8-1-2022

Balance Recovery Response in Community-Dwelling Adults with Unilateral Transtibial Amputation and the Potential Benefits of a Weight-Shifting Balance Training Intervention

Hui-Ting Shih

BALANCE RECOVERY RESPONSE IN COMMUNITY-DWELLING ADULTS WITH UNILATERAL TRANSTIBIAL AMPUTATION AND THE POTENTIAL BENEFITS OF A WEIGHT-SHIFTING BALANCE TRAINING INTERVENTION

By

Hui-Ting Shih

Bachelor of Science – Physical Therapy
National Taiwan University
2011

Master of Science – Physical Therapy
National Taiwan University
2013

A dissertation submitted in partial fulfillment
of the requirements for the

Doctor of Philosophy – Interdisciplinary Health Sciences

Graduate College

University of Nevada, Las Vegas
August 2022

UNLV | GRADUATE COLLEGE

Dissertation Approval

The Graduate College
The University of Nevada, Las Vegas

June 14, 2022

This dissertation prepared by

Hui-Ting Shih

entitled

Balance Recovery Response in Community-Dwelling Adults with Unilateral Transtibial Amputation and the Potential Benefits of a Weight-Shifting Balance Training Intervention

is approved in partial fulfillment of the requirements for the degree of

Doctor of Philosophy – Interdisciplinary Health Sciences
Graduate College

Szu-Ping Lee, PT, Ph.D.
Examination Committee Chair

Gaby Wulf, Ph.D.
Examination Committee Member

Robert Gregor, Ph.D.
Examination Committee Member

Mark Grabiner, Ph.D.
Examination Committee Member

Lung-Chang Chien, Dr.PH
Graduate College Faculty Representative

Alyssa Crittenden, Ph.D.
Vice Provost for Graduate Education &
Dean of the Graduate College

Abstract

Previous studies have shown that individuals with various physical, sensory and neuromuscular impairments are at higher risks of falls. Individuals with unilateral transtibial amputation (UTTA) suffered from all these impairments, and tripping not surprisingly caused a considerable number of falls in this population. To study falls, researchers have to put participants in a well-protected environment and reproduce tripping fall scenarios. Furthermore, the perturbation delivery needs to be precise in terms of temporo-spatial timing. These features would ensure the quality of responses elicited and reproducibility of the results. Thus, in Chapter 2, we developed a treadmill-based perturbation delivery protocol and confirmed that by referencing ground reaction force, the system was able to consistently and precisely deliver perturbations in early stance phase to elicit tripping falls.

Because tripping usually arrests only one side of the limb, individuals with UTTA may respond differently when encountering trips with their prosthetic versus non-prosthetic limb. Understanding the biomechanical differences in fall recovery response between these two tripping conditions will facilitate ideas for patient-specific intervention targeting tripping fall prevention. Therefore, in Chapter 3, we utilized the protocol developed in Chapter 2 to deliver destabilizing perturbations to the participants in order to examine the limb-to-limb differences during fall recovery. We found that while the gross fall recovery strategies (i.e. the stepping response) were similar, there existed key biomechanical differences. Perturbation during a static standing condition was typically arrested with the perturbed limb making the recovery step. Dynamic perturbation condition was recovered with the contralateral (non-perturbed) limb making the first

recovery step followed by the ipsilateral limb making the reciprocal second recovery step. We observed that certain defined response times were longer when the recover step was executed by the prosthetic limb in both static and dynamic perturbation conditions, suggesting the impaired sensory detection or motor execution of the prosthetic limb.

Currently, clinical practitioners are encouraged to include balance training in post amputation rehabilitation. A balance training that focuses on weight-shifting may prepare individuals with lower limb loss the essential ability to make successful recovery step when encountering destabilizing scenarios. However, it is currently unknown if a training program focusing solely on balance control can improve fall recovery response. Hence, in Chapter 4, we examined the effects of a 2-day weight-shifting balance training using protocols developed in Chapters 2 and 3. We found that certain biomechanical variables relevant to weight-shifting and weight-bearing during fall recovery were altered by the training. For instance, the duration for unloading the prosthetic limb before taking the recovery step during static perturbation condition were improved after training. Another example was that when the prosthetic limb was perturbed, the duration of the <u>first</u> recovery step increased; meanwhile, when the non-prosthetic limb was perturbed, the duration of the <u>second</u> recovery step increased. These two durations were the non-prosthetic limb executing the recovery step in which the prosthetic limb providing the stance support, and the stance time increased.

Overall, our findings suggest that sensorimotor deficits related to UTTA may lead to longer duration of step time when the prosthetic limb executed the recovery step. This is a promising direction to intervene in the future. Our balance training protocol appears to improve components that were related to participants' weight-shifting ability. Whilst for altering the global fall recovery responses, we concluded that a more perturbation-based approach may be required and should be investigated in the future.

Acknowledgements

I would like to first and wholeheartedly acknowledge my advisor and mentor, Dr. Szu-Ping Lee. He mentored and guided me throughout the past 6 years. The rigor, passion, intelligence, diligence he showed to research and life makes him my role model professionally and personally. In addition, as an international student, I would not be able to navigate the environment and culture here without him. Knowing that he believed in me and in the meantime was always willing to support me empowered me throughout the doctoral program. He is the best advisor and mentor I could ask for. I would also like to express my sincerest appreciation to dissertation committee. Dr. Lung-Chang Chien generously and patiently spent countless time on meeting with me and replying my emails with random statistical questions even at midnights. My memory about statistics now binds with the sunshine in Dr. Chien's office. It is one of the most fortunate things happened in my doctoral journey that UNLV has Dr. Robert Gregor as our research consultant, so do I have Dr. Gregor in my committee. He is the most humble, gentle, and wise man I have ever met who applied those spectacular characteristics to research and to the discussion we have. Dr. Gregor never stopped listening to me and, more importantly, inspiring me with his biomechanical expertise and experience. He re-defines the word "mentorship" in my dictionary. My dissertation is about falls, so the fact that Dr. Mark Grabiner graciously agreed to join the committee was exceptionally encouraging to me. Since Dr. Grabiner has done extensive work in the area of postural perturbation and fall recovery research, he facilitated a lot of in-depth and focused discussion that pointing out my blind spots which was extremely critical for the periods from developing the research ideas to analyzing and interpreting the results. Dr. Gabriele Wulf played a key role in my doctoral life. Motor learning has long been my research interest and Dr. Wulf supported the projects, including but not limited to my dissertation project, unconditionally. She guided me to incorporate two different perspectives, biomechanics and motor learning together, granting me a flashlight to light up a road that was easily unnoticed by biomechanists.

I would like to shout out to my precious colleagues, Lee-Kuen Chua, Takehiro Iwatsuki, Devin Kelly, Haehyun Lee, Harjiv Singh, my lab mates, Jason Ciccotelli, Lauren Dicky, Michelle Samuel. Because of their company, my life in UNLV was not only fruitful but also enjoyable. My sincerest gratitude goes to Yo Shih and Kit Leong, the loveliest couple in the world, Yen-Ling Chen, my angelic roommate for the past 5 years, Kuan-Che Lan, Chao-Chin Yang, Ming-Tai Wu, and Selene Tung. Your genuine friendship is an unexpected but priceless gift. Words cannot express how grateful I am for the technical assistance that Dr. Szu-ting Yi provided. Without her expertise in programming, I would conduct none of the study in this dissertation. Dr. Ya-Ting Hsu deserves all the positive adjectives I can think of for describing a person. She is considerate, courageous, and enthusiastic, taking care of me in every way just like my older sister in the US. Last but not least, I would like to send my special thanks to my family and friends in Taiwan. You all teach me how powerful a virtual hug could be; how much joy a post card can carry; and how impossible I survive the journey (attacked by COVID) without you all. Thank you for loving me as I am.

Table of Contents

List of Tables

List of Figures

Chapter 1: Introduction

In the United States, the population living with limb loss was estimated to be 1.6 million in 2005. Despite the progress of medical technology and medicine, this population is expected to more than double to 3.6 million by 2050 (Ziegler-Graham, MacKenzie, Ephraim, Travison, & Brookmeyer, 2008). It was estimated that there are about 300 to 500 amputations taking place each day in the United States (T. P. Sheehan & Gondo, 2014). Non-congenital lower limb loss may be due to trauma, cancer (i.e. tumor removal), or more commonly progression of chronic diseases (i.e. diabetes mellitus and peripheral vascular diseases) (Murdoch & Bennett-Wilson, 1996). Typically, amputations resulting from the former two causes are grouped as non-vascular amputation whereas the latter causes are termed vascular amputation. The number of non-vascular amputation has been declining so has the probability of amputations due to diabetes (Varma, Stineman, & Dillingham, 2014). However, the ageing society largens the total diabetic population substantially and prevails individuals with multiple comorbidities, therefore the total number of amputations remains high and increasing. As a result, researchers have reported that most amputations now occur in the elderly, concurrent with the recent observations that the mean ages of community-dwelling amputees range from 54 to 72.(Steinberg, Gottlieb, Siev-Ner, & Plotnik, 2019; Ziegler-Graham et al., 2008)

Falls occur frequently after lower limb amputation. Individuals with lower limb amputation fall more often than age-matched non-amputees.(Miller, Deathe, Speechley, & Koval, 2001) It was estimated that 16.5% of the unilaterally amputated patients in the surgical ward and 20.5% in inpatient rehabilitation units have fallen at least once during their hospitalization.(Pauley, Devlin, & Heslin, 2006; Yu, Lam, Nettel-Aguirre, Donald, & Dukelow, 2010) After returning to the community, the incidence of falls range from 50% to up to 80%.(Furtado et al., 2017; Christiane Gauthier-Gagnon, Grisé, & Potvin, 1999; Ülger, Topuz, Bayramlar, Erbahçeci, & Sener, 2010; Wong, Chen, Blackwell, & Rahal, 2015) There are some common risk factors of falls across different stages of recovery, for example, the level of amputation and numbers of comorbidities. There were also some community-exclusive risk factors such as increased gait variability, poorer balance, stump problems, back pain, and joint pain.(B. G. Hordacre, Barr, Patritti, & Crotty, 2015; Miller, Speechley, & Deathe, 2001; Parker, Hanada, & Adderson, 2013) Most of the community-dwelling participants were fitted with prostheses and had used the prostheses for indoor and outdoor mobility. If the developed level of mobility is lower, the more likely they might fall. Along with their movement performance becoming stable, some long-term problems evolved such as stump skin irritation, low back pain and knee pain in the non-prosthetic side. If there are more secondary comorbidities, it is also more likely that they will fall. The consequences of falls can be severe in various aspects. Amputee fallers significantly reduced their social activities and participation.(B. Hordacre, Barr, & Crotty, 2015) For falls that required hospitalization, the mean 6-month direct cost to a faller with lower limb amputation was on average 26,000 U.S. dollars.(Mundell, Maradit Kremers, Visscher, Hoppe, & Kaufman, 2017)

Across lifespan, from young to middle-aged to older adults, the majority of falls occur during walking.(Heijnen & Rietdyk, 2016; Talbot, Musiol, Witham, & Metter, 2005) A study further found that older residents in long-term care facilities fell mostly due to incorrect weight-shifting (41%) and tripping (21%).(Robinovitch et al., 2013) Prosthetic users are also prone to trips.(Ülger et al., 2010) In a prospective study, the authors tracked the falls of 8 below-the-knee amputees for one year (Rosenblatt, Bauer, & Grabiner, 2017), during which five of them reported multiple trips accumulating a total of 44 events. The most common circumstance of trips was walking on level ground; the second most common was walking on uneven surface, followed by walking downhill. The reason that amputees trip more often on level ground may be due to being less cautious and not paying attention to the required toe clearance.(Hunter, Higa, Frengopoulos, Viana, & Payne, 2018) An unanticipated obstacle or toe catch could lead to tripping falls even during level ground walking.

Trip Recovery Response

Definition of a Trip

The definition of a trip or stumble varies in the literature. The general consensus of a trip is the occurrence of an abrupt obstruction of a foot in swing during gait.(Arena, Davis, Grant, & Madigan, 2016; Pijnappels, Kingma, Wezenberg, Reurink, & Van Dieën, 2010; Potocanac, Pijnappels, Verschueren, van Dieën, & Duysens, 2016; Schulz, 2017; Sessoms et al., 2014) Typically only one foot is obstructed. In biomechanical studies examining tripping, researchers also focused on the trunk forward momentum which continues while a foot is obstructed. And it is this sudden, unexpected forward acceleration of body center of mass with respect to the disrupted base of support that contributes to a tripping fall.(B.-C. Lee, Martin, Thrasher, & Layne, 2016; Zhang, D'Andrea, Nunnery, Kay, & Huang, 2011)

Trip Recovery Strategy

Tripping fall recovery requires rapid responses immediately after the onset of the trip (Grabiner & Jahnigen, 1992; Stelmach & Worringham, 1985), and the responses consist of three main conceptual components: the reactive control of the forward rotation of trunk (Grabiner, Feuerbach, & Jahnigen, 1996; Grabiner, Koh, Lundin, & Jahnigen, 1993), the use of a limb in slowing the fall (Dietz, Quintern, Boos, & Berger, 1986; Eng, Winter, & Patla, 1994), and the recovery stepping of sufficient length to regain the base of support.(Fukagawa & Schultz, 1995; Grabiner et al., 1993) Encompassing these components, three different categories of tripping fall recovery strategies have been identified in able-bodied persons. They are described here.

- Elevating Strategy: When performing an elevating strategy, the tripped limb is typically flexed to prepare for a recovery step while the contralateral (non-tripped) limb support and elevate the body center of mass. It is the predominant strategy if the trip occurs during early swing phase of the tripped limb.
- Lowering Strategy: During lowering strategy, the tripped limb is quickly lowered to the ground immediate after the obstruction. The tripped limb

serves as the support limb during the perturbation to enable the recovery step from the contralateral, non-tripped limb. It is usually a response to a trip occurring during late swing phase of the tripped limb.

- Reaching Strategy: If the trip occurs during late swing phase, an individual may alternatively use a reaching strategy, prolonging or maximizing the flexion of the tripped limb, especially the hip, to execute an extra big forward step to go over the obstacle. In this case, the contralateral, non-tripped limb is the support limb.

The onset timing of the tripping event largely determines what strategy will be selected.(Eng et al., 1994) If the onset is during mid-swing, elevating and lowering strategies are both possible.(Schillings, Van Wezel, Mulder, & Duysens, 2000) Furthermore, no matter what strategies are used, the reciprocal stepping pattern remains. Specific to individuals with above-the-knee amputation, two additional fall arrest strategies have been identified. These strategies are unique to individuals with a unilateral amputation as they feature interruption to the reciprocal stepping pattern after the obstruction.(Shirota, Simon, & Kuiken, 2015)

- Hopping Strategy: In response to the trip, a hop is performed with both legs to clear the obstacle.
- Skipping Strategy: In response to the trip, the tripped limb (typically the non-prosthetic limb) is lowered quickly after the obstruction and then the same limb is used to skip and clear the obstacle.

Representative Variables to Quantify Trip Recovery Strategies

Several variables relevant to successful trip recovery have been investigated. In healthy older adults, probable factors were examined and shown that across different mechanisms of falling, the extent of trunk forward orientation is a significant predicting factor of the success/failure of fall recovery. The variables associated to this factor were peak trunk flexion angle and peak trunk flexion velocity.(Pavol, Owings, Foley, & Grabiner, 2001) These two variables have also been used to successfully discerned below-the-knee amputated fallers and non-fallers.(Sessoms et al., 2014) Longer reaction latency from trip onset to recovery limb load-off and shorter step length have also been found in failed recovery trials.(Owings, Pavol, & Grabiner, 2001)

Trip Simulation Methodology

Necessary Characteristics of the Methodology

Given tripping and the resulting falls are critical problems in the amputee population, biomechanical researchers have been looking for a tool that can replicate the trips so that enables us to safely evaluate the fall arrest response. An adequate tripping methodology should be able to attain certain technical characteristics.(King, Eveld, Martínez, Zelik, & Goldfarb, 2019) First, it should allow precision control of the timing of the perturbation event during gait including the onset and perturbation duration. As indicated previously, the gross fall recovery strategy selection is determined by the timing of the tripping onset (early vs. late swing phase).(Eng et al., 1994; Shirota, Simon, & Kuiken, 2014) Thus, controlling the onset timing of the perturbation during gait and keeping the perturbation duration consistent are key to reproduce the fall conditions. For research purposes, we need to repeatedly reproduce the controlled perturbations to the same person or to different persons to make unbiased within- and between-subject comparisons. Second, the tripping perturbations delivered need to be unanticipated so that the response is close to purely reactive to the perturbation as in real life. Anticipation undoubtedly alter the gait because of cautious guarding and preparation for coming perturbations, and the effect could influence the fall recovery response. In this sense, unanticipated perturbation delivery would ensure that the gait pattern is minimally impacted so are the responses. Third, the magnitude of the perturbation should ideally be adjustable to allow observation of a range of responses. Adjustable perturbation magnitude is also useful for intervention purposes. Last but not least, a tripping perturbation protocol should deliver the perturbation unilaterally to take account for the rotations of the trunk and limbs in all three planes after a foot obstruction as in real life.

Existing Protocols

Currently in the literature, three main methodologies to deliver tripping perturbation have been developed, they are: 1) obstacle-based, 2) tether-based, and 3) treadmill deceleration/acceleration-based methods. The former two methods can be applied overground or on a treadmill. The last method relies on changing the treadmill belt movement so is bound to the treadmill. In general, the overground methods have the problem of the walkway being too short so a steady-state gait may not be assured before entering the perturbation.(King et al., 2019) It is also difficult to control gait velocity of different participants during overground walking. In contrast, methods delivered on a treadmill may interfere the natural acceleration-deceleration pattern of walking and fall recovery.(Rossignaud, Oliveira, Lara, Mayor, & Rodacki, 2019)

The obstacle-based method was the first to be implemented and produced the most currently available data on simulated falls. It typically involves utilizing a concealed obstacle as a physical foot obstruction to produce a trip. The obstacle could be placed on the ground for overground walking (Arena et al., 2016; Crenshaw, Kaufman, & Grabiner, 2013b; Eng et al., 1994; Pavol et al., 2001; Pavol, Owings, Foley, & Grabiner, 2002; Pijnappels, Bobbert, & van Dieën, 2004, 2005; Pijnappels et al., 2010; Potocanac et al., 2016; Potocanac, Smulders, Pijnappels, Verschueren, & Duysens, 2015; Roos, McGuigan, Kerwin, & Trewartha, 2008; Roos, McGuigan, & Trewartha, 2010; Schulz, 2017; van den Bogert, Pavol, & Grabiner, 2002; Van der Burg, Pijnappels, & van Dieen, 2007) or dropped onto a treadmill (Haridas, Zehr, & Misiaszek, 2008; Schillings, Mulder, & Duysens, 2005; Schillings, Van Wezel, & Duysens, 1996; Schillings, Van Wezel, Mulder, & Duysens, 1999; Schillings et al., 2000). There are two main concerns regarding this method: onset timing and anticipation. It was hard to control at what gait phase the participants' foot would contact the obstacle. As perturbation onset timing is related to the fall recover strategy, if the perturbation timing cannot be controlled, the fall recover responses would be erratic. Moreover, variables exerted by different fall recovery strategies would be incomparable. Thus, successful trials must be further grouped by the fall recovery strategies before analysis. In other words, fall responses from any given individual may not be reproducible. The concern of anticipation when using this method was due to that the participants might be able to see the obstacle out of the corner of their eyes, hear the sound, or perceive the vibration made by the obstacle. Some studies provided devices such goggles and headphones to block participant's vision and hearing. However, these devices also interfere the normal visual and auditory flow while walking, potentially alter the fall recovery response.

The tether-based method could also be applied either overground (Aziz, Park, Mori, & Robinovitch, 2014; Blumentritt, Schmalz, & Jarasch, 2009) or on a treadmill (Forner-Cordero, Ackermann, & de Lima Freitas, 2011; Forner-Cordero, van der Helm, Koopman, & Duysens, 2015; J. K. Lee, Robinovitch, & Park, 2014; Shirota et al., 2014, 2015). The tether is typically attached to a participant's leg around the ankle or the foot. When the tether is tightened, it obstructs normal leg motion and simulates the trip. To what extent the presence of the tether interferes normal gait, and to what extent the tensioned tether disrupts ordinary leg movement associated with trip recovery remain unclear. Additionally, the response elicited by the tether-based method does not agree with that from obstacle-based methods.(Schillings et al., 2000; Shirota et al., 2014) Specifically, greater hip and knee flexion during lowering strategy was found using tether-based method. Increased flexion of the hip and knee followed by greater extension during elevating and lowering strategies was also observed with the tether-based method.

The treadmill deceleration/acceleration methods emerged within the last 10 years.(Crenshaw, Kaufman, & Grabiner, 2013a; B.-C. Lee et al., 2016; Sessoms et al., 2014; Zhang et al., 2011) Owings et al. introduced the concept of using a motorized treadmill to generate perturbation, whereas the protocol starts with the subject standing on a fully stopped treadmill and the sudden start of the treadmill belt served as the perturbation.(Owings et al., 2001) Zhang et al. modified the protocol to deliver the perturbation during walking.(Zhang et al., 2011) They perturbed individuals with above-the-knee amputation using two types of belt movements. The treadmill belt moved at the walking speed of the subject and suddenly accelerated then decelerated back to the walking speed or vice versa (i.e. decelerated and then accelerated). (**Figure 1**)

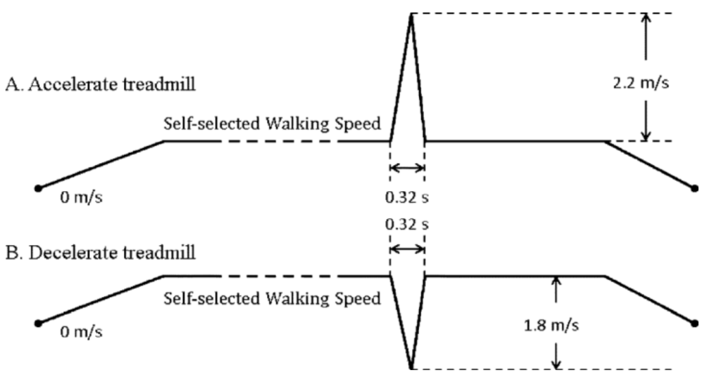


Figure 1. Belt Velocity Profiles of the Protocols by Lee et al.

Based on electromyography (EMG) activities, researchers claimed that this protocol was able to elicit responses comparable to what is occurring in daily life.(Berger, Dietz, & Quintern, 1984) Moreover, the method can minimize anticipatory reactions because no additional physical object was used and the responses can be examined in a reproducible manner. While this group of researchers succeeded in eliciting recovery responses, however, the method was criticized because the magnitude of the perturbation was not strong enough compared to the perturbations encountered in real life involving a foot-stopping obstacle. Sessoms et al. improved the method by decelerating the treadmill belt first to simulate the foot obstruction followed by belt acceleration to simulate the forward trunk motion in quick succession.(Sessoms et al., 2014) (**Figure 2**)

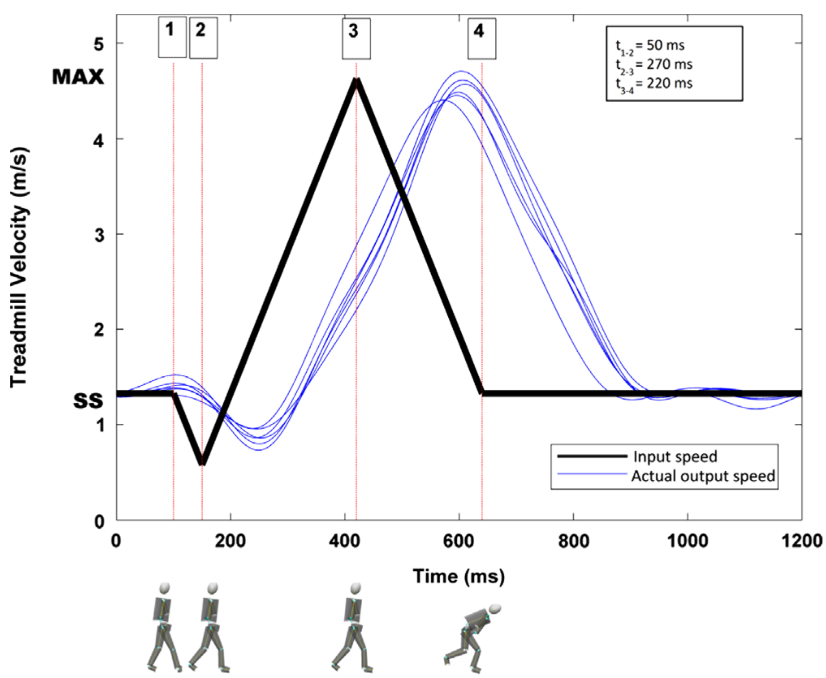


Figure 2. Belt Velocity Profile of the Protocol by Sessoms et al.

Ground reaction force and foot position were referred for delivering the perturbation so the onset timing can be precisely controlled. Nevertheless, the limitation is that the method typically perturbs both feet simultaneously using a single-belt treadmill in most currently published studies. Furthermore, this method applies the perturbation to the stance limb rather than the swing limb, rendering it less realistic.

Differential Responses to Prosthetic Limb vs. Non-Prosthetic Limb Trips

Due to the altered musculature, sensorimotor integration, cortical adaptation, and the prosthetic componentry, it is likely that people with lower limb amputation respond to trips differently when the prosthetic limb and the non-prosthetic limb are obstructed. Understanding the biomechanical mechanism will provide insight on future directions of fall prevention training and prosthetic design. If the fall recovery responses from the prosthetic limb, non-prosthetic limb, and the non-amputee limb are different, researchers may need to develop specialized rehabilitation program that targets the different trip recovery strategies.

Studies have examined the trip recovery response in people with above-the-knee amputation and knee disarticulation amputation.(Bellmann, Schmalz, & Blumentritt, 2010; Crenshaw et al., 2013b; Shirota et al., 2015) Crenshaw et al. found that when the prosthetic limb was tripped, all participants exhibited a lowering strategy whilst when the non-prosthetic limb was tripped, lowering, elevating, and hopping strategies were observed.(Crenshaw et al., 2013b) Similar findings were reported by Shirota et al. but the group recognized one more strategy, the skipping strategy, being used when the non-prosthetic limb was obstructed.(Shirota et al., 2015) As mentioned earlier, individuals with lower limb amputation develop unique fall recovery strategies, for example, hopping and skipping strategies, that non-amputees do not typically perform. Moreover, these two strategies were both observed when the non-prosthetic limb was obstructed and thus suggested that the pattern of the strategies used by the non-prosthetic limb is not only different from the prosthetic limb but also different from an intact limb of a non-amputee.(Shirota et al., 2015) Shirota et al. also found that regardless of the side of obstruction, even the non-prosthetic limb, people with transfemoral amputation used delayed-lowering strategy less frequently than those without amputation. Comparing the responses from when the prosthetic, non-prosthetic, and non-amputee limbs were perturbed, they observed that it takes longer to complete the fall arrest when the prosthetic limb was perturbed. Some differences of joint angles were mentioned but it was inconsistent across strategies and across limbs. The rate of fall was either similar or not examined between the sides of obstruction.(Crenshaw et al., 2013b; Shirota et al., 2015)

Additionally, literature suggested that prosthetic knee joint might be the key to effective recovery in above-the-knee amputees because different prosthetic knee joints allow various functionality and capability of trip recovery and fall prevention. In a study of 19 participants with above-the-knee amputation, Kahle et al. showed that the use of a microprocesser knee can reduce the average number of trips to 3 and the number of falls to 1 in two months, in contrast to using a non-microprocessor knee which resulted in on average 7 trips and 3 falls during the same time period.(Kahle, Highsmith, & Hubbard, 2008) Bellmann et al. simulated the fall recovery response by having the participants stumble on their prosthetic limb. They found that the falls were primarily due to the rapid, excessive knee flexion. In other words, the uncontrolled knee buckling on the prosthetic leg is the main reason leading to the eventual fall. Even though participants attempted to arrest the fall by making compensatory steps and upper extremity movement, the knee joint still collapsed, resulting in unavoidable falls.(Bellmann et al., 2010)

Individuals with below-the-knee amputation preserve, at least in part, the knee control on the amputated side. With the knowledge that the knee is important in trip recovery,(Bellmann et al., 2010; Crenshaw et al., 2013b) it is possible that people with below-the-knee amputation manifest differently from their above-the-knee counterparts. Few studies have investigated the differential trip recovery responses following the prosthetic and non-prosthetic tripping in people with below-the-knee amputation. In a training study, Kaufman et al. applied trip simulation to 14 male military service members and reported their trunk control manifestation. Regardless of training, the participants showed greater peak trunk flexion angle and higher peak trunk flexion velocity when the prosthetic limb was tripped in contrast to non-prosthetic limb trips.(Kaufman, Wyatt, Sessoms, & Grabiner, 2014) The results indicated better trunk control and lower likelihood of fall when the non-prosthetic limb was obstructed. Another study of active-duty members of the US military showed opposite results that perturbation of the non-prosthetic limb led to fewer numbers of fall but greater peak trunk flexion angle and velocity.(Sessoms et al., 2014)

Studies of fall recovery strategy in individuals with below-the-knee amputation are currently limited and demonstrate inconsistent results. In addition, participants from the two above-mentioned studies were relatively young (mean age 26 and 24.3 years, respectively), active (military service members), and with trauma-caused amputation. These characteristics are very different from community-dwelling amputees who are typically older, less active, and have more comorbidities. It is likely that the community-dwelling individuals with below-the-knee amputation will perform differently when encountering a trip. The fall recovery strategies are also likely different between trips to the prosthetic vs. non-prosthetic limb in this population. We recognize that knowledge regarding the differential fall responses will be useful in developing prosthetic componentry and rehabilitation strategies, the current evidence is inconclusive. This warranted the current research.

Motor Adaptations after Transtibial Amputation

Structural changes post lower limb amputation may further contribute to consequent physical and psychological challenges. Asymmetric weight-bearing and postural reorganization have been extensively reported in individuals with limb loss.(Rougier & Bergeau, 2009; Vanicek, Strike, McNaughton, & Polman, 2009b) It is well-understood that persons with lower limb amputation utilize more asymmetric strategies to perform basic daily activities and to retain balance when they are perturbed. In this section, the asymmetric weight-bearing and asymmetric movement patterns in individuals with lower limb amputation will be reviewed, from basic activities, gait, to unperturbed and perturbed postural control situations.

Asymmetric Movement Pattern during Basic Activities

The prosthetic and non-prosthetic limbs contribute to posture during standing unequally. In quiet standing, people with amputation tend to shift their whole-body center of pressure to the non-prosthetic side.(Rougier & Bergeau, 2009) Rougier and Bergeau additionally reported forward shift of the center of pressure under the prosthetic limb in below-the-knee amputees. During standing, Hlavackova et al. also observed that the non-prosthetic limb had a preferentially larger body weight distribution.(Hlavackova, Franco, Diot, & Vuillerme, 2011) Less variable center of pressure trajectory, greater center of pressure trajectory amplitude and velocity under the non-prosthetic side indicated larger contribution to quiet standing. Taken together, the evidence of over-reliance on the non-prosthetic limb during standing was clear.

Sit-to-stand is a basic movement that a person does every day. It was observed that individuals with below-the-knee amputation place the prosthetic limb more forward (away from the chair) than the non-prosthetic limb when executing a sit-to-stand.(Šlajpah, Kamnik, Burger, Bajd, & Munih, 2013) Thus, comparing the lower extremity joint angle between sides, amputees showed greater bilateral joint angle asymmetry than non-amputees. The forward foot placement allows the prosthetic limb to bear less weight and was confirmed by measuring the force exerted by each limb.(Özyürek, Demirbüken, & Angın, 2014) Clinical populations with hemiparesis use the same strategy to unload their affected side.(Roy et al., 2006) Moreover, Šlajpah et al. assessed the sit-to-stand performance at different chair heights.(Šlajpah et al., 2013) They found that the hip, knee and ankle of the prosthetic limb consistently stayed more extended than the non-amputated limb throughout the whole time regardless of the chair height. When standing from a taller chair, there was less joint work required and less joint range of motion excursion needed but the participants still chose to place the prosthetic limb more forward. It indicated that the manifestation was more of a choice rather than lack of capability. Further, amputees inclined the trunk forward more than non-amputees and exhibited greater center of gravity sway velocity.(Özyürek et al., 2014) The pattern suggested that they rely on the forward momentum created by the trunk to stand up from a seated position so that lessen the demand of knee extensors.

Asymmetric movement pattern has also been found during prosthetic gait. For example, below-the-knee amputees walk slower with higher step frequency and wider step width than non-amputees.(Hak et al., 2013) In terms of work, generally, the residual knee of the prosthetic limb contributed less when compared to the knee on the non-prosthetic side in both the eccentric and concentric phases.(Prinsen, Nederhand, & Rietman, 2011) On the other hand, the hip on the prosthetic side contributed more work than the non-prosthetic side in both concentric and eccentric phases. A previous study showed asymmetric patterns in joint work and energy recovery in prosthetic gait.(Tesio, Lanzi, & Detrembleur, 1998) In the cases of below-the-knee amputations, in comparisons to the non-prosthetic limb, the steps made by the prosthetic limb showed 21% lower combined work from the hip, knee, and ankle joints in the lower extremity than the non-amputated limb. Side to side comparison further showed an interesting finding that energy recovery, i.e. the percent of energy conversion between gravitational potential energy and kinetic energy, was 8% greater in the prosthetic limb. The authors speculated that amputees tend to unload the prosthetic limb which leads to an increased reliance on the pendulum-like gait pattern over the prosthetic steps which further results in lower joint work and greater energy recovery. However, this energy efficiency on the prosthetic side during gait likely comes at a cost of the limb being stiffer and less accommodating. Thus, when reacting to perturbations, individuals with below-the-knee amputation tend to rely heavily on the non-prosthetic limb to respond and being more susceptible to fall in certain situations.(R. C. Sheehan, Beltran, Dingwell, & Wilken, 2015) In agreement with this theory, previous literature also showed that below-the-knee amputees reduced the demand of their knee extensors by decreasing knee flexion during loading response; smaller knee moment was also observed during early stance.(Sanderson & Martin, 1997; Winter & Sienko, 1988)

Response to Postural Perturbation

Interlimb asymmetry was also exhibited when responding to external perturbation. When confronting multi-directional support surface translation, middle-aged below-the-knee amputees were able to sustain whole-body center of mass displacement comparable to age-matched non-amputees.(Bolger, Ting, & Sawers, 2014) However, the authors further analyzed the center of pressure margin of stability for each leg. A smaller margin often translates to higher risk to lose balance. They found that the margin of stability for the prosthetic limb was smaller during anterior-posterior perturbation than the non-prosthetic limb. The lack of ankle musculature and the associated reduction in sagittal plane degrees of freedom control likely attribute to this. Curtze et al. found similar results regarding the direction-specific compensatory pattern.(Curtze, Hof, Postema, &

Otten, 2012) Below-the-knee amputees appear to cope with anterior-posterior perturbation by increasing the utilization of their non-prosthetic limb, specifically with greater ankle moment. Despite the great amount of mentioned compensation in anterior-posterior perturbation, they resisted medio-lateral perturbation in a way that is very close to non-amputees, mainly by recruiting both hips.

Further, unilateral amputation results in altered sensory and neuromuscular control in detecting and responding to perturbations. The alteration exists both in distal and proximal, i.e. lower extremities and trunk, measured by electromyography (EMG) activities. Rusaw et al. asked people with below-the-knee amputation to place their prosthetic limb or non-prosthetic limb on a tilting force platform.(Rusaw, Hagberg, Nolan, & Ramstrand, 2013) The platform was positioned to elicit either toe-up or toe-down perturbations. For each direction of perturbation, the participants were further asked to put 25%, 50%, or 75% of their body weight to the foot on the platform. The study had two main findings. First, the dampened sensory input on the prosthetic limb may have detrimental sensorimotor effects in detecting and responding to a perturbation. This was indicated by that when only placing the prosthetic limb on the platform to receive the perturbation, the non-prosthetic limb showed significant delayed EMG response. Second, asymmetric weight-bearing pattern observed in amputees may translate to delayed EMG response patterns when encountering postural perturbations. In general, participants displayed delayed EMG response to the perturbation in both the prosthetic and non-prosthetic limbs comparing to non-amputees. Different levels of weight-bearing had an effect on the response latency of the non-prosthetic limb. The condition of applying 75% weight on the non-prosthetic limb had significantly shorter EMG latency than the 25% and 50% conditions. Furthermore, not only the lower extremity but also the trunk neuromuscular control was altered by unilateral amputation. Hendershot et al. applied antero-posterior and medio-lateral perturbation to the trunk with pelvis immobilized and measured the biomechanical and EMG response in erector spinae and external oblique.(Hendershot, Bazrgari, & Nussbaum, 2013) Persons with amputation showed 22-24% lower trunk stiffness, 23-27% smaller maximum reflex EMG amplitude, and 8% greater latency to maximum reflex EMG than people without amputation. Comparing between sides in persons with amputation, in lateral perturbation that pushed participants to bend to their prosthetic side, the trunk was 20% less stiff, and the latency to maximum reflex force of the contralateral trunk muscles was 9% greater. These reduced and asymmetric trunk behaviors may be owing to the altered musculature and sensory input, therefore altered sensorimotor integration and altered muscle recruitment. The repeated exposure of asymmetric movement patterns may contribute to these learned trunk motor behaviors, too.

When comparing fallers and non-fallers with below-the-knee amputation, Vanicek et al. showed that the fallers exhibit greater vertical ground reaction force during initial contact (the first peak vertical ground reaction force, vGRF) and loading rate on their prosthetic limb.(Vanicek, Strike, McNaughton, & Polman, 2009a) This is indicative of a stiffer and uncontrolled gait pattern. The non-fallers exhibited greater knee extensor moment in contrast to the fallers. Specifically, the knee extensor moment on their prosthetic limb was significantly greater during mid stance than fallers, which indicated that the non-fallers actively recruited the muscles around the residual knee joint to maintain the stability of the prosthetic limb. This observed pattern may enable the non-fallers to be more adaptive when dealing with uneven terrain and responding to perturbation than the stiff limb pattern observed in fallers.

Sensorimotor Adaptations after Amputation and Implications to Fall Recovery Training

It is important to recognize that the musculoskeletal system is not the only system affected by amputation. Lower limb amputation also alters the sensorimotor function, resulting in neuromuscular adaptations that alter how people with lower limb amputation move and learn.

As mentioned earlier, amputation removes sensory organs and alters sensory input from below the amputation level. Amputated individuals need to reprogram the sensorimotor integration in order to perform locomotor tasks. Early studies found that to compensate for the loss of sensory input, below-the-knee prosthetic users showed greater visual dependency after amputation but such dependency can be decreased with rehabilitation.(C Gauthier-Gagnon, 1986) Geurts and colleagues found reduced visual dependency during quiet standing after a post-amputation rehabilitation program based on the significantly reduced postural sway in eye-closed condition.(Geurts, Mulder, Nienhuis, & Rijken, 1992) Beurskens et al. further tested experienced below-the-knee prosthetic users in a Computer Assisted Rehabilitation Environment (CAREN) which could create a perturbed visual optic flow while participants walking.(Beurskens, Wilken, &

Dingwell, 2014) (**Figure 3**) They reported that the participants showed greater trunk movement variability, greater step width variability during visually perturbed walking, but the extend of change was not significantly different from non-amputees. In this study, these young military service members have undergone comprehensive amputation care and did not show greater reliance on vision than non-amputees to sustain walking balance.

Figure 3. CAREN System

Researchers hypothesized that other sensory adaptation occurs to replace visual dependency. In line with the hypothesis, evidence showed increased or altered reliance on proprioception. Entropic Half-Life (EnHL) is a tool that can be used to analyze the center of pressure excursion and to quantify how much the previous postural position is used for current postural adjustment. Also known as the memory effect of postural control, it generalizes multiple postural control systems and feedback loops.(C. Pasluosta et al., 2018; C. F. Pasluosta et al., 2017) The sample entropy of center of pressure excursion represents the temporal regularity of the signal, with higher value indicating more irregular.(Richman & Moorman, 2000) EnHL is the time scale when the sample entropy equals 0.5. The shorter the EnHL, the less time it takes to transit from a regular to a more irregular pattern. A previous study has found that the EnHL during eyes-open foam standing was greater than eyes-open rigid surface standing and eyes-closed rigid surface standing.(Baltich, von Tscharner, Zandiyeh, & Nigg, 2014) It suggested greater EnHL value in vision-dependent postural control condition and smaller EnHL value in proprioception-reliant condition. Claret et al. exploited EnHL to differentiate the feedback control loops being utilized during eye-open and eye-closed unperturbed standing in unilateral above-the-knee amputees.(Claret et al., 2019) In the eye-open condition, comparing to non-amputees, the whole-body center of pressure EnHL value from amputees was not significantly different. However, analyzing the limbs separately, the EnHL value of the non-prosthetic limb was lower than the non-amputee limb and even lower than the prosthetic limb. The authors interpreted this finding in support of a greater reliance of proprioception for non-prosthetic limb control. Moving to eyes-closed condition, greater reliance on proprioception (i.e. lower EnHL value) was also found in the non-prosthetic limb than during the eye-open condition. However, the prosthetic limb remained at the same EnHL levels between the eyes-open and eyes-closed conditions, indicating no change of control strategy. Taken together, the whole-body balance control shifted to more proprioception dependent, suggesting that in the eyes-closed condition, the non-prosthetic limb dominated the postural control. The findings echo the motor adaptations mentioned above, that is individuals with amputation predominantly rely on the non-prosthetic side to maintain standing postural control. Moreover, they preferentially utilize the proprioceptive feedback from the non-prosthetic limb to achieve such control.

In addition to the loss of peripheral afferent signal, the motor control and motor learning alteration may originate at cortical level as well. It has been found that after amputation, the primary (M1) and secondary (M2) motor areas as well as primary (S1) and secondary (S2) somatosensory areas exhibited decreased inter-hemispheric task-state functional connectivity and increased intra-hemispheric functional connectivity. This is particularly prominent in the contralateral hemisphere to the amputation side.(Bramati et al., 2019) The findings indicate the amputees may reduce inter-limb cortical functions, suggesting a more disconnected or uncoordinated control of the lower extremities.

Decreased cortical inhibition exist in people with unilateral amputation and may have motor implications. Chen et al. showed that in individuals with unilateral lower limb amputation, the maximum output of transcranial magnetic stimulation (TMS) recruited a higher percentage of motor neuron pool on the prosthetic side than the non-prosthetic side but meanwhile similar amount of recruitment by spinal stimulation.(Chen, Corwell, Yaseen, Hallett, & Cohen, 1998) Furthermore, the threshold of the muscle activation of the rectus femoris on the prosthetic side was lower than that on the non-prosthetic side. These two findings indicated that, first, neuroplasticity after amputation occurred predominantly at the cortical level instead of spinal level. Second, the contralateral M1 (the hemisphere that controls the prosthetic side) became more excitatory. Short-latency intracortical inhibition (SICI) is a variable measured by TMS that quantifies the extent of intracortical inhibition. The mechanisms underlying SICI is GABAergic inhibition. GABA modulation greatly contributes to the motor cortical plasticity.(Bütefisch et al., 2000; Ziemann, Muellbacher, Hallett, & Cohen, 2001) Moreover, it plays an important role in motor learning.(Floyer-Lea, Wylezinska, Kincses, & Matthews, 2006; Tegenthoff, Witscher, Schwenkreis, & Liepert, 1999) Decreased GABA increases the cortical excitability and may facilitate motor learning. Stagg and colleagues primed the concentration of GABA and found that decreased GABA correlated to higher degrees of motor learning.(Stagg, Bachtiar, & Johansen-Berg, 2011) Localized functional magnetic resonance image double confirmed the response to GABA in M1 during motor learning. Reduced SICI has been found in amputees' bilateral hemispheres. Hordacre et al. conducted a longitudinal study in which they tracked the SICI changes in participants with unilateral amputation from admission to prosthesis fitting (median 32, range 14-41 days), first walk (median 50, range 32-75 days), until discharge (median 87, range 47-99 days).(B. Hordacre, Bradnam, Barr, Patritti, & Crotty, 2015) Reduced inhibition in the contralateral M1 was correlated with less gait variability, which was considered as an improvement of gait function. However, reduced SICI in the ipsilateral M1 was correlated with excessive gait variability.(B. Hordacre, Bradnam, Barr, Patritti, & Crotty, 2014) Reduced SICI is a form of GABAergic increased excitability. Since acquiring skills of how to use a prosthesis is also a type of motor learning, exposure to such demand and ongoing motor learning possibly prompts individuals with amputation to modulate the brain to a generally more excitatory status which benefits acquisition of prosthetic skills.

Structural changes in individuals with lower limb amputation at the cortical level have been located in several areas.(Molina-Rueda et al., 2019) The thickness of the grey matter decreases in premotor cortex, visual cortex, and somatosensory cortex. Regarding the white matter, the integrity of corona radiata decreased, so did the connection between premotor cortices and corpus callosum level. Neuroanatomically, these areas are associated with sensorimotor integration and motor planning. However, the motor indications of the cortical structural changes after limb amputation have not yet been directly proven.

Balance and Fall Recovery Training in Individuals with Lower Limb Amputation

Prosthetic rehabilitation has long aimed to assist prosthetic users to regain functional mobility, balance, postural control, and to prevent falls.(Geertzen, Martina, & Rietman, 2001) Weight-bearing activities and walking are the most applied interventions and evaluated outcomes in post-amputation rehabilitation. Rau and Bonvin employed lower limb strengthening, weight-bearing exercises, gait training, coordination exercises, obstacle management, and functional training to prosthetic users.(Rau, Bonvin, & De Bie, 2007) They found significant improvement in 2-minute walk test, maximum weight borne on the prosthetic limb, and lowered physiological cost for walking determined by the change of heart rate.

A study by Darter et al. assigned a 8-week home-based treadmill training to 8 above-the-knee amputees.(Darter, Nielsen, Yack, & Janz, 2013) Participants increased their self-selected walking speed, maximum walking speed, and 2-minute walk distance after training. In addition, the temporal-spatial gait symmetry improved at self-selected walking speed. Energy consumption decreased by 10% across all walking speeds. Some previous studies employed weight-bearing exercise and gait training but did not perform related outcome measurements, making it hard to conclude the effects of their training on mobility function.(Hershkovitz, Dudkiewicz, & Brill, 2013; Munin et al., 2001)

Balance and postural control are key rehabilitation concerns for this population. Andrysek et al. recruited children and adolescents with lower limb amputation, providing them video games that included balance and postural control components.(Andrysek et al., 2012) After 4 weeks of training, the center of pressure displacement during quiet standing decreased concurrently with improved functional balance measured by the Community Balance and Mobility Scale. Sethy et al. compared the effects of conventional balance training alone to balance training plus Phyaction exercise.(Sethy, Kujur, & Sau, 2009) The Phyaction exercise requires the participant to balance themselves standing on a pre-programmed moving platform. Participants who received augmented Phyaction exercise improved their reaching performance while the conventional group did not. Although the weight-bearing asymmetry remained after Phyaction exercise, comparing to baseline the Phyaction exercise group increased the weight distribution on the prosthetic side. Another study exploited BalanceReTrainer, a program that requires the standing person moving their center of pressure to 8 different targets around them in the transverse plane.(Matjačić & Burger, 2003) The device can provide external standing support from 0% to 100% depending on the person's ability. After training, the 14 individuals with below-the-knee amputation exhibited increased duration of single leg standing on the prosthetic side, decreased Timed-Up-and-Go time, and faster 10-meter walk.

Some post-amputation rehabilitation paradigms apply balance and fall prevention training during walking. A case report by Sheehan et al. delivered a perturbation-based gait training in a virtual environment.(R. C. Sheehan, Rábago, Rylander, Dingwell, & Wilken, 2016) The system featured a continuously changing walking surface angle that simulated the complexity of various outdoor surface terrains such as up/down slope and cross slope. A 43-year-old, above-the-knee amputated veteran improved his self-selected walking speed, reduced the time to complete a four-square step test, and showed decreased step width, decreased step width variability after undergoing this training. Additionally, the improvement persisted 5 weeks after the conclusion of the training. Specifically to prevent trip-related falls, Crenshaw et al. developed a 6-day compensatory-step training protocol and tested it on 5 above-the-knee or knee disarticulated participants.(Crenshaw et al., 2013a) The participants initially stood on a treadmill and then needed to respond to unanticipated posterior accelerations of the treadmill belts by making a few forward steps until the treadmill stopped. With practice, participants performed more successful recovery trials in terms of fewer falls and less hopping strategy used by the non-prosthetic limb. In the last day of training, comparing to the first day, trunk flexion angles at the first step significantly decreased when the recovery step was made by the prosthetic limb. Moreover, the distance between the center of mass and the first stepping limb increased. Taken together, these two kinematic variables marked improvements in recovery response to postural disturbance. A task-specific fall prevention training has also been studied.(Kaufman et al., 2014) Three types of perturbation were delivered: static perturbation, static walking, and e-trip. Static perturbation required participants started from standing still on a treadmill. The treadmill belt moved anteriorly or posteriorly, delivering a discrete perturbation that necessitated participants to take at least one step to re-balance themselves. Static walking started from standing on a treadmill as well. The treadmill belt unanticipatedly began to move, persisted at a target speed for a few seconds and stopped. After the perturbation, participant has to continue walking for a few more steps. E-trip used a treadmill deceleration/acceleration simulation method to deliver tripping perturbation during walking. Participants were all young military service members with below-the-knee amputation. After training, they were tested for response to simulated trips in a virtual environment, CAREN (Figure 3). The testing surroundings were different from the training context so that the performance observed was not merely adaptation to the training context but also hinted skill transfer. Reduced peak trunk flexion angle and trunk flexion velocity were found after training, indicating improved trip recovery response. This is one of the few studies showing the possibility of reducing fall risk and improving fall recovery response via a clinical training protocol.

As reviewed, asymmetric weight-bearing and asymmetric movement patterns typically exist in persons with lower limb amputation. They tend to overly rely on the non-prosthetic side to perform daily activities, to ambulate, and to respond to postural perturbation. Meanwhile, they tend to avoid shifting their weight to the prosthetic side. In tasks that they cannot avoid bearing weight on the prosthetic limb, the response is slower. In trip recovery strategies, those amputee-specific alterations disrupt reciprocal stepping patterns by relying on the non-prosthetic side to skip or hop. In order to regain the reciprocal pattern of typical trip recovery, the ability to quickly shift weight between limbs is one prerequisite to make steps. Hence, dampened weight-shifting ability could be a root cause for failed trip recovery. Evidence supported sensorimotor and neural plasticity after amputation, indicating the patients have the potential to learn motor tasks despite their amputation. Weight-bearing intervention has been commonly implemented to this population to mainly engage the involvement of the prosthetic side. Further, some rehabilitation exercises emphasize not only bearing weight on the prosthetic side but also the ability to shift weight to and from the prosthetic limb. This type of balance training programs has the advantage of clinical feasibility because it requires only affordable instrument that most of the clinical settings already have and a relatively small space. However, there is no direct evidence regarding the effect of balance training on trip recovery response. In other words, the transferring of weight-shifting ability to more effective fall recovery response in individuals with lower limb loss is currently unproven. Retaining some knee control and being free from the concerns of controlling a prosthetic knee, individuals with unilateral below-the-knee amputation may benefit from a weight-shifting training to improve their response to tripping.

Specific Aims

Individuals with lower limb amputation fall more frequently than their age-matched counterparts; the consequences of falls are detrimental. The fact that community-dwelling amputees are typically older leaves falls an even more important topic that warrants research. In order to precisely evaluate the fall recovery response, we hope to develop a tripping fall simulation protocol that incorporates most of the strengths and accounts for most of the weakness in existing protocols. A ground-reaction-force-based treadmill acceleration/deceleration method enables precise and reproducible tripping perturbation delivery. We further equipped this treadmill acceleration/deceleration method with unilateral tripping perturbation feature enabled by the dual-belt treadmill to better simulate the nature of tripping. In Aim 1 (Chapter 2), we validated this protocol by examining if the protocol could be used to differentiate the fall recovery responses from young and middle-aged to older adults.

The fall recovery responses of prosthetic versus non-prosthetic tripping has been studied mostly in people with above-the-knee amputation. It is expected that transtibial amputees would manifest differently from transfemoral amputees due to the partially preserved knee control of the amputated limb. However, the current knowledge base on persons with transtibial amputation is inconclusive and was acquired mostly from young military service members. We hoped to understand the biomechanical performance of prosthetic and non-prosthetic tripping in community-dwelling individuals with transtibial amputation (Aim 2, Chapter 3).

Lastly, asymmetric weight-bearing and movement patterns are well-documented issues in the amputee population. It alters how individuals with unilateral amputation perform daily activities and how they react to perturbations and other environmental challenges. Weight-shifting balance training is a common intervention in amputation rehabilitation. Most of the clinical settings has the instrument to apply such intervention and has the potential to change the fall recovery responses. However, the effect of a weight-shifting training on trip recovery response had not yet been specifically investigated. We employed a weight-shifting intervention to a group of transtibial amputees. We investigated whether the effects of weight-shifting training would transfer to their trip recovery responses (Aim 3, Chapter 4).

Chapter 2: Description, Reliability and Validity of a Ground-Reaction-Force-Triggered Protocol for Precise Delivery of Unilateral Trip-Like Perturbations During Gait

Abstract

Tripping is a common cause of falls. Concerns regarding the precision of delivery of simulated-fall protocols reside in the current biomechanical literature. This study aimed to develop a treadmill-based protocol that generated unanticipated trip-like perturbations during walking with high timing precision. The protocol utilized a side-by-side split-belt instrumented treadmill. Treadmill belt acceleration profiles (two levels of perturbation magnitude) were delivered unilaterally when the tripped leg bore 20% of the body weight during early stance. Peak trunk flexion angle was used to represent fall recovery responses and likelihood of falls. Test-retest reliability of the fall responses was examined in 10 young participants; validity was examined as whether the protocol can differentiate the fall responses between young and older adults (n=10 per group). Results showed that the perturbations could be precisely delivered during early stance phases (10-45 ms after initial contact). The protocol elicited excellent reliability of responses during both perturbation severities (ICC=0.944 and 0.911). Older adults exhibited significantly greater peak trunk flexion than young adults (p=0.035), indicating that current protocol was valid in differentiating individuals with different levels of fall risks. This protocol addressed some issues of previous simulated-fall protocols and may be useful for future fall research and clinical intervention.

Introduction

Falls frequently happen when normal walking cycle is disrupted, such as during a tripping event.(Berg, Alessio, Mills, & Tong, 1997; Heijnen & Rietdyk, 2016; Li et al., 2006) Whilst trips happen in people of all ages, they are seldomly considered a hazard for serious injuries in younger people due to their ability to arrest the fall. In contrast, the prevalence of falls has been surveyed as high as 62.1% in adults older than 50 years of age.(Painter, Elliott, & Hudson, 2009) A previously study found that residents in long-term care facilities fell mostly due to incorrect weight-shifting (41%) and tripping (21%).(Robinovitch et al., 2013)

The definition of a trip varies in the literature. The current consensus defines it as an abrupt obstruction of a foot followed by a loss of dynamic postural control during gait.(Arena, Davis, Grant, & Madigan, 2016; Pijnappels, Kingma, Wezenberg, Reurink, & van Dieën, 2010; Potocanac, Pijnappels, Verschueren, van Dieën, & Duysens, 2016; Schulz, 2017; Sessoms et al., 2014) In most real-life scenarios, only one foot is obstructed. And it is this sudden stop of the progression of a leg, coupled with the unexpected forward acceleration of body center of mass relative to the base of support that causes a trip and fall.(B. C. Lee, Martin, Thrasher, & Layne, 2016; Zhang, D'Andrea, Nunnery, Kay, & Huang, 2011) Researchers have focused on the trunk kinematics and kinetics in relation to leg motions and dynamic base of support to study mechanisms underlying trip-related falls and fall recovery responses.(Aviles et al., 2019; Pavol, Owings, Foley, & Grabiner, 2001)

Currently in the literature, one of the most commonly used methodologies to simulate tripping perturbations is the treadmill deceleration/acceleration method.(Berger, Dietz, & Quintern, 1984; Sessoms et al., 2014) Berger et al. first developed the idea of simulating tripping during walking by suddenly accelerating or decelerating the treadmill belt velocity while a participant walks on it.(Berger et al., 1984) This is the currently preferred method, as it requires no additional objects, and the walking surface-based perturbations can be delivered unexpectedly to minimize anticipatory reactions.(Zhang et al., 2011) Moreover, the magnitude of the perturbation is easily adjustable to allow observations of a range of fall recovery responses, which is typically achieved by manipulating the displacement duration and/or magnitude of the walking surface.(Pai, Maki, Iqbal, McIlroy, & Perry, 2000; Szturm & Fallang, 1998) Based on the analyses of electromyography activities and recovery responses, the researchers concluded that this type of protocol was able to elicit responses comparable to what occurs in daily life.(Berger et al., 1984; Owings, Pavol, & Grabiner, 2001) Sessoms et al. further improved the method by adding a brief deceleration phase of the treadmill belt to simulate the foot obstruction followed by belt acceleration to evoke the forward trunk motion (**Figure 5B**).(Sessoms et al., 2014)

While treadmill-based perturbation protocol is widely used in research, a number of limitations has been discussed, particularly regarding the precision of perturbation delivery. An adequate tripping protocol should allow precise control of the timing of perturbation including the onset and perturbation duration.(King, Eveld, Martínez, Zelik, & Goldfarb, 2019) This is critically important because the trip recovery strategy deployed by the walker is highly dependent on these two factors.(Eng, Winter, & Patla, 1994; Shirota, Simon, & Kuiken, 2014) Hence, there is a need for an experimental protocol that can consistently and precisely reproduce the perturbations to make valid within- or between-subject comparisons. Previous studies have exploited the walker's kinematics or a combination of kinematics and ground reaction force (GRF) to as triggering conditions for perturbation delivery.(Debelle, Harkness-Armstrong, Hadwin, Maganaris, & O'Brien, 2020; Ilmane, Croteau, & Duclos, 2015; Yoo, Seo, & Lee, 2019) However, these studies did not focus on timing control, and therefore the precision of perturbation delivery remains unknown. For example, in the protocol described by Sessom and colleagues, the tripping perturbation was delivered to the limb in single-limb support phase but the exact timing (i.e. early, mid, or late stance) was not prescribed. This leaves room for improvement regarding the precision and reproducibility in the fall research protocols.

The purpose of this study was to further improve the biomechanical evaluations of falls and fall recovery responses by describing a GRF-triggered treadmill perturbation method that emphasizes precise and reproducible timing control of the tripping perturbation delivery. In this study, we aimed to quantify the timing of perturbation delivery, establish the protocol's ability to elicit reliable fall recovery responses, and validate the protocol by examining if it can be used to differentiate trip recovery responses between populations with different levels of fall risk (i.e. younger vs. older adults).

Methods

System Apparatus Design

A Bertec side-by-side split-belt instrumented treadmill (Model ITC-11-20L-4, Bertec Corp., Columbus, OH, USA) is capable of separate control of the movements of the two treadmill belts. Each belt was equipped with one force plate capturing ground reaction force data from the walker's left and right foot contacts. The force data was sampled at 1000Hz, time-synchronized with the VICON (Oxford Metrics, Oxfordshire, UK) motion data and streamed by the Software Development Kit (SDK) to the MATLAB software (MathWorks Inc., Natick, MA, USA) on a personal computer. The MATLAB code (**Supplementary materials**) served as the interface to communicate between Datastream SDK and the treadmill controller. Specifically, the program read the vGRF from Datastream SDK and executed the pre-programmed perturbations via the treadmill controller. The treadmill controller received remote control commands from MATLAB via the Transmission Control Protocol/Internet Protocol (TCP/IP) port, through which the program delivered the prescribed tripping perturbation by accelerating/decelerating the treadmill motors (**Figure 4 and 5B**).

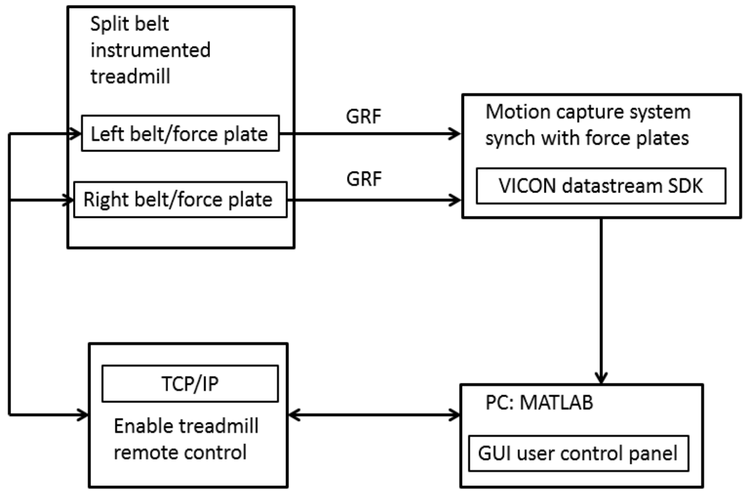


Figure 4. System Apparatus

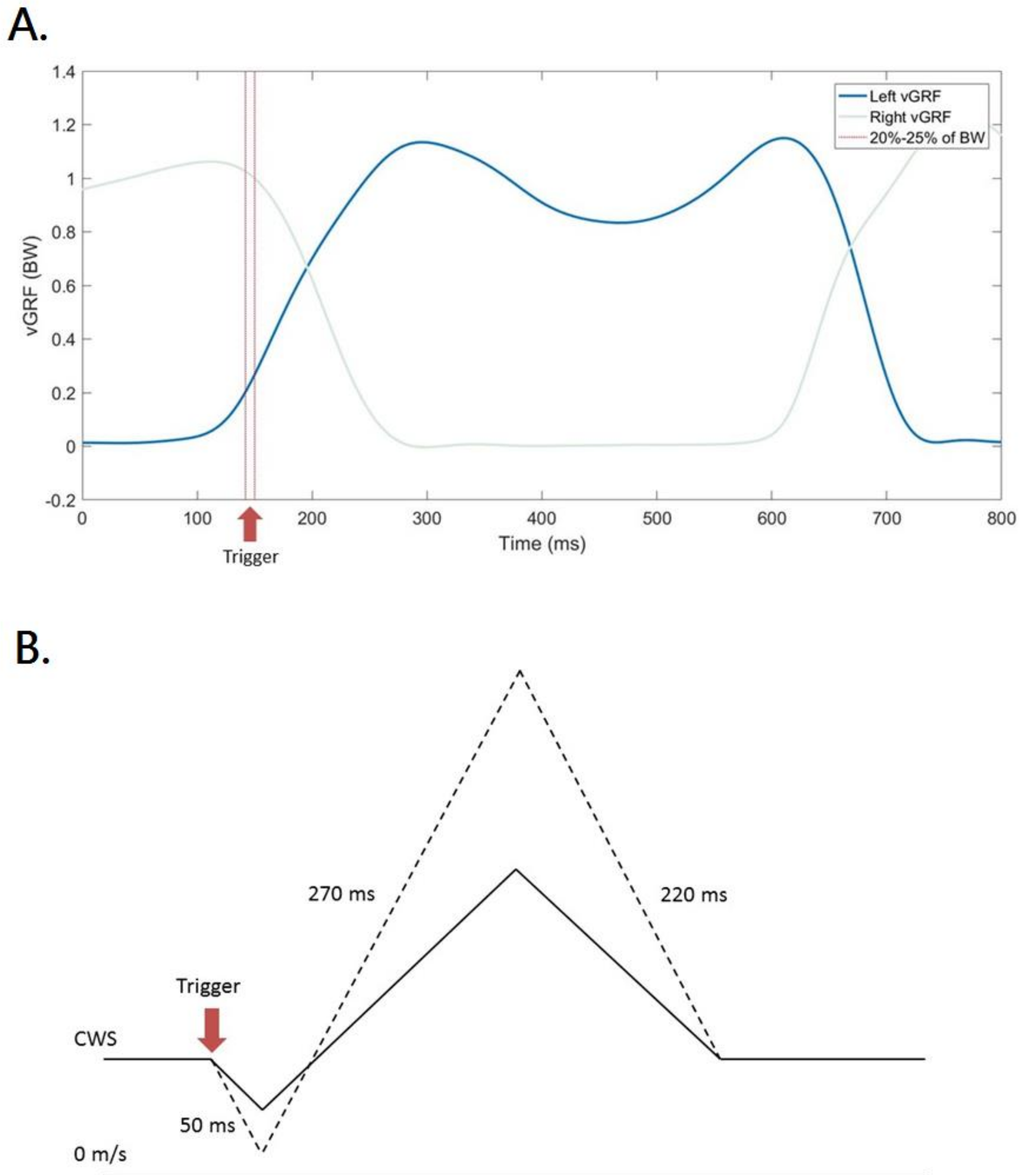


Figure 5. Perturbation Triggering Criteria and Treadmill Velocity Profile
(A) GRF from both limbs and perturbation triggering criteria. Blue vGRF line is from the tripped left limb; red lines enclose the designated window of a triggering event. (B) Treadmill velocity profile. Solid line stands for a tripping perturbation with a small acceleration; dash line represents a tripping perturbation with a large acceleration.

Perturbation Design

The treadmill-based tripping perturbation protocol began with establishing the participants' comfortable walking speed (CWS). During a perturbation the designated treadmill belt, either left or right, decelerated for 50ms, followed by 270ms of acceleration, and then decelerated again for 220ms to return to the CWS (**Figure 5B**). Two acceleration levels were used to simulate tripping perturbations of two levels of magnitude (small vs. large).(Sessoms et al., 2014) The acceleration magnitude utilized in the protocol was linearly scaled by the CWS. For a CWS at 1 m/s, the acceleration/deceleration was either ± 6 m/s^2 (small tripping perturbation) or ± 12 m/s^2 (large tripping perturbation). This allows the delivery of more realistic magnitudes of perturbation for individuals with slower walking speeds.

The automatic triggering criteria were based on the vGRF profile with the intention to deliver the perturbation during early stance phase of the tripped limb to reproduce a lowering strategy and ensure that the tripped limb went through the full course of the velocity changes of the treadmill belt. The perturbation is triggered when the following conditions are jointly met: First, the vGRF on the tripped side had to be between 20-25% of the person's body weight which was determined by asking the participant to stand quietly on the treadmill before the walking trials. Second, vGRF that met the first condition had to be greater than the vGRF 10ms prior to ensure that the trigger would occur in the ascending phase of the vGRF typical of during the early stance phase (**Figure 5A**).

Electro-Mechanical Delay

Since the control system coordinates 3 main components, there is likely electro-mechanical delay residing in the system. We recorded the timing when MATLAB delivered the perturbation as well as the onset of designated treadmill belt velocity change. The time difference between these two time points was the delay. (**Figure 6**) In addition, we measured the foot position at which when the perturbation took place (actual treadmill velocity change) to examine that given the delay, whether the foot position was still in the instance during gait as we planned.

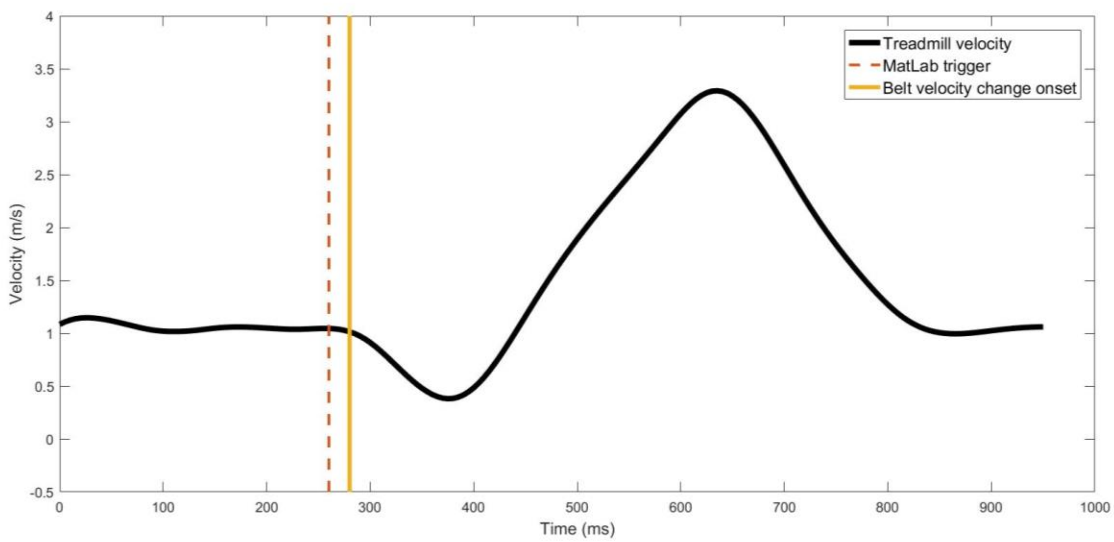


Figure 6. Example of a Trip at 12 m/s^2 and Electro-mechanical Delay

To measure the electro-mechanical delay, two additional markers were placed directly on the treadmill belts to capture the actual movement of the belts. MATLAB output the vGRF values of the instance that it triggered the trips. We traced back to the corresponding frame number of this specific vGRF as the onset of MATLAB trigger. The onset of treadmill belt velocity change was based on the velocity of the reflective markers on the belts. The first frame that the belt velocity was below 2 standard deviations (SD) of the CWS would be marked as belt velocity change onset.

<u>Protocol Reliability and Protocol Validation</u>

To validate our system, we examined the following parameters: 1) the timing consistency in delivering the perturbation in relationship to the gait phase percentages; 2) the test-retest reliability of the fall recovery responses elicited by the protocol; and 3) whether the protocol can be used to differentiate the trip recovery trunk kinematic responses between younger and older adults. The research protocol and procedures were approved by the Biomedical IRB of University of Nevada, Las Vegas.

For reliability, we recruited participants between ages of 18-30 from a university student population with no known diseases, injuries, or impairments that may influence their gait. Participants for reliability tests visited the laboratory twice, two weeks apart. Procedures of the two visits were identical. After provided consents for participation, participants were tested with the abovementioned protocol and with the kinematics captured using a 12-camera VICON motion capture system (sampling rate = 200Hz). The kinematic data were lowpass-filtered at 10Hz. A marker set for lower extremities and trunk was applied by the same investigator (**Figure 7A**). A harness attaching to a load cell was adjusted to prevent the participants from hitting the ground but not interrupt their gait (**Figure 7B**). Falls were defined as when the participants had to grab the supporting struts, or when the load cell/harness supported more than 50% of their body weight. Four different tripping incidences were delivered, including: small perturbation on the left limb and on the right limb, large perturbation on the left limb and on the right limb. The four perturbations were delivered in random order of timing, magnitudes, and sides.

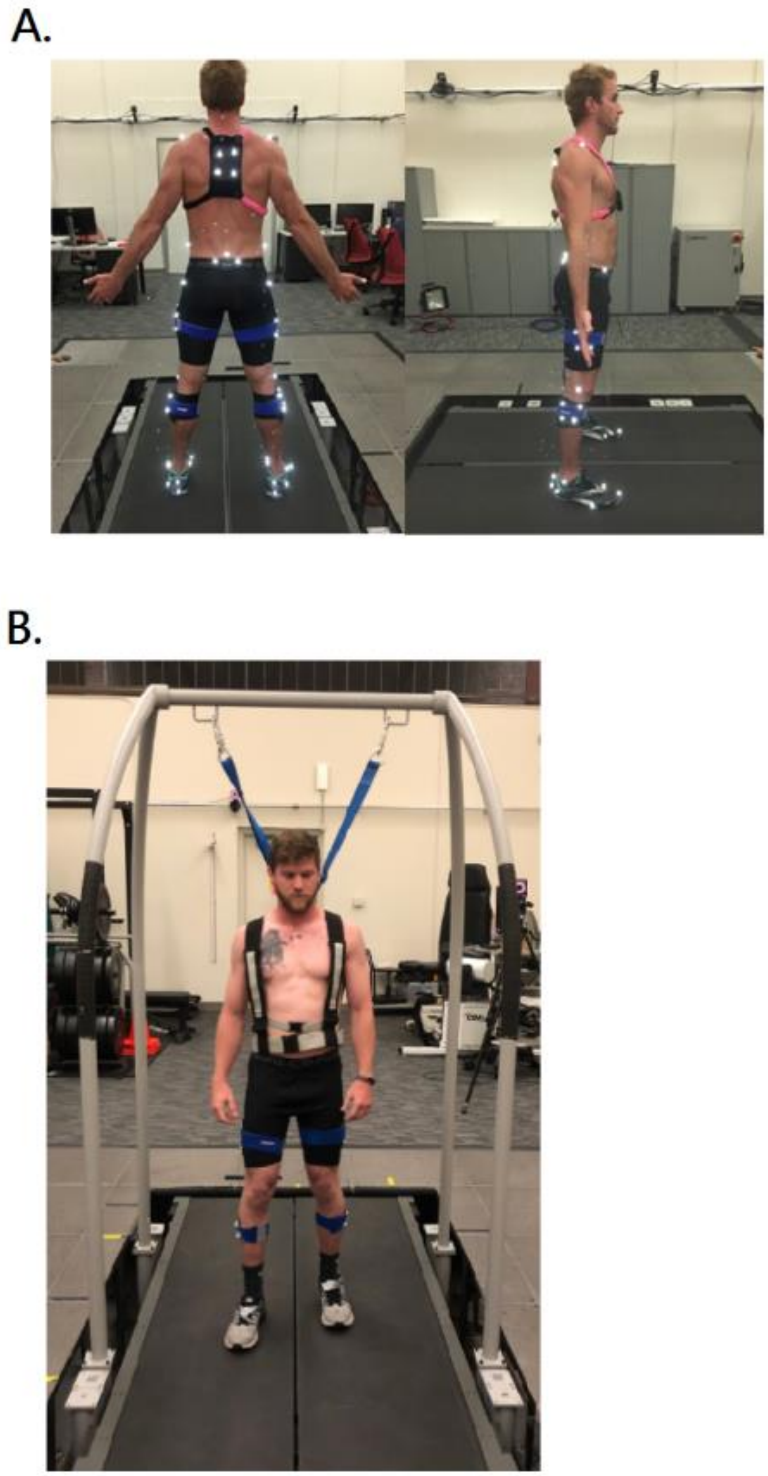


Figure 7. Experimental Settings
(A) Passive reflective marker set. (B) Hardware settings.

Peak trunk flexion angle during a tripping event have been shown to be predictive of a fall(Mark D Grabiner et al., 2008; Owings et al., 2001). Four events, initial contact, perturbation onset, contralateral limb contact, and tripped limb contact were chronologically identified in reference to foot position and vGRF data. Positive trunk flexion angles indicate trunk flexion. Peak trunk flexion angle, after perturbation onset and before tripped limb contact, was located and used for analysis. To confirm the consistency of perturbation delivery timing, the perturbation onset timing in percentage of stance phase was calculated. The length of stance phase was obtained from the normal walking period as an average value from 3 strides.

A group of older participants (>50 years of age) was recruited from the university and the local community for the purpose of protocol validation. They had no neurological, cardiovascular, and current musculoskeletal diseases that preclude walking on the treadmill. This older group visited the laboratory once. We compared their trip recovery trunk kinematic responses against the values from younger participants' first visit to eliminate the possibility of learning effects.

Statistical Analysis

Electro-mechanical delay is presented with mean, SD, and range. Test-retest reliability of the peak trunk flexion angle was estimated by intraclass correlation coefficients (ICC). Two-way mixed-effects model with absolute agreement definition for single measurement was selected ($ICC_{3.1}$)(Koo & Li, 2016). Two-way ANOVAs were used for examining the effects of group (younger vs. older) and perturbation accelerations (small vs. large) on peak trunk flexion angle. Significant level was set at 0.05. Statistical analyses were all conducted using SPSS version 24 (IBM Corp., Armonk, NY, USA).

Results

Ten younger individuals and twelve older individuals participated. One person in the older group reported hip pain and one other expressed fear of falling after the first tripping perturbation. These two persons withdrew from the study and were not included in the analysis (**Table 1**). One example of trunk flexion angle during perturbation and recovery was shown in **Figure 8**. For capturing electro-mechanical delay, 24 trips were collected and analyzed. The duration of delay was averagely 26.5ms, ranging from 5-35ms with a SD of 7.74ms. Furthermore, the heel position at the onset of perturbation delivery was consistently in the early stance phase consistent to our target timing of perturbation delivery (**Figure 9**).

Table 1. Demographics

	Age (y)	Sex (M/F)	BW (kg)	CWS (m/s)
Younger (n = 10)	20.90±1.66	3/7	62.79±9.63	0.90±0.17
Middle-aged (n = 10)	57.10±4.70	7/3	84.20±12.96	0.95±0.28

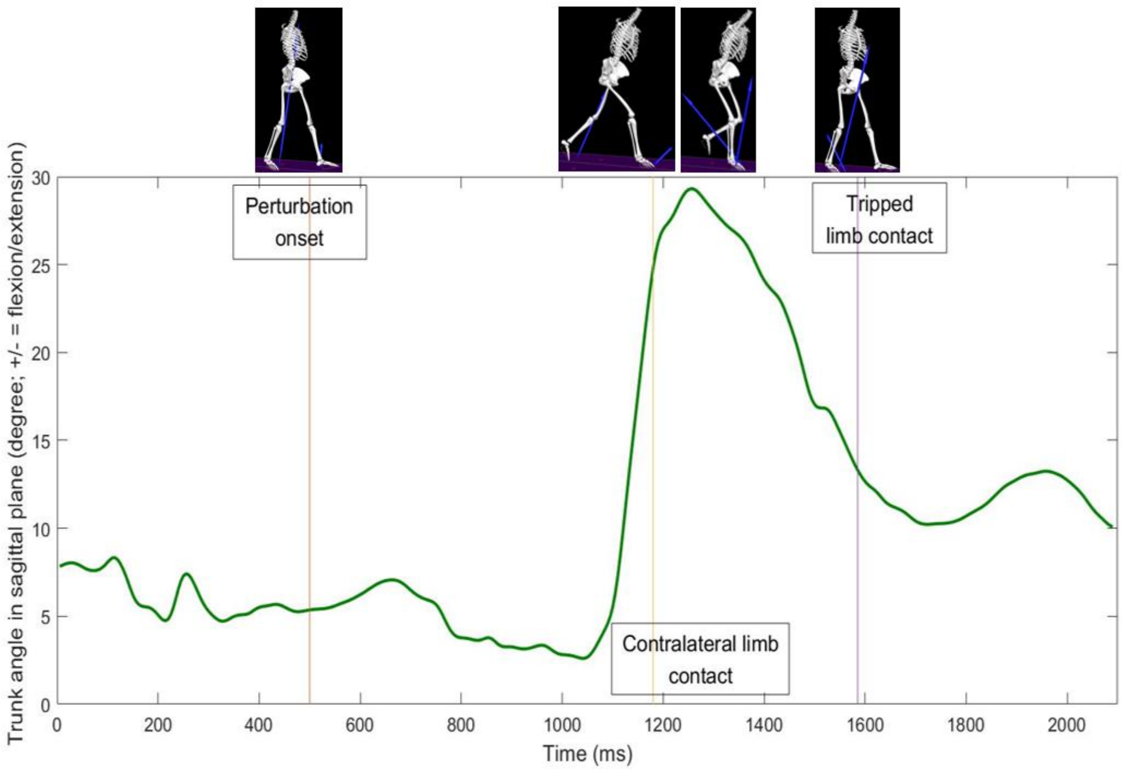


Figure 8. Example of Sagittal Trunk Flexion Angle in Response to the Tripping Perturbation and Recovery.
Left limb was the tripped limb.

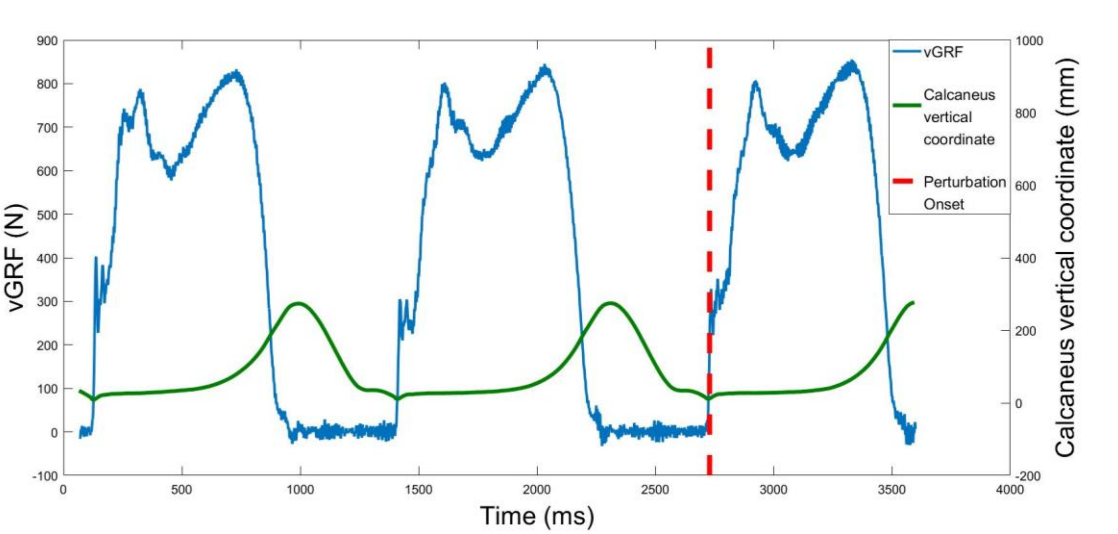


Figure 9. Real-time Raw GRF and Calcaneus position of Tripped Limb at One Perturbation

<u>Test-Retest Reliability</u>

In the younger group, the length of the stance phases ranged from 635ms to 810ms. Perturbations were consistently delivered at 10-45ms after initial contact, approximately 2.5% (median, range [1.42-4.55%]) of the stance phase. Peak trunk flexion angles during both the small and large perturbations were consistent between the visits (**Table 2**). The protocol yielded excellent test-retest reliability of the trunk kinematics during both small (ICC=0.944, 95% CI [0.792, 0.986]) and large perturbations (ICC=0.901, 95% CI [0.650, 0.975]).

Table 2. Fall Incidence and Peak Trunk Flexion Angle from Younger and Older Groups, During Small and Large Tripping Perturbations.

	Small acceleration			Large acceleration		
	Fall frequency (%)	Peak trunk flexion angle (°)		Fall frequency (%)	Peak trunk flexion angle (°)	
		Visit 1	Visit 2		Visit 1	Visit 2
Younger (n = 10)	0 (0%)	13.37±6.89	13.55±6.72	0 (0%)	23.50±10.00	23.20±8.19
Middle-aged (n = 10)	3 (15%)	19.88±6.44		6 (30%)	33.70±10.05	

Protocol Validation

Of the 20 included participants, the older group was significantly heavier (p<0.001; **Table 1**). There was no significant difference in CWS between groups (p=0.63). None of the participants in the younger group fell in any tripping trials. The older group had 3 falls under the small perturbation condition and 6 falls under large perturbation condition, yielding percentages of fall 15% and 30% respectively (**Table 2**). Fell trials were not included in the comparisons for trunk angles. Significant group main effect (p=0.035) and significant perturbation magnitude main effect (p<0.001) were found in peak trunk flexion angles. Large perturbations elicited significantly greater peak trunk flexion angles than small perturbations did by 72%. Older adults recovered from trips exhibiting 45% greater peak trunk flexion angle comparing against young adults (**Table 2**).

Discussion

The proposed protocol was successful in delivering precise and reproducible perturbations that simulates tripping falls. The two acceleration profiles simulated two levels of magnitude of tripping perturbations as intended. The protocol was capable of differentiating the trip recovery responses between younger and older adults.

We confirmed the consistency of the protocol in two aspects. First, we observed that all perturbations were precisely delivered before the first 5% of the stance phase as intended in a range of different CWS and lengths of stance phase. This high level of consistency was attributed to the GRF-based triggering criteria. Second, the test-retest reliability of the protocol was confirmed by the participants' reproducible trunk kinematics in response to the tripping perturbation. With a two-week interval between tests, this protocol was able to elicit similar and reliable test-retest responses from the participants. A protocol that provides precise and reproducible perturbation timing can enable further investigations into the mechanisms of falls and successful recovery.

In previously established simulated tripping research protocols based on obstacles, a concealed object is either placed on the ground to obstruct the foot or dropped onto the treadmill in order to produce a trip.(Arena et al., 2016; Crenshaw, Kaufman, & Grabiner, 2013b; Schillings, Mulder, & Duysens, 2005) There is the concern of anticipation due to that the participants might be able to see the obstacle out of the corner of their eyes, hear the sound, or perceive the vibration made by the obstacle. Our protocol used GRF-based triggering criteria to precisely invoke the trip by taking advantage of the modern force plate-instrumented treadmills. GRF has been shown to remain consistent within one's gait cycle and most consistent at one's self-selected walking-speed(Masani, Kouzaki, & Fukunaga, 2002; Patel & Bhatt, 2015). For these reasons, our protocol delivers trips triggered by GRF while the participant walks at a self-selected CWS. Furthermore, two magnitude levels of perturbation were delivered to either the left or the right leg with random timing. In doing so, the tripping perturbations were induced at precisely the same phase in the gait cycle (i.e. the early stance phase) and elicited consistent trunk kinematic responses. Therefore, our data supports that the novel GRF-based triggering protocol can deliver reproducible tripping perturbation results likely due to the precise timing control. Our novel protocol with more consistent responses may reduce the need for multiple trials and the potential confounding effect of learning and anticipation.

Two treadmill belt acceleration profiles were utilized to simulate two levels of magnitude of tripping perturbations. We observed the magnitude main effect on peak trunk flexion angle in both age groups. In response to the larger perturbations, both groups exhibited higher peak trunk flexion angles. Our results aligned with the previous findings that more severe falls is associated with a loss of trunk control.(Bourke, O'Donovan, & Olaighin, 2008; Liu & Lockhart, 2014) The magnitude main effect we found also echoed with the results by Lee et al. who examined the slip recovery responses after different severities of perturbation.(A. Lee, Bhatt, & Pai, 2016) Altering the duration of the tripping perturbation may be another way to manipulate the magnitude of perturbations.(Tokuno, Cresswell, Thorstensson, & Carpenter, 2010) However, manipulating the duration of the perturbation may influences the gross fall recovery strategy selection.(Shirota et al., 2014) Since the acceleration profile (i.e. both the acceleration and duration) can be easily manipulated in our novel protocol, this enables researchers and clinicians to provide different levels of perturbation for potential research and training purposes.(Crenshaw, Kaufman, & Grabiner, 2013a)

Tripping fall recovery requires rapid responses immediately after the onset of the trip,(Mark D Grabiner & Jahnigen, 1992; Stelmach & Worringham, 1985) and the responses consist of three main conceptual components: the reactive control of the forward rotation of trunk,(Mark D Grabiner, Feuerbach, & Jahnigen, 1996; Mark D Grabiner, Koh, Lundin, & Jahnigen, 1993) the use of a support limb in slowing the fall,(Dietz, Quintern, Boos, & Berger, 1986; Eng et al., 1994) and the recovery stepping of sufficient length.(Fukagawa & Schultz, 1995; Mark D Grabiner et al., 1993) In studies on older adults, the peak trunk flexion angles from fell trials were 37.0 to 58.5 degrees, and those from recovery trials were 22.0 to 37.3 degrees.(M. D. Grabiner, Bareither, Gatts, Marone, & Troy, 2012; Pavol et al., 2001) Consistent with previous findings in older adults, the current protocol elicited similar trunk kinematics to what other protocols prompted in the recovery trials. Moreover, this protocol has the feature of unilateral foot perturbation enabled by the use of the side-by-side dual belt treadmill.

We validated that the proposed protocol can serve as an assessment tool to differentiate the trip recovery responses between younger and older adults. The protocol successfully induced more falls in the older group as well as more falls in the large acceleration condition. Lack of control of the trunk forward motion has been identified as a risk factor of higher rate of fall and poorer balance recovery after a trip.(Pavol et al., 2001) Our results showed that older adults demonstrated greater peak trunk flexion angle relative to their younger counterparts. These findings showed that the novel protocol may be useful in detecting the disparate trip recovery responses between different age and clinical populations. Since the compensatory postural responses evoked by trips are not solely under volitional control, conventional balance measurements might not be responsive to such balance insufficiency.(Nashner, 1976, 1977, 1980) Hence this task-specific tool with the ability of differentiating populations in high risks of tripping falls may be used as a screening instrument.

The main limitation of this study was that the tripping perturbations were delivered during stance phase as oppose to the swing phase during trips in real life scenarios. This is an inherent limitation of the treadmill-based tripping protocol given that perturbation can only be delivered to the leg in stance. However, our data showed that at the instant of the perturbation (i.e. early stance phase), approximately 80% the body weight was still bore by the non-tripped limb. Our success in consistent delivering the perturbation at this early stance phase may be as realistic as it can be when it comes to treadmill-based simulated fall protocols.

Chapter 3: Comparison of Fall Recovery Responses of the Prosthetic versus Non-Prosthetic Limb in Individuals with Unilateral Transtibial Amputation

Abstract

Individuals with unilateral transtibial amputation (UTTA) are at high risks of falling especially when they encounter destabilizing perturbations. This population exhibited asymmetric movement patterns when performing various activities, and these patterns may manifest during postural perturbation and subsequent recovery. Hence the study aims to compare the responses between the fall recovery responses between conditions where the prosthetic and non-prosthetic limbs were perturbed in individuals with UTTA. Twelve participants were exposed to perturbations delivered during static standing and dynamic walking conditions. We extracted the durations that comprised the fall recovery; the unloading and swinging durations of each step as well as knee and hip joint excursion during perturbations for our comparisons. During the static perturbation condition, the duration of the recovery step and the swing proportion of the step were significantly longer when the prosthetic limb was perturbed. During dynamic perturbation condition, the duration of the first recovery step and the swing proportion of this step were significantly longer when the non-prosthetic limb was perturbed in which the movement was executed by the prosthetic limb. No significant differences were observed in the second recovery step. The knee and hip excursion range were not significantly different side to side. Hence, our findings indicated that the prosthetic

limb spent longer time to cover similar joint excursion and the slower movement was mainly attributed to the swing phase of the recovery stepping. Our findings suggested that the prosthetic limb stepping performance may be a target for improvement with therapeutic intervention. Furthermore, individuals with UTTA may be at a higher risk of falls when the prosthetic limb is perturbed during standing, and at a higher risk of tripping falls when the non-prosthetic limb is perturbed during walking.

Introduction

A global estimation revealed that in 2019, there were 176 million individuals with amputation who need rehabilitation service.(Cieza et al., 2020) Globally, 5.5 million years of life lived with disability have been attributed to amputation(Cieza et al., 2020), which placed it as one of the most impactful conditions to human health loss and burden on society. It has been reported that 50 to 80% community-dwelling persons with lower limb amputation (LLA) experienced at least one fall in the past year.(Ülger, Topuz, Bayramlar, Erbahçeci, & Sener, 2010) Tripping is one of the leading causes of falls in individuals with LLA, including in individuals with transtibial amputation.(Rosenblatt, Bauer, Rotter, & Grabiner, 2014)

Asymmetric weight-bearing and postural reorganization have been extensively reported in individuals with LLA.(Rougier & Bergeau, 2009; Vanicek, Strike, McNaughton, & Polman, 2009b) An inevitable consequence of LLA is altered sensory and neuromuscular control in detecting and responding to postural perturbations. A previous study showed that individuals with unilateral transtibial amputation (UTTA) were able to sustain whole-body center of mass displacement similar to age-matched non-amputees when confronting multi-directional perturbations induced by support surface translation.(Bolger, Ting, & Sawers, 2014) The whole-body level postural control scheme comparable to non-amputees may be achieved through greater contribution from the intact limb, demonstrating a coping strategy that characterized limb to limb asymmetry.(Curtze, Hof, Postema, & Otten, 2012) To demonstrate this, Rusaw et al. asked people with UTTA to stand with their prosthetic limb or non-prosthetic limb on a tilting force platform.(Rusaw, Hagberg, Nolan, & Ramstrand, 2013) The platform was positioned to elicit either toe-up or toe-down perturbations unexpectedly. They found that participants with LLA displayed generally delayed electromyography responses to postural perturbations comparing to non-amputees. The author speculated that the delayed responses may have been caused by the dampened sensory input on the prosthetic limb which prolongs the detection and response to perturbations. The dampened sensory input from the prosthetic limb was confirmed by the observation that when the prosthetic limb was perturbed, the non-prosthetic limb also showed significantly delayed EMG response compared to conditions that the perturbations were delivered to the non-prosthetic side.

Individuals with UTTA typically develop and adopt different gait patterns comparing to individuals without amputation even when accounting for gait velocity.(Varrecchia et al., 2019) A previous study revealed significantly decreased swing duration on the non-prosthetic side in contrast to individuals without LLA.(Nolan et al., 2003) Furthermore, shorter stance duration and longer swing duration were found in the prosthetic side, indicating a reliance on the non-prosthetic, intact limb to provide stability during gait.(Varrecchia et al., 2019) However, the temporal asymmetry often accompanies load and joint kinetic asymmetry that were believed to ultimately jeopardize balance control and gait efficiency.(J. Engsberg, Lee, Tedford, & Harder, 1993; J. R. Engsberg, Lee, Patterson, & Harder, 1991)

Kinetic gait patterns in joint work and energy recovery (i.e. the percent of energy conversion between gravitational potential energy and kinetic energy) in individuals with UTTA may be an indicator for fall risks in response to perturbations during walking. For example, Tesio et al. showed that energy recovery was 8% greater in the prosthetic limb.(Tesio, Lanzi, & Detrembleur, 1998) The authors speculated that individuals with UTTA tend to adopt an inverted pendulum-like gait pattern on the prosthetic side. However, this movement pattern of the prosthetic leg during gait likely comes at a cost of the limb being stiffer and less accommodating. Thus, when reacting to perturbations, individuals with transtibial amputation may rely heavily on the non-prosthetic limb to arrest falls, and in certain situations making them more susceptible to falls when taking a step with the prosthetic limb to arrest a fall is unavoidable.(Sheehan, Beltran, Dingwell, & Wilken, 2015) In agreement with this theory, when comparing fallers and non-fallers with UTTA, Vanicek et al. showed that the fallers exhibit greater peak vertical ground reaction force and loading rate during initial contact (the first peak vertical ground reaction force, vGRF) on their prosthetic limb.(Vanicek, Strike, McNaughton, & Polman, 2009a) This is indicative of a stiffer and uncontrolled gait pattern.

Despite the previous studies on the asymmetric movement patterns in individuals with UTTA, most studies on dynamic postural perturbations such as tripping and slipping during gait were limited to people with transfemoral amputation.(Bellmann, Schmalz, & Blumentritt, 2010; Crenshaw, Kaufman, & Grabiner, 2013; Shirota, Simon, & Kuiken, 2015) These studies showed it took longer time to complete the fall arrest when the prosthetic limb was perturbed.(Crenshaw et al., 2013; Shirota et al., 2015) However, it is possible that individuals with UTTA may perform differently when encountering a dynamic perturbation due to having a certain level of control of their residual knee. The duration of each component of the recovery responses and joint kinematics in this population have not yet been well investigated. Therefore, the purpose of the present study is to compare the fall recovery biomechanical responses when simulated trips were delivered to the prosthetic and the non-prosthetic limbs in individuals with UTTA. Understanding the side-to-side difference in tripping and fall recovery response can be valuable in developing more effective rehabilitation protocols and prosthetic components to reduce fall risks in individuals with UTTA.

Methods

Participants

We recruited participants from prosthetic clinics, physical therapy clinics, and local amputee support group in the Southern Nevada region of the United States. Community-dwelling individuals with unilateral transtibial amputation, 18 to 80 years of age, with at least 6 months of prosthesis use experience, and being able to walk independently for at least 5 minutes were eligible to participate in the study. Those who had other health conditions that may influence the balance and safety, such as eye diseases, vestibular dysfunction, and other neurological conditions were excluded from participation. Powered ankle prosthesis users were also excluded. This research protocol was approved by the Institutional Review Board of University of Nevada, Las Vegas.

Trip Recovery Response Test

After participants provided consent for participation and filled out health and medical questionnaires including Activities-specific Balance Confidence (ABC) scale,(Mandel et al., 2016; Miller, Deathe, & Speechley, 2003) Fear of Falling Avoidance Behavior Questionnaire (FFABQ)(Landers, Durand, Powell, Dibble, & Young, 2011), and Prosthetic Limb Users Survey of Mobility (PLUS-M). Two types of perturbation were used to examine participants' fall recovery responses which were during standing and walking (i.e. static vs. dynamic perturbation condition). The system for perturbation delivery and data collection consisted of three main components: 1) Bertec side-by-side split-belt instrumented treadmill (Model ITC-11-20L-4, Bertec Corp., Columbus, OH), 2) VICON motion capture and data collection system and Datastream Software Development Kit (SDK; Oxford Metrics, Oxfordshire, UK), and 3) a customized controller program written in MATLAB (MathWorks Inc., Natick, MA) to coordinate the data input and treadmill control.

During static perturbation condition, participants started from standing still on the walking surface of the treadmill. The perturbation was created by one of the treadmill belts moving posteriorly for 200 ms at an acceleration rate of 6 m/s^2. Each perturbation was delivered unexpectedly and randomly to either the prosthetic or non-prosthetic limb. For the walking perturbation, our research group has developed a novel, GRF-based protocol that is able to produce reliable and repeatable trip-like perturbations during gait. (see Chapter 2) In this condition, participants were asked to walk on the treadmill at their self-selected, comfortable walking speed (CWS). A sudden perturbation to one limb was delivered during the early stance phase defined as when vGRF was between 20-25% of their body weight. When the triggering condition was met, one side of the split-belt treadmill would go through a prescribed deceleration-acceleration profile, before returning to CWS.(Sessoms et al., 2014) The belt acceleration rate was linearly scaled based on participant's CWS. In total, 2 standing and 2 walking perturbations (to the prosthetic limb [P] and to the non-prosthetic limb [NP]) were delivered in random order. Participants' kinematic responses to the perturbations were recorded using a 12-camera motion capture system (sampling rate = 250Hz). A lower extremity plus trunk marker set was applied.(Shih et al., 2019) A harness attaching to a load cell was provided to prevent the participants from hitting the ground when a fall occurs but not interrupting their gait.

Data Process and Analysis

One of the dependent variables was frequency of falls out of the four perturbation trials, which was defined by either the participants holding onto the handlebar or the harness supported more than 50% of the body weight, based on the reading of the load cell. Temporal variables during the fall recovery event were also analyzed. During static perturbation, most fall recovery responses involved the perturbed limb making a forward step. Trials in which an ipsilateral stepping strategy was used in the analyses. Correspondingly, three events, perturbation onset, perturbed limb unloading, and ipsilateral limb contacting the ground were identified based on the GRF signal and the heel marker location, and the durations between these three events were calculated as limb unloading, swinging, and recovery stepping (sLU, sSwing, sRS; **Figure 10A**).

During dynamic perturbation, the typical recovery response consists of using the tripped side as a transient support limb before making the first recovery step with the contralateral limb, followed by a second recovery step with the ipsilateral, tripped limb (**Figure 10B**). The timing of the stepping events was determined using GRF and heel marker location. These events included perturbation onset (Onset), contralateral limb off (Contra Off; when the contralateral limb is completely lifted off the ground); contralateral limb contact (Contra On; when the contralateral limb makes the first ground contact of the first recovery step), ipsilateral limb off (Ipsi Off), and ipsilateral limb contact (Ipsi On; Figure 1B). Six temporal duration variables were calculated accordingly: 1) limb off (dLO), from Onset to Contra Off; 2) first swing phase (dSwing1) of the contralateral limb, from Contra off to Contra on; 3) second swing phase (dSwing2) made by the ipsilateral limb, from Ipsi off to Ipsi on; 4) first recovery step (dRS1) made by the contralateral limb, from Onset to Contra On; 5) second recovery step (dRS2) made by the ipsilateral limb, from Contra On to Ipsi On; 6) full recovery course (FR) including two recovery steps,(Pijnappels, Bobbert, & van Dieën, 2004) from Onset to Ipsi On (**Figure 10B**).

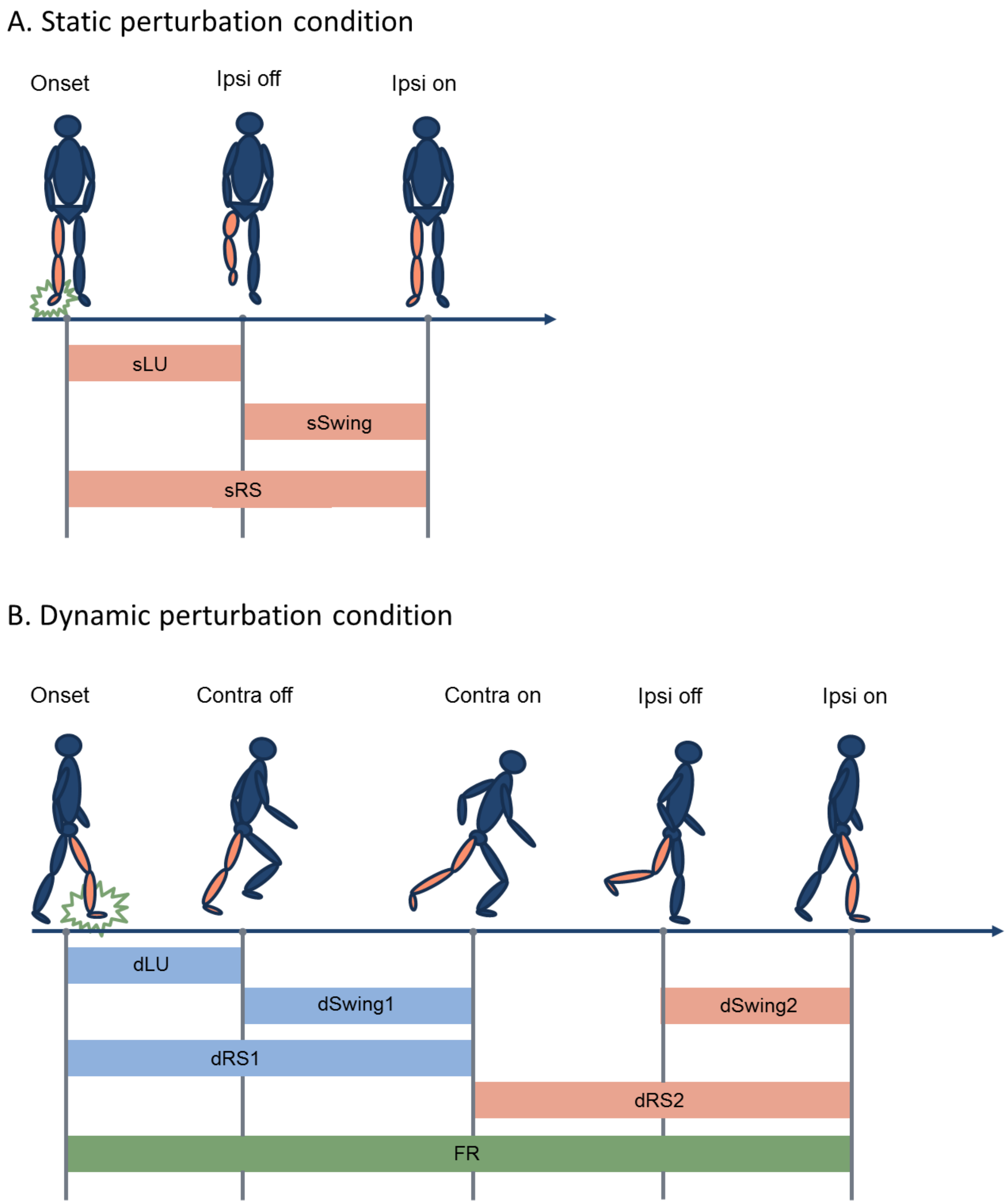


Figure 10. Typical Recovery Responses and Temporal Dependent Variables from Static and Dynamic Perturbation Conditions

(A) Static perturbation trials. The limb in orange is the perturbed/ipsilateral limb, and the limb in blue is the contralateral limb. (B) For a dynamic perturbation trial, 5 events are illustrated, and 6 durations were calculated as dependent variables. The limb in orange is the perturbed/ipsilateral limb, and the limb in blue is the contralateral limb. Five events were perturbation onset (Onset), contralateral limb off (Contra Off; when the contralateral limb is completely lifted off the ground); contralateral limb contact (Contra On; when the contralateral limb makes the first ground contact of the first recovery step), ipsilateral limb off (Ipsi Off), and ipsilateral limb contact (Ipsi On).

Finally, 3D marker trajectories were lowpass-filtered at 10 Hz. Hip and knee peak flexion (HipP and KneeP) and ranges of motion (HipR and KneeR) angles in the sagittal plane during sRS, dRS1, and dRS2 were extracted from the stepping limbs. Specifically, during sRS, the ipsilateral limb was the limb of interest (sHipP, sKneeP, sHipR, sKneeR); during dRS1 and dRS2, the limbs were contralateral (dHipP1, dKneeP1, dHipR1, dKneeR1) and ipsilateral (dHipP2, dKneeP2, dHipR2, dKneeR2) respectively.

Number of falls were tallied and presented as frequencies and compared using Chi-square tests. Depending on data distribution tested by skewness and kurtosis, paired t tests or Wilcoxon Signed Rank tests were used for comparing the nine temporal variables and twelve kinematic variables between prosthetic and non-prosthetic limbs within subject. Alpha level was at 0.05.

Results

We recruited 12 individuals with UTTA. One of the participants did not complete the static standing trials due to discomforts of the osseointegration implant and was excluded. **Table 3** shows each participant’s demographic information. Owing to the high inter-individual variance, we presented median and interquartile range in all tables and figures if not otherwise indicated.

Table 3. Demographics and Scores of Fall-related and Amputee-Specific Mobility Surveys

ID	Amputation side	Amputation cause	Sex	Age, y	Height, cm	BW, kg	CWS, m/s	ABC	FFABQ	PLUS-M
001	Left	Non-vascular	M	73	195.6	81.7	0.50	84.4	8	58.4
002	Right	Non-vascular	F	59	179.1	60.3	0.40	85.0	2	61.0
003	Right	Vascular	M	58	182.0	92.1	0.70	73.8	10	51.2
004	Left	Vascular	M	24	175.3	76.4	0.95	98.8	0	67.1
005	Left	Non-vascular	M	63	172.7	89.0	0.64	86.9	6	52.0
006	Left	Non-vascular	M	57	182.9	109.7	0.50	65.9	2	51.2
007	Right	Non-vascular	M	50	180.3	76.3	0.70	94.4	2	67.1
008	Left	Non-vascular	F	45	165.1	102.0	0.65	87.5	10	58.4
009	Right	Non-vascular	F	29	162.6	61.2	0.60	92.5	10	53.6
010	Right	Non-vascular	F	58	170.2	113.4	0.35	63.1	8	49.8
011	Left	Non-vascular	M	62	167.6	77.5	0.29	85.0	19	49.1
012	Right	Vascular	F	67	160.0	80.4	0.10	75.6	1	56.3
Mean	L/R 6/6	Vas/Nonvas 3/9	M/F 7/5	53.8	174.5	85.0	0.53	82.7	6.5	56.3
SD				14.7	10.2	17.0	0.23	11.0	5.5	6.3

BW: body weight; CWS: comfortable walking speed; ABC: Activity-specific Balance Confidence Scale, presented in average percentage score from 16 questions; FFABQ: Fear of Falling Avoidance Behavior Questionnaire, presented in sum score on 14 questions with a total possible score 0-56 (higher score indicates more fear); PLUS-M: Prosthetic Limb Users Survey of Mobility, presented in T-score with a total possible score 21.8-71.4.

Response to Static Perturbation

Two (16.7%) and three (25%) unsuccessful recoveries were observed when the prosthetic and non-prosthetic sides were perturbed respectively (χ^2=0.253, p=0.615). Overall, sLO was similar between sides (P vs. NP; 420 [165] vs. 350 [79] ms, p= 0.128; **Figure 11A**) whereas sRS and sSwing were significantly longer when the perturbation was delivered to the prosthetic limb than the intact limb (P vs. NP; sSwing 212 [90] vs. 156 [71.5] ms, p= 0.027; sRS 630 [195.5] vs. 530 [95] ms, p=0.028; **Figure 11B-C**). Hip and knee peak flexion, hip and knee range of motion were not significantly different between sides (P vs. NP; sHipP, 32 [8.16] vs. 35.84 [15.42] degree, p=0.176; sKneeP, 52.75 [14.91] vs. 59.43 [12.63] degree, p=0.069; sHipR, 43.55 [20.8] vs. 39.62 [13.02] degree, p=0.735; sKneeR, 34.88 [18.72] vs. 47.23 [15.45] degree, p=0.104; **Figure 12**).

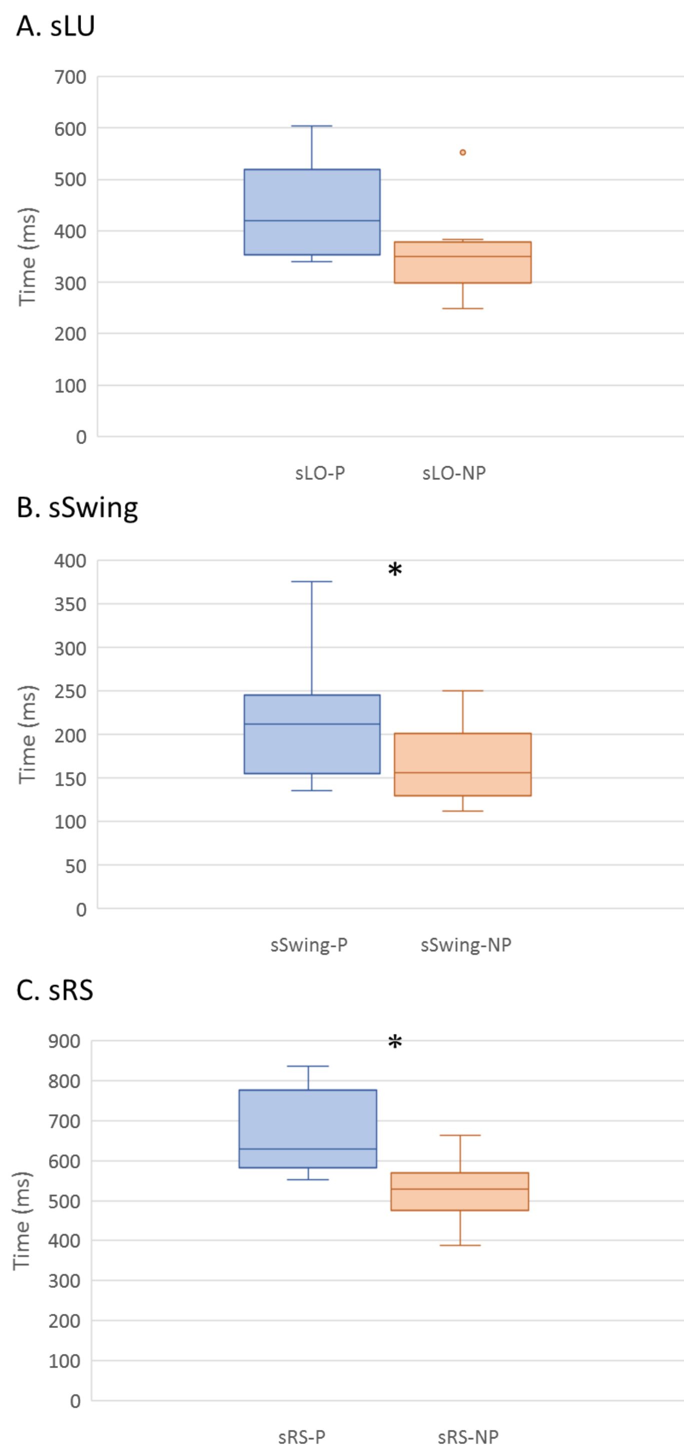


Figure 11. Temporal Variables from Static Perturbation Condition
(A) Duration from perturbation onset to the perturbed ipsilateral limb completely leaving the ground. (B) Swing phase of the recovery response. (C) Recovery step durations. The blue bars stand for tripping perturbations delivered to the prosthetic side; the orange bars stand for tripping perturbations delivered to the non-prosthetic side. Star sign indicates significant side to side difference.

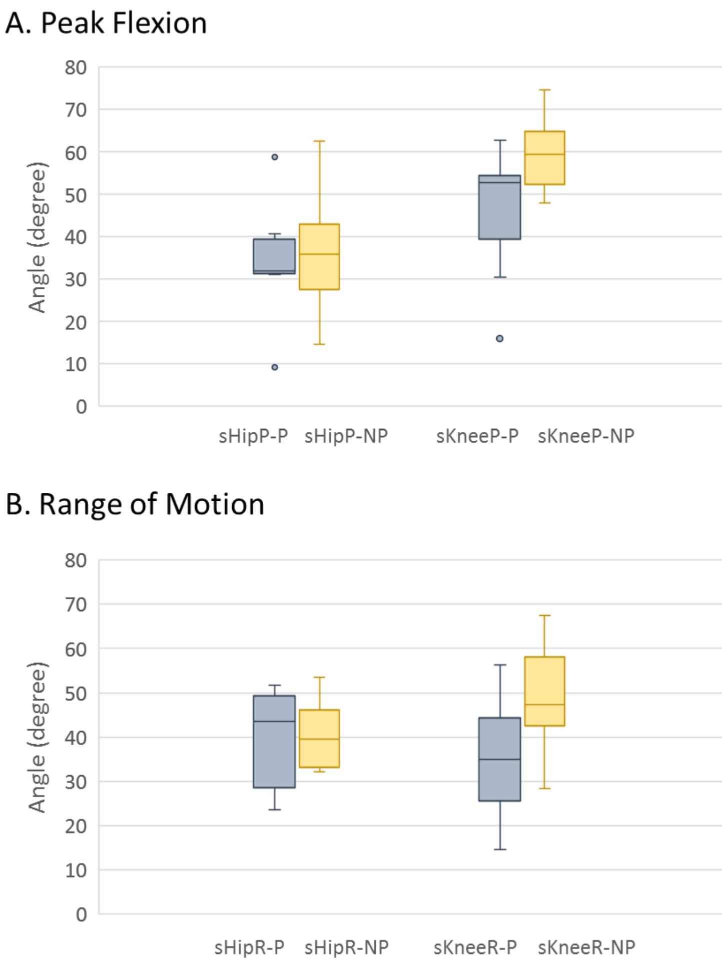


Figure 12. Kinematic Variables from Static Perturbation Condition
(A) Peak flexion angle of hip and knee during the recover step. (B) Range of motion of hip and knee during the recovery step. The limb of interests was the ipsilateral perturbed limb. The blue bars stand for tripping perturbations delivered to the prosthetic side; the yellow bars stand for tripping perturbations delivered to the non-prosthetic side.

Response to Dynamic Perturbation

A total of 3 falls occurred during the walking condition, including 1 fall in the P and 2 falls in the NP trials. The frequency of falls was not statistically significantly different between sides (χ^2=0.381, p=0.537). Comparing the stepping response timing characteristics between perturbed limbs, no significant difference was found in dLO (P vs. NP; 195.5 [127] vs. 222 [126.75] ms, p=0.754; **Figure 13A**). However, dRS1 and dSwing1 were significantly longer when the non-prosthetic limb was perturbed (P vs. NP; dRS1, 495.5 [100] vs. 556 [133.5] ms, p=0.011; dSwing1, 280.5 [57.25] vs. 352 [56] ms, p=0.009; **Figure 13B-C**). No difference in the second recovery step durations (P vs. NP; dRS2, 572.5 [75] vs. 538 [169.25] ms, p=0.147; dSwing2, 422 [85.75] vs. 348 [125] ms, p=0.346; **Figure 4D-E**) was detected. Lastly, there was no difference in terms of the total recovery duration (P vs. NP; 1092 [183.75] vs. 1100 [87] ms, p=0.724; **Figure 13F**). During dRS1, neither the peak flexions nor the range of motion of hip and knee were different side to side (P vs. NP; dHipP1, 32.03 [15.88] vs. 36.78 [18.89] degree, p=0.269; dKneeP1, 57.73 [15.66] vs. 56.88 [10.81] degree, p=0.648; dHipR1, 37.58 [15.09] vs. 36.57 [13.47] degree, p=0.347; dKneeR1, 48.61 [20.24] vs. 46.42 [14.76] degree, p=0.718; **Figure 14A-B**). Similar results were observed in dRS2, where none of the hip and knee kinematics were significantly different between sides (P vs. NP; dHipP2, 34.3 [11.14] vs. 26.6 [13.28] degree, p=0.075; dKneeP2, 62.45 [8.12] vs. 57.99 [21.87] degree, p=0.433; dHipR2, 35.93 [8.04] vs. 34.17 [17.97] degree, p=0.070; dKneeR2, 48.46 [10.51] vs. 52.99 [25.62] degree, p=0.388; **Figure 14C-D**).

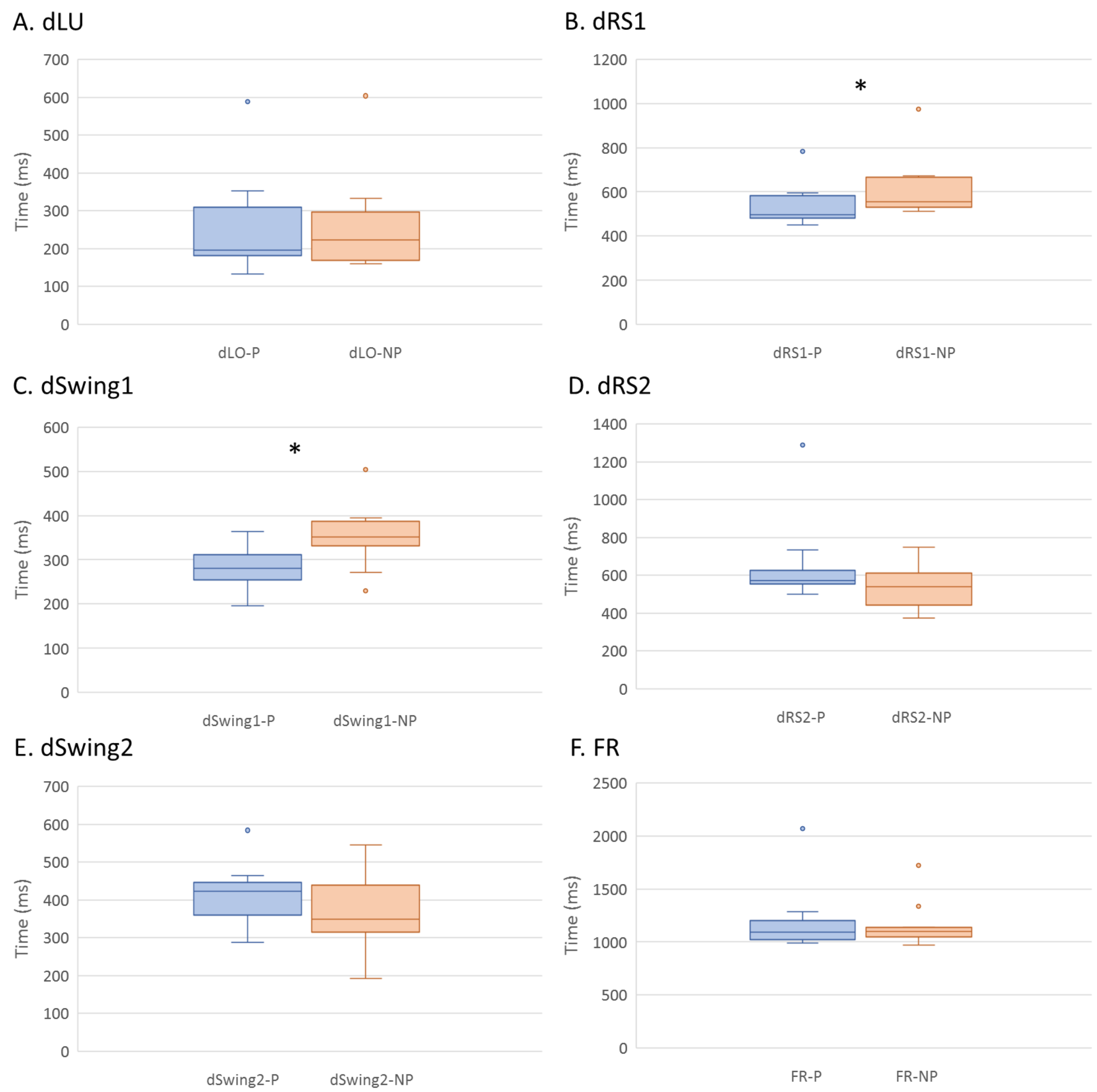


Figure 13. Temporal Variables from Dynamic Perturbation Condition
(A) Duration from perturbation onset to the contralateral limb completely leaving the ground. (B) Duration of the first recovery step. (C) Duration of the first swing phase. (D) Duration of the second recovery step. (E) Duration of the second swing phase. (F) Duration of the full course of recovery including two recovery steps. The blue bars stand for tripping perturbations delivered to the prosthetic side; the orange bars stand for tripping perturbations delivered to the non-prosthetic side. Star sign indicates significant side to side difference.

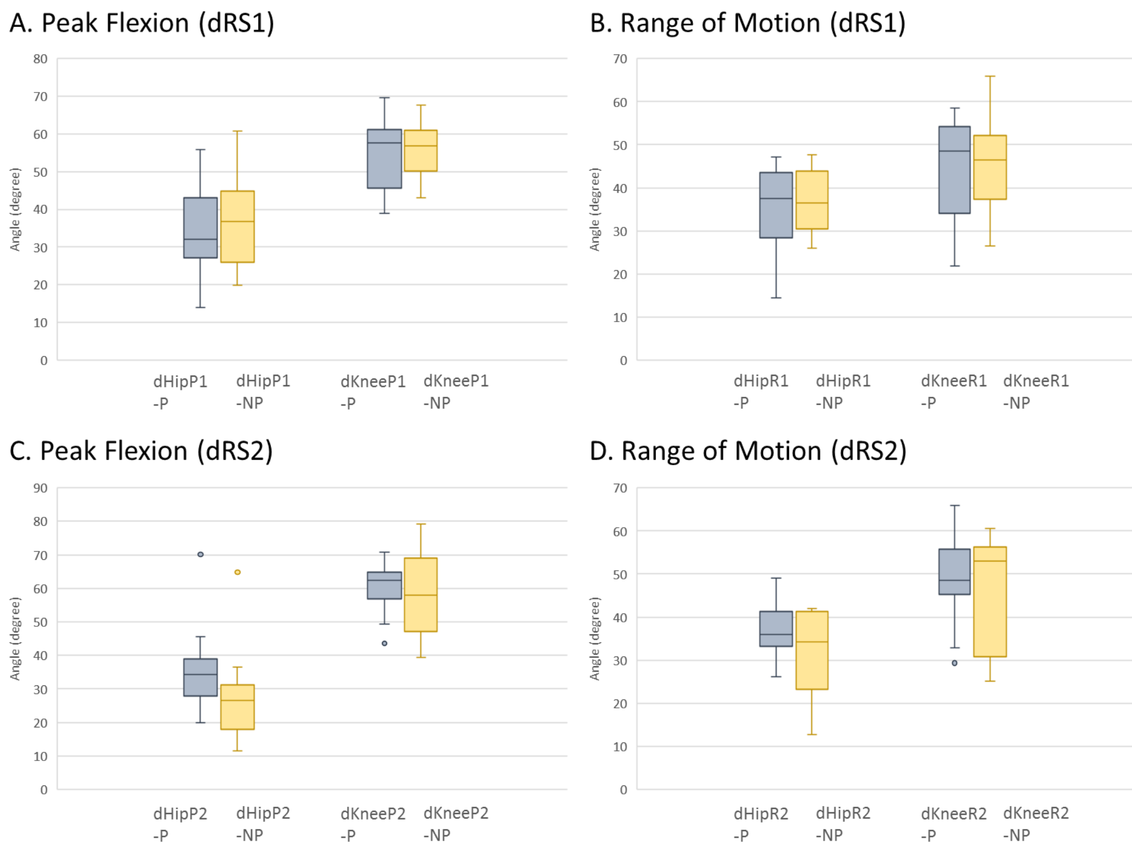


Figure 14. Kinematic Variables from Dynamic Perturbation Condition
(AB) Hip and knee peak flexion and range of motion during the first recovery step. The limb of interests was the contralateral limb. (CD) Hip and knee peak flexion and range of motion during the second recovery step. The limb of interests was the ipsilateral limb. The blue bars stand for tripping perturbations delivered to the prosthetic side; the yellow bars stand for tripping perturbations delivered to the non-prosthetic side.

Discussion

Our findings suggested that there are key biomechanical differences in fall recovery responses when tripping perturbations were delivered to the prosthetic or the non-prosthetic limb. Under the static perturbation condition, sRS was significantly longer when the perturbation was delivered to the prosthetic limb.

However, under the dynamic perturbation condition, dRS1 and dSwing1 were significantly longer when the non-prosthetic limb was perturbed. Our results indicated that fall recovery stepping response is not only different between prosthetic and non-prosthetic limbs in individuals with UTTA, the temporo-biomechanical differences are also context-dependent.

To our knowledge, our study was the first to examine fall recovery stepping reaction in response to static and dynamic perturbations in individuals with UTTA. Based on our observation, the gross recovery strategies used by individuals with UTTA were similar to individuals without amputation and dissimilar to individuals with transfemoral amputation.(Eng, Winter, & Patla, 1994; Shirota et al., 2015) During static standing perturbations, the limb that was perturbed was typically the side making a quick forward recovery step. During dynamic walking perturbations, the typical response was using the perturbed limb as a transient support when making a recovery step using the contralateral limb, followed by one more recovery step using the ipsilateral (perturbed) limb. Previous studies using a comparable perturbation apparatus (i.e. treadmill belt velocity change at early stance phase) have shown that this response strategy was exhibited by non-amputated young and older adults.(Golyski, Vazquez, Leestma, & Sawicki, 2022; Jensen, Brown, & Woollacott, 2001; Maki & McIlroy, 2006) However, in individuals with transfemoral amputation, special recovery strategies that do not require alternating steps such as skipping and hopping with the non-prosthetic limb have been observed.(Shirota et al., 2015) These unconventional and perhaps higher risk strategies were executed likely due to participant's inability to initiate the first recovery step with the prosthetic limb. In the current study, we did not observe such responses in our participants with UTTA regardless the limb they were perturbed. This highlights the need to understand how individuals with different levels of lower limb amputation may adopt different fall recovery strategies. This knowledge is important for developing individualized fall prevention and rehabilitation protocols after amputation.

Our findings also showed that in both standing and walking conditions, the recovery step duration is longer when the step was made by the prosthetic limb. For example, sRS measures the time between sensing the perturbation onset to completing a forward step during the static condition. Our results showed that sRS-P as executed by the prosthetic limb was significantly longer than sRS-NP. Similar results were found in the first recovery step in the dynamic conditions where dRS1-NP and dSwing1-NP were significantly longer when the contralateral (i.e. prosthetic) limb was used to make the recovery step. The abovementioned longer durations may be attributed to impaired sensorimotor function of the prosthetic limb, including dampened sensory detection and lack of active control of the ankle joint as suggested by Rusaw et al.(Rusaw et al., 2013)

One of the contributing factors to the generally longer fall recovery duration was motor impairment of the prosthetic limb. In both static and dynamic conditions, we divide the recovery step duration (sRS and dRS1) into two components, LO and Swing. The LO component involved the perturbation detection and reaction time whereas Swing was mostly motor execution. We observed significant differences between sides in sRS, sSwing, as well as in dRS1 and dSwing1. This suggested that the longer time it took to execute the recovery steps may explain the differences in duration of the fall recovery. We further explored the hip and knee joints excursions but did not find pronounced differences. However, it is possible that more time is needed for a prosthetic limb to cover a similar range than a non-prosthetic limb. The slower movement of the prosthetic limb may be of concern in tripping situations that involve a physical obstacle where the toe clearance, knee flexion, and hip flexion demand is higher. The slower knee and hip flexion may underly the observed higher risks of tripping falls in individuals with UTTA.

Other motor deficit has been investigated in the literature that may also contribute to the longer duration of the prosthetic limb. Even though the biological knees were preserved in individuals with UTTA, the knee on the prosthetic limb contributed less work than the knee on the non-prosthetic limb in both the eccentric (i.e. energy absorption) and concentric (i.e. energy generation) phases of gait(Prinsen, Nederhand, & Rietman, 2011), and thus may be weaker. The lack of active ankle control in the prosthetic limb caused reduced toe clearance in swing phase in this population.(Johnson, De Asha, Munjal, Kulkarni, & Buckley, 2014) In the non-prosthetic limb and individuals without amputation, minimum toe clearance was related to the peak forward velocity of the swing foot, and therefore increased with higher walking velocity.(Schulz, 2011) However, this velocity related change was absent in the prosthetic limb due to the lack of an articulated ankle in most current prosthetic designs.(De Asha & Buckley, 2015) Taken together, the prosthetic swing phase has been shown to be longer than the non-prosthetic swing during walking(Jaegers, Arendzen, & de Jongh, 1995; Varrecchia et al., 2019), which may also contribute to their higher fall risk.

In walking conditions, the differences in duration in the first step of recovery did not retain in the second step. In individuals without amputation, it has been reported that most individuals were able to arrest the forward momentum of trunk when the support limb (i.e. ipsilateral limb in the present study) performing the push-off, before the Ipsi Off.(Pijnappels et al., 2004) Hence, having the ability to execute the first recovery step in a timely manner might be key for successful fall recovery. The focus of future trip recovery research paradigm should be on the first recovery step, and particularly when this step was executed by the prosthetic limb.

While we believe that the stepping response observed and reported in the current study is typical of individuals with UTTA, these results might not be generalized to individuals who prefer special strategies such as skipping or hopping. In addition, the intensity of perturbation may alter the responses as well. Comparing to previous protocols, the accelerations that we used elicited a gentler perturbation than the ones in the literature due to the participants' characteristics. Our participants were older, less active, and walked slower than the military participants in previous studies. We believe more research on older adults with different levels of lower limb amputation is needed, due to the growing number of this population.

Chapter 4: Effects of a 2-Day Weight-Shift Balance Training Program on Biomechanics of Fall Recovery in Community Dwelling Individuals with Unilateral Transtibial Amputation

Abstract

When individuals were perturbed to fall, the fall recovery responses frequently involve making one or multiple steps. Individuals with unilateral transtibial amputation (UTTA) tend to rely on the non-prosthetic limb more and consistently bearing more weight on that limb. However, timely weight-shifting is a necessary ability underlying effective stepping response and fall arrest. Therefore, the current study examines the effect of a short-term weight-shifting training program and whether the learned skills transfer to altering fall recovery responses. Ten participants with UTTA completed the procedures consisting of a 2-day training with pre/post-training fall recovery evaluations. During training, we instructed the participants to practice 40 weight-shifting trials, 30 seconds each, while standing on an instrumented stabilometer. The goal of the task was keeping a balance platform horizontal for as long as possible. Balance performance was quantified by the duration that the balance platform was horizontal. Fall recovery response was evaluated by delivering unanticipated perturbations during static standing and dynamic walking conditions. During each condition, perturbations with small and large magnitudes were delivered to their prosthetic and non-prosthetic limb randomly for a total of 4 perturbations. Biomechanical variables extracted during perturbations were the duration to unload the stepping limb,

duration to complete the recovery step, and peak trunk flexion angle. Participants improved their weight-shifting balance performance after training by 17.9%. Moreover, the duration for unloading the prosthetic limb in static condition significantly shortened after training. When the prosthetic limb was perturbed, the first recovery step duration, and when the non-prosthetic limb was perturbed, the second recovery step duration both significantly lengthened after training. These were the durations of recovery steps where the prosthetic limb serving as the support limb in dynamic condition. No training effect was found on peak trunk flexion angle. Perturbation with greater magnitude required participants to complete the recovery in a shorter period of time. Our findings suggested that certain weight-shifting components important to the fall recovery response improved after the intervention, but the global trunk kinematics did not change. The task specificity should be considered when developing future training programs aimed at reducing fall risk in this population.

Introduction

Falls occur frequently after lower limb amputation (LLA). Individuals with LLA have been shown to fall more often than age-matched non-amputees.(Miller, Speechley, & Deathe, 2001) Previous studies have shown that up to 80% of community-dwelling persons living with LLA have experienced at least one fall each year.(Furtado et al., 2017; Ülger, Topuz, Bayramlar, Erbahçeci, & Sener, 2010; Wong, Chen, Blackwell, & Rahal, 2015) Documented risk factors that are predictive of falls in individuals with LLA include increased gait variability, poor balance, stump problems, back pain, and joint pain.(Hordacre, Barr, Patritti, & Crotty, 2015; Miller, Speechley, et al., 2001; Parker, Hanada, & Adderson, 2013)

Previous studies showed that majority of the falls in community-dwelling adults occur during walking.(Heijnen & Rietdyk, 2016; Talbot, Musiol, Witham, & Metter, 2005) An observational study of falls in real-life found that older residents in long-term care facilities fell mostly due to incorrect weight-shifting (41%) and tripping (21%) while traversing level ground.(Robinovitch et al., 2013) Individuals with LLA are at even higher risks for tripping falls due to the greater required toe clearance from prosthesis use.(Hunter, Higa, Frengopoulos, Viana, & Payne, 2020) Diminished and disrupted lower limb proprioception may also contribute to the increased risk.(Ülger et al., 2010) In a prospective study by Rosenblatt et al., the authors tracked the falls from 8 persons with transtibial amputation for one year, during which 5 of them reported multiple trips accumulating a total of 44 falls.(Rosenblatt, Bauer, & Grabiner, 2017) The authors also reported the most common circumstance where tripping occurs was walking on level ground, followed by walking on uneven surfaces.

A typical trip recovery requires a destabilized individual to make one or more alternating steps to regain balance (i.e. recovery steps) regardless of the recovery strategies utilized.(Eng, Winter, & Patla, 1994) To enable the reciprocal stepping pattern, having the ability to quickly shift weight between limbs (i.e. from the intact limb to the prosthetic limb and vis-versa) is a prerequisite to executing successful reactive responses to tripping perturbations. Persons with LLA tend to favor the non-prosthetic leg to stand, ambulate, and respond to perturbations.(Hlavackova, Franco, Diot, & Vuillerme, 2011; Rusaw, Hagberg, Nolan, & Ramstrand, 2013; Šlajpah, Kamnik, Burger, Bajd, & Munih, 2013; Tesio, Lanzi, & Detrembleur, 1998) The disuse motor patterns may be developed due to shift of body center of mass, pain, weakness, loss of sensory input, and general lack of confidence regarding the prosthetic limb.(Hlavackova et al., 2011; Rusaw et al., 2013; Šlajpah et al., 2013; Tesio et al., 1998) Previous research showed that response time in individuals with LLA is longer than age-matched non-amputees, when executing motor tasks that require the prosthetic leg to detect perturbations.(Rusaw et al., 2013) The slower stepping response may contribute to the less effective trip recovery in individuals with LLA. Hence, dampened weight-shifting ability, from a sensorimotor perspective, could be a root cause for failed trip recovery. Broadly speaking, such weight-shifting ability encompasses one's ability of bearing weight on the prosthetic limb before executing the subsequent fall arresting steps in a timely manner.(Pijnappels, Bobbert, & van Dieën, 2005) A previous study has showed that weight-shift balance training using an unstable platform is effective to improve postural control of single leg balance test in individuals with LLA.(Erbahçeci, Bayramlar, Şener, Bayar, & Ülger, 2001) However, it is currently unknown whether a training program designed to enhance side-to-side weight-shifting for individuals with LLA can lead to improved stepping response during a tripping event.

The purpose of this study was to investigate the effect of a 2-day weight-shifting balance training program using an unstable platform on balance performance. In addition, we examined whether such training effect can be transferred to enhance trip recovery responses in individuals with unilateral transtibial amputation (UTTA).

Methods

Participants

We recruited individuals with UTTA who were 18 years or older. They must have had and used their current prosthesis for more than 6 months and be able to walk independently for 5 minutes without assistive devices. Potential participants with health conditions that can influence their balance such as acute stump pain, open wound, vision impairment, and neurological conditions were excluded. Eligibility was screened by emails or phone calls prior to the visits. Potential participants who were eligible were invited to the Clinical Locomotion Neuromechanics Lab at the University of Nevada, Las Vegas (UNLV). Informed consents for participating in the study were obtained. All protocols and procedures were approved by the UNLV Institutional Review Board for Biomedical Research. The study was registered on ClinicalTrials.gov (NCT03989063).

Procedure

This training study involved three sessions: baseline evaluation on day 1, a total of 40 balance practice trials on days 1 and 2, and post-training evaluation on day 3 (Figure 1). On days 1 and 3, we examined the fall recovery responses to tripping perturbations during standing (static) and walking (dynamic) conditions. Participants' basic demographic information, mobility and fall-related behaviors were assessed through self-reported questionnaires. The fall-related surveys and amputee-specific mobility questionnaire included the Activity-specific Balance Confidence Scale (ABC), Fear of Falling Avoidance Behavior Questionnaire (FFABQ), and Prosthetic Limb Users Survey of Mobility (PLUS-M).(Hafner, Morgan, Askew, & Salem, 2016; Landers, Durand, Powell, Dibble, & Young, 2011; Mandel et al., 2016; Miller, Deathe, & Speechley, 2003) The post-training evaluation on day 3, which took place at least 24 hours after day 2 to allow learning consolidation, included a retention test of the balance training (see below for details).(Song, 2009) Changes in balance performance over the course of training and the retention test performance were used to determine the training effects. The trip recovery evaluation was conducted again on day 3 (**Figure 15**).

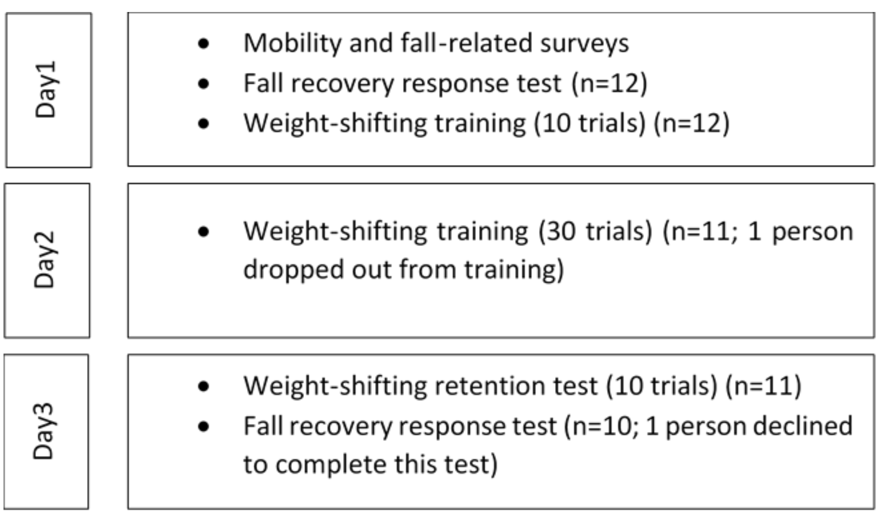


Figure 15. Study Testing and Training Schedule

The fall recovery response test was conducted with a 12-camera VICON motion capture system (Oxford Metrics, Oxfordshire, UK; sampling frequency = 250 Hz) and 1 video camera (sampling frequency = 125 Hz). A standardized lower extremity plus trunk marker set was used to capture kinematics of bilateral lower extremities (including the prosthesis) and the trunk, which was modeled as a single segment connected to the pelvis.(Shih et al., 2019)

<u>Weight-Shifting Balance Training</u>

The weight-shifting training was conducted using a Stabilometer (Lafayette Instruments, Model 16030; **Figure 16**). The maximal range of motion was 15° of tilt to each side centered around a free-moving hinge. Following a beeping sound from the device, the angular position of the platform was recorded over the 30-sec trial duration. The participants were instructed that the goal of the training task was to “keep level/horizontal for as long as possible”. After every practice block, which comprised of 5 trials, a researcher reported the averaged time-in-balance (out of 30 seconds) to the participant as feedback of performance. Time-in-balance was defined as the accumulated time during which the platform stayed within ±5° from horizontal. The participants practiced the task for 40 trials (8 blocks). The retention test on day 3 included 10 trials (2 blocks) without instruction and feedback. Self-efficacy was measured after the first block on day 1 and after retention test by a visual analog scale 1 to 10 asking how confident the participant was to keep themselves in balance on the stabilometer for 10, 15, 20, and 25 seconds.

Figure 16. The Stabilometer-Based Weight-Shifting Balance Task

Fall Recovery Response Test

Participants' fall recovery response to perturbations during standing and walking were assessed. Customized MATLAB (2017b, MathWorks Inc., Natick, MA, USA) scripts were used for precise delivering the prescribed perturbations. In both conditions, the participant wore a harness with a load cell to protect them from falling and to measure the amount of support that the harness provided. In the static perturbation condition, perturbations were delivered while participants stood still on an instrumented side-by-side split-belt treadmill (Model ITC-11-20L-4, Bertec Corp., Columbus, OH, USA). The perturbation was delivered as one of the treadmill belts moved posteriorly for 200 ms at two different magnitudes of accelerations which translates to standing surface displacements of 0.12 m and 0.24 m (small vs. large perturbations). The two magnitudes of perturbations were each applied randomly and unexpectedly to the prosthetic (small prosthetic, SP; large prosthetic, LP) and non-prosthetic legs (small non-prosthetic, SNP; large non-prosthetic, LNP). A perturbation was delivered when the researcher visually confirmed the readings from the force platforms that the participant was standing upright with the body weight evenly distributed between the legs.

The perturbation under a dynamic perturbation condition simulates a trip by modifying the protocol described by Sessoms et al.(Sessoms et al., 2014), and has been described and validated previously (Chapter 2). Briefly, the protocol began by establishing participants' comfortable walking speed (CWS) which was defined as the self-select walking speed during a 1-minute walking trial. The MATLAB customized script was designed to automatically apply the perturbation once the vertical ground reaction force (vGRF) exceeded 20-25% of body weight during early stance phase. During the perturbation, one of the belts, either left or right, decelerated first to simulation the obstruction of foot, then accelerated to induce the recovery responses such as trunk movement commonly seen in tripping, before returning to CWS.(Sessoms et al., 2014) Similar to the static perturbation condition, two magnitudes of acceleration were used to simulate small and large trips, so the four collected trials were also denoted as SP, LP, SNP, and LNP. The accelerations utilized in the protocol were linearly scaled by the CWS to accommodate the participants' different CWS. The timing, magnitude and side of perturbation was randomly selected and applied.

Data Extraction and Analysis

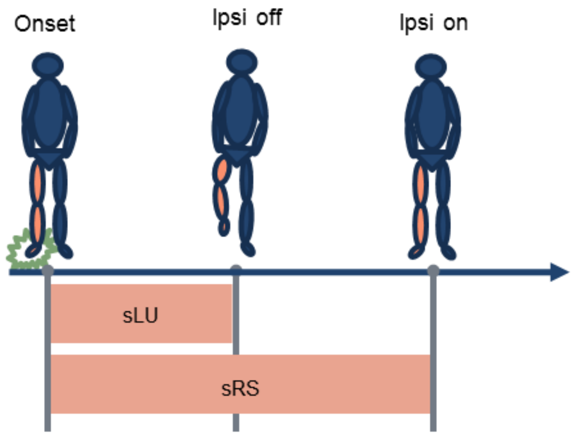
All fall response test trials were labeled in VICON Nexus before imported to Visual 3D (C-Motion, Rockville, MD, USA). Marker trajectories were low-pass filtered at 10 Hz using the Butterworth filter function in Visual 3D, before joint kinematics were computed. Trials when participants touched or grabbed the handrails or when the harness provided more than 50% of body weight support after the perturbations were marked as trials of unsuccessful recovery. The criteria applied to both standing and walking fall recovery response tests.

Participants typically utilized an ipsilateral stepping strategy to recover from perturbations encountered during standing (**Figure 17A**). Therefore, the duration spent to unload the ipsilateral limb (sLU) was computed as from the perturbation onset to the ipsilateral tripped limb leaving the ground; the duration spent to make the recovery step (sRS) was computed as from the perturbation onset to the ipsilateral tripped limb contacting the ground (**Figure 17A**). Peak trunk flexion angle (sTF) was extracted in the window from perturbation onset to the ipsilateral limb contacting the ground.(Kurtzer, 2015; Pruszynski & Scott, 2012; Sloot, van den Noort, van der Krogt, Bruijn, & Harlaar, 2015)

In the walking condition, perturbation onset timing was identified based on the heel marker and vGRF data. For all successful recovery trials in this study, the perturbation (i.e. to the ipsilateral limb) was followed by a first recovery step made by the contralateral limb, and a second recovery step with the ipsilateral limb (**Figure 17B**). Therefore, as shown in **Figure 17B**, the following sequential events were identified: perturbation onset (Onset), contralateral limb lifting off the ground

(Contra-Off), contralateral limb contact (Contra-On; first recovery step), ipsilateral limb lifting off the ground (Ipsi-Off), and the subsequent ipsilateral limb contact (Ipsi-On; second recovery step). A step contact was defined as when vGRF exceeded 40 N to account for the vGRF signal noise introduced by the moving treadmill belt. Based on these 5 time stamps, we calculated the following durations (**Figure 17B**): 1) limb unloading (dLU; from Onset to Contra-Off), 2) the first recovery step time (dRS1; from Onset to Contra-On), 3) the second recovery step time (dRS2; from Contra-On to Ipsi-On), and 4) the full recovery time (FR; from Onset to Ipsi-On). In addition, we identified the peak trunk flexion angle (dTF) from Onset to Ipsi-On for it has been reported as a significant indicator for the likelihood of tripping falls.(Owings, Pavol, & Grabiner, 2001; Pavol, Owings, Foley, & Grabiner, 2001) In order to account for individual variance in walking pattern and speed, the reported peak trunk flexion angle during fall recovery was normalized for each participant as the difference from their trunk angle during normal walking at their CWS. The peak trunk flexion angle during normal walking was obtained as an average value from three steps in each trial to account for outlying values. Timing of perturbation onset was directly exported from the MATLAB program.

A. Static perturbation condition



B. Dynamic perturbation condition

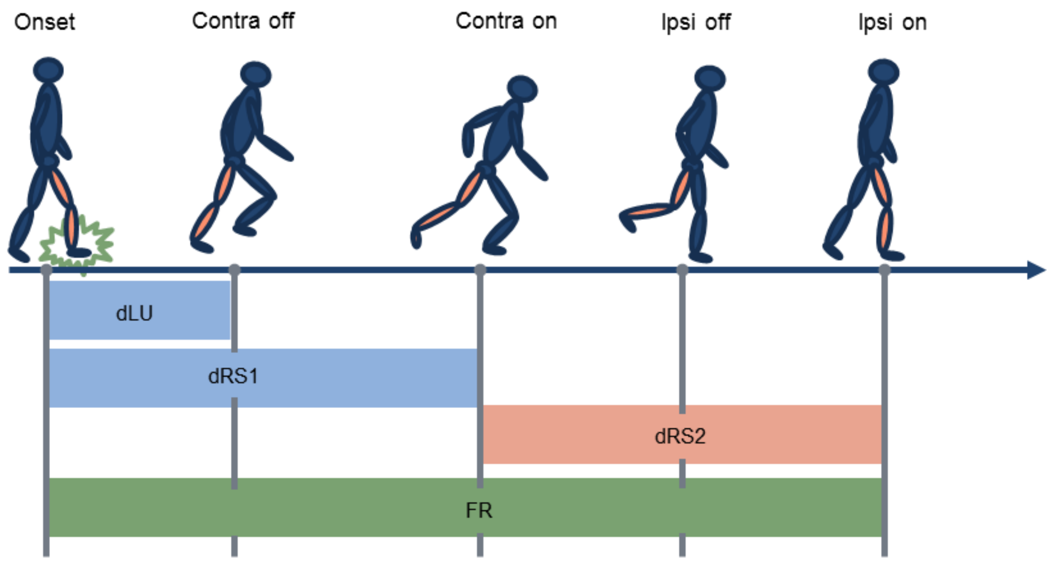


Figure 17. Definitions for Temporal Variables in Static and Dynamic Perturbation conditions

<u>Statistical Analysis</u>

Normality of all data was examined by skewness and kurtosis. Non-parametric tests were performed when normality assumption was violated.

Changes in balance performance during training was quantified. Paired t test was used to compare their balance performance and self-efficacy at retention relative to their baseline performance (first block of practice) to examine if there was significant improvement.(Chiviacowsky, Wulf, & Wally, 2010) Wilcoxon Signed Rank tests were used to examine the effects of training (pre vs. post) and the effect of perturbation magnitude (small acceleration vs. large perturbation) on the above-mentioned temporal and kinematic variables during fall recovery.(Pereira, Afonso, & Medeiros, 2015; Sheskin, 2003)

Results

Twelve persons participated in the study. One of them withdrew on day 2 and did not complete the training. Another withdrew on day 3 and did not complete the post-training evaluations (**Figure 15**). Participants' demographic information and ABC, FFABQ, PEQ-MS, PLUS-M scores on day 1 are presented in **Table 4**.

Table 4. Demographics and Scores of Fall-Related and Amputee-Specific Mobility Surveys

ID	Amputation side	Amputation cause	Sex	Age, year	Height, cm	BW, kg	CWS, m/s	ABC	FFABQ	PLUS-M
001	Left	Non-vascular	M	73	195.6	81.7	0.50	84.4	8	58.4
002	Right	Non-vascular	F	59	179.1	60.3	0.40	85.0	2	61.0
003	Right	Vascular	M	58	182.0	92.1	0.70	73.8	10	51.2
004	Left	Vascular	M	24	175.3	76.4	0.95	98.8	0	67.1
005	Left	Non-vascular	M	63	172.7	89.0	0.64	86.9	6	52.0
006 (drop)	Left	Non-vascular	M	57	182.9	109.7	0.50	65.9	2	51.2
007 (drop)	Right	Non-vascular	M	50	180.3	76.3	0.70	94.4	2	67.1
008	Left	Non-vascular	F	45	165.1	102.0	0.65	87.5	10	58.4
009	Right	Non-vascular	F	29	162.6	61.2	0.60	92.5	10	53.6
010	Right	Non-vascular	F	58	170.2	113.4	0.35	63.1	8	49.8
011	Left	Non-vascular	M	62	167.6	77.5	0.29	85.0	19	49.1
012	Right	Vascular	F	67	160.0	80.4	0.10	75.6	1	56.3
Mean	L/R 6/6	Vas/Nonvas	M/F	53.8	174.5	85.0	0.53	82.7	6.5	56.3
SD		3/9	7/5	14.7	10.2	17.0	0.23	11.0	5.5	6.3

BW: body weight; CWS: comfortable walking speed; ABC: Activity-specific Balance Confidence Scale, presented in percentage on 16 questions with a total possible score 0-1600, a higher ABC score indicated higher balance confidence; FFABQ: Fear of Falling Avoidance Behavior Questionnaire, presented in sum score on 14 questions with a total possible score 0-56, a higher FFABQ score indicated more avoidance behavior due to fear of falling; PLUS-M: Prosthetic Limb Users Survey of Mobility, presented in T-score with a total possible score 21.8-71.4, a higher PLUS-M score indicated greater mobility of the prosthetic user.

Weight-Shifting Training

A total of 11 participants completed the weight-shifting training. Their changes in balance performance over time were presented in **Figure 18**.

Stabilometer performance significantly improved from 6.6 seconds to 9.8 seconds after training (p=0.015). Self-efficacy, on the other hand, did not change significantly (pre-training vs. post-training, 13.4 vs. 15.8, p=0.334).

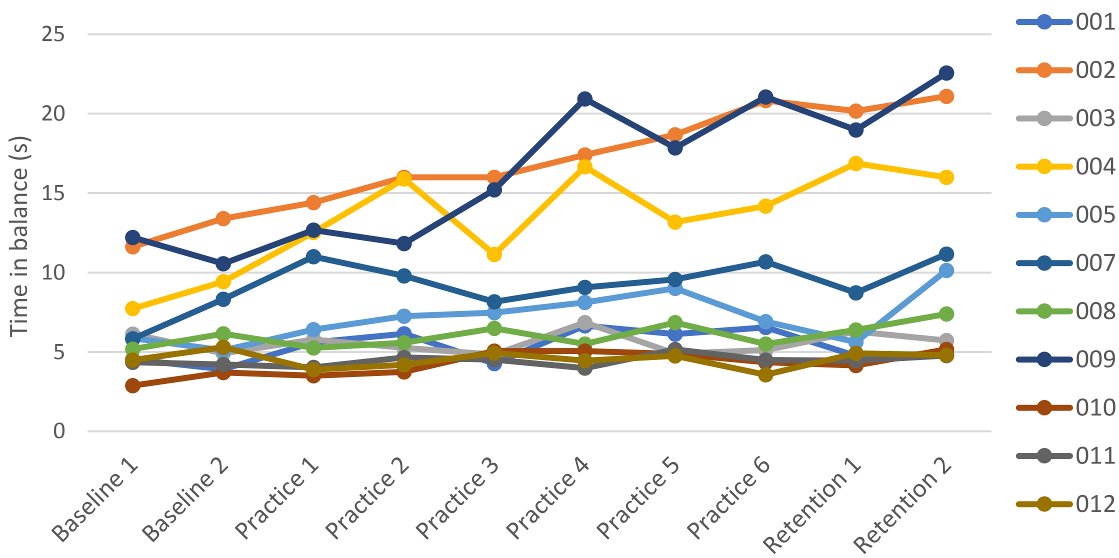


Figure 18. Changes in Balance Performance During the Weight-Shifting Training

Static Perturbation Condition

Before training, 5 (20.8%) and 4 (16.7%) trials were not successfully recovered at small and large perturbations respectively. After training, the unsuccessful recovery trials reduced to 2 (10%) at small perturbation and 3 (15%) at large perturbation (**Table 5**). Wilcoxon Signed Rank tests showed differences before and after training in unloading time (sLU) and step time (sRS) when the small perturbations were delivered to the prosthetic limb (**Appendix**). Nevertheless, we did not find such differences when the acceleration was large, nor when the perturbations were employed to the non-prosthetic limb (**Appendix**). Lastly, we did not observe training effect on peak trunk flexion (sTF) (**Appendix**). Large perturbation magnitude elicited shorter unloading time both when the prosthetic limb (pre-training, p=0.012; post-training, p=0.043) and non-prosthetic limb (pre-training, p=0.038; post-training, p=0.028) were perturbed. Additionally, larger perturbation elicited greater peak trunk flexion when the non-prosthetic limb was perturbed (pre-training, p=0.003; post-training, p=0.007).

<u>Dynamic Perturbation Condition</u>

Three (12.5%) unsuccessful recoveries during small perturbation and 6 (25%) during large perturbation occurred before training. After training, the numbers reduced to 1 (5%) and 4 (20%) during small and large perturbation, respectively (**Table 6**). First recovery step duration when the prosthetic limb (dRS1-P) was perturbed by small perturbation, second recovery step duration and full recovery duration when the non-prosthetic limb (dRS2-NP & FR-NP) was perturbed by large perturbation were significantly longer after training (**Appendix**). Other temporal variables on day 3 did not differ from day 1 (**Appendix**). Additionally, the peak trunk flexion angle (dTF) did not significantly change after training (**Appendix**). Comparing to small perturbation, the first recovery step durations (dRS1) were significantly shorter when the perturbation magnitude was large regardless of the perturbed limb (prosthetic, pre-training, p=0.004; post-training, $p=0.003$; non-prosthetic, pre-training, $p=0.008$; post-training, $p=0.005$). Similarly, large acceleration induced greater peak trunk flexion than small acceleration did (prosthetic, pre-training, $p=0.002$; post-training, $p=0.021$; non-prosthetic, pre-training, $p=0.005$; post-training, $p=0.003$).

Table 5. Success and Recovery Strategies Used in the Static Standing Condition

	Day 1				Day 3			
	SP	LP	SNP	LNP	SP	LP	SNP	LNP
001	Unsuccessful				Contralateral stepping			Non-stepping
002		Unsuccessful			Contralateral stepping	Contralateral stepping		
003								
004	Contralateral stepping				Contralateral stepping			
005	Unsuccessful	Unsuccessful			N/A	N/A	N/A	N/A
006			Unsuccessful		N/A	N/A	N/A	N/A
007					Contralateral stepping			
008								
009								
010	Contralateral stepping	Unsuccessful	Unsuccessful	Unsuccessful	Unsuccessful	Unsuccessful	Unsuccessful	Unsuccessful
011								
012			Unsuccessful			Unsuccessful		

Blank cells represent recovery via stepping strategy of the ipsilateral limb (i.e. the side that was perturbed). Unsuccessful: unsuccessful recovery; Non-stepping: recovery via a strategy that was not stepping; Contralateral stepping: recovery via stepping using the contralateral limb; N/A: not applicable

Table 6. Unsuccessful Recovery Trials in the Dynamic Walking Condition

	Day 1				Day 3			
	SP	LP	SNP	LNP	SP	LP	SNP	LNP
001	Unsuccessful	Unsuccessful	Unsuccessful	Unsuccessful	Unsuccessful	Unsuccessful		
002								
003								
004								
005					N/A	N/A	N/A	N/A
006					N/A	N/A	N/A	N/A
007								
008			Unsuccessful	Unsuccessful		Unsuccessful		Unsuccessful
009								
010		Unsuccessful		Unsuccessful				
011								Unsuccessful
012				Unsuccessful				

Blank cells represent recovery via lowering strategy. Unsuccessful: unsuccessful recovery; N/A: not applicable.

Discussion

The current study examined whether a short-term weight-shifting balance training can improve the balance performance and further change perturbation recovery responses in individuals with UTTA. The hypothesis was partially supported based on our results from a 2-day training program in 12 participants. The 2-day training significantly improved balance performance. Regarding the training effect, in static condition, we found shorter unloading time (sLU-P) and shorter recovery step time (sRS-P) when the prosthetic limb was disturbed with small perturbation; in dynamic condition, we found longer first recovery step time (dRS1-P) when the prosthetic limb was perturbed, and longer second recovery step time (dRS2-NP), longer full recovery time (FR-NP) when the non-prosthetic limb was perturbed. However, we did not find statistically significant differences in the rest of the temporal and kinematic variables before and after the weight-shifting balance training. Regarding the effect of perturbation magnitude, large perturbation induced shorter unloading time and greater trunk flexion in static condition. In dynamic condition, large perturbation elicited shorter first recovery step time and again, greater trunk flexion angle.

Participants in the present study learned the weight-shifting task after the 2-day training. Our participants increased the time that they were able to keep themselves in balance on the stabilometer. Similar findings were reported in individuals with UTTA that a 3-week balance training utilizing a balance board was superior than a conventional program which consisted of strengthening, postural exercise, and activities in parallel bars.(Erbahçeci et al., 2001) In the present study, we observed that the fall recovery responses were different after training. For example, in static perturbation condition, participants unloaded their prosthetic limb quicker when that limb was perturbed (sLU-P). Further, it required shorter time for the prosthetic limb to complete the recovery step (sRS-P) after training. In order to unload the prosthetic limb, participants needed to conduct weight-shifting, which is the component that we intended to train using the stabilometer. However, we did not find the other direction of weight-shifting (sLU-NP) changed after training. On the other hand, in dynamic perturbation condition, dRS1-P and dRS2-NP were significantly longer post-training. Noted that when conducting these two recovery steps, it was both the prosthetic limb served as the support limb to allow the non-prosthetic limb to step forward. Therefore, the longer duration may suggest that the prosthetic limb provided more stable support so there was more time for the non-prosthetic limb making the step. In the abovementioned study by Erbahceci et al., the effects of their balance training transferred to increased single leg stance time in eye open as well as eye-closed conditions.(Erbahçeci et al., 2001) Overall, the weight-shifting intervention altered some components in fall recovery responses. Moreover, the changed components were different when it came to static and dynamic perturbation conditions.

The selection of intervention and to what extent the effects of the intervention would transfer to different mobility functions need to be considered and examined. A recent meta-analysis favored exercise intervention (i.e. supervised balance, strengthening programs plus home exercise and education) over usual care in improving walking distance, speed, and endurance for individuals with amputation.(Abou, Fliflet, Zhao, Du, & Rice, 2022) In previous Clinical Practice Guidelines for post-amputation care, balance training has been recommended as a part of rehabilitation for patients with lower limb loss.(Webster et al., 2019) Current prosthetic rehabilitation programs typically include weight-bearing and balance training for individuals who are recovering from lower LLA.(Esquenazi & DiGiacomo, 2001; Hershkovitz, Dudkiewicz, & Brill, 2013; Munin et al., 2001; Ülger, Yıldırım Şahan, & Çelik, 2018; Wong et al., 2016) For example, Van de Meent and colleagues implemented a progressive loading program to gradually increase the amount of body weight bore by the osseointegration implant during the early stage after surgery to improve bone strength and single leg balance ability.(Van de Meent, Hopman, & Frölke, 2013) Rau, Bonvin and Bie used therapeutic exercises such as the rice planting position (i.e. squat with emphasis on bearing weight on the prosthetic leg) to encourage weight-bearing.(Rau, Bonvin, & De Bie, 2007) Wii Fit that includes weight-bearing and balance video games has also been utilized as an intervention.(Andrysek et al., 2012) Other balance training such as moving center of pressure to different directions during standing or remaining standing when on a transversely moving platform has also been examined in this population.(Damayanti Sethy, Kujur, & Sau, 2009; Matjacić & Burger, 2003)

Despite the abundance of training protocols, most of the prosthetic balance and fall prevention interventions involve relatively static tasks that lack the reactive movement elements that are paramount to effectively improve fall recovery responses. To our knowledge, this study is the first to examines whether the effect of weight-shifting balance training can be translated to improve fall recovery. Our findings showed that participants did improve their balance performance, but it did not significantly change the fall recovery trunk biomechanics which reflect the general likelihood of falls. Moreover, the shorter sLU, dRS1, and greater sTF, dTF elicited by the large acceleration indicated that it could be even more challenging when responding to the larger magnitude of perturbation as it required faster reactive responses and increased the likelihood of falls. To date, there is no consensus regarding how therapy and prosthetic training can reduce the risks of falls in the LLA population. Clemens et al. reviewed evidence and concluded that the optimal practice that could address this fall issue has not yet been established.(Clemens, Doerger, & Lee, 2020)

Task specificity could be a contributing factor to how much the training effect may carry, thereby task-specific training may be required. The training program we chose in this study focused on the voluntary weight-shifting ability between the two sides which may only contribute to a part of the reactive responses. Consequently, we observed training effect in the unloading duration as well as in the support that the prosthetic limb could provide. A task specific training may be necessary for the global improvement of reactive responses of the unexpected perturbations. Specific to prevent trip-related falls, Crenshaw et al. developed a 6-day compensatory-step training protocol and tested it on 5 participants with transfemoral or knee disarticulation amputation.(Crenshaw, Kaufman, & Grabiner, 2013) In this training, participants stood on a treadmill and were instructed to respond to unanticipated accelerations of the treadmill belt by making a few forward steps until the treadmill stopped. With practice, participants performed more successful recovery resulting in fewer falls and less hopping strategy used by the non-prosthetic limb. Trunk flexion angles significantly decreased when the recovery step was successfully made. Moreover, the distance between the center of mass and the first stepping limb increased, indicting larger and perhaps more effective stepping response for fall arrest. These two kinematic variables marked improvements in recovery response to the postural disturbance in lower limb prosthesis users. Another treadmill-based training program has also shown positive outcomes for young military service members with transtibial amputation.(Kaufman, Wyatt, Sessoms, & Grabiner, 2014) Reduced peak trunk flexion angle and trunk flexion velocity were found after training, indicating improved trip recovery response. This is one of the few studies showing the possibility of reducing fall risks and improving fall recovery response after amputation via a clinical training protocol.

In the broader field of fall research, the representativeness and specificity of measurements and variables have attracted increasing attention. A recent systematic review revealed that exercise interventions were effective in improving the population’s balance ability.(Abou et al., 2022) It was often claimed that better balance ability led to reduced risks of fall since such risks were usually indirectly measured by self-reported surveys or performance of certain mobility tasks.(Miller, Deathe, Speechley, & Koval, 2001; Steinberg, Gottlieb, Siev-Ner, & Plotnik, 2019) In contrast, we directly measured the participants’ recovery responses during standing and walking simulated fall scenarios. We believe this approach is superior in detecting true fall recovery performance, as it has been shown that measures of postural stability (i.e., maximum excursion and center of pressure velocity during the limit of stability tests) cannot predict the recovery responses to postural perturbations.(Owings, Pavol, Foley, & Grabiner, 2000) Researchers started to advocate the importance of developing and selecting the more sensitive and representative biomechanical variables for measuring the risks of falls during gait (i.e., trips, slips, and side falls) and for quantifying effectiveness of an intervention.(Grabiner & Kaufman, 2021) The efficacy and effectiveness of perturbation-based training and assessment in the limb loss population are not well understood, leaving room for future research.

The primary limitation of the present study is the dearth of a control group. Some of the findings may be due to that the participants became familiarized with the perturbation protocol, despite that the perturbations were delivered randomly and unexpectedly. It is also possible the effect of the weight-shift training was not recognized from our way of evaluation; it may be showing in other variables such as EMG timing measurement. The present study mapped out critical considerations for establishing the training paradigm that should be taken into account when designing future fall prevention studies for individuals with LLA.

Chapter 5: Executive Summary

Falls threaten people's health and wellbeing across various populations, and individuals with unilateral transtibial amputation are at an increased risk. In order to achieve our ultimate rehabilitation goal of reducing falls, researchers need to have a reliable and valid testing apparatus that can simulate the fall scenario in a realistic and reproducible manner. Utilizing such apparatus would allow us to investigate the biomechanics of falls as well as examining whether certain interventions may alter the fall recovery responses. This dissertation followed this fundamental premise by firstly developing a protocol that was able to precisely deliver unilateral perturbations in the desired phase during gait, using ground reaction force as reference. Taking advantages of the simulation protocol, we investigated the fall recovery responses of the prosthetic and non-prosthetic limbs to identify the key between-side biomechanical differences. In the 3rd study, we examined whether clinically feasible weight-shifting training program may be used to improve fall recover biomechanical responses.

Our primary finding was that the response duration for the prosthetic limb to execute the recovery step is generally longer, in both static standing and dynamic walking conditions. We further confirmed that the differences were mainly attributed to the longer time it takes to advance the prosthetic limb when making a recovery step. When the perturbation magnitude was bigger, it required even faster responses, and that pinpointed the importance of being able to perform a functional recovery step in a timely manner. The weight-shifting balance training was a novel task for our participants with unilateral transtibial amputation. The hypothesis that improving the weight-shifting ability with training can lead to a concurrent improvement in fall recovery response was partially supported. Considering the key elements of fall recovery stepping response, the training enhanced certain temporo-spatial performance. First, we found that participants were able to unload their prosthetic limb significantly faster after training. Second, recovery step time, where the support limb must bear the body weight long enough to allow effective stepping to take place, was also improved. Specifically, we found that after training when the prosthetic limb was used as the support limb, the stance time was significantly improved. Nevertheless, our balance training protocol consisted of only one exercise and should only be considered as a preliminary step to building a comprehensive intervention program.

For future interventions, seeking training programs that target the fast-reactive motor function of the prosthetic limb may be a means to change their fall recovery response and to help individuals with limb loss to handle static and dynamic perturbations. Each training component should have its specific goals. Thus, depending on the purpose of intervention, we can choose to emphasize certain components of the fall recovery responses (i.e. timing, stepping magnitude, confidence in bearing weight on the prosthetic limb…etc.) by taking a task-oriented approach that might improve the patient's global fall recovery responses. In summary, I believe the three studies improved our understanding of the fall recovery responses of individuals with transtibial amputation and brought useful tools and ideas to the fall assessment and intervention paradigm for this clinical population.

Appendix

Dependent Variables from Standing and Walking Conditions

	Perturbation Side	Acceleration	Pre-Training	Post-Training	p Value (Pre vs. Post)
sLU, ms	Prosthetic	Small	420.0 (165.0)	358.0 (133.0)*	0.028
		Large	299.5 (188.5)†	301.5 (166.8)†	0.575
	Non-prosthetic	Small	350.0 (79.0)	365.0 (124.0)	0.090
		Large	280.0 (112.0)†	262.0 (43.0)†	0.093
sRS, ms	Prosthetic	Small	630.0 (195.5)	608.0 (75.5)*	0.039
		Large	608.0 (284.0)	557.5 (64.0)	0.528
	Non-prosthetic	Small	530.0 (95.0)	512.0 (185.5)	0.575
		Large	512.0 (88.0)	527.5(105.0)	0.779
sTF, degree	Prosthetic	Small	24.0 (13.7)	30.8 (19.5)	0.059
		Large	25.3 (7.3)	27.3 (17.6)	0.203
	Non-prosthetic	Small	19.7 (11.1)	22.4 (12.6)	0.878
		Large	29.7 (11.3)†	29.6 (14.1)†	0.508
dLU, ms	Prosthetic	Small	195.5 (127)	230 (144)	0.074
		Large	215.5 (122)	264 (90)	0.444
	Non-prosthetic	Small	222 (126.8)	224 (114)	0.262
		Large	186 (108.8)	230 (159)	0.126
dRS1, ms	Prosthetic	Small	495.5 (100)	520 (132)*	0.017
		Large	464 (65.5)	488 (88)	0.386
	Non-prosthetic	Small	556 (133.5)	545 (101)	0.767
		Large	518 (131.5)	535 (88)	0.799
dRS2, ms	Prosthetic	Small	572.5 (75)	595 (236)	0.799
		Large	514 (262.5)	528 (105)	0.799
	Non-prosthetic	Small	538 (169.3)	552 (224)	0.683
		Large	416.5 (212)	505 (216)*	0.028
FR, ms	Prosthetic	Small	1092 (183.8)	1100 (203)	0.332
		Large	988 (335)	1036 (140)	0.475
	Non-prosthetic	Small	1100 (87)	1084 (229)	0.646
		Large	900 (329.5)†	1032 (235)*	0.047

dTF, degree	Prosthetic	Small	4.1 (5.5)	7.6 (6.1)	0.445
		Large	12.4 (6.7)†	11.8 (18.6)†	0.333
	Non-prosthetic	Small	3.6 (5.3)	4.5 (6.9)	0.114
		Large	11.4 (9.2)†	13 (9)†	0.285

Values are presented in Median (Interquartile range) unless otherwise specified. sRS: the duration from perturbation onset to the ipsilateral limb contacting the ground in trials using an ipsilateral stepping strategy to recover; sTF: the peak trunk flexion angle in standing conditions. dLU: the duration from perturbation onset to the contralateral limb lifting off the ground. dRS1: the duration from perturbation onset to the contralateral limb contacting the ground. dRS2: the duration from contralateral limb contacting the ground to the ipsilateral limb contacting the ground. FR: the duration from perturbation onset to the ipsilateral limb contacting the ground. dTF: the peak trunk flexion angle in walking conditions. Star signs indicate statistically significant differences between pre- and post-training. Cross signs indicate statistically significant differences between small and large accelerations.

References

Chapter 1

Andrysek, J., Klejman, S., Steinnagel, B., Torres-Moreno, R., Zabjek, K. F., Salbach, N. M., & Moody, K. (2012). Preliminary evaluation of a commercially available videogame system as an adjunct therapeutic intervention for improving balance among children and adolescents with lower limb amputations. Archives of physical medicine and rehabilitation, 93(2), 358-366.

Arena, S. L., Davis, J. L., Grant, J. W., & Madigan, M. L. (2016). Tripping elicits earlier and larger deviations in linear head acceleration compared to slipping. PloS one, 11(11).

Aziz, O., Park, E. J., Mori, G., & Robinovitch, S. N. (2014). Distinguishing the causes of falls in humans using an array of wearable tri-axial accelerometers. Gait & posture, 39(1), 506-512.

Bütefisch, C. M., Davis, B. C., Wise, S. P., Sawaki, L., Kopylev, L., Classen, J., & Cohen, L. G. (2000). Mechanisms of use-dependent plasticity in the human motor cortex. Proceedings of the national academy of sciences, 97(7), 3661-3665.

Baltich, J., von Tscharner, V., Zandiyeh, P., & Nigg, B. M. (2014). Quantification and reliability of center of pressure movement during balance tasks of varying difficulty. Gait & posture, 40(2), 327-332.

Bellmann, M., Schmalz, T., & Blumentritt, S. (2010). Comparative biomechanical analysis of current microprocessor-controlled prosthetic knee joints. Archives of physical medicine and rehabilitation, 91(4), 644-652.

Berger, W., Dietz, V., & Quintern, J. (1984). Corrective reactions to stumbling in man: neuronal co-ordination of bilateral leg muscle activity during gait. The Journal of physiology, 357(1), 109-125.

Beurskens, R., Wilken, J. M., & Dingwell, J. B. (2014). Dynamic stability of individuals with transtibial amputation walking in destabilizing environments. Journal of biomechanics, 47(7), 1675-1681.

Blumentritt, S., Schmalz, T., & Jarasch, R. (2009). The safety of C-leg: biomechanical tests. JPO: Journal of Prosthetics and Orthotics, 21(1), 2-15.

Bolger, D., Ting, L. H., & Sawers, A. (2014). Individuals with transtibial limb loss use interlimb force asymmetries to maintain multi-directional reactive balance control. Clinical Biomechanics, 29(9), 1039-1047.

Bramati, I. E., Rodrigues, E. C., Simões, E. L., Melo, B., Höfle, S., Moll, J., . . . Tovar-Moll, F. (2019). Lower limb amputees undergo long-distance plasticity in sensorimotor functional connectivity. Scientific reports, 9(1), 1-10.

Chen, R., Corwell, B., Yaseen, Z., Hallett, M., & Cohen, L. G. (1998). Mechanisms of cortical reorganization in lower-limb amputees. Journal of Neuroscience, 18(9), 3443-3450.

Claret, C. R., Herget, G. W., Kouba, L., Wiest, D., Adler, J., von Tscharner, V., . . . Pasluosta, C. (2019). Neuromuscular adaptations and sensorimotor integration

following a unilateral transfemoral amputation. Journal of neuroengineering and rehabilitation, 16(1), 115.
Crenshaw, J. R., Kaufman, K. R., & Grabiner, M. D. (2013a). Compensatory-step training of healthy, mobile people with unilateral, transfemoral or knee disarticulation amputations: A potential intervention for trip-related falls. Gait & posture, 38(3), 500-506.
Crenshaw, J. R., Kaufman, K. R., & Grabiner, M. D. (2013b). Trip recoveries of people with unilateral, transfemoral or knee disarticulation amputations: Initial findings. Gait & posture, 38(3), 534-536.
Curtze, C., Hof, A. L., Postema, K., & Otten, B. (2012). The relative contributions of the prosthetic and sound limb to balance control in unilateral transtibial amputees. Gait & posture, 36(2), 276-281.
Darter, B. J., Nielsen, D. H., Yack, H. J., & Janz, K. F. (2013). Home-based treadmill training to improve gait performance in persons with a chronic transfemoral amputation. Archives of physical medicine and rehabilitation, 94(12), 2440-2447.
Dietz, V., Quintern, J., Boos, G., & Berger, W. (1986). Obstruction of the swing phase during gait: phase-dependent bilateral leg muscle coordination. Brain research, 384(1), 166-169.
Eng, J. J., Winter, D. A., & Patla, A. E. (1994). Strategies for recovery from a trip in early and late swing during human walking. Experimental brain research, 102(2), 339-349.
Floyer-Lea, A., Wylezinska, M., Kincses, T., & Matthews, P. M. (2006). Rapid modulation of GABA concentration in human sensorimotor cortex during motor learning. Journal of neurophysiology, 95(3), 1639-1644.
Forner-Cordero, A., Ackermann, M., & de Lima Freitas, M. (2011). A method to simulate motor control strategies to recover from perturbations: Application to a stumble recovery during gait. Paper presented at the 2011 Annual International Conference of the IEEE Engineering in Medicine and Biology Society.
Forner-Cordero, A., van der Helm, F. C., Koopman, H., & Duysens, J. (2015). Recovery response latencies to tripping perturbations during gait decrease with practice. Paper presented at the 2015 37th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC).
Fukagawa, N. K., & Schultz, A. B. (1995). Muscle function and mobility biomechanics in the elderly: an overview of some recent research. The Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 50(Special_Issue), 60-63.
Furtado, S., Briggs, T., Fulton, J., Russell, L., Grimer, R., Wren, V., . . . Gerrand, C. (2017). Patient experience after lower extremity amputation for sarcoma in England: a national survey. Disability and rehabilitation, 39(12), 1171-1190.
Gauthier-Gagnon, C. (1986). Augmented sensory feedback in the early training of standing balance of below-knee amputees. Physiother Can, 38, 137-142.
Gauthier-Gagnon, C., Grisé, M.-C., & Potvin, D. (1999). Enabling factors related to prosthetic use by people with transtibial and transfemoral amputation. Archives of physical medicine and rehabilitation, 80(6), 706-713.
Geertzen, J., Martina, J., & Rietman, H. (2001). Lower limb amputation part 2: Rehabilitation-A 10 year literature review. Prosthetics and orthotics international, 25(1), 14-20.

Geurts, A., Mulder, T., Nienhuis, B., & Rijken, R. (1992). Postural reorganization following lower limb amputation. Possible motor and sensory determinants of recovery. Scandinavian journal of rehabilitation medicine, 24(2), 83-90.

Grabiner, M. D., Feuerbach, J. W., & Jahnigen, D. W. (1996). Measures of paraspinal muscle performance do not predict initial trunk kinematics after tripping. Journal of biomechanics, 29(6), 735-744.

Grabiner, M. D., & Jahnigen, D. (1992). Modeling recovery from stumbles: preliminary data on variable selection and classification efficacy. Journal of the American Geriatrics Society, 40(9), 910-913.

Grabiner, M. D., Koh, T. J., Lundin, T. M., & Jahnigen, D. W. (1993). Kinematics of recovery from a stumble. Journal of gerontology, 48(3), M97-M102.

Hak, L., van Dieën, J. H., van der Wurff, P., Prins, M. R., Mert, A., Beek, P. J., & Houdijk, H. (2013). Walking in an unstable environment: strategies used by transtibial amputees to prevent falling during gait. Archives of physical medicine and rehabilitation, 94(11), 2186-2193.

Haridas, C., Zehr, E. P., & Misiaszek, J. E. (2008). Adaptation of cutaneous stumble correction when tripping is part of the locomotor environment. Journal of neurophysiology, 99(6), 2789-2797.

Heijnen, M. J. H., & Rietdyk, S. (2016). Falls in young adults: Perceived causes and environmental factors assessed with a daily online survey. Human movement science, 46, 86-95.

Hendershot, B. D., Bazrgari, B., & Nussbaum, M. A. (2013). Persons with unilateral lower-limb amputation have altered and asymmetric trunk mechanical and neuromuscular behaviors estimated using multidirectional trunk perturbations. Journal of biomechanics, 46(11), 1907-1912.

Hershkovitz, A., Dudkiewicz, I., & Brill, S. (2013). Rehabilitation outcome of post-acute lower limb geriatric amputees. Disability and rehabilitation, 35(3), 221-227.

Hlavackova, P., Franco, C., Diot, B., & Vuillerme, N. (2011). Contribution of each leg to the control of unperturbed bipedal stance in lower limb amputees: new insights using entropy. PloS one, 6(5).

Hordacre, B., Barr, C., & Crotty, M. (2015). Community activity and participation are reduced in transtibial amputee fallers: a wearable technology study. BMJ Innovations, 1(1), 10-16.

Hordacre, B., Bradnam, L. V., Barr, C., Patritti, B. L., & Crotty, M. (2014). Ipsilateral corticomotor excitability is associated with increased gait variability in unilateral transtibial amputees. European Journal of Neuroscience, 40(2), 2454-2462.

Hordacre, B., Bradnam, L. V., Barr, C., Patritti, B. L., & Crotty, M. (2015). Intracortical inhibition is modulated by phase of prosthetic rehabilitation in transtibial amputees. Frontiers in human neuroscience, 9, 276.

Hordacre, B. G., Barr, C., Patritti, B. L., & Crotty, M. (2015). Assessing gait variability in transtibial amputee fallers based on spatial-temporal gait parameters normalized for walking speed. Archives of physical medicine and rehabilitation, 96(6), 1162-1165.

Hunter, S. W., Higa, J., Frengopoulos, C., Viana, R., & Payne, M. W. (2018). Evaluating knowledge of falls risk factors and falls prevention strategies among lower extremity amputees after inpatient prosthetic rehabilitation: a prospective study. Disability and rehabilitation, 1-10.

Kahle, J., Highsmith, M., & Hubbard, S. (2008). Comparison of non-microprocessor prosthetic knee mechanisms and the C-Leg in the PEQ, stumbles and falls, walking tests, stair descent, and knee preference. J Rehabil Res Dev, 45(1), 1-13.

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.

King, S. T., Eveld, M. E., Martínez, A., Zelik, K. E., & Goldfarb, M. (2019). A novel system for introducing precisely-controlled, unanticipated gait perturbations for the study of stumble recovery. Journal of neuroengineering and rehabilitation, 16(1), 69.

Lee, B.-C., Martin, B. J., Thrasher, T. A., & Layne, C. S. (2016). The effect of vibrotactile cuing on recovery strategies from a treadmill-induced trip. IEEE transactions on neural systems and rehabilitation engineering, 25(3), 235-243.

Lee, J. K., Robinovitch, S. N., & Park, E. J. (2014). Inertial sensing-based pre-impact detection of falls involving near-fall scenarios. IEEE transactions on neural systems and rehabilitation engineering, 23(2), 258-266.

Matjacić, Z., & Burger, H. (2003). Dynamic balance training during standing in people with trans-tibial amputation: a pilot study. Prosthetics and orthotics international, 27(3), 214-220.

Miller, W. C., Deathe, A. B., Speechley, M., & Koval, J. (2001). The influence of falling, fear of falling, and balance confidence on prosthetic mobility and social activity among individuals with a lower extremity amputation. Archives of physical medicine and rehabilitation, 82(9), 1238-1244.

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.

Molina-Rueda, F., Navarro-Fernández, C., Cuesta-Gómez, A., Alguacil-Diego, I. M., Molero-Sánchez, A., & Carratalá-Tejada, M. (2019). Neuroplasticity Modifications Following a Lower-Limb Amputation: A Systematic Review. PM&R, 11(12), 1326-1334.

Mundell, B., Maradit Kremers, H., Visscher, S., Hoppe, K., & Kaufman, K. (2017). Direct medical costs of accidental falls for adults with transfemoral amputations. Prosthetics and orthotics international, 41(6), 564-570.

Munin, M. C., Espejo-De Guzman, M. C., Boninger, M. L., Fitzgerald, S. G., Penrod, L. E., & Singh, J. (2001). Predictive factors for successful early prosthetic ambulation among lower-limb amputees. J Rehabil Res Dev, 38(4), 379-384.

Murdoch, G., & Bennett-Wilson, A. (1996). Amputation: surgical practice and patient management: CRC Press.

Owings, T. M., Pavol, M. J., & Grabiner, M. D. (2001). Mechanisms of failed recovery following postural perturbations on a motorized treadmill mimic those associated with an actual forward trip. Clinical Biomechanics, 16(9), 813-819.

Özyürek, S., Demirbüken, İ., & Angın, S. (2014). Altered movement strategies in sit-to-stand task in persons with transtibial amputation. Prosthetics and orthotics international, 38(4), 303-309.

Parker, K., Hanada, E., & Adderson, J. (2013). Gait variability and regularity of people with transtibial amputations. Gait & posture, 37(2), 269-273.
Pasluosta, C., Hannink, J., Gaßner, H., Von Tscharner, V., Winkler, J., Klucken, J., & Eskofier, B. (2018). Motor output complexity in Parkinson's disease during quiet standing and walking: analysis of short-term correlations using the entropic half-life. Human movement science, 58, 185-194.
Pasluosta, C. F., Steib, S., Klamroth, S., Gaßner, H., Goßler, J., Hannink, J., . . . Klucken, J. (2017). Acute neuromuscular adaptations in the postural control of patients with Parkinson's disease after perturbed walking. Frontiers in aging neuroscience, 9, 316.
Pauley, T., Devlin, M., & Heslin, K. (2006). Falls sustained during inpatient rehabilitation after lower limb amputation: prevalence and predictors. American journal of physical medicine & rehabilitation, 85(6), 521-532.
Pavol, M. J., Owings, T. M., Foley, K. T., & Grabiner, M. D. (2001). Mechanisms leading to a fall from an induced trip in healthy older adults. The Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 56(7), M428-M437.
Pavol, M. J., Owings, T. M., Foley, K. T., & Grabiner, M. D. (2002). Influence of lower extremity strength of healthy older adults on the outcome of an induced trip. Journal of the American Geriatrics Society, 50(2), 256-262.
Pijnappels, M., Bobbert, M. F., & van Dieën, J. H. (2004). Contribution of the support limb in control of angular momentum after tripping. Journal of biomechanics, 37(12), 1811-1818.
Pijnappels, M., Bobbert, M. F., & van Dieën, J. H. (2005). Push-off reactions in recovery after tripping discriminate young subjects, older non-fallers and older fallers. Gait & posture, 21(4), 388-394.
Pijnappels, M., Kingma, I., Wezenberg, D., Reurink, G., & Van Dieën, J. H. (2010). Armed against falls: the contribution of arm movements to balance recovery after tripping. Experimental brain research, 201(4), 689-699.
Potocanac, Z., Pijnappels, M., Verschueren, S., van Dieën, J., & Duysens, J. (2016). Two-stage muscle activity responses in decisions about leg movement adjustments during trip recovery. Journal of neurophysiology, 115(1), 143-156.
Potocanac, Z., Smulders, E., Pijnappels, M., Verschueren, S., & Duysens, J. (2015). Response inhibition and avoidance of virtual obstacles during gait in healthy young and older adults. Human movement science, 39, 27-40.
Prinsen, E. C., Nederhand, M. J., & Rietman, J. S. (2011). Adaptation strategies of the lower extremities of patients with a transtibial or transfemoral amputation during level walking: a systematic review. Archives of physical medicine and rehabilitation, 92(8), 1311-1325.
Rau, B., Bonvin, F., & De Bie, R. (2007). Short-term effect of physiotherapy rehabilitation on functional performance of lower limb amputees. Prosthetics and orthotics international, 31(3), 258-270.
Richman, J. S., & Moorman, J. R. (2000). Physiological time-series analysis using approximate entropy and sample entropy. American Journal of Physiology-Heart and Circulatory Physiology, 278(6), H2039-H2049.

Robinovitch, S. N., Feldman, F., Yang, Y., Schonnop, R., Leung, P. M., Sarraf, T., . . . Loughin, M. (2013). Video capture of the circumstances of falls in elderly people residing in long-term care: an observational study. The Lancet, 381(9860), 47-54.

Roos, P. E., McGuigan, M. P., Kerwin, D. G., & Trewartha, G. (2008). The role of arm movement in early trip recovery in younger and older adults. Gait & posture, 27(2), 352-356.

Roos, P. E., McGuigan, M. P., & Trewartha, G. (2010). The role of strategy selection, limb force capacity and limb positioning in successful trip recovery. Clinical Biomechanics, 25(9), 873-878.

Rosenblatt, N. J., Bauer, A., & Grabiner, M. D. (2017). Relating minimum toe clearance to prospective, self-reported, trip-related stumbles in the community. Prosthetics and orthotics international, 41(4), 387-392.

Rossignaud, R., Oliveira, A. C., Lara, J. P., Mayor, J. J., & Rodacki, A. L. (2019). Methodological tools used for tripping gait analysis of elderly and prosthetic limb users: a systematic review. Aging clinical and experimental research, 1-8.

Rougier, P. R., & Bergeau, J. (2009). Biomechanical analysis of postural control of persons with transtibial or transfemoral amputation. American journal of physical medicine & rehabilitation, 88(11), 896-903.

Roy, G., Nadeau, S., Gravel, D., Malouin, F., McFadyen, B. J., & Piotte, F. (2006). The effect of foot position and chair height on the asymmetry of vertical forces during sit-to-stand and stand-to-sit tasks in individuals with hemiparesis. Clinical Biomechanics, 21(6), 585-593.

Rusaw, D., Hagberg, K., Nolan, L., & Ramstrand, N. (2013). Bilateral electromyogram response latency following platform perturbation in unilateral transtibial prosthesis users: Influence of weight distribution and limb position. Journal of rehabilitation research and development, 50(4), 531-544.

Sanderson, D. J., & Martin, P. E. (1997). Lower extremity kinematic and kinetic adaptations in unilateral below-knee amputees during walking. Gait & posture, 6(2), 126-136.

Schillings, A., Mulder, T., & Duysens, J. (2005). Stumbling over obstacles in older adults compared to young adults. Journal of neurophysiology, 94(2), 1158-1168.

Schillings, A., Van Wezel, B., & Duysens, J. (1996). Mechanically induced stumbling during human treadmill walking. Journal of neuroscience methods, 67(1), 11-17.

Schillings, A., Van Wezel, B., Mulder, T., & Duysens, J. (1999). Widespread short-latency stretch reflexes and their modulation during stumbling over obstacles. Brain research, 816(2), 480-486.

Schillings, A., Van Wezel, B., Mulder, T., & Duysens, J. (2000). Muscular responses and movement strategies during stumbling over obstacles. Journal of neurophysiology, 83(4), 2093-2102.

Schulz, B. W. (2017). A new measure of trip risk integrating minimum foot clearance and dynamic stability across the swing phase of gait. Journal of biomechanics, 55, 107-112.

Sessoms, P. H., Wyatt, M., Grabiner, M., Collins, J.-D., Kingsbury, T., Thesing, N., & Kaufman, K. (2014). Method for evoking a trip-like response using a treadmill-based perturbation during locomotion. Journal of biomechanics, 47(1), 277-280.

Sethy, D. M., Kujur, E. S., & Sau, K. (2009). Effect of balance exercise on balance control in unilateral lower limb amputees. Indian Journal of Occupational Therapy, 41(3), 63-68.

Sheehan, R. C., Beltran, E. J., Dingwell, J. B., & Wilken, J. M. (2015). Mediolateral angular momentum changes in persons with amputation during perturbed walking. Gait & posture, 41(3), 795-800.

Sheehan, R. C., Rábago, C. A., Rylander, J. H., Dingwell, J. B., & Wilken, J. M. (2016). Use of perturbation-based gait training in a virtual environment to address mediolateral instability in an individual with unilateral transfemoral amputation. Physical therapy, 96(12), 1896-1904.

Sheehan, T. P., & Gondo, G. C. (2014). Impact of limb loss in the United States. Physical Medicine and Rehabilitation Clinics, 25(1), 9-28.

Shirota, C., Simon, A. M., & Kuiken, T. A. (2014). Trip recovery strategies following perturbations of variable duration. Journal of biomechanics, 47(11), 2679-2684.

Shirota, C., Simon, A. M., & Kuiken, T. A. (2015). Transfemoral amputee recovery strategies following trips to their sound and prosthesis sides throughout swing phase. Journal of neuroengineering and rehabilitation, 12(1), 79.

Šlajpah, S., Kamnik, R., Burger, H., Bajd, T., & Munih, M. (2013). Asymmetry in sit-to-stand movement in patients following transtibial amputation and healthy individuals. International Journal of Rehabilitation Research, 36(3), 275-283.

Stagg, C. J., Bachtiar, V., & Johansen-Berg, H. (2011). The role of GABA in human motor learning. Current Biology, 21(6), 480-484.

Steinberg, N., Gottlieb, A., Siev-Ner, I., & Plotnik, M. (2019). Fall incidence and associated risk factors among people with a lower limb amputation during various stages of recovery–a systematic review. Disability and rehabilitation, 41(15), 1778-1787.

Stelmach, G. E., & Worringham, C. J. (1985). Sensorimotor deficits related to postural stability: implications for falling in the elderly. Clinics in geriatric medicine, 1(3), 679-694.

Talbot, L. A., Musiol, R. J., Witham, E. K., & Metter, E. J. (2005). Falls in young, middle-aged and older community dwelling adults: perceived cause, environmental factors and injury. BMC public health, 5(1), 86.

Tegenthoff, M., Witscher, K., Schwenkreis, P., & Liepert, J. (1999). Pharmacological modulation of training-induced plastic changes in human motor cortex. Electroencephalography and clinical neurophysiology. Supplement, 51, 188-196.

Tesio, L., Lanzi, D., & Detrembleur, C. (1998). The 3-D motion of the centre of gravity of the human body during level walking. II. Lower limb amputees. Clinical Biomechanics, 13(2), 83-90.

Ülger, Ö., Topuz, S., Bayramlar, K., Erbahçeci, F., & Sener, G. (2010). Risk factors, frequency, and causes of falling in geriatric persons who has had a limb removed by amputation. Topics in Geriatric Rehabilitation, 26(2), 156-163.

van den Bogert, A. J., Pavol, M., & Grabiner, M. D. (2002). Response time is more important than walking speed for the ability of older adults to avoid a fall after a trip. Journal of biomechanics, 35(2), 199-205.

Van der Burg, J., Pijnappels, M., & van Dieen, J. H. (2007). The influence of artificially increased trunk stiffness on the balance recovery after a trip. Gait & posture, 26(2), 272-278.
Vanicek, N., Strike, S., McNaughton, L., & Polman, R. (2009a). Gait patterns in transtibial amputee fallers vs. non-fallers: Biomechanical differences during level walking. Gait & posture, 29(3), 415-420.
Vanicek, N., Strike, S., McNaughton, L., & Polman, R. (2009b). Postural responses to dynamic perturbations in amputee fallers versus nonfallers: a comparative study with able-bodied subjects. Archives of physical medicine and rehabilitation, 90(6), 1018-1025.
Varma, P., Stineman, M. G., & Dillingham, T. R. (2014). Epidemiology of limb loss. Physical Medicine and Rehabilitation Clinics, 25(1), 1-8.
Winter, D. A., & Sienko, S. E. (1988). Biomechanics of below-knee amputee gait. Journal of biomechanics, 21(5), 361-367.
Wong, C. K., Chen, C. C., Blackwell, W. M., & Rahal, R. T. (2015). Balance ability measured with the Berg Balance Scale: a determinant of fall history in community-dwelling adults with leg amputation. Journal of rehabilitation medicine, 47(1), 80-86.
Yu, J. C., Lam, K., Nettel-Aguirre, A., Donald, M., & Dukelow, S. (2010). Incidence and risk factors of falling in the postoperative lower limb amputee while on the surgical ward. PM&R, 2(10), 926-934.
Zhang, F., D'Andrea, S. E., Nunnery, M. J., Kay, S. M., & Huang, H. (2011). Towards design of a stumble detection system for artificial legs. IEEE transactions on neural systems and rehabilitation engineering, 19(5), 567-577.
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.
Ziemann, U., Muellbacher, W., Hallett, M., & Cohen, L. G. (2001). Modulation of practice-dependent plasticity in human motor cortex. Brain, 124(6), 1171-1181.

Chapter 2

Arena, S. L., Davis, J. L., Grant, J. W., & Madigan, M. L. (2016). Tripping elicits earlier and larger deviations in linear head acceleration compared to slipping. PLoS One, 11(11).
Aviles, J., Allin, L. J., Alexander, N. B., Van Mullekom, J., Nussbaum, M. A., & Madigan, M. L. (2019). Comparison of treadmill trip-like training versus Tai Chi to improve reactive balance among independent older adult residents of senior housing: a pilot controlled trial. The Journals of Gerontology: Series A, 74(9), 1497-1503.
Berg, W. P., Alessio, H. M., Mills, E. M., & Tong, C. (1997). Circumstances and consequences of falls in independent community-dwelling older adults. Age and ageing, 26(4), 261-268.
Berger, W., Dietz, V., & Quintern, J. (1984). Corrective reactions to stumbling in man: neuronal co-ordination of bilateral leg muscle activity during gait. The Journal of physiology, 357(1), 109-125.

Bourke, A. K., O'Donovan, K. J., & Olaighin, G. (2008). The identification of vertical velocity profiles using an inertial sensor to investigate pre-impact detection of falls. Medical Engineering & Physics, 30(7), 937-946.

Crenshaw, J. R., Kaufman, K. R., & Grabiner, M. D. (2013a). Compensatory-step training of healthy, mobile people with unilateral, transfemoral or knee disarticulation amputations: A potential intervention for trip-related falls. Gait & Posture, 38(3), 500-506.

Crenshaw, J. R., Kaufman, K. R., & Grabiner, M. D. (2013b). Trip recoveries of people with unilateral, transfemoral or knee disarticulation amputations: Initial findings. Gait & posture, 38(3), 534-536.

Debelle, H., Harkness-Armstrong, C., Hadwin, K., Maganaris, C. N., & O'Brien, T. D. (2020). Recovery from a forward falling slip: measurement of dynamic stability and strength requirements using a split-belt instrumented treadmill. Frontiers in sports and active living, 2.

Dietz, V., Quintern, J., Boos, G., & Berger, W. (1986). Obstruction of the swing phase during gait: phase-dependent bilateral leg muscle coordination. Brain research, 384(1), 166-169.

Eng, J. J., Winter, D. A., & Patla, A. E. (1994). Strategies for recovery from a trip in early and late swing during human walking. Experimental brain research, 102(2), 339-349.

Fukagawa, N. K., & Schultz, A. B. (1995). Muscle function and mobility biomechanics in the elderly: an overview of some recent research. The Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 50(Special_Issue), 60-63.

Grabiner, M. D., Bareither, M. L., Gatts, S., Marone, J., & Troy, K. L. (2012). Task-specific training reduces trip-related fall risk in women. Medicine and science in sports and exercise, 44(12), 2410-2414. doi:10.1249/MSS.0b013e318268c89f

Grabiner, M. D., Donovan, S., Bareither, M. L., Marone, J. R., Hamstra-Wright, K., Gatts, S., & Troy, K. L. (2008). Trunk kinematics and fall risk of older adults: translating biomechanical results to the clinic. Journal of Electromyography and Kinesiology, 18(2), 197-204.

Grabiner, M. D., Feuerbach, J. W., & Jahnigen, D. W. (1996). Measures of paraspinal muscle performance do not predict initial trunk kinematics after tripping. Journal of biomechanics, 29(6), 735-744.

Grabiner, M. D., & Jahnigen, D. (1992). Modeling recovery from stumbles: preliminary data on variable selection and classification efficacy. Journal of the American Geriatrics Society, 40(9), 910-913.

Grabiner, M. D., Koh, T. J., Lundin, T. M., & Jahnigen, D. W. (1993). Kinematics of recovery from a stumble. Journal of gerontology, 48(3), M97-M102.

Heijnen, M. J. H., & Rietdyk, S. (2016). Falls in young adults: Perceived causes and environmental factors assessed with a daily online survey. Human movement science, 46, 86-95.

Ilmane, N., Croteau, S., & Duclos, C. (2015). Quantifying dynamic and postural balance difficulty during gait perturbations using stabilizing/destabilizing forces. Journal of biomechanics, 48(3), 441-448.

King, S. T., Eveld, M. E., Martínez, A., Zelik, K. E., & Goldfarb, M. (2019). A novel system for introducing precisely-controlled, unanticipated gait perturbations for the study of stumble recovery. Journal of neuroengineering and rehabilitation, 16(1), 69.

Koo, T. K., & Li, M. Y. (2016). A Guideline of Selecting and Reporting Intraclass Correlation Coefficients for Reliability Research. Journal of Chiropractic Medicine, 15(2), 155-163. doi:https://doi.org/10.1016/j.jcm.2016.02.012

Lee, A., Bhatt, T., & Pai, Y. C. (2016). Generalization of treadmill perturbation to overground slip during gait: effect of different perturbation distances on slip recovery. Journal of biomechanics, 49(2), 149-154.

Lee, B. C., Martin, B. J., Thrasher, T. A., & Layne, C. S. (2016). The effect of vibrotactile cuing on recovery strategies from a treadmill-induced trip. IEEE Transactions on Neural Systems and Rehabilitation Engineering, 25(3), 235-243.

Li, W., Keegan, T. H., Sternfeld, B., Sidney, S., Quesenberry Jr, C. P., & Kelsey, J. L. (2006). Outdoor falls among middle-aged and older adults: a neglected public health problem. American journal of public health, 96(7), 1192-1200.

Liu, J., & Lockhart, T. E. (2014). Trunk Angular Kinematics During Slip-Induced Backward Falls and Activities of Daily Living. Journal of Biomechanical Engineering, 136(10). doi:10.1115/1.4028033

Masani, K., Kouzaki, M., & Fukunaga, T. (2002). Variability of ground reaction forces during treadmill walking. Journal of Applied Physiology, 92(5), 1885-1890. doi:10.1152/japplphysiol.00969.2000

Nashner, L. M. (1976). Adapting reflexes controlling the human posture. Experimental brain research, 26(1), 59-72. doi:10.1007/bf00235249

Nashner, L. M. (1977). Fixed patterns of rapid postural responses among leg muscles during stance. Exp Brain Res, 30(1), 13-24. doi:10.1007/bf00237855

Nashner, L. M. (1980). Balance adjustments of humans perturbed while walking. Journal of Neurophysiology, 44(4), 650-664. doi:10.1152/jn.1980.44.4.650

Owings, T. M., Pavol, M. J., & Grabiner, M. D. (2001). Mechanisms of failed recovery following postural perturbations on a motorized treadmill mimic those associated with an actual forward trip. Clinical Biomechanics, 16(9), 813-819.

Pai, Y.-C., Maki, B., Iqbal, K., McIlroy, W., & Perry, S. (2000). Thresholds for step initiation induced by support-surface translation: a dynamic center-of-mass model provides much better prediction than a static model. Journal of biomechanics, 33(3), 387-392.

Painter, J. A., Elliott, S. J., & Hudson, S. (2009). Falls in community-dwelling adults aged 50 years and older prevalence and contributing factors. Journal of Allied Health, 38(4), 201-207.

Patel, P., & Bhatt, T. (2015). Adaptation to large-magnitude treadmill-based perturbations: improvements in reactive balance response. Physiological reports, 3(2), e12247.

Pavol, M. J., Owings, T. M., Foley, K. T., & Grabiner, M. D. (2001). Mechanisms leading to a fall from an induced trip in healthy older adults. The Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 56(7), M428-M437.

Pijnappels, M., Kingma, I., Wezenberg, D., Reurink, G., & van Dieën, J. H. (2010). Armed against falls: the contribution of arm movements to balance recovery after tripping. Experimental brain research, 201(4), 689-699. doi:10.1007/s00221-009-2088-7

Potocanac, Z., Pijnappels, M., Verschueren, S., van Dieën, J., & Duysens, J. (2016). Two-stage muscle activity responses in decisions about leg movement adjustments during trip recovery. Journal of Neurophysiology, 115(1), 143-156.
Robinovitch, S. N., Feldman, F., Yang, Y., Schonnop, R., Leung, P. M., Sarraf, T., . . . Loughin, M. (2013). Video capture of the circumstances of falls in elderly people residing in long-term care: an observational study. The Lancet, 381(9860), 47-54.
Schillings, A., Mulder, T., & Duysens, J. (2005). Stumbling over obstacles in older adults compared to young adults. Journal of Neurophysiology, 94(2), 1158-1168.
Schulz, B. W. (2017). A new measure of trip risk integrating minimum foot clearance and dynamic stability across the swing phase of gait. Journal of biomechanics, 55, 107-112. doi:https://doi.org/10.1016/j.jbiomech.2017.02.024
Sessoms, P. H., Wyatt, M., Grabiner, M., Collins, J.-D., Kingsbury, T., Thesing, N., & Kaufman, K. (2014). Method for evoking a trip-like response using a treadmill-based perturbation during locomotion. Journal of biomechanics, 47(1), 277-280.
Shirota, C., Simon, A. M., & Kuiken, T. A. (2014). Trip recovery strategies following perturbations of variable duration. Journal of biomechanics, 47(11), 2679-2684.
Stelmach, G. E., & Worringham, C. J. (1985). Sensorimotor deficits related to postural stability: implications for falling in the elderly. Clinics in geriatric medicine, 1(3), 679-694.
Szturm, T., & Fallang, B. (1998). Effects of varying acceleration of platform translation and toes-up rotations on the pattern and magnitude of balance reactions in humans. Journal of Vestibular Research, 8(5), 381-397.
Tokuno, C. D., Cresswell, A. G., Thorstensson, A., & Carpenter, M. G. (2010). Age-related changes in postural responses revealed by support-surface translations with a long acceleration–deceleration interval. Clinical Neurophysiology, 121(1), 109-117.
Yoo, D., Seo, K.-H., & Lee, B.-C. (2019). The effect of the most common gait perturbations on the compensatory limb's ankle, knee, and hip moments during the first stepping response. Gait & posture, 71, 98-104.
Zhang, F., D'Andrea, S. E., Nunnery, M. J., Kay, S. M., & Huang, H. (2011). Towards design of a stumble detection system for artificial legs. IEEE Transactions on Neural Systems and Rehabilitation Engineering, 19(5), 567-577.

Chapter 3

Bellmann, M., Schmalz, T., & Blumentritt, S. (2010). Comparative biomechanical analysis of current microprocessor-controlled prosthetic knee joints. Archives of physical medicine and rehabilitation, 91(4), 644-652.
Bolger, D., Ting, L. H., & Sawers, A. (2014). Individuals with transtibial limb loss use interlimb force asymmetries to maintain multi-directional reactive balance control. Clinical Biomechanics, 29(9), 1039-1047.
Cieza, A., Causey, K., Kamenov, K., Hanson, S. W., Chatterji, S., & Vos, T. (2020). Global estimates of the need for rehabilitation based on the Global Burden of Disease study 2019: a systematic analysis for the Global Burden of Disease Study 2019. The Lancet, 396(10267), 2006-2017.

Crenshaw, J. R., Kaufman, K. R., & Grabiner, M. D. (2013). Trip recoveries of people with unilateral, transfemoral or knee disarticulation amputations: Initial findings. Gait & posture, 38(3), 534-536.

Curtze, C., Hof, A. L., Postema, K., & Otten, B. (2012). The relative contributions of the prosthetic and sound limb to balance control in unilateral transtibial amputees. Gait & posture, 36(2), 276-281.

De Asha, A. R., & Buckley, J. G. (2015). The effects of walking speed on minimum toe clearance and on the temporal relationship between minimum clearance and peak swing-foot velocity in unilateral trans-tibial amputees. Prosthetics and orthotics international, 39(2), 120-125.

Eng, J. J., Winter, D. A., & Patla, A. E. (1994). Strategies for recovery from a trip in early and late swing during human walking. Experimental brain research, 102(2), 339-349.

Engsberg, J., Lee, A., Tedford, K., & Harder, J. (1993). Normative ground reaction force data for able-bodied and below-knee-amputee children during walking. Journal of pediatric orthopedics, 13(2), 169-173.

Engsberg, J. R., Lee, A. G., Patterson, J. L., & Harder, J. A. (1991). External loading comparisons between able-bodied and below-knee-amputee children during walking. Archives of physical medicine and rehabilitation, 72(9), 657-661.

Golyski, P. R., Vazquez, E., Leestma, J. K., & Sawicki, G. S. (2022). Onset timing of treadmill belt perturbations influences stability during walking. Journal of biomechanics, 130, 110800. doi:https://doi.org/10.1016/j.jbiomech.2021.110800

Jaegers, S. M. H. J., Arendzen, J. H., & de Jongh, H. J. (1995). Prosthetic gait of unilateral transfemoral amputees: A kinematic study. Archives of physical medicine and rehabilitation, 76(8), 736-743. doi:https://doi.org/10.1016/S0003-9993(95)80528-1

Jensen, J. L., Brown, L. A., & Woollacott, M. H. (2001). Compensatory stepping: the biomechanics of a preferred response among older adults. Exp Aging Res, 27(4), 361-376. doi:10.1080/03610730109342354

Johnson, L., De Asha, A. R., Munjal, R., Kulkarni, J., & Buckley, J. G. (2014). Toe clearance when walking in people with unilateral transtibial amputation: effects of passive hydraulic ankle.

Landers, M. R., Durand, C., Powell, D. S., Dibble, L. E., & Young, D. L. (2011). Development of a scale to assess avoidance behavior due to a fear of falling: the Fear of Falling Avoidance Behavior Questionnaire. Physical therapy, 91(8), 1253-1265.

Maki, B. E., & McIlroy, W. E. (2006). Control of rapid limb movements for balance recovery: age-related changes and implications for fall prevention. Age Ageing, 35 Suppl 2, ii12-ii18. doi:10.1093/ageing/afl078

Mandel, A., Paul, K., Paner, R., Devlin, M., Dilkas, S., & Pauley, T. (2016). Balance confidence and activity of community-dwelling patients with transtibial amputation. Journal of Rehabilitation Research & Development, 53(5).

Miller, W. C., Deathe, A. B., & Speechley, M. (2003). Psychometric properties of the Activities-specific Balance Confidence Scale among individuals with a lower-limb amputation. Archives of physical medicine and rehabilitation, 84(5), 656-661.

Nolan, L., Wit, A., Dudziñski, K., Lees, A., Lake, M., & Wychowañski, M. (2003). Adjustments in gait symmetry with walking speed in trans-femoral and trans-tibial amputees. Gait & posture, 17(2), 142-151.

Pijnappels, M., Bobbert, M. F., & van Dieën, J. H. (2004). Contribution of the support limb in control of angular momentum after tripping. Journal of biomechanics, 37(12), 1811-1818. doi:https://doi.org/10.1016/j.jbiomech.2004.02.038

Prinsen, E. C., Nederhand, M. J., & Rietman, J. S. (2011). Adaptation strategies of the lower extremities of patients with a transtibial or transfemoral amputation during level walking: a systematic review. Archives of physical medicine and rehabilitation, 92(8), 1311-1325.

Rosenblatt, N. J., Bauer, A., Rotter, D., & Grabiner, M. D. (2014). Active dorsiflexing prostheses may reduce trip-related fall risk in people with transtibial amputation. J Rehabil Res Dev, 51(8), 1229-1242.

Rougier, P. R., & Bergeau, J. (2009). Biomechanical analysis of postural control of persons with transtibial or transfemoral amputation. American journal of physical medicine & rehabilitation, 88(11), 896-903.

Rusaw, D., Hagberg, K., Nolan, L., & Ramstrand, N. (2013). Bilateral electromyogram response latency following platform perturbation in unilateral transtibial prosthesis users: Influence of weight distribution and limb position. Journal of rehabilitation research and development, 50(4), 531-544.

Schulz, B. W. (2011). Minimum toe clearance adaptations to floor surface irregularity and gait speed. Journal of biomechanics, 44(7), 1277-1284.

Sessoms, P. H., Wyatt, M., Grabiner, M., Collins, J.-D., Kingsbury, T., Thesing, N., & Kaufman, K. (2014). Method for evoking a trip-like response using a treadmill-based perturbation during locomotion. Journal of biomechanics, 47(1), 277-280.

Sheehan, R. C., Beltran, E. J., Dingwell, J. B., & Wilken, J. M. (2015). Mediolateral angular momentum changes in persons with amputation during perturbed walking. Gait & posture, 41(3), 795-800.

Shih, H.-T., Teng, H.-L., Gray, C., Poggemiller, M., Tracy, I., & Lee, S.-P. (2019). Four weeks of training with simple postural instructions changes trunk posture and foot strike pattern in recreational runners. Physical Therapy in Sport, 35, 89-96.

Shirota, C., Simon, A. M., & Kuiken, T. A. (2015). Transfemoral amputee recovery strategies following trips to their sound and prosthesis sides throughout swing phase. Journal of neuroengineering and rehabilitation, 12(1), 1-11.

Tesio, L., Lanzi, D., & Detrembleur, C. (1998). The 3-D motion of the centre of gravity of the human body during level walking. II. Lower limb amputees. Clinical Biomechanics, 13(2), 83-90.

Ülger, Ö., Topuz, S., Bayramlar, K., Erbahçeci, F., & Sener, G. (2010). Risk factors, frequency, and causes of falling in geriatric persons who has had a limb removed by amputation. Topics in Geriatric Rehabilitation, 26(2), 156-163.

Vanicek, N., Strike, S., McNaughton, L., & Polman, R. (2009a). Gait patterns in transtibial amputee fallers vs. non-fallers: Biomechanical differences during level walking. Gait & posture, 29(3), 415-420.

Vanicek, N., Strike, S., McNaughton, L., & Polman, R. (2009b). Postural responses to dynamic perturbations in amputee fallers versus nonfallers: a comparative study

with able-bodied subjects. Archives of physical medicine and rehabilitation, 90(6), 1018-1025.
Varrecchia, T., Serrao, M., Rinaldi, M., Ranavolo, A., Conforto, S., De Marchis, C., . . . Silvetti, A. (2019). Common and specific gait patterns in people with varying anatomical levels of lower limb amputation and different prosthetic components. Human movement science, 66, 9-21.

Chapter 4

Abou, L., Fliflet, A., Zhao, L., Du, Y., & Rice, L. (2022). The Effectiveness of Exercise Interventions to Improve Gait and Balance in Individuals with Lower Limb Amputations: A Systematic Review and Meta-analysis. Clinical Rehabilitation, 02692155221086204.
Andrysek, J., Klejman, S., Steinnagel, B., Torres-Moreno, R., Zabjek, K. F., Salbach, N. M., & Moody, K. (2012). Preliminary evaluation of a commercially available videogame system as an adjunct therapeutic intervention for improving balance among children and adolescents with lower limb amputations. Archives of physical medicine and rehabilitation, 93(2), 358-366.
Chiviacowsky, S., Wulf, G., & Wally, R. (2010). An external focus of attention enhances balance learning in older adults. Gait & posture, 32(4), 572-575.
Clemens, S., Doerger, C., & Lee, S.-P. (2020). Current and emerging trends in the management of fall risk in people with lower limb amputation. Current geriatrics reports, 9(3), 134-141.
Crenshaw, J. R., Kaufman, K. R., & Grabiner, M. D. (2013). Compensatory-step training of healthy, mobile people with unilateral, transfemoral or knee disarticulation amputations: A potential intervention for trip-related falls. Gait & posture, 38(3), 500-506. doi:https://doi.org/10.1016/j.gaitpost.2013.01.023
Damayanti Sethy, M., Kujur, E. S., & Sau, K. (2009). Effect of balance exercise on balance control in unilateral lower limb amputees. Indian Journal of Occupational Therapy, 41(3), 63-68.
Eng, J. J., Winter, D. A., & Patla, A. E. (1994). Strategies for recovery from a trip in early and late swing during human walking. Experimental brain research, 102(2), 339-349.
Erbahçeci, F., Bayramlar, K., Şener, F., Bayar, K., & Ülger, Ö. (2001). Balance training in amputees: Comparison of the outcome of two rehabilitation approaches. Eklem Hastalik Cerrahisi, 12(2), 194-198.
Esquenazi, A., & DiGiacomo, R. (2001). Rehabilitation after amputation. Journal of the American Podiatric Medical Association, 91(1), 13-22.
Furtado, S., Briggs, T., Fulton, J., Russell, L., Grimer, R., Wren, V., . . . Gerrand, C. (2017). Patient experience after lower extremity amputation for sarcoma in England: a national survey. Disability and rehabilitation, 39(12), 1171-1190.
Grabiner, M. D., & Kaufman, K. (2021). Developing and establishing biomechanical variables as risk biomarkers for preventable gait-related falls and assessment of intervention effectiveness. Frontiers in sports and active living, 268.

Hafner, B. J., Morgan, S. J., Askew, R. L., & Salem, R. (2016). Psychometric evaluation of self-report outcome measures for prosthetic applications. Journal of rehabilitation research and development, 53(6), 797.

Heijnen, M. J. H., & Rietdyk, S. (2016). Falls in young adults: Perceived causes and environmental factors assessed with a daily online survey. Human movement science, 46, 86-95.

Hershkovitz, A., Dudkiewicz, I., & Brill, S. (2013). Rehabilitation outcome of post-acute lower limb geriatric amputees. Disability and rehabilitation, 35(3), 221-227.

Hlavackova, P., Franco, C., Diot, B., & Vuillerme, N. (2011). Contribution of each leg to the control of unperturbed bipedal stance in lower limb amputees: new insights using entropy. PLoS One, 6(5), e19661.

Hordacre, B. G., Barr, C., Patritti, B. L., & Crotty, M. (2015). Assessing gait variability in transtibial amputee fallers based on spatial-temporal gait parameters normalized for walking speed. Archives of physical medicine and rehabilitation, 96(6), 1162-1165.

Hunter, S. W., Higa, J., Frengopoulos, C., Viana, R., & Payne, M. W. (2020). Evaluating knowledge of falls risk factors and falls prevention strategies among lower extremity amputees after inpatient prosthetic rehabilitation: a prospective study. Disability and rehabilitation, 42(16), 2252-2261.

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. doi:10.1007/s11999-014-3664-0

Kurtzer, I. L. (2015). Long-latency reflexes account for limb biomechanics through several supraspinal pathways. Frontiers in Integrative Neuroscience, 8. doi:10.3389/fnint.2014.00099

Landers, M. R., Durand, C., Powell, D. S., Dibble, L. E., & Young, D. L. (2011). Development of a scale to assess avoidance behavior due to a fear of falling: the Fear of Falling Avoidance Behavior Questionnaire. Physical therapy, 91(8), 1253-1265.

Mandel, A., Paul, K., Paner, R., Devlin, M., Dilkas, S., & Pauley, T. (2016). Balance confidence and activity of community-dwelling patients with transtibial amputation. Journal of Rehabilitation Research & Development, 53(5).

Matjacić, Z., & Burger, H. (2003). Dynamic balance training during standing in people with trans-tibial amputation: a pilot study. Prosthetics and orthotics international, 27(3), 214-220.

Miller, W. C., Deathe, A. B., & Speechley, M. (2003). Psychometric properties of the Activities-specific Balance Confidence Scale among individuals with a lower-limb amputation. Archives of physical medicine and rehabilitation, 84(5), 656-661.

Miller, W. C., Deathe, A. B., Speechley, M., & Koval, J. (2001). The influence of falling, fear of falling, and balance confidence on prosthetic mobility and social activity among individuals with a lower extremity amputation. Archives of physical medicine and rehabilitation, 82(9), 1238-1244.

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.

Munin, M. C., Espejo-De Guzman, M. C., Boninger, M. L., Fitzgerald, S. G., Penrod, L. E., & Singh, J. (2001). Predictive factors for successful early prosthetic ambulation among lower-limb amputees. J Rehabil Res Dev, 38(4), 379-384.
Owings, T. M., Pavol, M. J., Foley, K. T., & Grabiner, M. D. (2000). Measures of postural stability are not predictors of recovery from large postural disturbances in healthy older adults. Journal of the American Geriatrics Society, 48(1), 42-50.
Owings, T. M., Pavol, M. J., & Grabiner, M. D. (2001). Mechanisms of failed recovery following postural perturbations on a motorized treadmill mimic those associated with an actual forward trip. Clinical Biomechanics, 16(9), 813-819.
Parker, K., Hanada, E., & Adderson, J. (2013). Gait variability and regularity of people with transtibial amputations. Gait & posture, 37(2), 269-273.
Pavol, M. J., Owings, T. M., Foley, K. T., & Grabiner, M. D. (2001). Mechanisms leading to a fall from an induced trip in healthy older adults. The Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 56(7), M428-M437.
Pereira, D. G., Afonso, A., & Medeiros, F. M. (2015). Overview of Friedman's test and post-hoc analysis. Communications in Statistics-Simulation and Computation, 44(10), 2636-2653.
Pijnappels, M., Bobbert, M. F., & van Dieën, J. H. (2005). How early reactions in the support limb contribute to balance recovery after tripping. Journal of biomechanics, 38(3), 627-634. doi:https://doi.org/10.1016/j.jbiomech.2004.03.029
Pruszynski, J. A., & Scott, S. H. (2012). Optimal feedback control and the long-latency stretch response. Experimental brain research, 218(3), 341-359.
Rau, B., Bonvin, F., & De Bie, R. (2007). Short-term effect of physiotherapy rehabilitation on functional performance of lower limb amputees. Prosthetics and orthotics international, 31(3), 258-270.
Robinovitch, S. N., Feldman, F., Yang, Y., Schonnop, R., Leung, P. M., Sarraf, T., . . . Loughin, M. (2013). Video capture of the circumstances of falls in elderly people residing in long-term care: an observational study. The Lancet, 381(9860), 47-54.
Rosenblatt, N. J., Bauer, A., & Grabiner, M. D. (2017). Relating minimum toe clearance to prospective, self-reported, trip-related stumbles in the community. Prosthetics and orthotics international, 41(4), 387-392.
Rusaw, D., Hagberg, K., Nolan, L., & Ramstrand, N. (2013). Bilateral electromyogram response latency following platform perturbation in unilateral transtibial prosthesis users: Influence of weight distribution and limb position. Journal of rehabilitation research and development, 50(4), 531-544.
Sessoms, P. H., Wyatt, M., Grabiner, M., Collins, J.-D., Kingsbury, T., Thesing, N., & Kaufman, K. (2014). Method for evoking a trip-like response using a treadmill-based perturbation during locomotion. Journal of biomechanics, 47(1), 277-280.
Sheskin, D. J. (2003). Handbook of parametric and nonparametric statistical procedures: Chapman and Hall/CRC.
Shih, H.-T., Teng, H.-L., Gray, C., Poggemiller, M., Tracy, I., & Lee, S.-P. (2019). Four weeks of training with simple postural instructions changes trunk posture and foot strike pattern in recreational runners. Physical Therapy in Sport, 35, 89-96.
Šlajpah, S., Kamnik, R., Burger, H., Bajd, T., & Munih, M. (2013). Asymmetry in sit-to-stand movement in patients following transtibial amputation and healthy individuals. International Journal of Rehabilitation Research, 36(3), 275-283.

Sloot, L. H., van den Noort, J. C., van der Krogt, M. M., Bruijn, S. M., & Harlaar, J. (2015). Can treadmill perturbations evoke stretch reflexes in the calf muscles? PLoS One, 10(12), e0144815.

Song, S. (2009). Consciousness and the consolidation of motor learning. Behavioural Brain Research, 196(2), 180-186. doi:https://doi.org/10.1016/j.bbr.2008.09.034

Steinberg, N., Gottlieb, A., Siev-Ner, I., & Plotnik, M. (2019). Fall incidence and associated risk factors among people with a lower limb amputation during various stages of recovery–a systematic review. Disability and rehabilitation, 41(15), 1778-1787.

Talbot, L. A., Musiol, R. J., Witham, E. K., & Metter, E. J. (2005). Falls in young, middle-aged and older community dwelling adults: perceived cause, environmental factors and injury. BMC public health, 5(1), 1-9.

Tesio, L., Lanzi, D., & Detrembleur, C. (1998). The 3-D motion of the centre of gravity of the human body during level walking. II. Lower limb amputees. Clinical Biomechanics, 13(2), 83-90.

Ülger, Ö., Topuz, S., Bayramlar, K., Erbahçeci, F., & Sener, G. (2010). Risk factors, frequency, and causes of falling in geriatric persons who has had a limb removed by amputation. Topics in Geriatric Rehabilitation, 26(2), 156-163.

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.

Van de Meent, H., Hopman, M. T., & Frölke, J. P. (2013). Walking ability and quality of life in subjects with transfemoral amputation: a comparison of osseointegration with socket prostheses. Archives of physical medicine and rehabilitation, 94(11), 2174-2178.

Webster, J. B., Crunkhorn, A., Sall, J., Highsmith, M. J., Pruziner, A., & Randolph, B. J. (2019). Clinical practice guidelines for the rehabilitation of lower limb amputation: an update from the Department of Veterans Affairs and Department of defense. American journal of physical medicine & rehabilitation, 98(9), 820-829.

Wong, C. K., Chen, C. C., Blackwell, W. M., & Rahal, R. T. (2015). Balance ability measured with the Berg balance scale: a determinant of fall history in community-dwelling adults with leg amputation. Journal of rehabilitation medicine, 47(1), 80-86.

Wong, C. K., Ehrlich, J. E., Ersing, J. C., Maroldi, N. J., Stevenson, C. E., & Varca, M. J. (2016). Exercise programs to improve gait performance in people with lower limb amputation: A systematic review. Prosthetics and orthotics international, 40(1), 8-17.

Curriculum Vitae

HUI-TING SHIH

b96408017@gmail.com

Ph.D. in Interdisciplinary Health Sciences Rehabilitation Science, Clinical Locomotion Neuromechanics Lab

University of Nevada, Las Vegas (UNLV), Las Vegas, NV, USA. 2016-2022
Advisor: Szu-Ping Lee, PhD

M.S. in Physical Therapy, Postural Control Lab

National Taiwan University (NTU), Taipei, Taiwan. 2011-2013
Advisor: Ming-Hsia Hu, PhD

B.S. in Physical Therapy

National Taiwan University (NTU), Taipei, Taiwan. 2007-2011

CONFERENCE PRESENTATIONS

1. Singh H, **Shih HT**, Kal E, Bennett T, Wulf G. Functional variability increases with a distal external focus. 26-28 May 2022, Waikoloa HI, USA, North American Society for the Psychology of Sports and Physical Activity.
2. Ho S, **Shih HT**, Chien LC, Clemens S, Jensen C, Lee SP. Returning to Work After Dysvascular Lower Limb Amputation: A Model Analysis of Functional and Contextual Predictors. 2-5 March 2022, Atlanta, Georgia, USA, American Academy of Orthotists and Prosthetists Annual Meeting.
3. Clemens S, Kershaw K, Ho S, **Shih HT**, Nagode N, Lee SP. A Pilot Study Identifying Racial Disparities in Prosthetic Mobility in People with Lower Limb Amputation. 2-5 February 2022, San Antonio, Texas, USA, APTA Combined Section Meeting.
4. **Shih HT**, Lee SP. Trip Recovery Responses during Trip-like Perturbation to Non-prosthetic vs Prosthetic Legs in Individuals with Unilateral Transtibial Amputation. 1-4 November 2021, Virtual, International Society of Prosthetics and Orthotics.
5. **Shih HT**, Anderson J, Fabian C, Hadley S, Ng D, Lee SP. A Novel Ground reaction force triggered Tripping Protocol Elicits Reliable Responses in Young Adults. 1-28 February 2021, Virtual, APTA Combined Section Meeting.
6. **Shih HT**, Yi ST, Shih Yo, Lee SP. Description and Validation of a Ground Reaction Force Triggered Treadmill Protocol for Simulation of Tripping Falls. 4-7 August 2020, Virtual, American Society of Biomechanics Annual Meeting.
7. Abdollahipour R, Iwatsuki T, **Shih HT**. Autonomy support enhances movement efficiency. 21-24 November 2019, Launceston, Australia, Australasian Cognitive Neuroscience Conference.
8. Lee SP, **Shih HT**, Amaya G, Gante R, Meinzer Z, Sadrossadat A. Physical Therapy After Lower Extremity Amputation: Predictors and Outcome Implications. 5-6 October 2019, Kobe, Hyogo, Japan, International Society of Prosthetics and Orthotics.
9. **Shih HT**, Lee SP. Socioeconomic Status Affects Motor Activity Endurance and Fear of Falling Avoidance Behaviour in People with Lower Limb Loss. 31 July-4 August 2019, Calgary, Alberta, Canada, American Society of Biomechanics Annual Meeting.
10. Iwatsuki T, **Shih HT**, Abdollahipour R, Wulf G. More bang for the buck: Autonomy support increases muscular efficiency. 6-8 June 2019, Baltimore MD, USA, North American Society for the Psychology of Sports and Physical Activity.

11. Kubo M, Horn L, Gorton J, Ferraro A, **Shih HT**, MacLeod T, Lee SP. Physical Therapy After Amputation in Community Dwelling Adults: A Quantitative and Qualitative Study. 12-15 February 2019, Denver, APTA Combined Section Meeting.
12. **Shih HT**, Lee SP. Effect of Age and Body Mass Index on Mobility in Middle-Aged to Older People With And Without Lower Limb Amputation. 8-11 August 2018, Rochester MN, USA, American Society of Biomechanics Annual Meeting.
13. **Shih HT**, Hsieh CT, Huang CY, Hu MH. Response to Different Balance Training Task Difficulties in Chronic Stroke: A Pilot Study. National Taiwan University Physical Therapy Student Conference 2013.

INVITED PRESENTATIONS

Exercise for Fall Prevention. 28 August, 2021, Las Vegas, Aging Wellness Expo.
Prosthetics and Post-amputation Physical Therapy. 3 January, 2020, Dept. of Physical Therapy, Asia University

SKILLS

- VICON Motion Capture System and Motion Analysis
- Transcranial Magnetic Stimulation
- Peripheral Nerve Stimulation
- Electromyography
- BrainSight Brain Navigation System
- HUMAC NORM Isokinetic Dynamometer
- Smart Balance Master/Bertec Balance Advantage
- Protokinetics Zeno Walkway for Gait Analysis
- DEXA Scan Body Composition Analyzer
- Nexus
- Visual 3D
- MATLAB
- BrainSight
- Signal
- SPSS
- MPlus

TEACHING EXPERIENCE

Teaching assistant: Dept. of Physical Therapy, University of Nevada, Las Vegas
DPT 749 & DPT 749L Applied Exercise Physiology and Laboratory
DPT 791 Applied Research Statistics
DPT 750 & DPT 750L Prosthetics and Orthotics and Laboratory
PTS 712 Physiological Bases of Rehabilitation (PhD level course)
HSC 703 Grant Writing in Health Sciences (PhD level course)
Guest lecturer: Dept. of Psychology, University of Nevada, Las Vegas
PSY 101 General Psychology – Biological bases of behaviors
Teaching assistant: Dept. of Physical Therapy, National Taiwan University
PT3010 Clinical Clerkship in Physical Therapy
PT2014, PT2015 Physical Agent Therapy and Practice
PT3011 Introduction to Motor Control and Learning
PT3029, PT3030 Neurological Physical Therapy and Practice
PT7077 Practical Statistics
Instructor: Jenteh Junior College of Medicine, Nursing and Management
Neurological Physical Therapy (lecture and practice)

MENTORING EXPERIENCE

Doctor of Physical Therapy Students, Student Research Mentor
Class of 2021: James Anderson, Catrina Anne Fabian, Samuel Hadley, Cheuk Wing Denise Ng
Doctor of Physical Therapy Students, Student Research Co-mentor

Class of 2019: Greg Amaya, Rosman Gante, Zachary Meinzer, Abbas Sadrossadat
Class of 2020: Michaela Kubo, Jennifer Gorton, Lindsey Horn, Amanda Ferraro

CLINICAL EXPERIENCE

Instructor: Community College of Mucha District, Taipei City Government
Instructor of group exercise class
Invited Speaker: Shin Kong Wu Ho-Su Memorial Hospital (Medical Center)
Instructor of community health workshops
Physical Therapist: National Taiwan University Hospital (Medical Center) & Shu Tien Clinic
Neurological, bedside acute care, orthopedic, burn care and cardiopulmonary physical therapist

FELLOWSHIP & AWARDS

International Society of Biomechanics Matching Dissertation Grant. 2021-2022. ($5,000)

American Physical Therapy Association Biomechanics Special Interest Group Student Travel Award. 2021. ($200)

University of Nevada, Las Vegas Barrick Graduate Fellowship. 2021-2022 ($15,000 plus tuition).

University of Nevada, Las Vegas Summer Doctoral Fellowship. Summer 2018 ($7,000).

Graduate & Professional Student Association (GPSA) at University of Nevada, Las Vegas Research Forum
1st Place Poster presentation. 2018
2nd Place Oral Presentation. 2019
1st Place Oral presentation. 2021
1st Place Oral presentation. 2022

University of Nevada, Las Vegas, Department of Physical Therapy Travel Grant
Travel Grant for International Society of Prosthetics and Orthotics Congress, Kobe, Hyogo, Japan. Fall 2019. ($1,200)
Travel Grant for International Society of Biomechanics and American Society of Biomechanics Combined Congress, Calgary, Alberta, Canada. Summer 2019. ($600)
Travel Grant for American Physical Therapy Association Combined Section Meeting, Washington DC, USA. Spring 2019 ($350)
Travel Grant for Regenerative Rehabilitation Symposium, Pittsburgh, Pennsylvania, USA. Fall 2017 ($800).
Travel Grant for American Physical Therapy Association Combined Section Meeting, San Antonio, TX, USA. Spring 2017 ($1,500)

University of Nevada, Las Vegas Access Grant.
Spring 2018, 2019 ($2,000).
Fall 2017, 2018 ($2,000).

Graduate & Professional Student Association at University of Nevada, Las Vegas (GPSA) Book Scholarship. Spring 2018 ($200).

Graduate & Professional Student Association at University of Nevada, Las Vegas (GPSA) Travel Funding
Travel funding for International Society of Prosthetics and Orthotics Congress, Kobe, Hyogo, Japan. Fall 2019 ($762.65)
Travel funding for International Society of Biomechanics and American Society of Biomechanics Combined Congress, Calgary, Alberta, Canada. Summer 2019 ($500).

Travel funding for American Society of Biomechanics Annual Meeting, Rochester, Minnesota, USA. Summer 2018 ($800).
Funding for International Society of Prosthetics and Orthotics Congress, virtual. Fall 2021

CERTIFICATIONS

Graduate College Research Certificate, University of Nevada Las Vegas
Collaborative Institutional Training Initiative (CITI Program), University of Nevada Las Vegas
Geriatrics and Long-Term Care Program, National Taiwan University

PROFESSIONAL MEMBERSHIPS

International Society of Biomechanics (ISB)
International Society of Prosthetics and Orthotics (ISPO)
American Society of Biomechanics (ASB)
American Physical Therapy Association (APTA)

SERVICES

Taiwanese Young Researcher Association (project Tyra)
Crew in charge of public relationship and membership affair. 2020-now
University of Nevada, Las Vegas, Taiwanese Student Association
Vice president. 2016-now
National Taiwan University Hospital
Volunteer physical therapy. Serving medical consultation in suburban areas in Taiwan. 2014-2016
WCPT-AWP & ACPT Combined Conference
Executive Assistant of World Confederation for Physical Therapy, Asia Western Pacific Region (WCPT-AWP) & Asia Confederation for Physical Therapy (ACPT) Combined Conference, Taichung, Taiwan. 2013.
National Taiwan University, Department of Physical Therapy
Chief editor of departmental newsletter. 2011-2013
Taiwan Physical Therapy Association
Master of ceremony in Taiwan Physical Therapy Association Academic Conference and Annual General Meeting. 2011-2012
Asia Physical Therapy Student Association (APTSA)
Minister of secretarial works, headquarters of Asia Physical Therapy Student Association. 2012-2013