed within these conditions, certain effects emerged, showing that a level of difficulty was effectively introduced through the perturbation.

An examination of the mean POC for the front (Fig. 4.4) and rear (Fig. 4.5) wheels showed a slight shift towards the centre of the roller following both speed-up and speed-down perturbations. This effect was especially pronounced for the rear wheel under NO conditions. One possible explanation is that participants grew more confident as each trial progressed. Alternatively, enhanced concentration post-perturbation could also have contributed, mirroring findings in gait research (Alizadehsaravi et al., 2022; Kurz et al., 2016; Ribeiro de Souza et al., 2023).

Regarding specific differences between speed-up and speed-down perturbations, bicycle accelerometer data (Fig. 4.9) recorded greater variability following the speed-down perturbation. However, the bicycle (Fig. 4.6) and trunk angles (Fig. 4.7) were largely unchanged between pre- and post-speed-down perturbation phases, as were body accelerometer standard deviations (Fig. 4.8), implying a steering-based strategy for balance when balancing became more difficult. Similar patterns have been observed in previous research (Cain et al., 2016), where less-experienced riders—comparable to leisure cyclists—employed more steer control rather than adjusting trunk or bicycle angles. However, in that study, steer control was employed more by the less-experienced riders at faster speeds.

### *4.4.2. Effects of simulated prosthesis conditions*

#### 4.4.2.1. Effects of simulated prosthesis conditions before speed perturbation

While the speed perturbation was designed to intensify potential effects of prosthesis use, differences between the O and NO conditions were evident even before the speed change. Under O conditions, the front (Fig. 4.4) and rear (Fig. 4.5) wheel POC were generally closer to the roller's centre, particularly before the speed perturbation. Nonetheless, POC variability increased across all conditions, suggesting higher instability. As previously noted, the perturbation may have prompted a more central POC by increasing concentration, leading participants to increase cognitive load in order to maintain balance while cycling with an impaired limb (Alizadehsaravi et al., 2022; Kurz et al., 2016; Ribeiro de Souza et al., 2023). This effect, although expected in a controlled laboratory setting where the primary task is balancing on the rollers, could pose considerable challenges in everyday leisure cycling—particularly in urban environments that demand divided attention (Barra et al., 2006).

Although the POC remained closer to the roller's centre, balance strategies under O conditions shifted slightly, as evidenced by changes in bicycle (Fig. 4.6) and trunk (Fig. 4.7) angles. Participants tended to lean more towards the safe side, suggesting heightened uncertainty about maintaining stability. This shift in posture reflects a well-documented phenomenon, in which deliberate attempts to preserve balance result in increased postural deviation (Masters & Maxwell, 2008; Parr et al., 2024; Wulf, 2013). Consequently, the altered bicycle and trunk angles reinforce the POC findings, implying that participants adopted more cautious balancing strategies under O conditions. Additionally, the differences between NO and O for these parameters varied between speed-up and speed-down conditions before these perturbations, despite both trials commencing at the same speed. Such findings suggest an anticipatory effect, as participants were informed of the speed condition to be carried out.

Anticipation has likewise been shown to influence results in gait research involving individuals with motor impairments (Batcir et al., 2021; Le Mouel et al., 2019; Tajali et al., 2018).

Finally, the power output (Fig. 4.10) between the intact and simulated prosthetic legs worsened under all O conditions before the perturbation and remained largely unchanged after. This demonstrates the shifting of force production toward the sound limb. These findings align with earlier research (Seratiuk Flores et al., 2023) and parallel the observations detailed in Chapter 2. Power distribution between affected and unaffected legs at lower saddle heights also matched the patterns seen in Chapter 2, as O conditions displayed left–right balance levels akin to those achieved at lower saddle heights. This outcome suggests that results obtained using an ergometer can be transferred to a static bicycle requiring active balance.

#### 4.4.2.2. Effects of speed perturbation on simulated prosthetic conditions

As discussed thus far, wearing the orthosis introduced a modest increase in instability during the cycling task. Nevertheless, the instability was further exacerbated, when compared to NO levels, at the speed-down O conditions.

Despite participants' increased trunk lean (Fig. 4.7), body accelerometer data (Fig. 4.8) remained broadly comparable before and after the perturbations. By contrast, bicycle accelerometer results (Fig. 4.9) revealed heightened instability: while speed-up trials exhibited mostly consistent variability, speed-down trials registered greater standard deviations under O conditions. This additional variability emerged even before the perturbation and may relate to the significant difference seen between O and NO conditions for bicycle angle (Fig. 4.6) in speed-up trials—a difference that did not appear in speed-down conditions.

Power output symmetry (Fig. 4.10) remained largely unchanged across speed-up and speed-down trials. However, the speed-down scenario displayed elevated standard deviations, indicating increased instability at reduced speeds. Altogether, these findings suggest that higher speeds can maintain pre-perturbation stability, whereas lower speeds further exacerbate imbalance under O conditions, compared to NO. This phenomenon underscores the potential difficulties individuals with unilateral transtibial prostheses may face when starting, stopping, or cycling slowly—activities that demand careful balance management.

### *4.4.3. Implications*

Although individuals with amputations frequently exhibit impaired gait balance, while persons with other motor impairments express concerns about balancing on bicycles, the current findings indicate no statistically significant difference in cycling balance ability between prosthetic and non-prosthetic conditions. Some compensatory strategies that differed from intact cycling were observed, particularly at lower speeds; however, overall stability remained intact. Nonetheless, this adaptation may be influenced by a fear of falling, owing to diminished motor capacity, which affects cyclists' confidence and balance.

From an applied perspective, these results underscore the increased cognitive load associated with prosthetic use, which can be a barrier for the adoption of cycling as a leisure activity. This heavier cognitive load caused by prosthetic use during cycling might be a barrier to its practice in complex environments, such as urban areas, where attention needs to be divided. This suggests that when advising individuals with prostheses, clinicians should consider not just their physical capability but also the cognitive load involved in balance and navigation while cycling in everyday settings.

### *4.4.4. Limitations*

Similar to the method used in Chapter 2, this study employed orthoses to simulate prosthetic conditions. While this approach enables controlled results, actual prosthesis users exhibit distinct motor adaptations that orthoses cannot fully replicate (Childers et al., 2014). The use of the orthosis also infers an increased weight on the affected limb, of about 800 g to 1 kg, whereas protheses are often lighter than biological limbs. However, about 70% of persons with amputations report perceiving their prosthetic limb as heavier than their biological limb (Handy Eone et al., 2018), probably due to impaired muscle function. Therefore, while the orthoses would have led to an accurate perceived weight disparity between intact and affected limbs, physically, this was not simulated.

Additionally, participants in this study were younger than the typical demographic of lower-limb prosthesis users who cycle (62.0 ± 13.0 years) (Poonsiri et al., 2021); however, involving older adults might introduce additional balance issues related to ageing. Another limitation is that while the study provided some insight into the presence of balancing challenges while cycling with prostheses, it focused on stabilised cycling at a relatively narrow speed range. This excluded other critical phases such as mounting, initiating movement, and stopping. A broader spectrum of speeds would help in more accurately characterising the balance challenges faced by unilateral prosthesis users. Additionally, exploring these challenges in an urban context could lead to significant insights into the barriers that might hinder individuals with unilateral transtibial amputations from adopting cycling as a leisure activity.

## 4.5. Conclusion

In conclusion, although using unilateral lower-limb prostheses can introduce specific balancing challenges, it did not substantially impair participants' cycling balance under simulated prosthetic conditions. Nevertheless, the study identified important considerations, including the elevated cognitive load associated with prosthesis use and the anticipation of potential balance difficulties.

# Chapter 5
# General Discussion

## 5.1. Main findings

Cycling for leisure can greatly enhance quality of life and facilitate rehabilitation for individuals with unilateral transtibial amputations. Previous biomechanical studies on professional cycling with prostheses have identified potential adaptations that may be necessary for recreational cycling, particularly for managing asymmetries between unaffected (UL) and affected (AL) legs (Childers et al., 2009, 2014; Childers & Gregor, 2011; Dyer, 2016; Koutny et al., 2013). However, research specifically examining cycling at lower power output levels remains limited. Consequently, specialised cycling products for this population are limited. Moreover, qualitative research reveals that many prosthesis users harbour reservations about using bicycles and integrating cycling into their routines (Poonsiri et al., 2018, 2021). To address this, the present thesis explores factors related to the bicycle, prosthesis, and cycling setting, with the aim of informing targeted interventions that support leisure cycling among individuals with unilateral transtibial amputations.

The first study examined the role of bicycle saddle height in leisure cycling, comparing intact cycling conditions and a simulated prosthetic condition. The objective was to identify saddle height adjustments that would render cycling with a simulated prosthesis more akin to regular cycling, diminishing asymmetries between lower limbs. As a main finding, lowering the saddle to a maximum knee extension in the AL of 37°–45° resulted in knee and hip joint movement resembling that of intact cycling. Additionally, muscle activation in the semitendinosus, gastrocnemius, and vastus lateralis became more aligned with intact cycling patterns. Another principal finding showed that the higher saddle heights—previously assumed to promote better joint engagement—in fact exacerbated power asymmetries, disrupted muscle activity, and increased perceived exertion.

The second study examined the impact of introducing ankle mobility to an otherwise rigid prosthesis through springs with varying spring constants attached onto the pylon. Two transtibial prosthesis users participated—one with a traumatic amputation, the other with a congenital amputation—differing in the cause of limb loss and duration of prosthesis use. The main finding revealed that incorporating a stiffer spring could benefit cycling performance and comfort by promoting a semitendinosus activation pattern closer to that of intact cycling, alongside enhanced power symmetry between the UL and AL. However, the congenital amputation participant did not experience similar improvements, highlighting the importance of individualised approaches in prosthetic design and function, particularly in light of how long an individual has used a prosthesis.

The third and final study explored the potential balance challenges associated with cycling using unilateral transtibial prostheses. As in previous investigations, data were compared between intact and simulated prosthetic conditions. Individuals with lower-limb prostheses often experience increased fear of falling and reduced confidence in their balance, frequently perceiving cycling as challenging and risky. However, this study found no significant difference in balancing ability between intact cycling and simulated prosthetic conditions once the cycling motion was established. Nonetheless, another key finding indicated a slight increase in instability alongside an altered strategy for bicycle and trunk leaning under simulated prosthetic conditions, possibly pointing to the need for further investigation.

## 5.2. Implications

The main aim of the research carried out in this thesis is to inform the design and settings of selected components used by individuals with unilateral transtibial amputations when leisure cycling. Consider, for example, a hypothetical case of a female patient who underwent a unilateral traumatic transtibial amputation one year earlier and is now adjusting to daily life with her prosthesis. This section explores her potential challenges that may arise in the case she decides to start leisure cycling, illustrating how the findings of this thesis may collectively enhance her experience, the equipment she uses, and the support she receives from healthcare professionals. Figure 5.1 visually represents her journey.

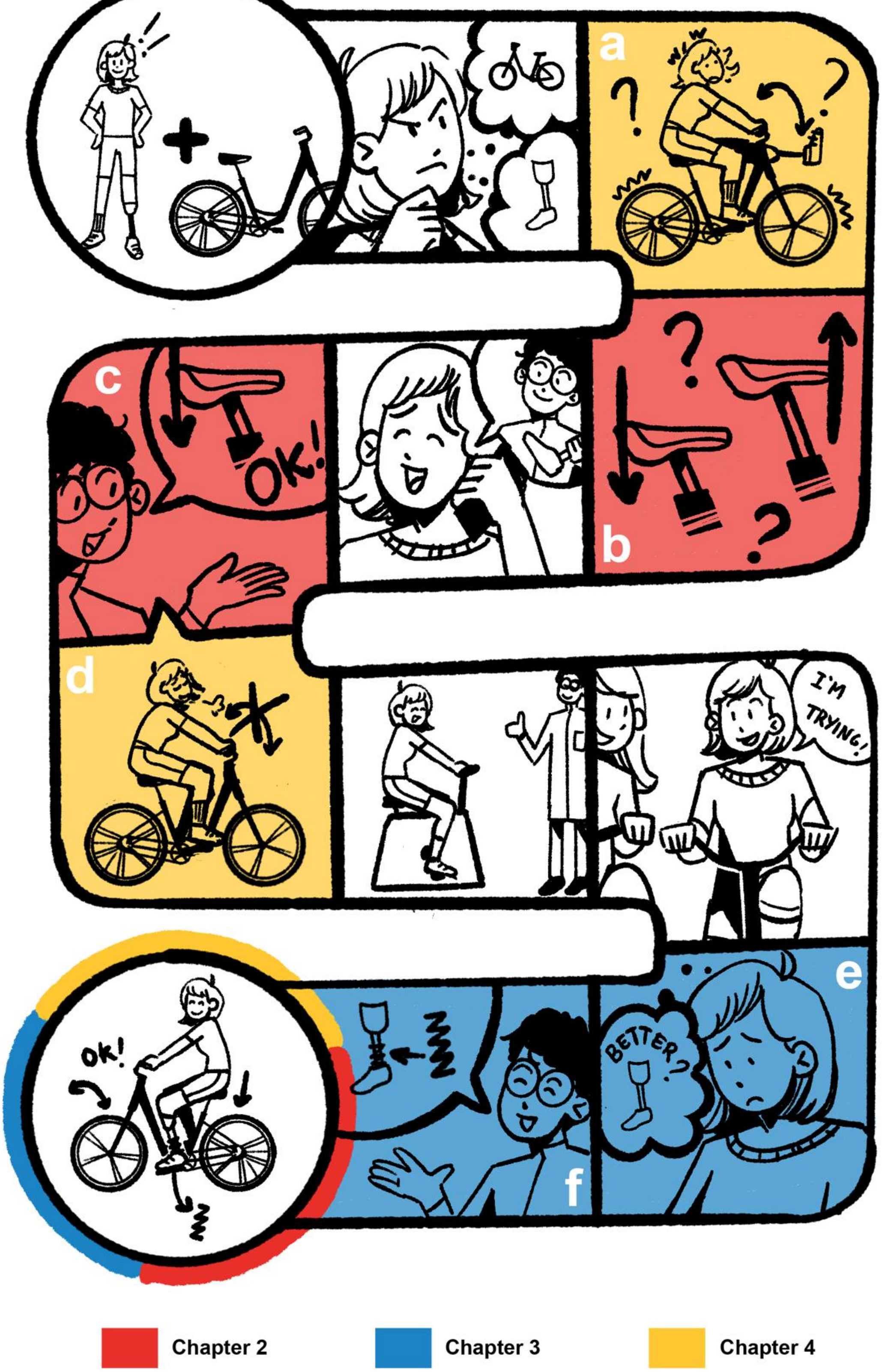


**Figure 5.1.** An example unilateral transtibial prosthesis user journey and its interactions with the findings of this thesis.

Initially, this patient, feeling somewhat limited by her new transtibial prosthesis, contemplates taking up leisure cycling—an activity she previously enjoyed—as she owns a typical city bicycle that she once used for short errands and park visits. Recognising the health benefits of cycling, she hopes it may also improve her social engagement by participating in an activity her friends without amputations regularly enjoy. Online research yields minimal guidance beyond information about professional cyclists with amputations who wear specialised prostheses, but regarding leisure cycling, she discovers a recommendation for mid-foot pedal placement (Seratiuk Flores et al., 2023).

Reassured by this information, she nonetheless remains concerned about potential balance issues (Fig. 5.1A). Her walking balance is already affected compared to her pre-amputation state, and she imagines that cycling may exacerbate this instability. Further research indicates that a lower saddle height can improve balance, yet she also encounters warnings about possible increased knee stress and general guidelines for persons without amputations indicating the benefits of higher saddle heights (Fig. 5.1B). Unsure, she consults her prosthetist, who has overseen her adaptation to the prosthesis.

During the appointment at the clinic, the prosthetist, aware of this thesis's research findings, explains that lowering the saddle height does more than simply lower the bicycle's centre of gravity, improving balance at lower speeds and manoeuvrability: it can also enhance comfort by minimising asymmetries between the intact and amputated limbs (Fig. 5.1C). Crucially, a lower saddle height also enables her to place both feet on the ground when initiating and halting the bicycle—an appealing feature, given her previous worry about the possibility of falling (Fig. 5.1D). The prosthetist notes that such a setup may replicate her pre-amputation cycling experience in terms of muscle activation and joint motion. He further explains that research into balancing with a unilateral prosthesis on a bicycle at a lower saddle height has revealed no significant deviations from cycling without a prosthesis, although a slight increase in instability remains possible (Fig. 5.1D).

She then tests cycling on the clinic's ergometer, with the saddle adjusted such that her prosthetic knee reaches a maximum extension of approximately 40°. Although the saddle is lower than what she recalls from her city bicycle, she finds the position comfortable. At her request, the prosthetist temporarily raises the saddle for comparison, leading her to notice greater asymmetries between her lower limbs. She remains somewhat anxious about potential issues during mounting and dismounting the bicycle, starting, stopping, and coping with reduced tactile feedback on her prosthetic foot, but these trials and adjustments bolster her confidence.

The following day, she decides to attempt outdoor cycling. After a brief period of adaptation, she finds the activity manageable and can now join friends on leisure rides, boosting her self-confidence and facilitating ongoing rehabilitation while engaging in physical exercise. Nonetheless, she still notices differences from how pre-amputation cycling felt, especially as she increases her riding frequency (Fig. 5.1E). During her next appointment, she expresses these concerns to the prosthetist, who suggests a minor pylon modification—a spring attachment at the ankle level that can be engaged during cycling (Fig. 5.1F). After practicing on the clinic's ergometer to try different spring settings, followed by a few days of outdoor cycling, she reports increased comfort. This is due to a more natural feeling regarding how the muscles in her residual limb and sound leg are being engaged, and how her joints move in her prosthetic leg, bringing it closer to what she remembers from before her amputation. This development further cements her decision to continue cycling for leisure and commuting, as well as for maintaining social connections.

Although this example is deliberately idealised, it underscores how a patient's encounter with leisure cycling and prosthetic use can be informed by the research findings presented in this thesis. This encounter can also take different shapes, be introduced by different agents, or be enacted by other professionals. However, the approach presented here provides clear insights into how specific interventions—like saddle height adjustments, spring attachments, or reassurance regarding balance—may improve the overall cycling experience and help clinicians tailor strategies for users adapting to life with a transtibial prosthesis.

## 5.3. Limitations and future studies

Two of the studies described in this thesis relied on simulated prosthesis conditions. While this strategy offers certain benefits—such as removing factors related to amputation level, time since first use of prosthesis, and variability in prosthesis models—it does not replicate the full neuromuscular adaptation observed with an actual amputation and habitual prosthesis use. Moreover, these orthosis-based studies involved younger participants, thereby reducing the potential influence of age-related musculoskeletal changes and not fully addressing the older demographic that constitutes a significant proportion of lower-limb prosthesis users who practice cycling (Poonsiri et al., 2019, 2021). Although Chapter 3 partially addressed this issue, it remained limited to a case study involving just two participants.

Another key limitation of these studies lies in their reliance on controlled laboratory conditions, which often involved stationary bicycles. While such environments enable more accurate assessments of factors such as muscle activity and power balance—free from external influences or increased balance demands—they do not fully mirror real-world cycling. Future work could address this gap by utilising wireless sensors in outdoor contexts, thus distinguishing prosthesis-related effects from practical, real-world constraints. In addition, virtual reality (VR) technology may offer a partial yet valuable simulation of real-life conditions, further bridging the gap between laboratory findings and everyday cycling experiences.

Despite these constraints, the research herein provides preliminary indications of key factors involved in leisure cycling for individuals with unilateral transtibial amputations—an under-explored field encompassing variables related to bicycle design, situational conditions, prosthesis functionality, and user-specific characteristics. Building on these findings, future research should pursue practical investigations, such as validating saddle height settings under varied real-world conditions, exploring whether time since initial prosthesis use affects the introduction of new prosthetic features, and examining balance challenges across different cycling phases (e.g. push-off, deceleration, stopping, or sudden braking).

## 5.4. Conclusion

In conclusion, this thesis provides new insights into various biomechanical factors and challenges affecting leisure cycling for individuals with unilateral transtibial amputations. It introduces the possible advantages of pairing lower saddle heights with ankle-level movement, both of which can reduce asymmetries and make muscle activation patterns more closely resemble those of intact limbs. Moreover, results suggest that lower saddle heights also influenced the bicycle manoeuvrability and facilitated balance, dispelling the common fear that prosthesis use invariably complicates cycling—as no significant differences emerged when compared to intact-limb cycling. Although numerous variables still require examination to ensure a safer, more effective experience, these findings offer a valuable foundation for future investigations and demonstrate considerable potential for practical application.

# References

Abbiss, C. R., Peiffer, J. J., & Laursen, P. B. (2009). Optimal cadence selection during cycling: Review article. *International SportMed Journal*, *10*(1), 1–15. https://doi.org/10.10520/EJC48370

Alizadehsaravi, L., Koster, R. A. J., Muijres, W., Maas, H., Bruijn, S. M., & van Dieën, J. H. (2022). The underlying mechanisms of improved balance after one and ten sessions of balance training in older adults. *Human Movement Science*, *81*, 102910. https://doi.org/10.1016/j.humov.2021.102910

Barra, J., Bray, A., Sahni, V., Golding, J. F., & Gresty, M. A. (2006). Increasing cognitive load with increasing balance challenge: Recipe for catastrophe. *Experimental Brain Research*, *174*(4), 734–745. https://doi.org/10.1007/s00221-006-0519-2

Barth, C. A., Wladis, A., Blake, C., Bhandarkar, P., Aebischer Perone, S., & O'Sullivan, C. (2021). Retrospective observational study of characteristics of persons with amputations accessing International Committee of the Red Cross (ICRC) rehabilitation centres in five conflict and postconflict countries. *BMJ Open*, *11*(12), e049533. https://doi.org/10.1136/bmjopen-2021-049533

Batcir, S., Lubovsky, O., Bachner, Y. G., & Melzer, I. (2021). The effects of bicycle simulator training on anticipatory and compensatory postural control in older adults: Study protocol for a single-blind randomized controlled trial. *Frontiers in Neurology*, *11*, 614664. https://doi.org/10.3389/fneur.2020.614664

Baughn, M., Arellano, V., Hawthorne-Crosby, B., Lightner, J. S., Grimes, A., & King, G. (2022). Physical activity, balance, and bicycling in older adults. *PLOS ONE*, *17*(12), e0273880. https://doi.org/10.1371/journal.pone.0273880

Bing, F., Zhang, G., Wang, Y., & Zhang, M. (2024). Effects of workload and saddle height on muscle activation of the lower limb during cycling. *BioMedical Engineering OnLine*, *23*(1), 6. https://doi.org/10.1186/s12938-024-01199-y

Bini, R., Hume, P. A., & Croft, J. L. (2011). Effects of bicycle saddle height on knee injury risk and cycling performance. *Sports Medicine*, *41*(6), 463–476. https://doi.org/10.2165/11588740-000000000-00000

Bini, R., Hume, P., Croft, J., & Kilding, A. (2013). Pedal force effectiveness in cycling: A review of constraints and training effects. *Journal of Science and Cycling*, *2*, 11–24.

Bini, R., & Priego-Quesada, J. (2022). Methods to determine saddle height in cycling and implications of changes in saddle height in performance and injury risk: A systematic review. *Journal of Sports Sciences*, *40*(4), 386–400. https://doi.org/10.1080/02640414.2021.1994727

Bini, R. R. (2020). Acute effects from changes in saddle height in perceived comfort during cycling. *International Journal of Sports Science & Coaching*, *15*(3), 390–397. https://doi.org/10.1177/1747954120918965

Bini, R. R., & Hume, P. A. (2014). Assessment of bilateral asymmetry in cycling using a commercial instrumented crank system and instrumented pedals. *International Journal of Sports Physiology and Performance*, *9*(5), 876–881. https://doi.org/10.1123/ijspp.2013-0494

Bini, R. R., Hume, P. A., Croft, J., & Kilding, A. E. (2013). Pedal force effectiveness in cycling: A review of constraints and training effects. *Journal of Science and Cycling*, *2*(1), Article 1.

Bini, R. R., Hume, P. A., & Crofta, J. L. (2011). Effects of saddle height on pedal force effectiveness. *Procedia Engineering*, *13*, 51–55. https://doi.org/10.1016/j.proeng.2011.05.050

Bini, R. R., Hume, P. A., & Kilding, A. E. (2014). Saddle height effects on pedal forces, joint mechanical work and kinematics of cyclists and triathletes. *European Journal of Sport Science*, *14*(1), 44–52. https://doi.org/10.1080/17461391.2012.725105

Boccia, M., Di Vita, A., Palermo, L., Nemmi, F., Traballesi, M., Brunelli, S., De Giorgi, R., Galati, G., & Guariglia, C. (2020). Neural modifications in lower limb amputation: An fMRI study on action and non-action oriented body representations. *Brain Imaging and Behavior*, *14*(2), 416–425. https://doi.org/10.1007/s11682-019-00142-3

Borg, G. (1970). Perceived exertion as an indicator of somatic stress. *Scandinavian Journal of Rehabilitation Medicine*, *2*(2), 92–98.

Bragaru, M., Dekker, R., Geertzen, J. H. B., & Dijkstra, P. U. (2011). Amputees and sports: A systematic review. *Sports Medicine*, *41*(9), 721–740. https://doi.org/10.2165/11590420-000000000-00000

Bramati, I. E., Rodrigues, E. C., Simões, E. L., Melo, B., Höfle, S., Moll, J., Lent, R., & Tovar-Moll, F. (2019). Lower limb amputees undergo long-distance plasticity in sensorimotor functional connectivity. *Scientific Reports*, *9*(1), 2518. https://doi.org/10.1038/s41598-019-39696-z

Brockett, C. L., & Chapman, G. J. (2016). Biomechanics of the ankle. *Orthopaedics and Trauma*, *30*(3), 232–238. https://doi.org/10.1016/j.mporth.2016.04.015

Burden, A. M., Trew, M., & Baltzopoulos, V. (2003). Normalisation of gait EMGs: A re-examination. *Journal of Electromyography and Kinesiology*, *13*(6), 519–532. https://doi.org/10.1016/S1050-6411(03)00082-8

Burke, E. (2003). *High-tech Cycling*. Human Kinetics.

Cain, S. M., Ashton-Miller, J. A., & Perkins, N. C. (2016). On the skill of balancing while riding a bicycle. *PLOS ONE*, *11*(2), e0149340. https://doi.org/10.1371/journal.pone.0149340

Carey, J. J., Toovey, R., Spittle, A. J., Imms, C., & Shields, N. (2023). Exploring adaptive cycling interventions for young people with disability: an online survey of providers in Australia. *Journal of Clinical Medicine*, *12*(17), 5523. https://doi.org/10.3390/jcm12175523

Caruso, M., & Harrington, S. (n.d.). *Prevalence of Limb Loss and Limb Difference in the United States: Implications for Public Policy*.

Carvallo, E. (Emmanuel). (1901). *Théorie du Mouvement du Monocycle et de la Bicyclette.*

Childers, L., Hodson-Tole, E., & Gregor, R. (2009). Activation changes in the gastrocnemius muscle: Adaptation to a new functional role following amputation. *Medicine and Science in Sports and Exercise*, *41*. https://doi.org/10.1249/01.mss.0000354161.24516.91

Childers, W. L., & Gregor, R. J. (2011). Effectiveness of force production in persons with unilateral transtibial amputation during cycling. *Prosthetics and Orthotics International*, *35*(4), 373–378. https://doi.org/10.1177/0309364611423129

Childers, W. L., Kistenberg, R. S., & Gregor, R. J. (2009). The biomechanics of cycling with a transtibial amputation: Recommendations for prosthetic design and direction for future research. *Prosthetics and Orthotics International*, *33*(3), 256–271. https://doi.org/10.1080/03093640903067234

Childers, W. L., Kistenberg, R. S., & Gregor, R. J. (2011). Pedaling asymmetries in cyclists with unilateral transtibial amputation: Effect of prosthetic foot stiffness. *Journal of Applied Biomechanics*, *27*(4), 314–321. https://doi.org/10.1123/jab.27.4.314

Childers, W. L., & Kogler, G. F. (2014). Symmetrical kinematics does not imply symmetrical kinetics in people with transtibial amputation using cycling model. *Journal of Rehabilitation Research and Development*, *51*(8), 1243–1254. https://doi.org/10.1682/JRRD.2013.11.0241

Childers, W. L., Prilutsky, B. I., & Gregor, R. J. (2014). Motor adaptation to prosthetic cycling in people with trans-tibial amputation. *Journal of Biomechanics*, *47*(10), 2306–2313. https://doi.org/10.1016/j.jbiomech.2014.04.037

Childers, W. L., & Takahashi, K. Z. (2018). Increasing prosthetic foot energy return affects whole-body mechanics during walking on level ground and slopes. *Scientific Reports*, *8*, 5354. https://doi.org/10.1038/s41598-018-23705-8

Chowdhury, S., & Costello, S. B. (2016). An examination of cyclists' and non-cyclists' mode choice under a new cycle network. *Road and Transport Research*, *25*(4), 50–61. Scopus.

Claret, C. R., Herget, G. W., Kouba, L., Wiest, D., Adler, J., von Tscharner, V., Stieglitz, T., & Pasluosta, C. (2019). Neuromuscular adaptations and sensorimotor integration following a unilateral transfemoral amputation. *Journal of NeuroEngineering and Rehabilitation*, *16*(1), 115. https://doi.org/10.1186/s12984-019-0586-9

Connick, M. J., & Li, F.-X. (2013). The impact of altered task mechanics on timing and duration of eccentric bi-articular muscle contractions during cycling. *Journal of Electromyography and Kinesiology*, *23*(1), 223–229. https://doi.org/10.1016/j.jelekin.2012.08.012

D, Handy Eone, E, Nseme Etouckey, M, Essi, T, Ngo Nyemb., B, Ngo Nonga, F, Ibrahima, & M, Sosso. (2018). Satisfaction of patients with amputated lower limb wearing external prostheses. *International Journal of Orthopaedics Sciences*, *4*(1), 368–372. https://doi.org/10.22271/ortho.2018.v4.i1f.52

da Silva, J. C. L., Tarassova, O., Ekblom, M. M., Andersson, E., Rönquist, G., & Arndt, A. (2016). Quadriceps and hamstring muscle activity during cycling as measured with intramuscular electromyography. *European Journal of Applied Physiology*, *116*, 1807–1817. https://doi.org/10.1007/s00421-016-3428-5

de Vey Mestdagh, K. (1998). Personal perspective: In search of an optimum cycling posture. *Applied Ergonomics*, *29*(5), 325–334.

Devan, H., Hendrick, P., Ribeiro, D. C., A Hale, L., & Carman, A. (2014). Asymmetrical movements of the lumbopelvic region: Is this a potential mechanism for low back pain in people with lower limb amputation? *Medical Hypotheses*, *82*(1), 77–85. https://doi.org/10.1016/j.mehy.2013.11.012

Dyer, B. (2016). Cycling with an amputation: A systematic review. *Prosthetics and Orthotics International*, *40*(5), 538–544. https://doi.org/10.1177/0309364615610659

Dyer, B., & Disley, B. X. (2020). The aerodynamic impact of a range of prostheses designs when cycling with a trans-tibial amputation. *Disability and Rehabilitation: Assistive Technology*, *15*(5), 577–581. https://doi.org/10.1080/17483107.2019.1594409

Elmer, S. J., Barratt, P. R., Korff, T., & Martin, J. C. (2011). Joint-specific power production during submaximal and maximal cycling. *Medicine and Science in Sports and Exercise*, *43*(10), 1940–1947. https://doi.org/10.1249/MSS.0b013e31821b00c5

Ericson, M. O., Nisell, R., Arborelius, U. P., & Ekholm, J. (1985). Muscular activity during ergometer cycling. *Scandinavian Journal of Rehabilitation Medicine*, *17*(2), 53–61.

Ericson, M. O., Nisell, R., & Németh, G. (1988). Joint motions of the lower limb during ergometer cycling. *Journal of Orthopaedic & Sports Physical Therapy*, *9*(8), 273–278. https://doi.org/10.2519/jospt.1988.9.8.273

Eshraghi, A., Safaeepour, Z., Geil, M. D., & Andrysek, J. (2018). Walking and balance in children and adolescents with lower-limb amputation: A review of literature. *Clinical Biomechanics*, *59*, 181–198. https://doi.org/10.1016/j.clinbiomech.2018.09.017

Fonda, B., & Sarabon, N. (2010). Biomechanics of cycling. *Sport Science Review*, *XIX*. https://doi.org/10.2478/v10237-011-0012-0

Garcia-Lopez, J., Ferrer, V., Palomo, V., Ogueta-Alday, A., & Rodriguez-Marroyo, J. (2017). Acute effects of small changes in crank length on gross efficiency and pedalling technique during submaximal cycling. *Journal of Sports Sciences*, *35*, 1328–1335. https://doi.org/10.1080/02640414.2016.1215490

Gloc, D., Nowak, Z., Nowak-Lis, A., Gabryś, T., Szmatlan-Gabrys, U., Valach, P., & Pilis, A. (2021). Indoor cycling training in rehabilitation of patients after myocardial infarction. *BMC Sports Science, Medicine and Rehabilitation*, *13*, 151. https://doi.org/10.1186/s13102-021-00379-w

Hermens, H. J., Freriks, B., Disselhorst-Klug, C., & Rau, G. (2000). Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of Electromyography and Kinesiology*, *10*(5), 361–374. https://doi.org/10.1016/s1050-6411(00)00027-4

*ISO 8548-2:2020*. (n.d.). ISO. Retrieved 25 September 2024, from https://www.iso.org/standard/76809.html

Kaufman, K. R., Wyatt, M. P., Sessoms, P. H., & Grabiner, M. D. (2014). Task-specific fall prevention training is effective for warfighters with transtibial amputations. *Clinical Orthopaedics and Related Research*, *472*(10), 3076–3084. https://doi.org/10.1007/s11999-014-3664-0

Koenig, K. D., Hall, M. J., Gormley, C., Kaleta, M., Munger, M., Laine, J., & Morgan, S. J. (n.d.). Clinical outcomes measurement in pediatric lower limb prosthetics: A scoping review. *Journal of Pediatric Rehabilitation Medicine*, *17*(2), 147–165. https://doi.org/10.3233/PRM-230014

Koutny, D., Paloušek, D., STOKLÁSEK, P., Rosický, J., Tepla, L., Prochazkova, M., Svoboda, Z., & Krejci, P. (2013). The biomechanics of cycling with transtibial prosthesis: A case study of a professional cyclist. *International Journal of Medical, Health, Biomedical, Bioengineering and Pharmaceutical Engineering*, *7*, 476–481.

Kurz, I., Gimmon, Y., Shapiro, A., Debi, R., Snir, Y., & Melzer, I. (2016). Unexpected perturbations training improves balance control and voluntary stepping times in older adults—A double blind randomized control trial. *BMC Geriatrics*, *16*(1), 58. https://doi.org/10.1186/s12877-016-0223-4

Le Mouel, C., Tisserand, R., Robert, T., & Brette, R. (2019). Postural adjustments in anticipation of predictable perturbations allow elderly fallers to achieve a balance recovery performance equivalent to elderly non-fallers. *Gait & Posture*, *71*, 131–137. https://doi.org/10.1016/j.gaitpost.2019.04.025

LeMond, G., & Gordis, K. (1987). *Greg LeMond's Complete Book of Bicycling*. Putnam.

Lowell, J., & McKell, H. D. (1982). The stability of bicycles. *American Journal of Physics*, *50*(12), 1106–1112. https://doi.org/10.1119/1.12893

*Lower Limb Amputations: Epidemiology and Assessment | PM&R KnowledgeNow*. (2017, March 1). https://now.aapmr.org/lower-limb-amputations-epidemiology-and-assessment/

Lukežič, A., Vojíř, T., Čehovin, L., Matas, J., & Kristan, M. (2018). Discriminative correlation filter with channel and spatial reliability. *International Journal of Computer Vision*, *126*(7), 671–688. https://doi.org/10.1007/s11263-017-1061-3

Macdermid, P. W., & Edwards, A. M. (2010). Influence of crank length on cycle ergometry performance of well-trained female cross-country mountain bike athletes. *European Journal of Applied Physiology*, *108*(1), 177–182. https://doi.org/10.1007/s00421-009-1197-0

*MagLOCK Bike Pedal—Magnetic Bike Pedals*. (n.d.). Retrieved 30 September 2024, from https://maglockbikepedal.com/?srsltid=AfmBOorKnlDo47oXNRW3oObfhAofEO1UOtjQb7Zl-TbfP7orcvfbL50c

Marinho-Buzelli, A. R., Zaluski, A. J., Mansfield, A., Bonnyman, A. M., & Musselman, K. E. (2019). The use of aquatic therapy among rehabilitation professionals for individuals with spinal cord injury or disorder. *The Journal of Spinal Cord Medicine*, *42*(Suppl 1), 158–165. https://doi.org/10.1080/10790268.2019.1647935

Masters, R., & Maxwell, J. (2008). The theory of reinvestment. *International Review of Sport and Exercise Psychology*, *1*(2), 160–183. https://doi.org/10.1080/17509840802287218

Mayer, A., Kudar, K., Bretz, K., & Tihanyi, J. (2008). Body schema and body awareness of amputees. *Prosthetics and Orthotics International*, *32*(3), 363–382. https://doi.org/10.1080/03093640802024971

Miller, W. C., Speechley, M., & Deathe, B. (2001). The prevalence and risk factors of falling and fear of falling among lower extremity amputees. *Archives of Physical Medicine and Rehabilitation*, *82*(8), 1031–1037. https://doi.org/10.1053/apmr.2001.24295

Moreno-Pérez, V., Courel-Ibáñez, J., Mateo-March, M., López-Samanes, Á., & Del Coso, J. (2021). Descriptive profile for lower-limb range of motion in professional road cyclists. *The Journal of Sports Medicine and Physical Fitness*, *61*(3). https://doi.org/10.23736/S0022-4707.20.11178-2

Morgenroth, D. C., Gellhorn, A. C., & Suri, P. (2012). Osteoarthritis in the disabled population: A mechanical perspective. *PM&R*, *4*(5, Supplement), S20–S27. https://doi.org/10.1016/j.pmrj.2012.01.003

Mornieux, G., Stapelfeldt, B., Gollhofer, A., & Belli, A. (2008). Effects of pedal type and pull-up action during cycling. *International Journal of Sports Medicine*, *29*(10), 817–822. https://doi.org/10.1055/s-2008-1038374

Nikitas, A. (2018). Understanding bike-sharing acceptability and expected usage patterns in the context of a small city novel to the concept: A story of 'Greek Drama'. *Transportation Research Part F: Traffic Psychology and Behaviour*, *56*, 306–321. https://doi.org/10.1016/j.trf.2018.04.022

O'Bryan, S. J., Brown, N. A. T., Billaut, F., & Rouffet, D. M. (2014). Changes in muscle coordination and power output during sprint cycling. *Neuroscience Letters*, *576*, 11–16. https://doi.org/10.1016/j.neulet.2014.05.023

Parr, J. V. V., Mills, R., Kal, E., Bronstein, A. M., & Ellmers, T. J. (2024). A "conscious" loss of balance: directing attention to movement can impair the cortical response to postural perturbations. *Journal of Neuroscience*, *44*(48). https://doi.org/10.1523/JNEUROSCI.0810-24.2024

Pierson-Carey, C. D., Brown, D. A., & Dairaghi, C. A. (1997). Changes in resultant pedal reaction forces due to ankle immobilization during pedaling. *Journal of Applied Biomechanics*, *13*(3), 334–346. https://doi.org/10.1123/jab.13.3.334

Poonsiri, J., Dekker, R., Dijkstra, P. U., Hijmans, J. M., & Geertzen, J. H. B. (2018). Bicycling participation in people with a lower limb amputation: A scoping review. *BMC Musculoskeletal Disorders*, *19*(1), 398. https://doi.org/10.1186/s12891-018-2313-2

Poonsiri, J., Dekker, R., Dijkstra, P. U., Hijmans, J. M., & Geertzen, J. H. B. (2021). Cycling in people with a lower limb amputation. *BMC Sports Science, Medicine & Rehabilitation*, *13*(1), 75. https://doi.org/10.1186/s13102-021-00302-3

Poonsiri, J., Dekker, R., Dijkstra, P. U., Nutchamlong, Y., Dismanopnarong, C., Puttipaisan, C., Suakonburi, S., Pimchan, P., Hijmans, J. M., & Geertzen, J. H. B. (2019). Cycling of people with a lower limb amputation in Thailand. *PLOS ONE*, *14*(8), e0220649. https://doi.org/10.1371/journal.pone.0220649

Priego Quesada, J. I., Pérez-Soriano, P., Lucas-Cuevas, A. G., Salvador Palmer, R., & Cibrián Ortiz de Anda, R. M. (2017). Effect of bike-fit in the perception of comfort, fatigue and pain. *Journal of Sports Sciences*, *35*(14), 1459–1465. https://doi.org/10.1080/02640414.2016.1215496

Prost, V., Johnson, W. B., Kent, J. A., Major, M. J., & Winter, A. G. (2022). Biomechanical evaluation over level ground walking of user-specific prosthetic feet designed using the lower leg trajectory error framework. *Scientific Reports*, *12*(1), 5306. https://doi.org/10.1038/s41598-022-09114-y

Reilly, K. T., & Sirigu, A. (2011). Motor cortex representation of the upper-limb in individuals born without a hand. *PLOS ONE*, *6*(4), e18100. https://doi.org/10.1371/journal.pone.0018100

Renzi, R., Unwin, N., Jubelirer, R., & Haag, L. (2006). An international comparison of lower extremity amputation rates. *Annals of Vascular Surgery*, *20*(3), 346–350. https://doi.org/10.1007/s10016-006-9044-9

Ribeiro de Souza, C., Ávila de Oliveira, J., Takazono, P. S., da Silva Rezende, L., Silva-Batista, C., Coelho, D. B., & Teixeira, L. A. (2023). Perturbation-based balance training leads to improved reactive postural responses in individuals with Parkinson's disease and freezing of gait. *The European Journal of Neuroscience*, *57*(12), 2174–2186. https://doi.org/10.1111/ejn.16039

Rissel, C., Passmore, E., Mason, C., & Merom, D. (2013). Two pilot studies of the effect of bicycling on balance and leg strength among older adults. *Journal of Environmental and Public Health*, 686412. https://doi.org/10.1155/2013/686412

Robergs, R., Peveler, W., Bishop, P., J, S., & M, R. (2005). Comparing methods for setting saddle height in trained cyclists. *Exerc Physiol*, *8*(51–55).

Rybarczyk, B., Edwards, R., & Behel, J. (2004). Diversity in adjustment to a leg amputation: Case illustrations of common themes. *Disability and Rehabilitation*, *26*(14–15), 944–953. https://doi.org/10.1080/09638280410001708986

Sarabon, N., Fonda, B., & Markovic, G. (2012). Change of muscle activation patterns in uphill cycling of varying slope. *European Journal of Applied Physiology*, *112*(7), 2615–2623. https://doi.org/10.1007/s00421-011-2236-1

Schaffalitzky, E., Gallagher, P., Maclachlan, M., & Ryall, N. (2011). Understanding the benefits of prosthetic prescription: Exploring the experiences of practitioners and lower limb prosthetic users. *Disability and Rehabilitation*, *33*(15–16), 1314–1323. https://doi.org/10.3109/09638288.2010.529234

Schneider, R. J., & Hu, L. (2015). Improving university transportation sustainability: Reducing barriers to campus bus and bicycle commuting. *International Journal of Sustainability Policy and Practice*, *11*(1), 24–34.

Seratiuk Flores, H., Yeoh, W. L., Loh, P. Y., Morinaga, K., & Muraki, S. (2023). Biomechanical analysis of recreational cycling with unilateral transtibial prostheses. *Prosthesis*, *5*(3). https://doi.org/10.3390/prosthesis5030052

Silva, J., Ekblom, M., Tarassova, O., Andersson, E., Rönquist, G., Grundström, H., & Arndt, A. (2018). Effect of increasing workload on knee extensor and flexor muscular activity during cycling as measured with intramuscular electromyography. *PLOS ONE*, *13*, e0201014. https://doi.org/10.1371/journal.pone.0201014

So, R. C. H., Ng, J. K.-F., & Ng, G. Y. F. (2005). Muscle recruitment pattern in cycling: A review. *Physical Therapy in Sport*, *6*(2), 89–96. https://doi.org/10.1016/j.ptsp.2005.02.004

Tajali, S., Rouhani, M., Mehravar, M., Negahban, H., Sadati, E., & Oskouei, A. E. (2018). Effects of external perturbations on anticipatory and compensatory postural adjustments in patients with multiple sclerosis and a fall history. *International Journal of MS Care*, *20*(4), 164–172. https://doi.org/10.7224/1537-2073.2016-098

Takeuchi, N., Izumi, S.-I., Ota, J., & Ueda, J. (2016). Neural plasticity on body representations: advancing translational rehabilitation. *Neural Plasticity*, 9737569. https://doi.org/10.1155/2016/9737569

Thevarajah, A., Wallen, M., Imms, C., Lonsdale, C., Carey, J. J., & Froude, E. H. (2023). Impact of adapted bicycle riding on outcomes for children and adolescents with disabilities: A systematic review. *Developmental Medicine & Child Neurology*, *65*(4), 456–468. https://doi.org/10.1111/dmcn.15446

Tiele, A., Soni-Sadar, S., Rowbottom, J., Patel, S., Mathewson, E., Pearson, S., Hutchins, D., Head, J., & Hutchins, S. (2020). Design and development of a novel transtibial cycling prosthesis. *JPO: Journal of Prosthetics and Orthotics*, *32*(2), 134. https://doi.org/10.1097/JPO.0000000000000274

Turpin, N. A., & Watier, B. (2020). Cycling biomechanics and its relationship to performance. *Applied Sciences*, *10*(12). https://doi.org/10.3390/app10124112

Ülger, Ö., Yıldırım Şahan, T., & Çelik, S. E. (2018). A systematic literature review of physiotherapy and rehabilitation approaches to lower-limb amputation. *Physiotherapy Theory and Practice*, *34*(11), 821–834. https://doi.org/10.1080/09593985.2018.1425938

van Bekkum, J. E., Williams, J. M., & Graham Morris, P. (2011). Cycle commuting and perceptions of barriers: Stages of change, gender and occupation. *Health Education*, *111*(6), 476–497. https://doi.org/10.1108/09654281111180472

van Melick, N., Meddeler, B. M., Hoogeboom, T. J., Nijhuis-van der Sanden, M. W. G., & van Cingel, R. E. H. (2017). How to determine leg dominance: The agreement between self-reported and observed performance in healthy adults. *PLoS ONE*, *12*(12), e0189876. https://doi.org/10.1371/journal.pone.0189876

Verma, R., Hansen, E. A., de Zee, M., & Madeleine, P. (2016). Effect of seat positions on discomfort, muscle activation, pressure distribution and pedal force during cycling. *Journal of Electromyography and Kinesiology*, *27*, 78–86. https://doi.org/10.1016/j.jelekin.2016.02.003

Watanabe, K., Yamaguchi, Y., Fukuda, W., Nakazawa, S., Kenjo, T., & Nishiyama, T. (2020). Neuromuscular activation pattern of lower extremity muscles during pedaling in cyclists with single amputation of leg and with two legs: A case study. *BMC Research Notes*, *13*, 299. https://doi.org/10.1186/s13104-020-05144-9

Whipple, F. J. W. (1899). The stability of the motion of a bicycle. *Quart. J. Pure Appl. Math.*, *30*, 312–348.

Wong, C. K., Chen, C. C., Benoy, S. A., Rahal, R. T., & Blackwell, W. M. (2014). Role of balance ability and confidence in prosthetic use for mobility of people with lower-limb loss. *Journal of Rehabilitation Research and Development*, *51*(9), 1353–1364. https://doi.org/10.1682/JRRD.2013.11.0235

Wulf, G. (2013). Attentional focus and motor learning: A review of 15 years. *International Review of Sport and Exercise Psychology*, *6*(1), 77–104. https://doi.org/10.1080/1750984X.2012.723728

Yang, H.-C., Lee, C.-L., Lin, R., Hsu, M.-J., Chen, C.-H., Lin, J.-H., & Lo, S. K. (2014). Effect of biofeedback cycling training on functional recovery and walking ability of lower extremity in patients with stroke. *The Kaohsiung Journal of Medical Sciences*, *30*(1), 35–42. https://doi.org/10.1016/j.kjms.2013.07.006

Yeoh, W. L., & Seratiuk Flores, H. (2024). *Motion Analysis 2D* [Python]. CPS Lab (Saga University). https://github.com/cps-lab-saga/motion-analysis-2d

Yum, H., Kim, H., Lee, T., Park, M. S., & Lee, S. Y. (2021). Cycling kinematics in healthy adults for musculoskeletal rehabilitation guidance. *BMC Musculoskeletal Disorders*, *22*(1), 1044. https://doi.org/10.1186/s12891-021-04905-2

Ziegler-Graham, K., MacKenzie, E. J., Ephraim, P. L., Travison, T. G., & Brookmeyer, R. (2008). Estimating the prevalence of limb loss in the United States: 2005 to 2050. *Archives of Physical Medicine and Rehabilitation*, *89*(3), 422–429. https://doi.org/10.1016/j.apmr.2007.11.005

# Appendix

## Appendix 1

Significant Dunnett's test results.

## Joint angles

| Variable | Saddle Height | Mean Difference | Significance |
|---|---|---|---|
| **Knee 0°** | -3.5% | -5.14 | .000 |
| | 0 | -8.55 | .000 |
| | +3.5% | -12.32 | .000 |
| | +7% | -15.27 | .000 |
| **Knee 90°** | -3.5% | -4.64 | .000 |
| | 0 | -9.13 | .000 |
| | +3.5% | -13.71 | .000 |
| | +7% | -17.89 | .000 |
| **Knee 180°** | -3.5% | -10.18 | .000 |
| | 0 | -18.76 | .000 |
| | +3.5% | -26.46 | .000 |
| | +7% | -33.05 | .000 |
| **Knee 270°** | -3.5% | -5.92 | .000 |
| | 0 | -10.73 | .000 |
| | +3.5% | -15.54 | .000 |
| | +7% | -19.32 | .000 |
| **Hip 0°** | -7% | -10.54 | .000 |
| | -3.5% | -7.50 | .000 |
| | 0 | -5.08 | .000 |
| | +3.5% | -2.04 | .004 |
| **Hip 90°** | -7% | -9.89 | .000 |
| | -3.5% | -6.58 | .000 |
| | 0 | -3.60 | .000 |
| | +3.5% | -1.67 | .004 |
| | +7% | 2.28 | .000 |
| **Hip 180°** | -7% | -5.37 | .000 |

| Variable | Saddle Height | Mean Difference | Significance |
|---|---|---|---|
| | 0 | 4.46 | .000 |
| | +3.5% | 9.02 | .000 |
| | +7% | 13.34 | .000 |
| **Hip 270°** | -7% | -7.01 | .000 |
| | -3.5% | -4.53 | .000 |
| | 0 | -1.97 | .001 |
| | +3.5% | 2.23 | .000 |
| | +7% | 4.06 | .000 |

## Instrumented pedals

| Variable | Saddle Height | Mean Difference | Significance |
|---|---|---|---|
| **L/R Balance** | -7% | 9.77 | .006 |
| | -3.5% | 14.33 | .000 |
| | 0 | 20.62 | .000 |
| | +3.5% | 20.88 | .000 |
| | +7% | 30.90 | .000 |
| **Unaffected Leg Torque Effectiveness** | -3.5% | 4.75 | .047 |
| | 0 | 7.89 | .000 |
| | +3.5% | 11.14 | .000 |
| | +7% | 17.15 | .000 |
| **Affected Leg Torque Effectiveness** | -7% | -8.25 | .004 |
| | -3.5% | -10.48 | .000 |
| | 0 | -11.79 | .000 |
| | +3.5% | -11.63 | .000 |
| | +7% | -14.70 | .000 |
| **Unaffected Leg Pedal Smoothness** | 0 | 1.83 | .017 |
| | +3.5% | 2.80 | .000 |
| | +7% | 3.98 | .000 |
| **Affected Leg Pedal Smoothness** | -7% | -2.09 | .008 |
| | -3.5% | -2.78 | .000 |

| Variable | Saddle Height | Mean Difference | Significance |
|---|---|---|---|
| | 0 | -2.84 | .000 |
| | +3.5% | -2.99 | .000 |
| | +7% | -3.60 | .000 |

## Subjective evaluation

| Variable | Saddle Height | Mean Difference | Significance |
|---|---|---|---|
| **Borg Scale** | -7% | 3.11 | .000 |
| | -3.5% | 2.21 | .001 |
| | 0 | 2.28 | .001 |
| | +3.5% | 3.55 | .000 |
| | +7% | 4.57 | .000 |