ABSTRACT

Yu, Zhenyuan. An Explorative Study on How to Improve Mediolateral Balance Using Hip Exoskeleton. (Under the direction of Dr. Ming Liu).

There is increasing evidence of the role of compromised mediolateral balance in falls and the need for assistance and rehabilitation specifically focused on mediolateral direction for various populations with motor deficits like amputees and the elderly. To address this need, we have investigated how mediolateral torque provided by a wearable hip exoskeleton could affect the Narrow Beam Walking Test (NBWT) performance and developed a neurorehabilitation platform by integrating the exoskeleton with a visual interface. The exoskeleton is expected to influence the kinematics of the subject during the Narrow Beam Walking Test (NBWT). The study was conducted on a subject with a left leg amputation under five different torque application conditions. The results indicate a decrease in maximum heel displacements and torso sway angles in the mediolateral direction during adduction torque trials. This effect is more pronounced on the second beam, highlighting varying task difficulties. Additionally, the different balance strategies employed by the subject's prosthetic and intact legs are analyzed. The neurorehabilitation platform aims to influence and rehabilitate the visuomotor mechanisms in users by engaging them in motion tasks based on visual feedback. During these tasks, the robot exerts various controlled resistances, managed by an admittance controller embedded within the system. A preliminary study involving three non-disabled participants was conducted to evaluate the system's performance and to observe any changes in hip joint kinematics and kinetics due to visuomotor training under four distinct admittance conditions. Results showed that all participants displayed improved motion consistency and enhanced interlimb coordination during training, illustrating the system's effectiveness. The study also delves into the analysis of human-robot torque interactions and

electromyography (EMG) signals, discussing their implications for neurorehabilitation, particularly for amputees and the elderly.

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An Explorative Study on How to Improve Mediolateral Balance Using Hip Exoskeleton

by
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A thesis submitted to the Graduate Faculty of
North Carolina State University
in partial fulfillment of the
requirements for the degree of
Master of Science

Biomedical Engineering

Raleigh, North Carolina
2024

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BIOGRAPHY

Zhenyuan Yu was born in Xi'an, China, in 1996. From an early age, he exhibited a strong interest in robots. In 2015, he enrolled in the School of Mechanical Engineering at Xi'an Jiaotong University, where he completed his undergraduate studies. During his second year of college, he spent a semester at the University of California, Los Angeles, where he developed an interest in wearable robotics. After completing a graduate project at Binghamton University, he decided to pursue further graduate studies at Xi'an Jiaotong University, focusing on the development of lower limb hip exoskeletons. After earning a Master's degree in Mechanical Engineering, he joined the joint Department of Biomedical Engineering at NC State University and UNC Chapel Hill. Under the guidance of Dr. Ming Liu and Dr. Helen Huang, he began his graduate studies in biomechanics and wearable robotics.

ACKNOWLEDGMENTS

I would first like to acknowledge Dr. Ming Liu and Dr. Helen Huang, who supported me and provided me the great research platform and the guidance on completing this work. I could not complete this work without their help and guidance. I would also like to acknowledge my lab mates in the NREL lab who is always willing to help and created a friendly environment. I would especially like to thank Ms. Vilma Berg and Dr. Matthew Fisher for supporting me when I struggled. Your patience and help are vital to me and I would always appreciate that. Finally, I would like to thank my parents for giving unconditional love and support through my ups and downs.

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CHAPTER 1: Introduction

Some paragraphs in Section 1.1 and Section 1.2 of this chapter are derived from a conference paper previously published by the author in 2023 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS) [54]. Permission for the use of this paper has been obtained from all co-authors of the paper.

1.1 Background

Falls are the leading cause of injury related deaths in older populations, and are caused by degenerating control of balance [1]. This problem is further exacerbated in people affected by amputation, which is considered one of the major risk factors of falls [2]. The asymmetry caused by amputation is found to result in higher dynamic instability, irregular step width modulation and general instability especially in the frontal plane, leading to falls [3,4].

Human balance involves complex mechanisms to maintain stability in both the sagittal and frontal planes. In the sagittal plane, balance is primarily managed through the coordination of muscles controlling forward and backward movements, such as the ankle plantarflexors and dorsiflexors, as well as hip and trunk muscles [9]. Maintaining sagittal plane balance often involves strategies such as adjusting step length, utilizing visual and vestibular inputs, and engaging postural control mechanisms to counteract forward or backward sway [10]. In contrast, frontal plane balance requires active control to prevent lateral falls, involving muscles responsible for side-to-side movements, including hip abductors and adductors, as well as trunk stabilizers [11]. The control of mediolateral stability is critical, particularly in populations with lower limb disabilities such as amputations that disrupt neuromuscular mechanisms [12].

Efforts to improve sagittal plane balance often involve strengthening exercises, balance training, and the use of assistive devices. Strengthening the ankle dorsiflexors and plantarflexors, as well as the hip extensors and flexors, can enhance sagittal plane stability [40]. Balance training programs that focus on proprioception, coordination, and reaction time have also shown promise in reducing fall risks [41]. Additionally, interventions such as treadmill training and virtual reality-based balance exercises are being explored to improve sagittal plane balance and reduce the incidence of falls [42].

Focusing on the frontal plane, there has been increasing evidence of the impact of diminishing mediolateral directional stability and balance in the occurrence of falls in older populations and could affect amputee populations similarly [5,6]. Moreover, falls in mediolateral direction are found to have more severe injuries with over 95% leading to hip fracture [7,8]. Mediolateral balance outcomes and variability in gait behavior, such as step width variability, have been shown to predict the likelihood of falls in individuals with lower limb disabilities [13]. Reduced muscle strength, sensory motor function, and loss of neural control are postulated to be root causes of dynamic instability [14]. Improved abduction-adduction control is a predictor of better gait performance in lower limb disability populations, highlighting the importance of enhancing mediolateral balance to reduce fall risks [15].

Clinical evidence supports the need for frontal plane assistance, particularly for individuals with lower limb disabilities. Studies have demonstrated that enhancing mediolateral stability can significantly reduce the risk of falls and associated injuries, such as hip fractures [43]. For instance, interventions aimed at improving hip abductor strength and lateral postural control have been shown to decrease fall rates and improve gait stability in elderly and amputee populations [44]. Moreover, the use of assistive devices, such as hip exoskeletons, has shown potential in providing additional support to enhance lateral balance, thereby reducing the incidence of falls in these vulnerable groups [45].

Hip exoskeletons have emerged as a promising solution in the field of assistive technology and rehabilitation engineering. These wearable devices are primarily designed as assistive devices to support and enhance the natural movement of the hip joints, thereby improving walking balance in individuals with mobility impairments [14,15]. Additionally, they have also been used as rehabilitation tools aimed at helping the populations with lower limb impairment improve their own motor balance ability [46,47,48]. Based on the background above, it is important to study how to improve mediolateral balance using hip exoskeleton as assistive device and rehabilitation device.

1.2 Current Approaches

In terms of assistive devices, the use of hip exoskeletons in enhancing walking balance has been extensively studied, demonstrating numerous benefits across various populations, including the amputees and the elderly. Several studies have shown that hip exoskeletons significantly improve stability during walking. Kao et al. [16] conducted a study where individuals with hip weakness exhibited reduced medial-lateral sway and improved step symmetry when using a hip exoskeleton. This reduction in sway is crucial for preventing falls, particularly in older adults who are at higher risk of fall-related injuries. Similarly, Di Natali et al. [17] found that hip exoskeletons improved dynamic stability in amputee users by providing targeted support during the gait cycle, enhancing their ability to maintain balance during walking. Postural control is critical for maintaining balance, especially during complex walking tasks such as navigating obstacles or walking on uneven surfaces. Awad et al. [18] demonstrated that stroke survivors using a hip exoskeleton showed improved postural control, which was evident through better alignment and reduced compensatory movements. This study highlighted that the exoskeleton's assistance in maintaining an upright posture and providing balanced support across the hips contributed significantly to enhanced stability.

The Narrow Beam Walking Test (NBWT) is a commonly utilized clinical method to evaluate balance and coordination during walking, particularly in amputees [19,20]. This test requires individuals to walk along a narrow beam, thus challenging their ability to maintain balance and stability. An improvement in the NBWT score signifies enhanced walking balance, making it an effective platform for assessing the impact of hip exoskeletons on walking performance. Despite its potential, there is a scarcity of studies investigating how hip exoskeletons influence subjects' performance on the NBWT.

In terms of rehabilitation systems, recent technological innovations have focused on inducing neuroplasticity in the chronic phase once initial therapy is performed during acute phases [21]. Several approaches including robot aided neurorehabilitation have shown to impact recovery and sensorimotor reintegration [22]. These approaches have shown that a combination of error augmentation and visual distortion can speed up learning [23,24]. Specifically, the role of virtual environments is found to be important in sensorimotor reintegration due to the recruitment of visuomotor networks combined with motor learning [25]. While these approaches have seen significant adoption for the upper limb both in research and commercial space [26], there have been limited adoption for lower limb [27]. Initial approaches for the lower limb have utilized foot plates for single joint based rehabilitation in seated positions [28] or bodyweight supported solutions for treadmill training [29]. These approaches focus only on strength training unlike sensorimotor rehabilitation approaches using visual feedback. While exoskeletons could have much higher promise for integration of robot aided rehabilitation with visual distortion, they have been mostly used to restore missing function rather provide rehabilitative effects [30]. Additionally, nearly all commercially available exoskeleton systems provide assistance in the sagittal plane (i.e., hip flexion/extension, plantarflexion/dorsiflexion of the ankle) and none provide therapeutic balance training or intervention in the frontal plane (i.e. hip abduction/adduction) [31-33]. The potential reasons for lack of frontal plane rehabilitation systems could be that there are limited relevant robotic systems that could form the basis to facilitate neurorehabilitation in mediolateral direction. Generally speaking, there's very few studies that focus the sensory motor control of the lower limb.

1.3 Objectives & Aims

The purpose of this study is to address the critical need for effective interventions that enhance mediolateral balance, particularly in populations with lower limb disabilities who are at a heightened risk of falls and related injuries. The interventions could be applied for assistance or rehabilitation purpose.

In terms of assistance purpose, by examining the impact of a hip exoskeleton on mediolateral stability during walking, this research seeks to provide empirical evidence on the potential benefits of such assistive devices in reducing fall risk and improving gait stability. The use of the Narrow Beam Walking Test (NBWT) as part of the first objective is particularly significant because it is a well-established clinical method for assessing dynamic balance and coordination in a challenging and controlled environment. The NBWT requires individuals to walk along a narrow beam, which accentuates the demands on mediolateral stability and mimics real-world scenarios where precise balance control is necessary. This makes the NBWT an ideal test to evaluate how the exoskeleton's mediolateral torque can influence balance performance. By focusing on this specific test, the study aims to provide a detailed analysis of the exoskeleton's effectiveness in enhancing balance under conditions that closely simulate daily activities that are prone to falls. This approach ensures that the findings are highly relevant and can be translated into practical applications for improving safety and mobility in amputee populations.

Additionally, in terms of rehabilitation purpose, by integrating visual feedback into a hip abduction/adduction exoskeleton, the study aims to explore innovative approaches for inducing neuroplasticity and improving sensorimotor control. This dual focus not only underscores the importance of enhancing immediate balance support but also highlights the potential for long-term rehabilitative benefits, ultimately contributing to improved quality of life and reduced healthcare costs for individuals with mobility impairments.

CHAPTER 2: Impact of Mediolateral Hip Torque on Narrow Beam Walking Test Performance

Section 2.1.1 and Section 2.1.3 of this chapter is derived from a conference paper previously published by the author in 2023 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS) [54]. Permission for the use of this paper has been obtained from all co-authors of the paper.

2.1 Method

The efficacy of hip exoskeletons in enhancing walking performance among amputees and older individuals has been demonstrated in many studies [37,38]. In this section, we aim to explore the impact of mediolateral hip torques provided by a hip exoskeleton on the walking balance of an amputee subject. Given the widespread use of the Narrow Beam Walking Test as a tool for assessing an individual's ability to maintain balance, we have chosen it for evaluating subject performance in this study. This section is structured as follows: firstly, we will introduce the experimental platform utilized, encompassing the robotic platform and the narrow beam. Subsequently, we will detail the experiment protocol and present the results obtained. Finally, a comprehensive analysis of the results and future scope will be provided to conclude this section.

2.1.1 Robotic Platform and Control

The robotic platform we used for this work is a bilateral hip abduction/adduction exoskeleton. This exoskeleton is actuated by two DC brushless motors on the back that are amplified by a harmonic gear of 100:1 ratio, providing the torque necessary for abduction-adduction assistance. The flexion-extension joints are designed to be passive hinges with low friction with a range of -30° to 150°. The load cells used on the robot are force sensors, which are mounted on the lever arms at 5.5 cm from the shaft of the actuator to measure interaction force that then generates the velocity profile for the controller. Thus, the interaction torque at the hip joint can be easily estimated, since the rotation joint and the load cell are fixed. Kinematics along abduction-adduction are obtained through encoders attached to the actuators, which have a resolution of 4551 counts/deg. The bandwidth of this exoskeleton is tested to be 5 Hz.

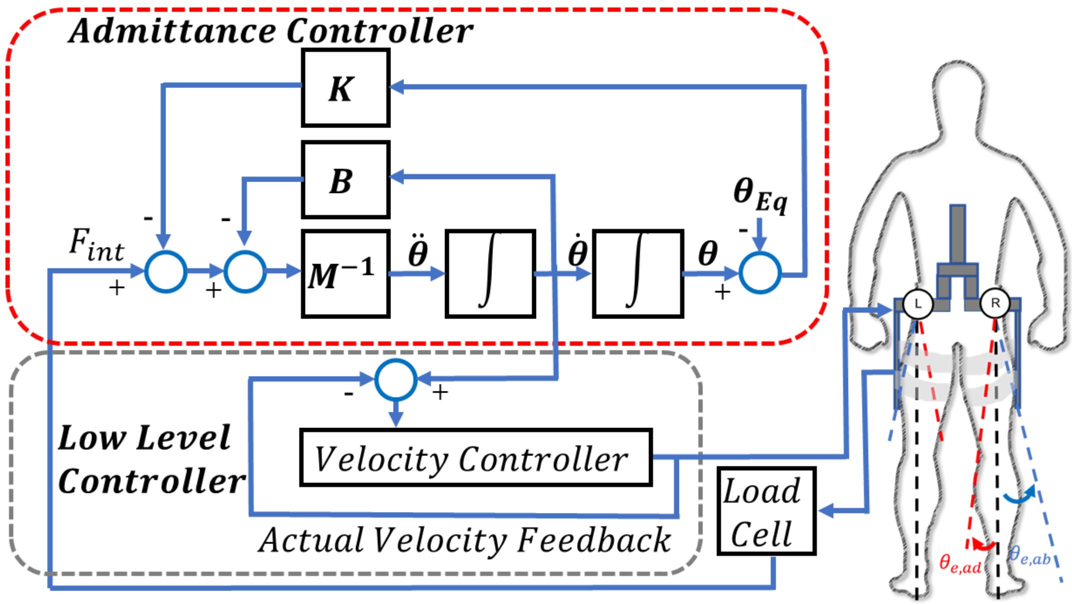


Figure 1. Diagram of the wearable robotic rehabilitation system. The left part shows the control method of the exoskeleton, where $F_{int}(t)$ is the interaction force, M, B, K are control parameters, $\ddot{\theta}$, $\dot{\theta}$, θ represent hip acceleration, velocity, position, correspondingly. θ_{Eq} is the equilibrium position.

The control method of the exoskeleton includes two levels. The low level controller is a velocity controller, as shown in Figure. 1. The actual motor velocity is obtained by a servo observer from the motor, and used as a reference for input velocity to generate the control signal. An admittance controller was implemented as the high level controller. Since one of the major functions of the robotic rehabilitation platform demonstrated in this paper is to provide personalized training program, it should have the ability to change the difficulty of training. Hence, admittance control is especially suitable, since we can easily change the training difficulty by changing the control parameters, specifically the stiffness. The interaction force is obtained from the load cell as the input of the controller. And it consists of three components, as shown in Eq. (1):

$$F_{int}(t) = M\ddot{\theta}(t) + B\dot{\theta}(\mathrm{t}) + K(\theta(t) - \theta_{Eq}) \tag{1}$$

where M, B, K are inertial, damping, stiffness parameters, correspondingly. $\theta(t)$ refers to the robot hip angle and θ_{Eq} is the equilibrium angle.

The target velocity obtained from the admittance controller is then used as a reference for the lower level controller to simulate admittance. By changing the four parameters, The resistive behavior of the robot can be instantaneously changed, as the controller is executed at a frequency of 1000 Hz. In this study, the stiffness has been varied to different levels to simulate different linear force fields while the other three parameters have been fixed.

2.1.2 Narrow Beam Walking Test

The Narrowing Beam-Walking Test (NBWT) is a clinical assessment tool created to evaluate the balance capabilities of ambulatory individuals with lower-limb impairments, particularly those prone to falls such as lower-limb prosthesis users. This test is structured to provide an objective measure of balance by utilizing minimal subjective judgment—participants are simply classified as either on or off the beam. Its progressive difficulty, achieved through decreasing beam widths, makes it adaptable to a wide range of skill levels and potential applications. Participants are tasked with walking along a series of four beams, each narrower than the previous one. Longer distances covered during the NBWT indicate better balance skills and a lower risk of falling.

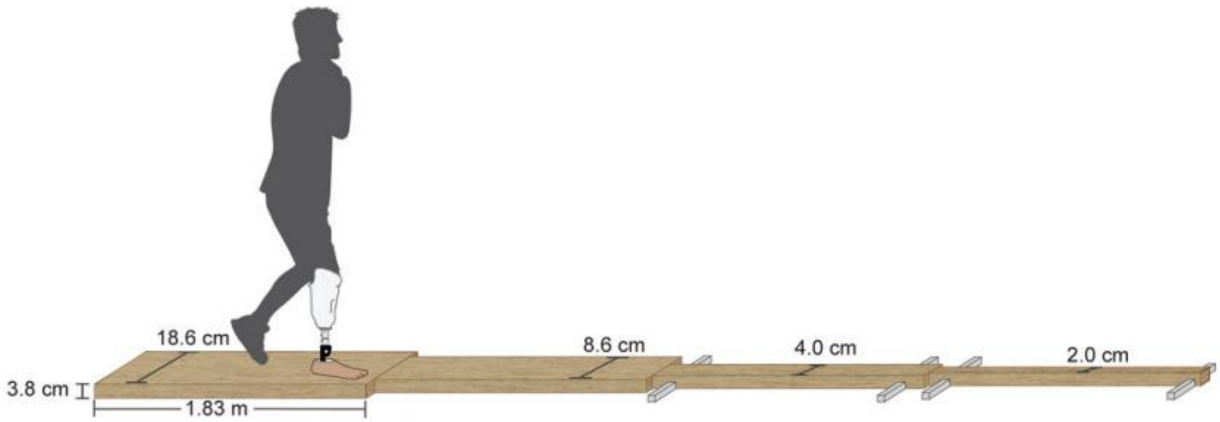


Figure 2. Narrow Beam Walking Test

The dimensions of the narrow beam platform are illustrated in Figure 2. The NBWT beams can be constructed using common materials found in construction, such as high-quality 2x8, 2x4, 2x2, and 2x1 boards. It is important to monitor for any warping of the boards over time and promptly replace them as needed. The beams should be arranged in a sequence from widest to narrowest, with the two narrowest beams equipped with lateral braces to prevent rotation and movement. Additionally, the beams should be marked with calibrated lines, starting at 2 feet along the widest beam and extending to the end of the narrowest beam.

Regarding space requirements, the fully assembled beam spans a length of 24 feet. Therefore, a minimum space of 30 feet in length and 8 feet in width is necessary to accommodate the NBWT adequately. This allows sufficient room for both the evaluator and spotter to be present near the patient during testing. Furthermore, the NBWT should be placed at least 0.91 meters (3.0

feet) away from walls or other structures to prevent participants from relying on them for support during the assessment.

2.1.3 Experimental Setup and Protocol

A pilot case study was conducted to investigate the performance on NBWT under influence of mediolateral hip torque. This study was conducted with the approval of the Institutional Review Board of the NC State University, with the subject providing informed consent. One left leg amputated subject (Sex: Male, Age:26, Height: 170cm, Weight: 75kg) participated in this study. The kinematic data of the subject is recorded using a wireless motion capture system (XSens MVN) during the experiments.

The subject was asked to wear the mediolateral hip exoskeleton during the whole experiments. The goal of the NBWT is to maximize the distance walked along the narrow beam. The procedure began with the subject positioned stationary at the onset of the narrow beam, with their preferred foot placed on the initial beam and the other foot off the beam. Upon initiation of the test, the subject was instructed to cross their arms and commence walking along the beam at their preferred pace and step length until the end of the test. Termination of the test occurred if the subject's foot strayed off the narrow beam or if their arms became uncrossed. The position of the subject's foot on the beam at the test's conclusion was documented as the final score. To ensure safety, a healthy individual accompanied the subject throughout the tests. The detailed NBWT instructions could be found in Appendix A [19,20].

The subject was asked to do the test under 5 conditions: zero impedance, abduction torque (K=80), abduction torque (K=50), adduction torque (K=80), adduction torque (K=50). The inertial and damping parameters of the exoskeleton remained unchanged in each condition. For each condition, the subject was asked to walk on the beam 9 times. The first 2 trials were used to let the subject get familiar with the test environment, and their data were not recorded. The subject has a 1 minute break between each trial and a 5 minute break between each condition.

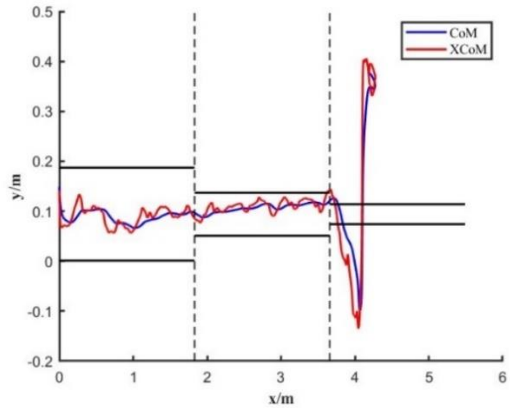
The test score and the body kinematics were recorded in each trial. In order to investigate the performance under different torques, we focused on the left and right heel position, center of mass and extrapolated center of mass, and the maximum torso sway angle. The results and analysis are demonstrated in the following sections.

2.2 Results

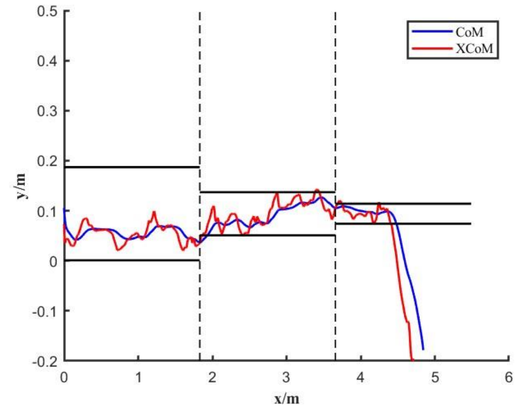
In this section, the primary objective is to examine the influence of mediolateral torques generated by the exoskeleton on the subject's performance during narrow beam walking. Initially, we focus on analyzing center of mass positions and the tracking of heel positions. Subsequently, we delve into the assessment of maximum heel positions and torso sway angles in the mediolateral direction, with particular attention to the falling side. This comprehensive approach allows us to gain insight into how the mediolateral torques exerted by the exoskeleton impact various aspects of the subject's performance on the narrow beam, including balance control and stability.

2.2.1 Center of Mass Positions

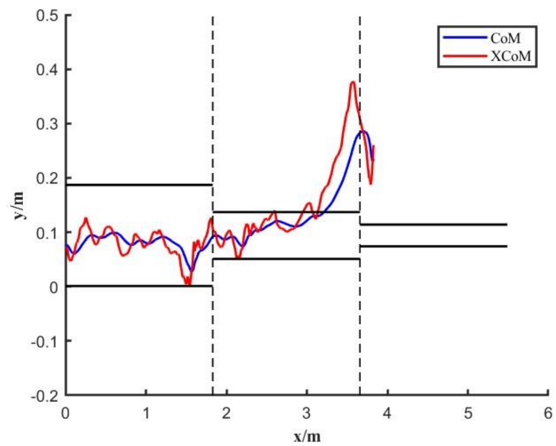
The center of mass (CoM) and extrapolated center of mass (XCoM) positions of three representative trials from the three conditions: zero impedance, adduction torque (K=-80), and abduction torque (K=80) are shown in Figure. 3, respectively. The position of the narrow beam is displayed in solid black line, and the center of mass and extrapolated center of mass positions are shown in blue and red curve, respectively.



(a) Zero impedance



(b) Adduction torque (K=-80)



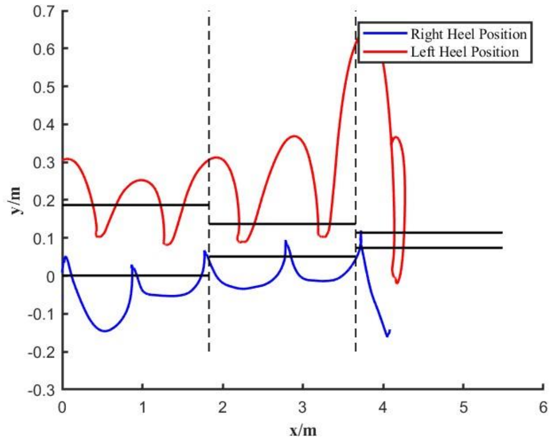
(c) Abduction torque (K=80)

Figure 3. Center of mass and extrapolated center of mass positions of three representative trials

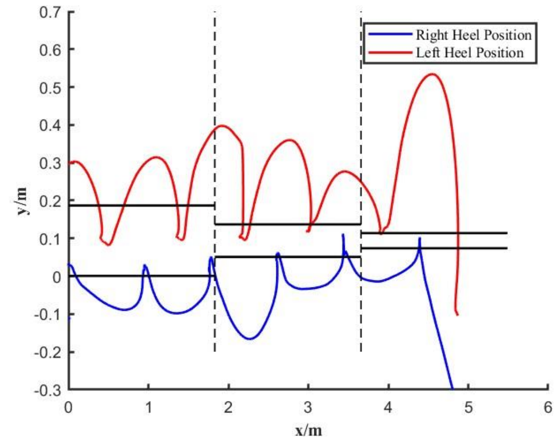
For all trials depicted in Figure 3, the subject successfully completed the first segment of the narrow beam. During the zero impedance trial, the subject fell at the end of the second segment. In the adduction torque trial (K=-80) and abduction trial (K=80), the subject fell at the first step of the third segment and the middle of the second segment, respectively. It is noteworthy that the width of the second segment is narrower than the subject's foot width, thereby making it the effective base of support. According to the literature [39], an individual can maintain balance if their extrapolated center of mass (XCoM) remains within their base of support. The results corroborate this, demonstrating that the subject maintained balance when their XCoM was within the confines of the narrow beam. Conversely, the subject fell if their XCoM extended beyond the beam. The subsequent section details the movement of the heel position.

2.2.2 Heel Position Tracks

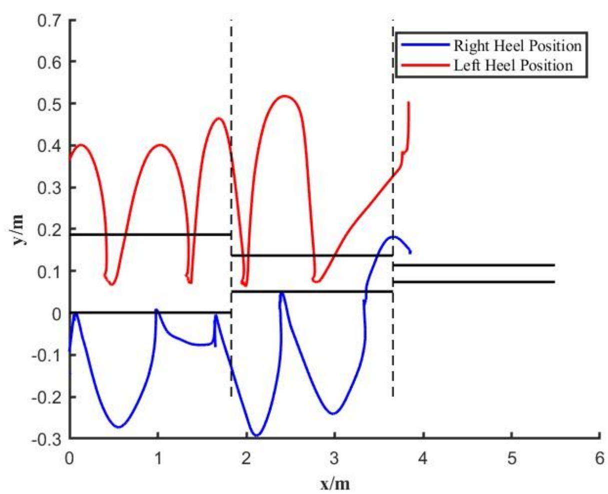
The left and right heel positions of three representative trials from the three conditions: zero impedance, adduction torque (K=-80), and abduction torque (K=80) are shown in Figure. 4, respectively. And the left and right heel positions are shown in red and blue curve, respectively. The comparasion of heel positions of the zero inpedance, adduction torque (K=-50), and abduction torque (K=50) are shown in Appendix B, which is not significant different. The potential reason for this effect is that the torque provided by the exoskeleton is not significant enough to affect the subject’s behavior.



(a) Zero impedance



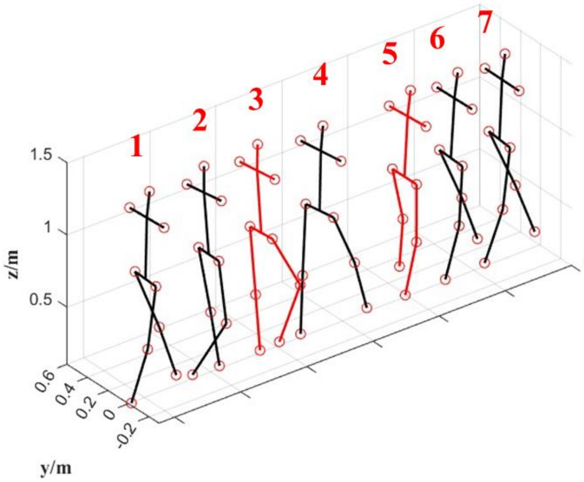
(b) Adduction torque (K=-80)



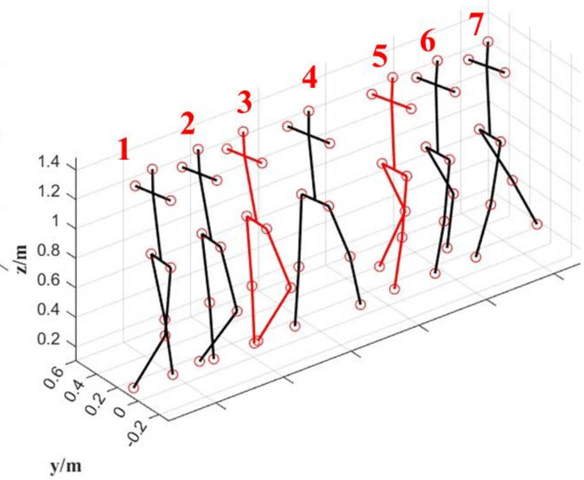
(c) Abduction torque (K=80)

Figure 4. Heel position tracks of three representative trials (K=0, 80, -80 Nm/rad)

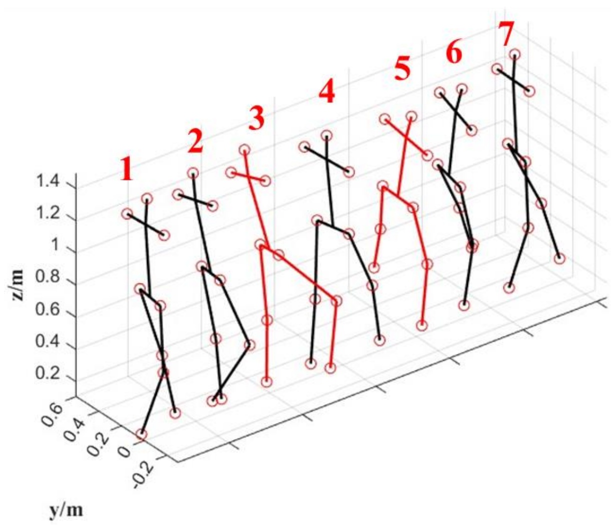
Since the subject is not permitted to use his arms for balance, leg movements in the mediolateral direction become crucial for maintaining stability. The greater the leg extension, the larger the heel's range of motion in the mediolateral direction. Figure 5 illustrates the body motion diagram during one representative gait of the three conducted trials. The most significant leg extension moments for the right and left legs in the mediolateral direction are shown in frames 3 and 5, respectively. The differing heel positions in the mediolateral direction for the left and right legs suggest that the subject employs different balance strategies for the amputated and intact sides. Further analysis of the maximum heel displacements in the mediolateral direction and the maximum torso sway angle will be provided in the subsequent section.



(a) Zero impedance



(b) Adduction torque (K=-80)



(c) Abduction torque (K=80)

Figure 5. One gait diagram of three representative trials

2.2.3 Maximum Heel Displacements and Torso Sway Angles in Mediolateral Direction

The boxplots of maximum heel displacements in mediolateral direction on segment 1 and segment 2 are shown in Figure 6 and Figure 7, respectively. In a boxplot, the outlier is determined based on the interquartile range (IQR). The IQR is the range between the first quartile (Q1) and the third quartile (Q3). In this work, the outliers in a boxplot are those values that lie more than 1.5 times the IQR below the first quartile or above the third quartile. These points are typically marked with small circles in the boxplot. A nonparametric one-way ANOVA test (Kruskal-Wallis test) and subsequent Dunn's multiple comparison test were applied to compare three or more groups of data. The difference with a p-value smaller than 0.05 was taken as significant.

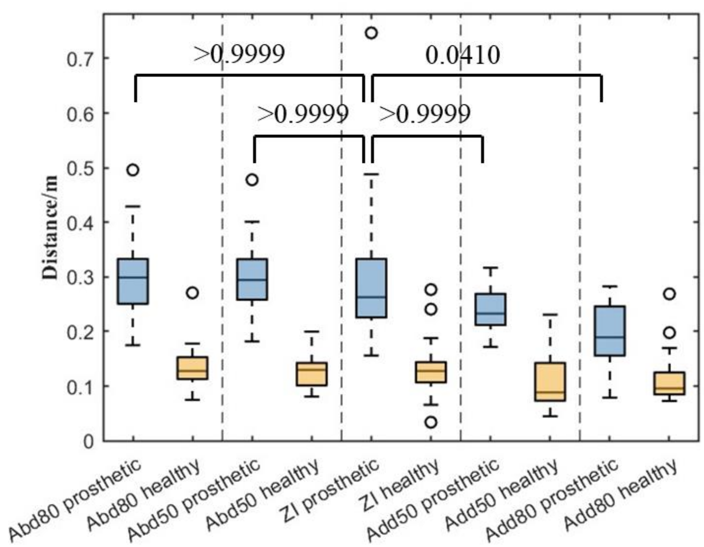


Figure 6. Maximum heel displacements in mediolateral direction on segment 1. N=12 for each group. The left side is the prosthetic side while the right side is the normal side. P-values between each group are marked above.

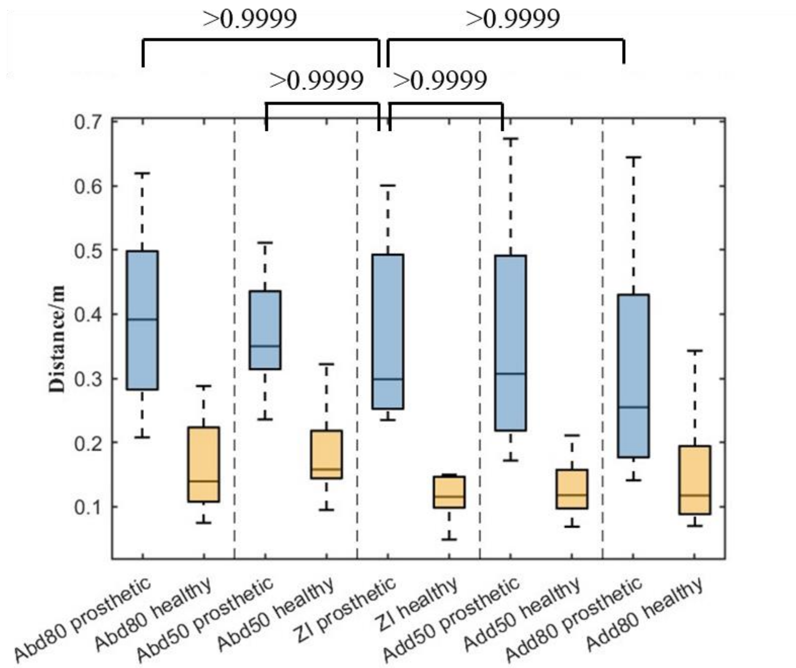


Figure 7. Maximum heel displacements in mediolateral direction on segment 2. N=12 for each group. The left side is the prosthetic side while the right side is the normal side. P-values between each group are marked above.

In general, the left leg (prosthetic side) exhibits larger maximum heel displacements than the right leg (intact side) across both segments. This indicates that the subject employs different strategies for using his prosthetic and intact leg during the swing phase. For both segments 1 and 2, the maximum heel displacements of the prosthetic leg in the mediolateral direction have a trend to decrease during adduction torque trials and increase during abduction trials, aligning with the directions of torque application. However, the statistic result showed that only zero impedance and adduction torque (K=80) are significantly different. The torque application effect is less pronounced on the intact leg side, suggesting better control of the subject's intact leg. The statistic results show that no group is significantly different from others. Additionally, on segment 2, the variation in maximum heel displacements in the mediolateral direction for both legs is greater compared to segment 1, highlighting the increased difficulty of walking on segment 2. The prosthetic side shows a more significant increase in variation than the intact side, further indicating better control ability of the intact leg.

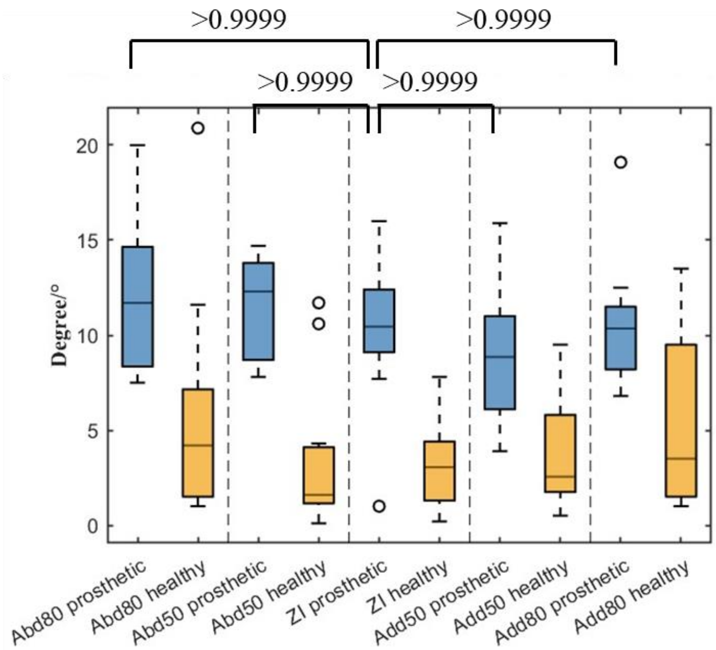


Figure 8. Maximum torso sway angles in mediolateral direction on segment 1. N=12 for each group. The left side is the prosthetic side while the right side is the normal side. P-values between each group are marked above.

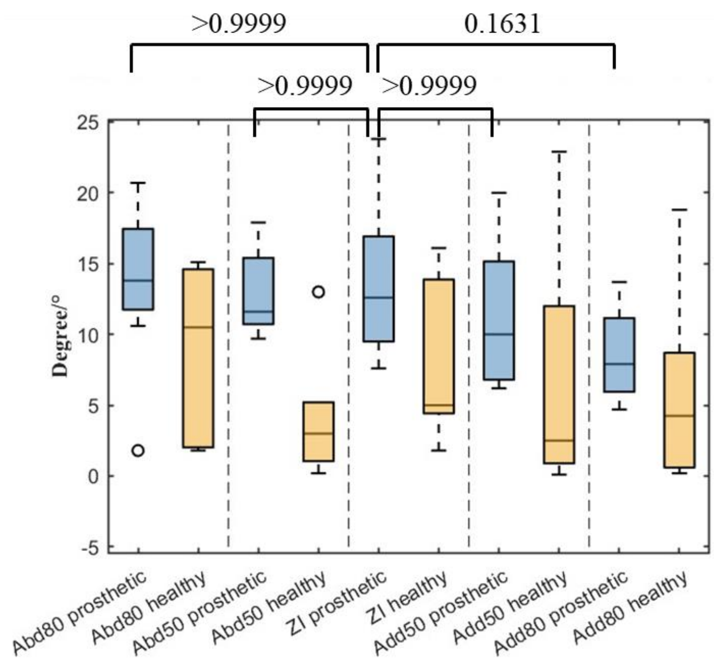


Figure 9. Maximum torso sway angles in mediolateral direction on segment 2. N=12 for each group. The left side is the prosthetic side while the right side is the normal side. P-values between each group are marked above.

The boxplots of maximum torso sway angles in the mediolateral direction for segments 1 and 2 are displayed in Figure 8 and Figure 9, respectively. Across both segments and various torque trials, the maximum trunk sway angles are smaller when the subject stands on his intact leg compared to when he stands on the prosthetic side. A significant reason for this is the role of the ankle in maintaining balance. The subject can use his intact ankle to stabilize himself, resulting in smaller maximum torso sway angles on the intact side. In contrast, when standing on the prosthetic side without active ankle support, the subject must rely more on torso sway to maintain balance.

On segment 2, the maximum torso sway angles in the mediolateral direction has a trend to decrease during adduction torque trials but remain relatively unchanged during abduction trials.

The statistic result also showed that only zero impedance and adduction torque (K=-80) trials are significantly different. However, on segment 1, the torque application effect is not as pronounced, where the statistic result showed that no group is significantly different others in the prosthetic side. This difference may be attributed to the greater challenge posed by segment 2 compared to segment 1. The subject's enhanced ability to maintain balance on segment 1 results in more stable torso sway, making it less susceptible to changes. The larger variation in maximum torso sway angles on segment 2 further indicates that walking on segment 2 is more challenging.

2.2.4 Falling Side and Test Score

The falling sides and legs are illustrated in Figure 10. The red bar indicates the number of times the subject falls on his healthy leg, while the blue bar indicates the number of times the subject falls on his prosthetic leg. The left side is where the prosthetic leg located and contribute to more falls than the right side. As shown in the figure, the subject is more likely to fall on his prosthetic leg and prosthetic side (left side).

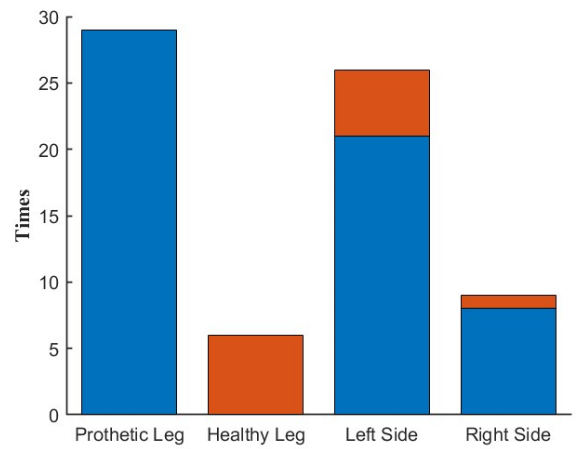


Figure 10. Falling side counts. The red bar indicates the number of times the subject falls on the healthy leg, while the blue bar indicates that on the prosthetic leg. The left side is the prosthetic side while the right side is the normal side.

The overall NBWT scores for the five conditions are presented in Figure 11. A nonparametric one-way ANOVA test (Kruskal-Wallis test) was applied to compare the five groups of data. The difference with a p-value smaller than 0.05 was taken as significant. Although torque application does affect the subject's walking performance to some extent, the overall test scores do not show significant differences among the conditions. However, the variations in test scores are larger in the torque application trials compared to the zero impedance trials.

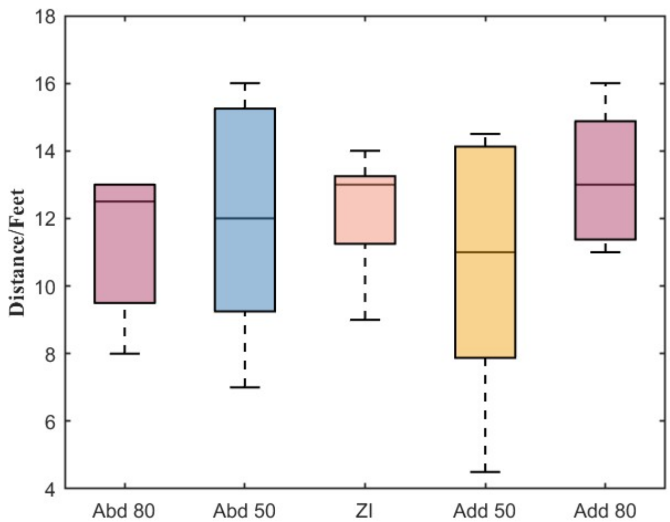


Figure 11. Test scores in each condition. N=**7** for each group. A nonparametric one-way ANOVA test (Kruskal-Wallis test) was applied to compare the five groups of data and the p-value was larger than 0.99.

2.3 Discussion and Future Scope

The main goal of this chapter is to investigate how mediolateral torque provided by hip exoskeleton could affect the NBWT performance. The findings of this study could be potentially used to further investigate how mediolateral hip exoskeleton could help improving walking balance as an assistive device.

2.3.1 Analysis of the Torque Application Effect on NBWT

The application of different assistive torques at the hip significantly influences the kinematic behavior of both the foot and torso during the Narrow Beam Walking Test (NBWT). Adduction torque, which aids in bringing the leg towards the midline during swing phase and pushing the torso towards the midline during stance phase, was observed to decrease the maximum heel displacements and torso sway angles in the mediolateral direction, particularly on the more challenging segment 2. This stabilization effect is more pronounced on the prosthetic side, indicating that the adduction torque helps the subject to move the prosthetic leg back to the beam during swing phase and mitigate the inherent instability caused by the lack of an active ankle joint in the prosthetic leg during stance phase. The feedback from the subject supports these observations.

Conversely, the abduction torque, which pushes the leg away from the midline during swing phase and pushes the torso away from the midline during stance phase, presented a greater challenge for the subject, especially on the prosthetic side. This difficulty is evidenced by increased mediolateral heel displacements and larger torso sway angles during abduction torque trials. The subject reported that controlling the prosthetic leg with abduction torque was more difficult, leading to less stable walking patterns. These findings suggest that while adduction torque is promising in enhancing stability, abduction torque may exacerbate instability, highlighting the need for tailored torque application strategies to improve walking balance in individuals using prosthetic limbs.

The dynamic relationship between the applied hip torques and the kinematic changes observed at the foot and torso can be explained by the mechanics of balance and gait. In normal gait, the hip joints produce both adduction and abduction torque in the frontal plane. These torques are essential for controlling the lateral movements of the body and ensuring that the legs can effectively transition from one phase of the gait cycle to the next [49,50]. For example, adduction torque stabilizes the pelvis during the transition from swing to stance phase, while abduction torque helps during the transition from stance to swing phase [50].

When the step width is reduced, for example when the subject is doing the NBWT, the requirement for precise control of these torques increases. Narrower step widths demand finer adjustments in mediolateral balance, thus increasing the reliance on both hip adduction and abduction torque to stabilize the body and prevent falls [51]. Specifically, a narrower step width brings the body's center of mass closer to the edge of the support base, necessitating stronger adduction and abduction torque to maintain balance [52]. Studies have shown that reduced step widths require greater mediolateral stability, thereby increasing the demand for hip adduction and abduction torques to maintain balance [51].

However, in our research, we observed that the participant preferred assistance in the form of adduction torque and found it beneficial in returning the leg to the midline. This preference was particularly noticeable on the prosthetic side. This is expected due to the fact that the participant is a trans-femoral amputee, who has the limitation of weaker adductor muscles on the prosthetic side [53]. This factor makes it more challenging for the participant to control the prosthetic leg, especially when abduction torque is applied.

The subjective feedback from the participant is aligned with our expectations. The participant reported difficulty in controlling the prosthetic leg when abduction torque was applied, particularly when returning the leg to the midline. Additionally, excessive abduction torque while moving the leg away from the midline further strained the adductor muscles. The poorer performance observed in the abduction torque trials suggests that the subject's adductor muscles are weaker, indicating that adduction torque could provide necessary support.

In summary, both adduction and abduction torque are essential for maintaining mediolateral balance during walking, especially when the step width is reduced. Our study's findings, along with the participant's feedback, highlight the specific challenges faced by a trans-femoral amputee and indicate the potential benefits of providing adduction torque assistance to improve gait stability and performance.

2.3.2 Future Scope

Since this is a very preliminary case study on how mediolateral torque could affect the NBWT performance, we only applied a few constant torque values in the experiments. In the future, we should investigate more torque application strategies. For instance, different gait cycles should be separated and different torque values and directions should be applied in different gait cycles, since the need for assistance could be different in different gait cycles. Besides, more subjects should be recruited to verify the consistency of the results.

CHAPTER 3: Development of a Wearable Rehabilitation System aimed at Enhancing Mediolateral Balance

The content of this chapter is derived from a conference paper previously published by the author in 2023 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS) [54]. Permission for the use of this paper has been obtained from all co-authors of the paper.

3.1 Methods

The aim of this section is to develop and verify a system consisting a wearable robotic platform integrated with visual feedback to be potentially used in neuro-rehabilitation for amputees and older population, specifically focusing on mediolateral balance. In order to achieve the goal of training users in hip abduction-adduction control, we developed a visual feedback interface integrated with a hip exoskeleton. In this section, we will describe the rehabilitation system, including the visual feedback user interface, and the experiment to validate the system. The rehabilitation system is shown in Figure. 12.

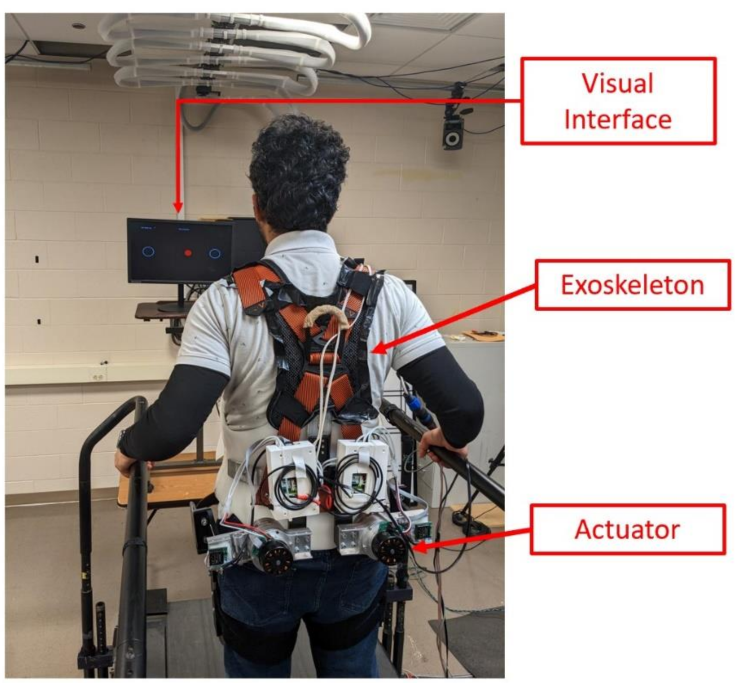


Figure 12. Hip abduction/adduction exoskeleton and experimental environment

3.1.1 Visual Feedback User Interface

An important aspect of neurorehabilitation is facilitating visuomotor integration, generally using visual feedback. In this study, a visual feedback was implemented to provide users their real time hip position, which is then used to help users move their hip to a target position while under the influence of the resistance provided by the robot. As shown in Figure 13, the control parameters, M, B, K, which have been introduced in chapter 2, are sent from the PC to the exoskeletons before training. The hip positions from the exoskeleton are sent to the PC in real time during training, which are displayed on the visual feedback interfaces.

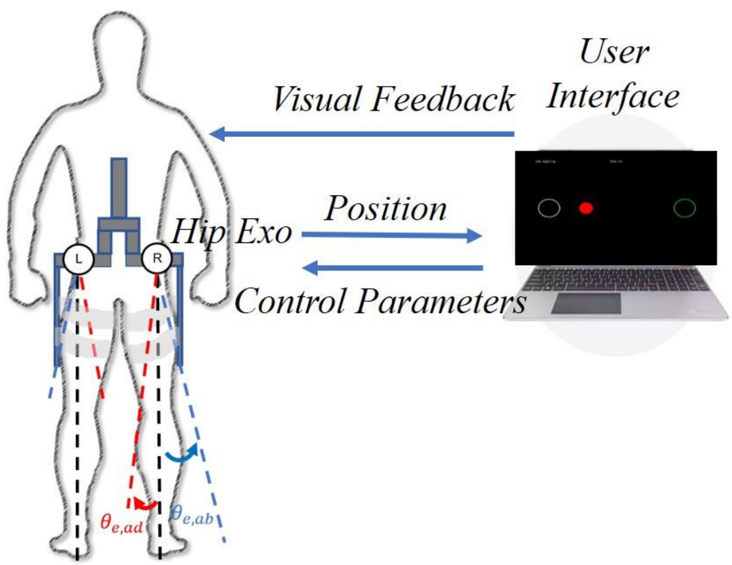


Figure 13. Diagram of the rehabilitation system

The interface, built using OpenGL and implemented in Python, is shown in Figure. 14. The visual interface is designed to facilitate training of both limbs and the current limb undergoing training is displayed on the top left. Additionally, the top of the interface has text that is used to instruct participants on different phases of motion during the rehabilitation protocol. The two hollow circles in the middle represent the start and target position, which are fixed. The border color of these circles is changed to green when subjects are expected to reach these positions by controlling the robot. A solid circle represents the current angle of the hip joint. The circle is shown in red during motion and will turn green once it is in bounds of the target position represented by one of the two hollow circles.

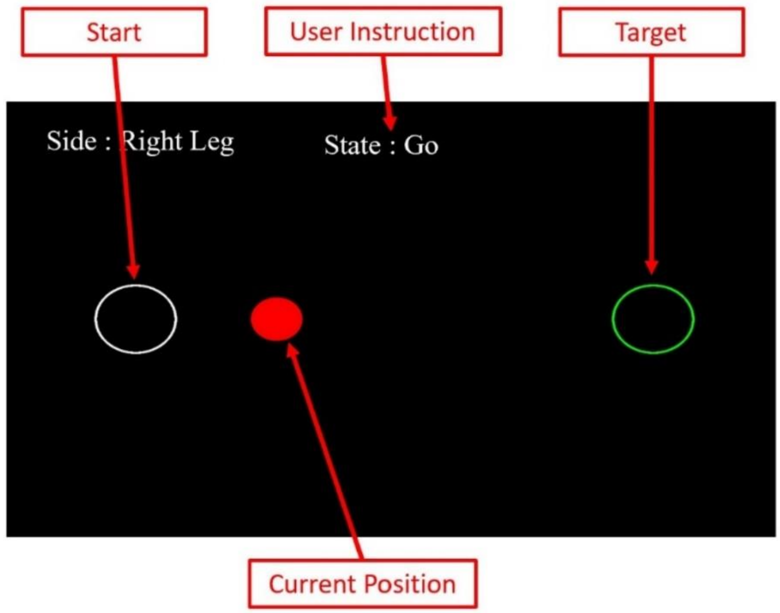


Figure 14. Screenshot of visual feedback user interface

The data communication between the user interface and the controller is accomplished through User Datagram Protocol (UDP) with an ethernet cable. In this paper, the communication frequency is 50Hz. As shown in Figure. 13, the control parameters are set at the beginning of each trial to the exoskeleton, and the hip position data is sent to the user interface.

3.1.2 Experimental Setup and Protocol

A pilot study was conducted to test the performance of the robotic rehabilitation system and observe user visuomotor adaptation as well as biomechanical responses. This study was conducted with the approval of the Institutional Review Board of the NC State University, with all three subjects providing informed consent. Three non disabled subjects (Sex: Male, Age:26-33, Height: 170-184cm, Weight: 70kg-85kg) participated in this study. The maximum voluntary contraction (MVC) of Glueteus Medius (GM) muscles is recorded prior to the training session. Each of them was instructed to perform training trials using the rehabilitation system for a total of 8 blocks of trials. In each block, the users were instructed to abduct their dominant leg hip joint to the target position and back, while the robot simulates different stiffnesses. The target position was set to 15° and the inertial and damping parameters were set to constants of 0.2 Kgm^2 and 0.3 Nm s/rad correspondingly. Participants were asked to do the training under 4 stiffness parameters: 20, 40, 60, and -10 Nm/rad. For each stiffness, users are instructed to undergo 20 trials, each split into a block of 10 trials resulting in 8 blocks.

Each trial lasts for a total of 20 seconds. For the first two seconds, the visual interface instructs the user to be "ready" for motion, while they maintain the stance phase. After 2 seconds of command "Ready", participants were alerted by a non-startling beep as well as the command "Go" on the interface, to instruct them to move their training hip joint to the target angle as swiftly and as accurately as possible. A total of 7 seconds is allocated to ensure users reach the target angle, highlighted by green hollow circle. At the end of 7 seconds, they are instructed to move back to the starting position with a command "Back", and are allocated 7 seconds. After that, a relaxation time of 4 seconds is provided before they are instructed to be ready for the next trial. The robot maintains consistent stiffness during the entire block of trials. Adequate rest is provided between the trials to prevent fatigue. During the training, participants are allowed to hold the handrail to aid their balance. The stiffness parameter of the side which is not being trained is set to 100 Nm/rad to provide a rigid support.

The hip position, velocity, and interaction torque of both sides of the exoskeleton were recorded. The peak and average interaction torque during motion is analyzed to understand control of movement. Additionally, GM muscle activity is recorded to analyze different activation strategies utilized by subjects in response to different stiffnesses. All the data is captured at 1000 Hz and is passed through a 4th order butterworth filter prior to analysis. In addition to the kinematics and kinetics, the variability of motion for the first 5 trials and last 5 trials for each stiffness are evaluated to analyze visuomotor adaptation.

3.2 Results

The main goals of this section are to understand user adaptation to different resistances simulated by the robot by utilizing the visual feedback. The kinematic, kinetic and EMG responses are analyzed to understand these adaptations.

3.2.1 Kinematics Response

Figure. 15 shows a representative hip position profile of an individual subject during trials involving different stiffnesses. The dashed vertical lines represent the start of abduction and adduction motion. As shown in Figure. 15, the subject was able to reach the target position in all the four different stiffness trails in about 3 seconds from the start of the movement. Overall, all three subjects managed to reach the target position for each of the four stiffnesses. It was observed that the position profile for stiffness condition of -10 Nm/rad resulted in a consistent overshoot, which could be due to the direction of force applied by the robot. However, all subjects were able to adapt to the unstable nature of the force and still reach the target. The settling times of three subjects across trials for both abduction and adduction motions are shown in Figure. 16(a) and Figure. 16(b). Subjects were found to reach steady state position of the target and back to reference starting point in under 2.5 sec, with the 60 Nm/rad trial being the slowest during target reaching and -10 Nm/rad being the slowest for adduction motion back to reference starting point.

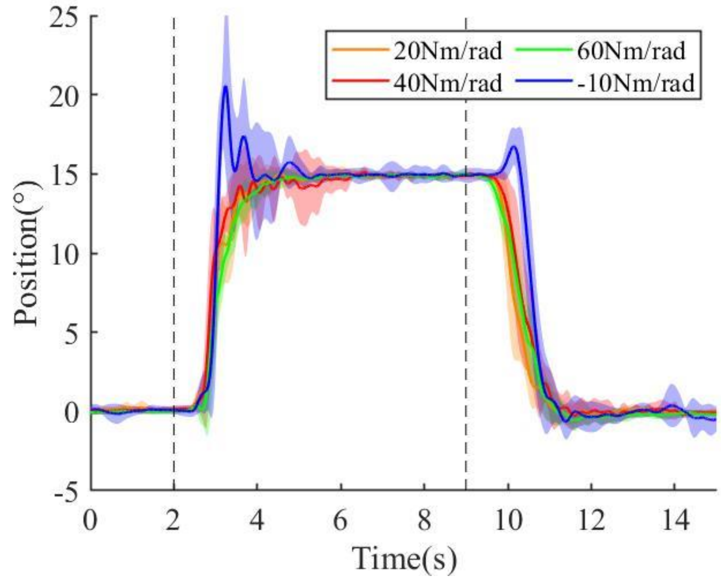
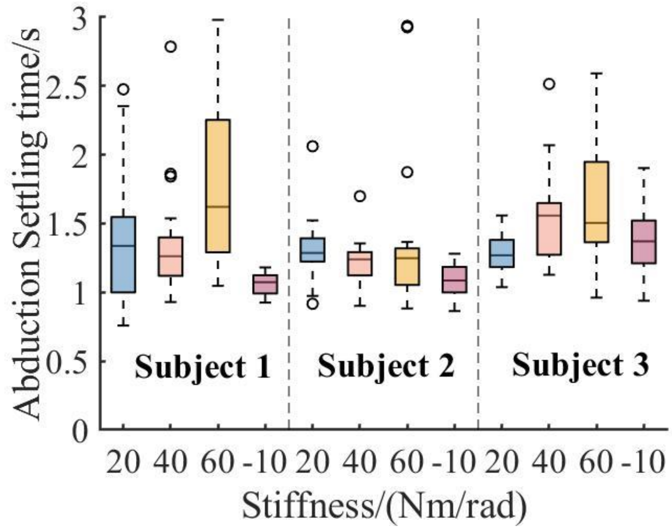
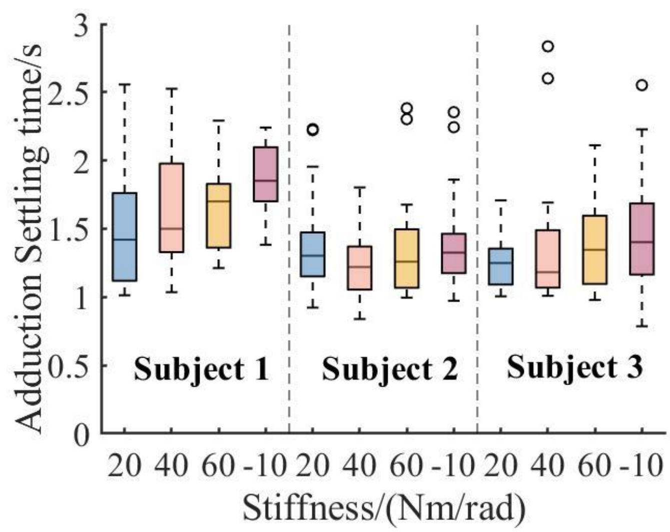


Figure 15. Hip Joint Angle during training. of an individual subject during trials involving different stiffnesses. The dashed vertical lines represent the start of abduction (2s) and adduction (7s) motion.



(a) Abduction settling time during trials involving different stiffnesses



(b) Adduction settling time during trials involving different stiffnesses

Figure 16. Boxplots of settling time response for all subjects. N=20 for each group. A nonparametric one-way ANOVA test (Kruskal-Wallis test) was applied to compare the five groups of data and the p-value was larger than 0.99.

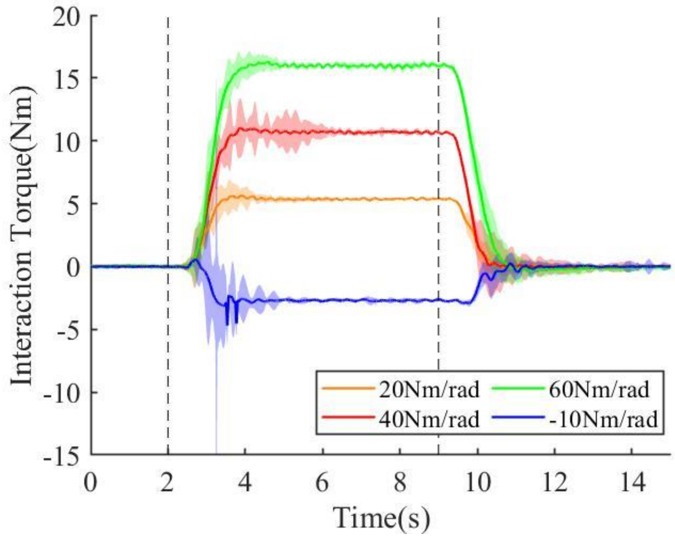
The mean standard deviation from 2s to 4s of the first and last five trials for each subject across the four stiffness values are shown in Table 1. The table shows that the motor performance of each subject has become more consistent during the later trials. Specifically, subject 2 showed the highest adaptation from the initial trials to the final ones, while subject 1 has shown the least. This data shows that there is a need to coordinate visual feedback and motion and these adaptations can be exploited to implement sensorimotor rehabilitation for people affected by amputation and the elderly.

Table 1. Standard deviation from 2s to 4s of first and last five movement trials for each stiffness

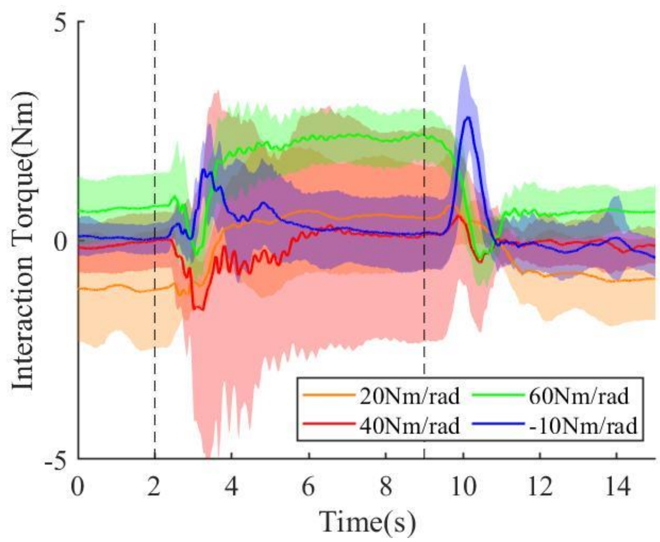
Stiffness (Nm/rad)	Subject 1		Subject 2		Subject3	
	1-5	16-20	1-5	16-20	1-5	16-20
20	1.51°	1.33°	2.33°	0.67°	1.49°	0.82°
40	1.87°	1.08°	2.58°	0.44°	2.71°	0.96°
60	1.32°	1.32°	2.29°	0.51°	1.96°	1.44°
-10	2.76°	1.95°	2.62°	1.45°	2.19°	2.18°

3.2.2 Kinetic Response

In addition to the kinematics, the interaction torque on both the training side and non training side are analyzed. Figure. 17(a) and Figure. 17(b) show the interaction torque for the training and no training side of a representative subject, where solid lines represent average interaction torque, and shaded areas represent standard deviation. As shown in Figure. 17(a), the static torque increases as the stiffness increases, which is reasonable to show the different training difficulties. The largest static torque is about 15 Nm, when stiffness is 60 Nm/rad. For the non training side, there is a higher torque observed during the 60 Nm/rad trials (Figure. 17(b)) and high variation during 40 Nm/rad trials, which suggests that users are using the contralateral limb to support their hip joint movement in order to reach the target during higher stiffness. In addition, the peak observed in the -10 Nm/rad trial further reinforces the contralateral limb coordination to perform the reaching tasks.

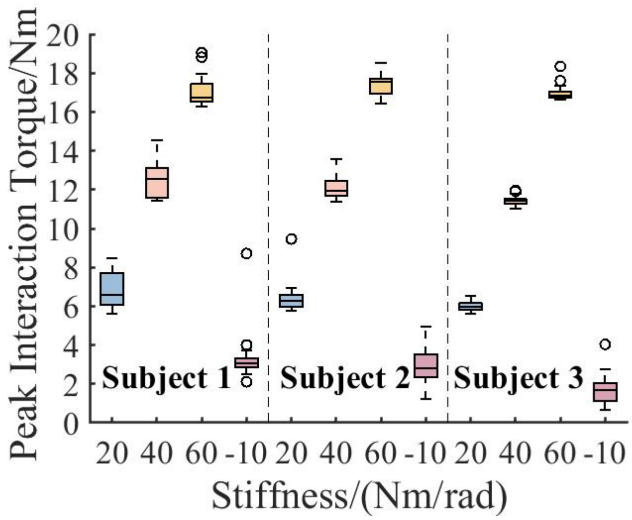


(a) Observed human-robot interaction torque during training

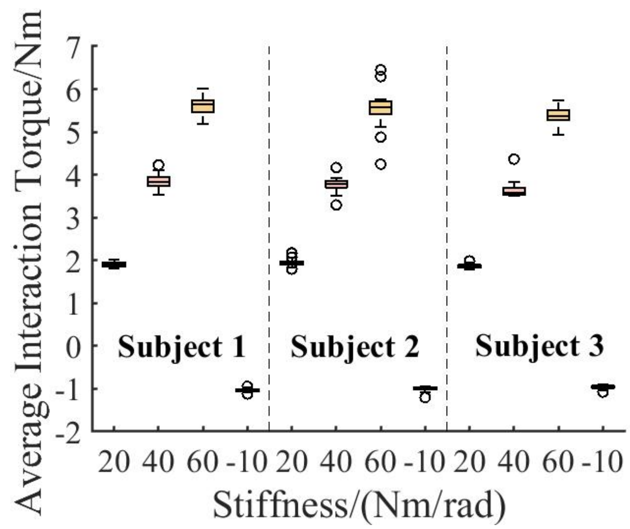


(b) Observed interaction torque on the non training side

Figure 17. Interaction torque of a single subject. during trials involving different stiffnesses. The dashed vertical lines represent the start of abduction (2s) and adduction (9s) motion.



(a) Peak interaction torque during trials involving different stiffnesses



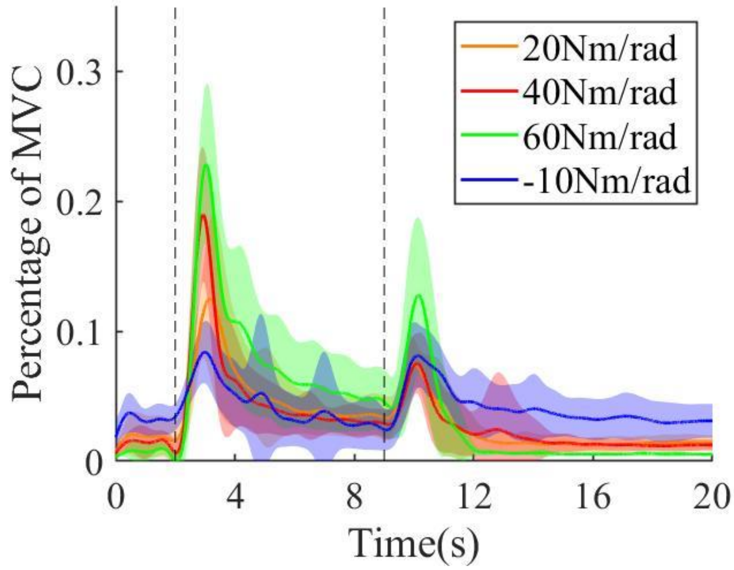
(b) Average interaction torque during trials involving different stiffnesses

Figure 18. Boxplots of peak and average interaction torque. N=20 for each group.

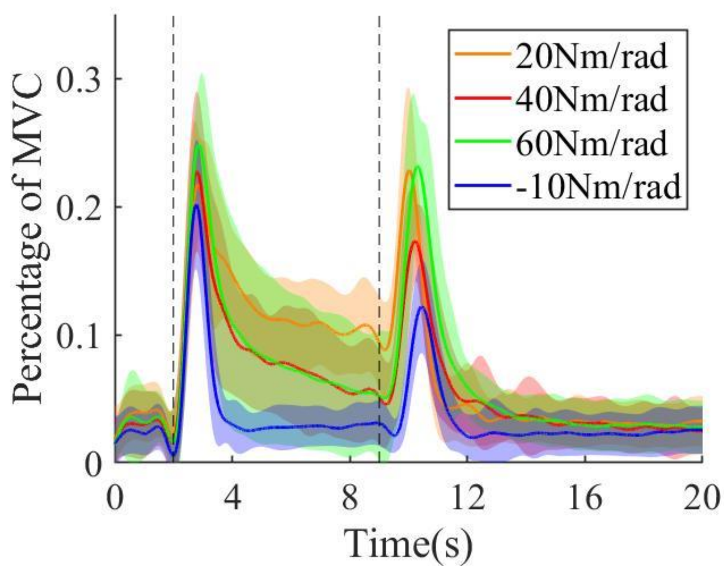
Figure. 18(a) and Figure. 18(b) represent the boxplots of peak interaction torque in the rising period and the average torque in the static period. The result shows a linear response with small variation, which shows users have high control over the torque response of their hip joint. Further, the lack of multiple peaks and the smooth increase in torque profile further imply consistent control of the joint torque generation capabilities of the users.

3.2.3 EMG Response

On analyzing EMG responses of subjects (Figure. 19), it is found that the activation increases with increase in stiffness as expected. It is observed that the GM muscle is activated during the static phases of the movement once the subjects reach the target during the 60 Nm/rad trial, while the activation is not as high during 20 and 40 Nm/rad trials. Additionally, it is observed that the EMG activity is higher during adduction part of the motion during -10 Nm/rad trials. These results show that by modulating the stiffness, the GM activity can be modulated during different parts of motion. More importantly, the EMG responses are also reflected in the non training side showing that there is activity in non training side to ensure stability during motion, showcasing interlimb coordination.

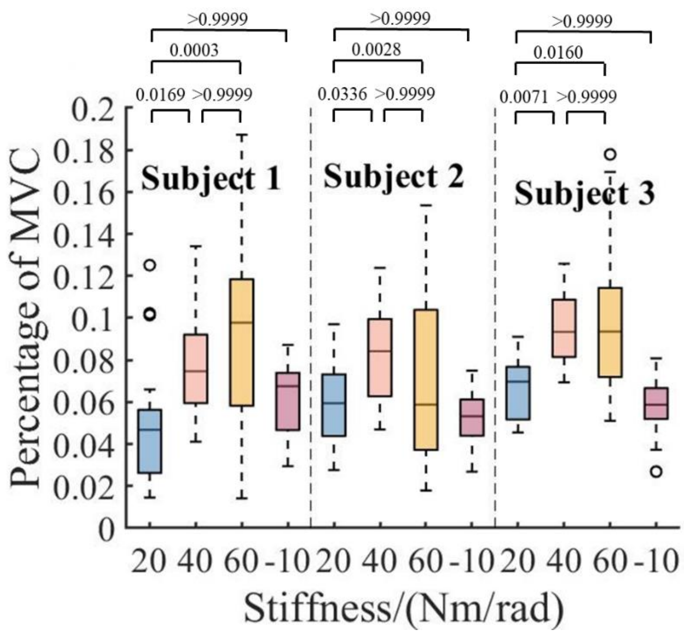


(a) Training side

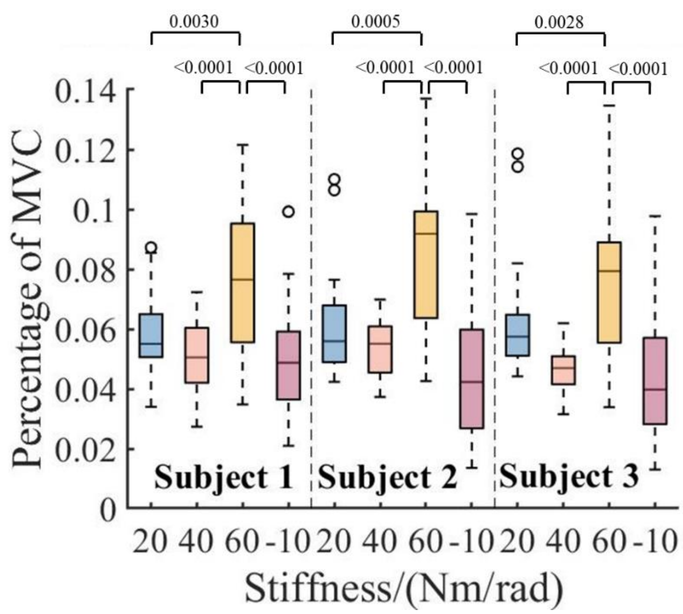


(b) Non training side

Figure 19. EMG response of training and non training side during trials involving different stiffnesses. The dashed vertical lines represent the start of abduction (2s) and adduction (9s) motion.



(a) Average EMG in rising period



(b) Average EMG in static period

Figure 20. Boxplots of average EMG responses of all subjects during trials involving different stiffnesses. N=20 for each group. A nonparametric one-way ANOVA test (Kruskal-Wallis test) and subsequent Dunn's multiple comparison test was applied, and the p-value was marked above.

The average EMG in the rising period and the static period for all the three subjects are shown in Figure. 20(a) and Figure. 20(b). A nonparametric one-way ANOVA test (Kruskal-Wallis test) and subsequent Dunn's multiple comparison test were applied to compare three or more groups of data. The difference with a p-value smaller than 0.05 was taken as significant. As expected, the average EMG is found to increase with increasing stiffness during motion. The 20Nm/rad trials shows significant difference with 40 and 60 Nm/rad trials through all the three subjects. However, the 40 and 60 Nm/rad trials are not significantly different. Once steady state is achieved, the EMG is found to be higher only for 60 Nm/rad stiffness, as observed that the 60Nm/rad trials are significantly different from the other three conditions across three subjects and the 20, 40, and-10 Nm/rad trials show no significant difference. Additionally, the variability in stiffness is found to increase during steady state for -10Nm/rad which could be a response to dealing with the instability in movement.

3.3 Discussion and Future Scope

The main goal of this chapter is to develop a neurorehabilitation system and analyze its ability to influence visuomotor mechanisms in non disabled populations. The findings from this study would aid in development of rehabilitation protocols for amputees and elder population.

3.3.1 Kinematic and EMG Analysis

The main focus behind kinematic analysis was to observe how users respond to different stiffnesses. It was expected that increasing stiffness would result in an increased rise and settling time, with the possibility that users would fail to reach the target at higher stiffnesses. Considering that the kinematic responses were similar to all three positive stiffnesses, it is possible that the resistance simulated was not challenging enough for non disabled users. In contrast, subjects showed that the negative stiffness environment was more challenging to control as seen by the increased overshoot and the higher variability during movement and steady state. Moreover, there is an increased settling time during abduction for 60 Nm/rad which reflects the additional effort needed for abduction while the -10 Nm/rad has higher settling time during adduction due to the unstable nature of the environment. The observed EMG responses correlate with the observed kinetic responses for positive stiffness trials, with higher activation required for trials involving higher torque. For the negative stiffness trial, it is observed that there is residual activation during normal stance phase, which could be a response to the unstable nature of the negative stiffness interaction. This response can be exploited for rehabilitation where GM muscle can be activated without the need for abduction in subjects with limited muscle activation control by exposing them to negative stiffness environment.

Looking at variability of motion during the first 5 trials and the last 5 trials, we observed that the participants were able to better control the motion as reflected by reduced variability in motion. This result shows that the users were able to coordinate the visual feedback with their hip joint motion to ensure more consistent motion. The ability to control movement based on visual feedback and systems that can facilitate that mechanism has been an important component of robot aided neurorehabilitation showing correlation with clinical outcomes [35]. Hence, the system could be a relevant solution for neurorehabilitation aimed at mediolateral balance.

3.3.2 Interaction Torque Analysis

The observed interaction torque on the training side shows that subjects were able to consistently reach the target and maintain the position once in the target. A more interesting observation is found by investigating the non training side interaction torque. While interaction torque on training side shows similar step response behavior with only varying magnitude across different stiffnesses, the non training side shows that subjects behaved differently for higher stiffnesses. It appears that subjects have utilized contralateral limb and coordinated motion to achieve the target at higher stiffness. In addition, there appears to be increased coordination between limbs in the -10 Nm/rad trials. Finally, the higher variation during 40 Nm/rad trials could possibly be due to users exploring different strategies employed to reach the target. Considering mediolateral balance involves inter limb coordination to ensure stability both during stance and locomotion [36], the fact that the certain stiffness trials induce coordinative behavior could be an important factor for rehabilitation. If the system could induce similar coordinative responses in people suffering from lower limb disability, the system could facilitate rehabilitation of interlimb coordination essential for gait and balance.

3.3.3 Future Scope

While the results from the current preliminary study are encouraging, it is uncertain how well the observed behaviors translate to amputee and older populations. However, this study could provide valuable feedback in development of rehabilitation protocols to address specific limitations in balance of people with poor balance ability. For example, a negative stiffness based protocol can be implemented in subjects with minimal EMG activity in GM muscle, while high stiffness based protocols can be implemented for muscle strengthening. Since the stiffness can be instantaneously changed, the stiffness value could further be fine-tuned to suit the individual needs, while still ensuring the neuromuscular behavior is trained. Investigating the evolution of kinematic and kinetic behavior of amputee and elder subjects and the corresponding clinical outcomes through a pilot study would provide further insights into the relevance of current solution in improving mediolateral balance and thereby reducing falls in these populations.

CHAPTER 4: Conclusion

This thesis investigated the potential of a hip abduction/adduction exoskeleton to improve mediolateral balance, specifically through two main parts: the impact of mediolateral hip torque on Narrow Beam Walking Test (NBWT) performance and the development of a wearable neurorehabilitation system integrated with a visual feedback interface.

The first part of the study (**Chapter2**) focused on the effect of mediolateral hip torque on NBWT performance, which is tested on a single subject with a left leg amputation. The exoskeleton's mediolateral torque influenced the kinematic performance of the subject during NBWT. The adduction torque trials showed a reduction in maximum heel displacements and torso sway angles, especially on the more challenging second beam. The distinct balance strategies employed by the prosthetic and intact legs were highlighted, demonstrating the potential of the exoskeleton to provide targeted assistance to improve walking balance.

The second part (**Chapter3**) involved developing and testing a neurorehabilitation system incorporating the exoskeleton and a visual feedback interface. Preliminary studies on non-disabled individuals showed increased motion consistency and improved interlimb coordination during training under various admittance conditions. The results indicated that the system could effectively facilitate visuomotor adaptation and could be a valuable tool in neurorehabilitation.

This research contributes to the growing body of knowledge on the use of wearable robotics for balance improvement and rehabilitation. The key contributions include: This first part of the study provides evidence that mediolateral hip torque can affect the subject’s behavior on NBWT, highlighting the exoskeleton's potential on improving the walking balance of amputees as an assistive device; the second part of the study indicates that the integration of the exoskeleton with a visual feedback system offers a novel approach to training and rehabilitation, emphasizing the importance of visuomotor integration in balance control.

While the findings are promising, further research is necessary to generalize the results across a broader population. For the first part of the study, additional amputee subjects should be recruited to verify the consistency of performance on the Narrow Beam Walking Test (NBWT). Moreover, different torque application strategies should be explored, as these have great potential to improve performance on the NBWT. For example, different gait cycles should be separated and torques in different values and directions should be applied, since the needs for assistance in different gait cycles could be different. For the second part of the study, it is essential to investigate more sophisticated control strategies for the exoskeleton, such as adaptive control algorithms that can adjust assistance levels in real-time based on the user's performance and needs. Additionally, conducting clinical trials to evaluate the system's effectiveness in real-world rehabilitation settings and its impact on reducing fall risks in target populations is crucial.

The results of this thesis underscore the potential of hip exoskeletons in enhancing mediolateral balance and providing a robust platform for neurorehabilitation. By addressing the specific needs of populations with impaired balance, such as amputees and the elderly, this technology holds promise for improving quality of life and reducing fall-related injuries. Continued research and development in this area are essential to fully realize the benefits of wearable robotics in rehabilitation and assistive applications.

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APPENDICES

Appendix A

Instructions of Narrow Beam Walking Test (NBWT) [19,20]

Resources

Time: 10 minutes

Personnel: Patient, test administrator, and spotter

Equipment: The Narrowing Beam Walking Test (NBWT) apparatus should be built to the establishing authors' specifications. 1 The NBWT consists of four, fixed-width beams:

• Beam 1: 6.0 ft (183 cm) long by 7.3 in (18.6 cm) wide

• Beam 2: 6.0 ft (183 cm) long by 3.4 in (8.6 cm) wide

• Beam 3: 6.0 ft (183 cm) long by 1.6 in (4.0 cm) wide

• Beam 4: 6.0 ft (183 cm) long by 0.8 in (2.0 cm) wide

Each board is approximately ¾-inches (1.5 cm) high. The NBWT beams can be made from common construction materials (i.e., high quality 2x8, 2x4, 2x2, and 2x1 boards), but be sure to replace any boards that warp over time. The beams should be connected in sequence from widest to narrowest. The narrowest two beams should have lateral braces to prevent rotation and movement. The beam should be scored with calibrated marks (i.e., lines), starting at 2 feet along the widest beam until the end of the narrowest beam.

Space: The fully constructed beam is 24 feet long. Therefore, you will require a space that is at least 30 feet long by 8 feet wide (to allow adequate room for the evaluator and spotter to stay near the patient). The NBWT should be placed at least 0.91m (3.0 ft) away from the wall or other structures so that participants cannot rely on them for support.

Cost: approx. $300

Administration

The Narrowing Beam Walking Test (NBWT) is administered by asking the patient to walk along the length of the connected beams with their arms crossed across their chest. The patient begins each trial with one foot on the wide end of the beam, and one foot on the ground next to the beam. The patient may use their assistive device to get into position, but should not use it when attempting the test. With their arms across their chest, patients are instructed to walk along the beam at any speed they choose. Five trials should be administered, though only the final three trials are used in scoring the measure. Begin the test by explaining the test to the patient:

• The goal of this test is to walk as far as possible along the beam. Speed is not being evaluated. Begin the test by standing with one foot on the wide end of the beam and the other foot on the ground to the side. You may choose which foot to put on the beam and which to put on the ground. Please cross both your arms across your chest.

Once you have provided an overview, provide a demonstration of how to walk the beam with arms crossed across your chest. Illustrate when the test will stop, both by stepping off the beam and by uncrossing your arms. After demonstrating, provide test instructions:

• When I say ‘begin,’ please walk along the beam as far as you can. Please walk at a comfortable speed. Remember to keep your arms crossed over your chest as you walk. Once you move your arms away from your body or step off the beam, I will ask you to stop. Are you ready? ‘Begin.’

Provide the test instructions above before the patient begins each trial. Spot or guard the patient, using a gait belt as appropriate, as they walk along the beam. Allow the participant to rest (at least 15 seconds) between each trial. Provide the test instructions before each trial and at least 15 seconds of rest after each trial.

Scoring

The test administrator should pay close attention to the patient as they walk along the beam, noting the location of the patient's feet when they either step off the beam or uncross their arms, whichever occurs first. Record the distance walked for each trial as the last 6-inch mark that the patient's forward foot crossed before they stepped off the beam or uncrossed their arms to regain their balance.

• For example, if the patient's right foot was between the 10-foot mark and 10-foot-6- inch mark when their left foot stepped off of the beam, the distance for that trial would be 10 feet.
• If the patient's forward-most foot was between the 11-foot-6-inch mark and the 12-foot mark when the patient/participant uncrossed their arms the distance for that trial would be 11.5 feet.
• If the patient's back foot is at the 13-foot-6-inch mark, and forward-most foot steps on the beam at the 14-foot-6-inch mark, but slides off to the side or the patient's foot then touches the ground, the distance for that trial would be 13.5 feet (the distance of the final foot to successfully accept body weight).

• Record the distance for the trial as ‘0’ if the participant cannot begin to walk along the beam, or if they do not pass the ‘0’ mark (which is located at a point 2 feet along the widest section of the beam). Once the patient’s forward-foot crosses the ‘0’ mark, record the distance as noted above.

Normalized distance, the overall score for the Narrowing Beam Walking Test, is calculated by dividing the distance walked in each of the three scored trials (the third, fourth, and fifth trials) by the calibrated length of the beam (i.e., 22.0ft (6.71m)). After dividing the distance of each trial by 22.0, find the average of the normalized distance for the three scored trials. This is the overall normalized distance for the NBWT.

Interpretation

The overall normalized distance for the NBWT will be a value between 0.0 and 1.0, where higher values (i.e., those closer to 1.0) represent better balance ability, and lower values (i.e., those closer to 0.0) represent poorer balance ability.

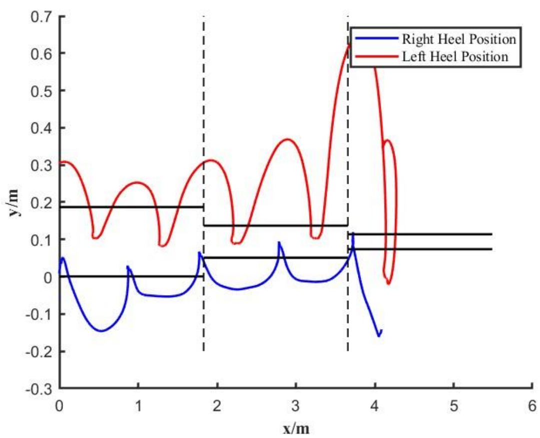
Documentation in Clinical Notes

When documenting the results of the test in clinical notes, first start with stating the name of test and the day the test was performed. Depending on what type of test was performed, describe the results (patient scored this normalized score). Compare these results to the results of the same test taken at an earlier time. Describe what these results mean in terms of patient health (improvement/deterioration, increase/decrease of patient’s balance abilities, etc...)

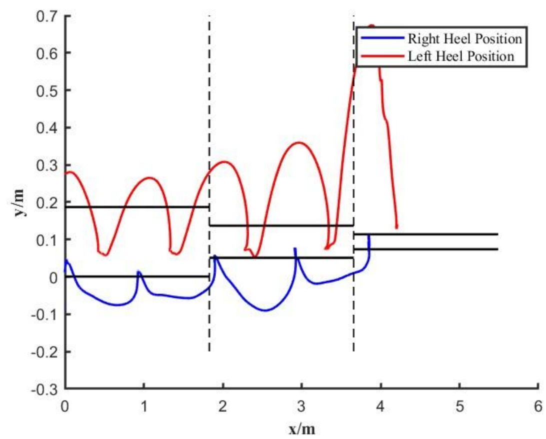
Example: When assessed with the Narrowing Beam Walking Test on (99/99/9999) the patient recorded a normalized score of (X.X). This score was [greater/less] than previously recorded. This represents an (X.X) point [decrease/increase] in the normalized score since last recorded on (99/99/9999), and therefore represents an [improvement/deterioration] in the patient's balance abilities.

Appendix B

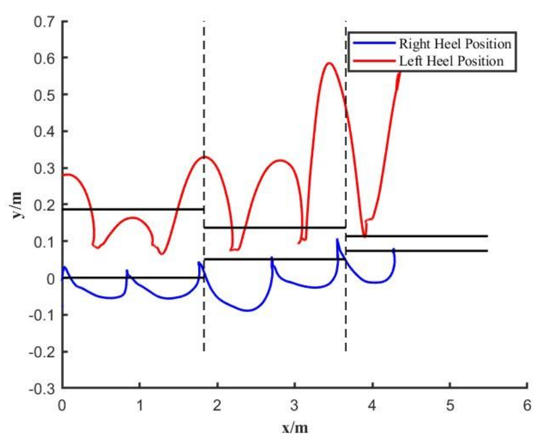
Comparision of Heel Position Tracks of Zero Impedance, K=50, and K=-50 Trials



(a) Zero Impedance



(b) Adduction Torque (K=-50)



(c) Abduction Torque (K=50)

Figure 21. Heel position tracks of three representative trials (K=0, 50, -50 Nm/rad)