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Fall risk is a concern for a variety of clinical populations, especially in lower-limb amputees. The risk of falling during walking is increased by an individual with pathology's diminished ability for obstacle negotiation. Virtual obstacle crossing environments offer a rehabilitation technique that is space and material efficient and may enhance obstacle crossing skill acquisition and retention though the use of task specificity, repetition, and feedback; while presenting an engaging and motivating challenge for participants. Current literature has not determined the response of an individual to virtual obstacle crossing in comparison to real environment over-ground obstacle crossing, nor whether aging influences this behavior. In a first step to determine the clinical viability of a virtual reality obstacle crossing environment, this task was tested using healthy able-bodied individuals (20 younger adults and 20 older adults) to determine an individual's expected crossing behavior during a single session of training. The purpose of this study was to (1) determine the biomechanical obstacle-crossing behavior of an able-bodied individual within a virtual environment, (2) determine if a learning effect exists with virtual obstacle crossing, and (3) determine if the learning effect will transfer to over-ground obstacle crossing and create performance changes. Dependent variables measured were foot placement before and after the obstacles for the both the lead and trail limbs, toe/heel clearance for both limbs in the vertical and radial directions, and the peak toe and heel elevation. The hypotheses were: (1) a training effect would be observed at the end of the virtual obstacle crossing training in the form of the

adoption of a safer obstacle crossing strategy in the virtual environment, (2) a safer obstacle crossing strategy in the real environment would be adopted in the post-test relative to the pre-test, and (3) the performance changes in the virtual environment would be correlated with the performance changes in the real environment, suggesting an association between motor learning in a virtual environment and transfer to a real environment task. It was also postulated that each hypothesized finding would be affected by age, with older adults showing less learning and transfer (albeit still significant) compared to the younger adults. Results indicate that participants learned to cross the virtual obstacle more safely and that change in behavior was transfer to the real environment. Further, while both age groups showed transfer to the real environment task, they differed on which limb their transfer effects applied to. The data suggest it is plausible to use virtual reality training as a way to enhance gait characteristics in the context of obstacle avoidance, potentially a leading to a novel way to reduce fall risk.

VIRTUAL OBSTACLE CROSSING AND THE CLINICAL IMPLICATIONS FOR REHABILITATION

by

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APPROVAL PAGE

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TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

CHAPTER I

INTRODUCTION

The number of amputees rehabilitating and learning to live in our society has rapidly increased over the years, and walking is considered to be one of the most important everyday living activities to establish independence (Sagawa et al., 2011). Learning to use a prosthetic device is a key factor in rehabilitation of gait patterns. Amputees display irregular biomechanical gait patterns, which put those individuals at an increased risk of degenerative conditions, falling, and injuries to other related parts of the body, such as the intact limb and the lower back (Gailey et al., 2008).

Obstacle crossing is a critical part of safe ambulation and a necessity for trans-tibial amputees, especially because their ability to adapt their gait to obstacles within their gait path is often low (Hofstad et al., 2006, 2009). Obstacle crossing poses a challenge, as often times this low gait adaptability during obstacle crossing translates to increased risk of falling when compared to able-bodied adults (Miller, Speechley, & Deathe, 2001). Training programs to enhance obstacle avoidance strategies have been developed with some success in clinical populations (Darter & Wilken, 2011; Mirelman et al., 2010; Shema et al., 2014). However, these programs often require physical space and materials that may not be available in many clinical settings. Training with virtual obstacles may be a path to get around these space and material barriers. Virtual reality provides a safe way to potentially learn how to cross obstacles using a prosthetic device by providing an engaging environment that can also provide performance feedback to promote motor learning.

Current gaps in the literature call for a need to compare the biomechanical behavior of an amputee crossing a virtual obstacle within a virtual environment to the well-established real world over-ground obstacle crossing behavior in amputees. However, the effect of virtual obstacle crossing must first be established within able-bodied adults. As amputations may happen across the lifespan, it is important to establish the profile of this task with younger and older adults. However, a necessary first step is to examine the effect of virtual obstacle crossing in able-bodied adults to establish normative data. A learning effect resulting from repeated virtual obstacle crossings and feedback from the virtual environment needs to be determined. Additionally, it is important to examine if this learning effect transfers from the virtual environment to the over-ground environment to determine the feasibility of using virtual reality obstacle crossing in rehabilitation settings.

The purpose of this study was to (1) determine the biomechanical obstacle crossing behavior of an able-bodied individual within a virtual environment, (2) determine if a learning effect existed with virtual obstacle crossing, and (3) determine if the learning effect would transfer to over-ground obstacle crossing. Based on the previous literature, the following hypotheses were made:

Hypothesis 1: A training effect would be observed at the end of the virtual obstacle crossing training in the form of the adoption of a safer obstacle crossing strategy in the virtual environment.

Hypothesis 2: A safer obstacle crossing strategy in the real environment would be adopted in the post-test relative to the pre-test.

Hypothesis 3: The performance changes in the virtual environment would be correlated with the performance changes in the real environment, suggesting an association between motor learning in a virtual environment and transfer to a real environment task.

It was also postulated that each hypothesized finding would be affected by age, with older adults showing less learning and transfer (albeit still significant) compared to the younger adults.

CHAPTER II

REVIEW OF THE LITERATURE

Overview

This literature review will discuss the basics of gait in able-bodied adults and what happens to gait after lower-limb amputation. Next, this literature review will discuss the biomechanics of obstacle crossing in able-bodied adults and trans-tibial amputees. Additionally, the literature review will discuss motor learning and a few key foundational components of learning a new skill or re-learning a well-known skill. Finally, how virtual reality could be used to develop more adaptive obstacle crossing strategies in order to avoid the potential of a trip and fall will be explored.

The Gait Cycle in Able-bodied Adults

Gait occurs in three planes of motion: sagittal, transverse, and frontal. Joint flexion and extension occur in the sagittal plane, rotation in the transverse plane, and abduction and adduction in the frontal plane (Vaughan, Davis, & O'Connor, 1999). Gait is described as cyclical and periodic with each lower extremity alternating between stance and swing elements to create a complete gait cycle (Vaughan et al., 1999). One full gait cycle is defined as heel strike to following heel strike of the same limb (Robertson, Hamill, Caldwell, Kamen, & Whittlesey, 2004). The stance portion of gait alternates from double stance with the support of both limbs to single leg stance with single limb support. Double leg stance occurs for 20% of the gait cycle and 40% accounts for each lower extremity during single leg stance (Winter, 2009) as single leg support occurs in identical time frames for each limb (Vaughan et al., 1999). Able-bodied gait may be characterized as the lower limbs moving in symmetry with one another during the course of the gait cycle. With a symmetrical gait pattern, an energy efficient mode of walking is achieved (Skinner & Effeney, 1985).

The sole purpose of gait is to transport the body safely and efficiently across the ground, including on level, uphill, and downhill surfaces (Winter, 1991). In order to accomplish safe and efficient ambulation, there are five main functions which the muscular system must perform during each gait cycle: (1) maintenance of support of the upper body during stance, (2) maintenance of upright posture and balance of the total body, (3) control of foot trajectory to achieve safe ground clearance and a gentle heel or toe landing, (4) generation of mechanical energy to maintain the present forward velocity or to increase the forward velocity, and (5) absorption of mechanical energy for shock absorption and stability or to decrease the forward velocity of the body (Winter, 1991). The muscular system contains several muscle groups by which gait is controlled. These include the plantarflexors, quadriceps, hamstrings, and gluteals. The plantar flexors are the major source of energy generation for push-off (Winter, 1983) as they are the key for propulsion, providing an explosive energy burst just before toe-off (Winter, 1991). The quadriceps are the major energy absorbers (Winter, 1983) as they provide knee extension and hip flexion during swing phase and control knee flexion during weight acceptance (Winter, 1991). The hamstring muscle group decelerates the forward swinging limb and, at heel-strike, controls the forward rotation of the thigh and stabilizes the pelvis by acting as a hip extensor during weight acceptance (Winter, 1991). The gluteal muscle group controls hip flexion during weight acceptance in addition to controlling forward thigh rotation and thus knee flexion (Winter, 1991). The hip plays a small and variable role in energy absorption and generation (Winter, 1983) while the muscles of the trunk serve to stabilize the trunk and control balance during motion (Winter, 1991). Gait control increases throughout childhood and adolescence until it reaches full maturity in early adulthood. Neurological changes that naturally occur across the lifespan, such as aging, will alter gait control along with structural changes, such as limb amputation.

Changes in Gait Due to Level of Amputation

The level of amputation affects the kinematics and kinetics of an individual and thus an individual's ability to adapt to able-bodied gait characteristics. The level of amputation refers to the point of the limb at which the amputation was performed. The level of amputation may be below the knee transversely crossing the tibia, also known as trans-tibial amputation, or above the knee transversely crossing the femur, also known as trans-femoral amputation. As the level of amputation increases (i.e., moves up the limb), the deviation from able-bodied gait becomes increasingly larger because the amount of asymmetry in an amputee's walking pattern is directly related to their stump length (Block, Alimusaj, Heitzmann, Korber, & Wolf, 2014). A decrease in stump length directly correlates to an increase in the duration of stance phase and a decrease in the duration of swing phase for both limbs (Jaegers, Arendzen, & de Jongh, 1995). The increase in stance phase duration relates to overall walking speed. An amputee has a significantly decreased walking speed when compared to an able-bodied individual with the speed becoming increasingly lower with subsequently high amputation levels (Waters, Perry, Antonelli, & Hislop, 1976). Specific to trans-tibial amputees, this population exhibits a decrease in cadence, stride length, and gait cycle that is one standard deviation below able-bodied gait. In these same parameters, trans-femoral amputees are two standard deviations below able-bodied gait (Skinner & Effeney, 1985). With regards to kinetic parameters, decreasing stump length relates to increasing force impulse in the intact limb (Pruziner, Werner, Copple, Hendershot, & Wolf, 2014). This is a result of insufficient power in the prosthetic limb to lift and propel the body's center of mass.

In reference to changes due to rehabilitation, trans-tibial amputees require less time to create adjustment strategies than trans-femoral and knee-disarticulation amputees (Vrieling et al., 2009). Improvements in limb loading force are seen in trans-tibial amputees two years after limb loss. However, trans-femoral amputees still exhibit an increased loading and increased ground reaction forces two years after limb loss (Pruziner et al., 2014). Thus, with the higher level of amputation, the ability to improve and adapt to able-bodied gait patterns is either much slower than trans-tibial amputees or does not occur. Because of the high variability in gait with trans-femoral amputees, published literature focuses mainly on trans-tibial amputees. Thus, the remainder of this literature review will focus on trans-tibial amputees.

Changes in the Gait Cycle and Biomechanical Variables Due to Amputation

Trans-tibial amputees exhibit an asymmetrical gait in reference to able-bodied gait and specifically between the intact limb and the prosthetic limb. This asymmetry is often a result of the lack of lower leg muscles that produce plantarflexion and dorsiflexion in the prosthetic limb. Trans-tibial amputees are missing a key component contributing to the control of gait because the ankle plantar flexors produce 80% of the propulsion power generated during able-bodied gait (Winter, 1983). Because of the missing plantarflexion power, time-distance variables of velocity, stride time, and stride length are lower in amputees relative to able-bodied individuals (Bateni & Olney, 2002; Robinson, Smidt, & Arora, 1977). With limited plantar flexion in the prosthetic limb, the major source of power generation is due to the hip extensors and flexors and the hip external rotators (Sadeghi, Allard, & Duhaime, 2001).

Amputees employ different motor strategies when their gait characteristics are compared to able-bodied individuals (Sanderson & Martin, 1997), such as displaying a lower mean knee flexion than able-bodied individuals during early and late stance (Breakey, 1976). This may be a function of the rigidity of their prosthetic or a compensatory mechanism to reduce loading and, therefore, discomfort in the prosthetic limb socket. In able-bodied gait, the peak center of mass velocity in the anterior direction usually synchronizes with the minimum center of mass height during double support stance (Alexander, 1984). This does not occur synchronously for amputees. Amputees maintain a net propulsive anterior-posterior ground reaction force impulse throughout double support until toe-off while the center of mass is elevating (De Asha, Munjal, Kulkarni, & Buckley, 2014). This may be a compensatory mechanism of propulsion in the intact limb and reduced braking in the prosthetic limb (De Asha et al., 2014). However, this disorganization may contribute to the higher metabolic cost of gait seen in amputees (Skinner & Effeney, 1985; Waters et al., 1976).

Amputees not only employ different motor strategies in reference to able-bodied limbs, but also within their own limbs as the prosthetic limb and the intact limb act differently to produce the forward propulsion needed for gait (Sanderson & Martin, 1997). In general, amputees spend less time on their prosthetic limb than their intact limb (Jaegers et al., 1995), thus making stance phase for the intact limb longer in duration than in the prosthetic limb (Bateni & Olney, 2002; Breakey, 1976). Single limb support for the prosthetic limb is 37% of the gait cycle compared to 43% for the intact limb (Breakey, 1976). This may be due to the lack of trust in the prosthetic side to safely and properly bear weight. This issue of lack of trust is due to that fact that when compared to a normal limb, a prosthetic limb does not have the appropriate biological sensors that gather information about joint position, joint acceleration, and limb position, which help guide our limbs in a proactive and reactive manner. This lack of biological sensors makes the prosthetic limb more difficult to control and deemed not as trustworthy as the intact limb. Therefore, an amputee's transfer of weight back to the intact limb occurs as quickly as possible (Bateni & Olney, 2002).

The asymmetry induces changes within each limb during the gait cycle. At heel contact of the prosthetic limb, an increase of the activity in the hip abductor may be seen to compensate for the lack of dorsiflexion achieved by the prosthetic (Sadeghi et al., 2001). In early stance phase of the prosthetic limb, an increase in hip flexion may be observed (Bateni & Olney, 2002). This may be a result of an increase in step length, the prosthetic being fit to the stump in a flexion angle, the amputee attempting to increase positive power at the knee, or a combination of these (Bateni & Olney, 2002). In early stance of the intact limb, an increase in hip flexion may also be observed, which may be a result of slightly increased trunk flexion in order to increase the positive power output at the knee (Bateni & Olney, 2002). The intact limb shows a decrease in hip abduction, which may be a result of the lack of ability of the prosthetic limb to transfer the weight of the body to the intact limb (Sadeghi et al., 2001). In mid-stance of the prosthetic limb, hip abductor activity increases in order to increase frontal plane movement and maintain balance during forward propulsion (Sadeghi et al., 2001). Additionally, the hip of the prosthetic leg retains a more extended position during stance phase; thus, resulting in a more vertical orientation of the thigh (Sanderson & Martin, 1997). During push-off, hip flexion increases to increase power for forward propulsion as a compensation for the lack of plantar flexors (Sadeghi et al., 2001). Swing phase of the prosthetic limb is similar to that of the intact limb and able-bodied gait. However, amputees significantly deviate during the 80-95% of the gait cycle as knee flexors decelerate the swing of the prosthesis and the hip extensors prepare the prosthetic limb for the beginning of another gait cycle (Sadeghi et al., 2001).

Asymmetry may also induce changes in force patterns. In general, amputees impose more force onto their intact limb compared to their prosthetic limb (Sanderson & Martin, 1997; Silverman & Neptune, 2014; Vrieling et al., 2008a, 2008b). In the anterior-posterior plane, the prosthetic limb has lower peak braking and propulsion forces than the intact limb. However, the differences in peak forces were greater for the propulsive stance phase than the braking stance phase (Sanderson & Martin, 1997). The anterior-posterior impulse during walking is larger in the prosthetic limb compared to the intact limb (Silverman & Neptune, 2014). However, the intact limb contains larger peak forces and impulse in the axial and medio-lateral directions when compared to the prosthetic limb (Silverman & Neptune, 2014; Vrieling et al., 2008a). In the vertical plane, the peak force of late stance is also lower for the prosthetic limb than for the intact limb (Sanderson & Martin, 1997).

Asymmetry between each limb may be seen at each of the lower limb joints. At the ankle, a greater dorsiflexor moment in early stance of the intact limb is observed. The dorsiflexor moment is greater in amplitude and duration (Sanderson & Martin, 1997). The moment of the ankle’s predominant role is the support and propulsion of the body during gait (Winter, 1983). Greater differences between limbs are seen at the knee, predominantly in early stance phase when the body weight is being accepted onto the limb (Sanderson & Martin, 1997). The knee flexor moments at the beginning of the stance phase are lower for the prosthetic limb (Sanderson & Martin, 1997). As stance progresses, the prosthetic limb exhibits a small peak extensor movement while remaining flexed (Sanderson & Martin, 1997). The intact limb exhibits a higher and longer peak extensor moment before returning to flexion (Sanderson & Martin, 1997). The moments of the knee remain similar for both limbs for the remainder of the stance phase (Sanderson & Martin, 1997). About the hip joint, the intact limb exhibits a decreased extensor moment in early stance phase when compared to the prosthetic limb in stance phase. The loading differences in limbs are due to the differences in contributions from the prosthesis relative to the ankle plantarflexors (Silverman & Neptune, 2014). These motor strategies aim to reduce loading on and about the knee joint to reduce stump loading and increase stability (Sanderson & Martin, 1997).

Gait Initiation

Gait initiation requires propulsion and balance control, which amputees tend to lack due to missing an intact ankle joint and the musculature surrounding that joint (Vrieling et al., 2008a). A posterior center of pressure displacement and an anterior ground reaction force are critical for adequate propulsion in gait initiation (Vrieling et al., 2008a). During gait initiation, amputees exhibit a decrease in peak anterior ground reaction force, a smaller or absent posterior center of pressure shift, and a lower gait initiation velocity when compared to able-bodied adults (Vrieling et al., 2008a). The stiffness of the prosthetic foot, missing ankle dorsiflexors, and deficient sensory feedback all contribute to the decreased posterior center of pressure shift in amputees (Vrieling et al., 2008a). This decreased posterior center of pressure shift, along with missing plantar flexors, causes a reduced peak anterior ground reaction force (Vrieling et al., 2008a). As a result, gait initiation velocity is decreased (Vrieling et al., 2008a). The main compensation strategies amputees use to initiate gait are to provide an increased loading on the intact limb, elongate the period of propulsive force production in the intact limb, and to actually initiate gait using the prosthetic limb (Vrieling et al., 2008a).

Gait Termination

In able-bodied individuals, the leading limb is responsible for the production of the essential braking ground reaction force to terminate gait. Trans-tibial amputees are only able to produce 33% of the braking force when compared to able-bodied individuals (Vrieling et al., 2008b). Amputees exhibit a decreased peak braking ground reaction force in the prosthetic limb, along with no anterior center of pressure shift during leading with the prosthetic limb and an increased mediolateral center of pressure shift (Vrieling et al., 2008b). Amputees are not able to increase the braking force and shift the center of pressure anteriorly in order to terminate gait because of the lack of ankle function (Vrieling et al., 2008b). The limitations of the prosthetic limb are due to the missing lower-limb musculature and the different properties exhibited by a prosthetic device compared to a human limb (Vrieling et al., 2008b). The prosthetic foot's stiffness contributes to a decrease in the smoothness of the anterior shift of the center of pressure (Vrieling et al., 2008b).

Compensatory strategies seen in gait termination with amputees are using the intact limb as their preferred leading limb, a longer duration of producing a braking force in the intact limb, decreased gait termination velocity, and bearing more weight on the intact limb (Vrieling et al., 2008b). The leading limb must provide stability in addition to slowing down the forward movement so this may be a reason for preference of the intact limb as the leading limb for gait termination as braking force is larger, anterior center of pressure moves in front of the center of mass, and the mediolateral center of pressure shift is smaller (Vrieling et al., 2008b). The longer production of the decreased braking force may be a compensation strategy as amputees spend more time on their intact limb in single-stance phase (Vrieling et al., 2008b).

Differences in Prosthetics

Amputees have a wide selection of prostheses to choose from once they have recovered from amputation surgery. According to the American Academy of Orthotists and Prosthetists, the selection of a prosthesis depends on what K-level the amputee is classified as. K-levels are defined by Medicare and based on an individual's ability or potential to ambulate and navigate through their environment. A K0 level indicates an individual does not have the ability to ambulate safely and a prosthesis does not enhance their quality of life, thus these individuals are not eligible for a prosthesis. A K1 level indicates an individual has the ability to use a prosthesis for transfers or ambulation on a level surface. K1 level prostheses utilize single axis ankle/feet and external keel feet. A K2 level indicates an individual has the ability to ambulate within a typical community and traverse low-level barriers such as curbs, stairs, and uneven surfaces. K2 level prostheses are flexible-keel feet and multi-axial ankle/feet. A K3 level indicates that an individual has the ability to ambulate with a variable cadence and the ability to traverse most environmental barriers and may have activity that demands prosthetic use beyond simple locomotion. K3 level prostheses are flex foot and flex-walk systems, energy-storing feet, multi-axial ankle/feet, or dynamic response feet. Finally, a K4 level indicates an individual has the ability for prosthetic ambulation beyond basic locomotion and that exhibits high impact, stress, or energy levels, such as children, active adults, or athletes.

K4 level prostheses are any appropriate ankle-foot systems deemed appropriate for that specific individuals (American Academy of Orthotists & Prosthetists).

In terms of functionality, these ankle-foot devices may be passive, energy-storing, or hydraulic. The differences in these prostheses may affect an amputee's gait patterns by altering the forces directed on the intact limb. Conventional ankle-foot devices produce a negligible amount of energy generation during push-off because the foam of the foot absorbs energy during mid-stance just before push-off; thus, a major source of energy generation from stance phase is lost (Gitter, Czerniecki, & DeGroot, 1991). Energy-storing foot-ankle devices increase energy generation during push-off when compared to a conventional foam foot, but the pattern and magnitude of the knee and hip power outputs are similar to a conventional foam foot (Gitter et al., 1991). This is due to the spring efficiency of the foot used in these devices. Energy-storing prosthetic feet are passive mechanical springs that absorb energy during mid-stance and translate that energy into push-off (Gitter et al., 1991). However, an increase in energy absorption is due to a loss of energy elsewhere the body as there is no external energy input during ambulation (Gitter et al., 1991). The hydraulic ankle-foot device reduces the braking effect during stance by creating a smoother and more rapid progression of the center of pressure beneath the foot and the center of mass of the body (De Asha et al., 2014). Additionally, mean minimum toe clearance increases in the prosthetic limb due to a reduced stance knee flexion and an increased swing-limb hip flexion (Johnson, De Asha, Munjal, Kulkarni, & Buckley, 2014). The hydraulic prosthesis itself is situated in a relatively dorsiflexed position during toe-off and swing phase (Johnson et al., 2014).

Furthermore, the use of a hydraulic ankle-foot device results in an increase in freely chosen walking speed and step length (De Asha et al., 2014; Johnson et al., 2014).

In addition to the type of ankle-foot device an amputee uses, the mass of the footwear can alter the gait pattern of an amputee. Footwear mass alters the swing phase symmetry in particular (Donn, Porter, & Roberts, 1989). Contrary to common belief, lighter weight footwear is not always better as it depends on the preference of the individual. When preference is taken into account, symmetry increases in all parameters (Donn et al., 1989).

Effect of Time Since Amputation

The time since amputation has an effect on an amputee's ability to ambulate. With a greater amount of time that has passed since rehabilitation, balance and coordination improve and gait patterns increase their adaptability. With reference to balance and coordination, overall balance improves, as does the utilization of somatosensory input (Barnett, Vanicek, & Polman, 2013a). Directional control and endpoint center of gravity excursion also increases (Barnett et al., 2013a). With regards to intact limb loading, mean and peak vertical ground reaction forces lessen with more time since amputation (Pruziner et al., 2014). However, even after two years since amputation, the loading forces are still much larger for an amputee compared to an able-bodied individual (Pruziner et al., 2014). The advancements an amputee makes during the time from amputation reflect a greater learning of how to control the prosthetic limb. However, these advancements are moderated by the natural changes in gait that occur with aging.

Until now, the previously mentioned changes to an amputee's gait pattern have concentrated on the differences between able-bodied individuals and amputees or the differences between the intact and prosthetic limbs of the amputee. However, it is critical to explore how these changes affect an amputee's ability to ambulate around in their environment and how they adapt to any potential obstacles in their gait paths. Understanding the interaction between an amputee and their environment will enlighten our knowledge on why amputees have a 50% chance of falling within a year (Miller et al., 2001) compared to 18% for young able-bodied adults (ages 20-45), 21% for middle-aged able-bodied adults (ages 45-65), and 35% for community dwelling able-bodied adults 65 years of age and older (Talbot, Musiol, Witham, & Metter, 2005).

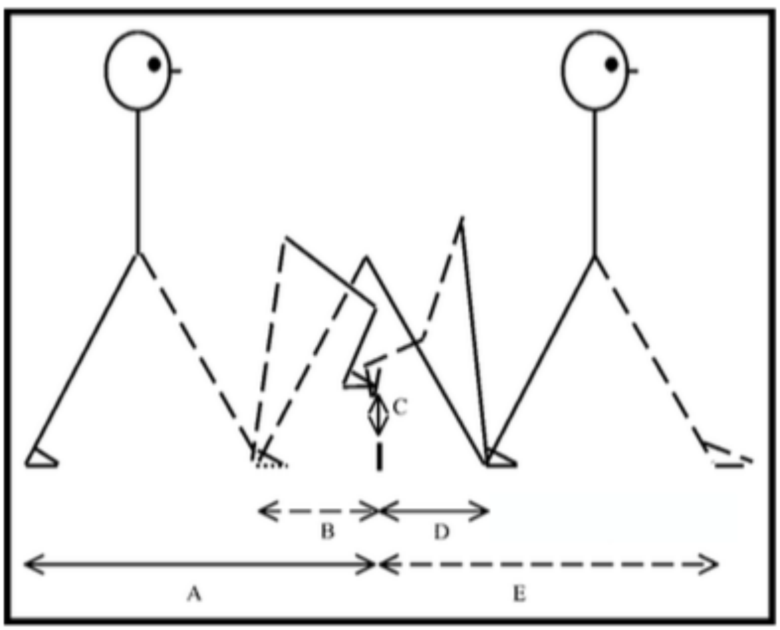
Obstacle Avoidance in Healthy Able-bodied Individuals

Obstacles are a naturally occurring part of our daily environment when defined as any physical object that requires the modulation of an individual's current gait pattern. Obstacles may be stationary and allow for a slow and early adaptation to the gait pattern such as stairs, curbs, and puddles. Obstacles may also be dynamic and require a sudden adaptation, such as a ball or an animal running into the predetermined gait path. High success rates of obstacle avoidance are due to an individual's ability to modify their limb trajectory within a step cycle (Patla, Prentice, Robinson, & Neufeld, 1991). Healthy able-bodied individuals contact obstacles 1-2% of the time (Rhea & Rietdyk, 2007; Rietdyk & Rhea, 2006) with the trail foot as the contactor 67-100% of the time (Heijnen, Romine, Stumpf, & Rietdyk, 2014; Mohagheghi, Moraes, & Patla, 2004; Rhea & Rietdyk, 2007; Rhea & Rietdyk, 2011, 2011; Rietdyk & Rhea, 2006) When crossing an obstacle, able-bodied individuals typically lead with their dominant limb (Chen, Ashton-Miller, Alexander, & Schultz, 1991; Sparrow, Shinkfield, Chow, & Begg, 1996).

There are two main measurements to analyze obstacle-crossing behaviors: limb elevation and foot placement. With unobstructed gait, the vertical displacements of specific anatomical landmarks are similar from stride to stride in able-bodied individuals (Winter, 1991). The heel reaches its highest elevation just after toe-off at the beginning of swing phase and descends downward for the subsequent heel strike (Winter, 1991). In comparison, the toe has two vertical displacement peaks. The first is shortly after toe-off and the other is concurrent with heel contact (Winter, 1991). However, in obstacle crossing, these vertical displacements change with the largest vertical displacement occurring with the toe for the lead limb and the heel for the trail limb. In the obstacle crossing literature, there is debate over which marker most accurately displays obstacle clearance ability (Sparrow et al., 1996). It may be suggested that the heel clearance is used for lead limb and toe clearance is used for trail limb. Additionally, movement trajectories of the leading and trailing limb are different (Patla, Rietdyk, Martin, & Prentice, 1996) as the lower limbs are independently controlled during obstacle crossing (Heijnen et al., 2014; Patla et al., 1996; Rhea & Rietdyk, 2011). When crossing an obstacle, the trail foot exhibits a greater vertical displacement than the lead foot (Sparrow et al., 1996). Foot clearance of the obstacle is due to an increased joint flexion of the swing limb particularly at the knee and the hip (Patla et al., 1996; Patla & Rietdyk, 1993).

Foot placement has been argued as a control strategy for negotiating obstacles (Patla et al., 1996). This strategy must take into consideration that the lead foot may be visually guided during traverse, whereas the trail limb cannot (Patla et al., 1996). In young healthy adults, obstacle contact is associated with placement of the trail foot (Chou & Draganich, 1998; Patla & Greig, 2006) with decreases in distance between the trail foot placement and the obstacle relating to increases in trail foot contact with the obstacle (Chou & Draganich, 1998). In all ages, trail foot placement before the obstacle across multiple trials is consistent in order to allow space and time for the trail foot to successfully clear the obstacle (Muir, Haddad, Heijnen, & Rietdyk, 2015).

Figure 1. Limb Elevation and Foot Placement Variables.



The solid line represents the lead limb and the dashed line represents the trail limb. The variables identified are: (A) lead-limb toe obstacle distance, (B) trail limb toe-obstacle distance, (C) lead limb heel elevation and trail limb toe elevation, (D) lead limb heel-obstacle distance, and (E) trail limb heel-obstacle distance. (Adapted from Said et al., 2005)

Obstacle Height

Adaptations of the limb trajectory in reference to obstacle height are evident at toe-off (Patla & Rietdyk, 1993). However, fine adjustments to trajectory are adapted during the swing phase (Patla et al., 1996). With increases in height of the obstacles, trail limb clearance increases and lead foot crossing time increases (Sparrow et al., 1996). However, obstacle-crossing distance is uninfluenced for either the lead or trail limb (Sparrow et al., 1996). Additionally, the percentage of normalized step, or the stride length at which the obstacle is crossed, is unaffected by obstacle height (Sparrow et al., 1996). With increasing heights, the lead foot crosses progressively earlier in the stride (Sparrow et al., 1996). One interpretation of this earlier lead-foot crossing is that as the height of the obstacle increases, the lead foot needs more time to successfully navigate the obstacles (Sparrow et al., 1996). Another interpretation is that a motor control strategy is initiated to keep the timing and positioning of the trail limb relative to the obstacle constant in order to reduce variation in trail foot placement and maintain stability (Sparrow et al., 1996). Crossing speed significantly decreases and foot clearance increases with increasing heights, but overall approach speed remains unaffected (Chen et al., 1991). Additionally, the horizontal braking impulse increases with increasing height as to slow the forward velocity of progression and allow for safer adaptations to the current obstacle (Patla et al., 1991).

Changes with Age

Older adults require the ability to safely ambulate and negotiate obstacles in order to maintain their independence and quality of life. Thirty-four percent of falls in older adults occur as a result of a trip with the majority of these falls occurring during obstacle negotiation (Berg, Alessio, Mills, & Tong, 1997). Because of this, older adults exhibit a more careful strategy of obstacle crossing. Older adults cross obstacles at a slower speed, with a shorter step length, and a shorter obstacle-heel strike distance (Chen et al., 1991; Muir et al., 2015). The shortened step length is a result of the lead toe being raised more vertically after toe-off, which leads the foot to extend beyond the landing position during swing phase, thus overshooting before retracting backwards to the final landing position (Muir et al., 2015). This lead foot overshoot becomes progressively increased with aging (Muir et al., 2015). Additionally, they cross the obstacle so that it is 10 percent further forward in their obstacle-crossing step (Chen et al., 1991). Furthermore, head angle is progressively lower with increasing age. This may be an attempt to gather more visual information during approach (Muir et al., 2015).

Visual Information

Visual information is used to continuously control locomotor tasks and create adaptations to gait (Patla, 1998). Visual information about stationary obstacles is usually gathered two steps ahead of the obstacle and used in a feed-forward manner (Patla & Vickers, 2003). An obstacle must be in sight during the approach to form a memory that can successfully guide the action (Heijnen et al., 2014). An individual may successfully cross an obstacle provided they have sight of the obstacle during approach until at least three steps beforehand (Mohagheghi et al., 2004) with the top and lower edge fully visible (Heijnen et al., 2014; Rhea & Rietdyk, 2011). Removing sight of the obstacle five steps before crossing lowers the success rate to 50% as foot placement becomes inaccurate for successful obstacle avoidance (Patla & Greig, 2006).

When stepping over an obstacle, the lead limb is visible in the lower visual field, creating online visual information to control the trajectory of the lead limb (Mohagheghi et al., 2004; Patla, 1998; Patla et al., 1996; Rietdyk & Rhea, 2006, 2011). In contrast, the trailing limb and the obstacle are not visually available when the trail limb is crossing the obstacle, thus the obstacle presents a greater challenge and the risk of falling increases (Hill et al., 1997; Patla et al., 1996). Because of the greater challenge the obstacle presents, a need for visual exteroceptive input arises.

Visual exteroceptive input refers to the information visually gathered from the external environment. Visual exteroceptive input is said to largely influence the gait characteristics when stepping over an obstacle. Individuals are capable of accurately judging dimensions of an obstacle and are able to proactively change their limb trajectory to safely cross it (Patla & Rietdyk, 1993). Visual exteroceptive input in obstacle crossing may be observed, such as the height and width of the obstacle, or inferred, such as the fragility or rigidity of the obstacle (Patla et al., 1996). However, the shape of the obstacle does not affect the limb trajectory if the height and the width of the obstacle are similar (Spaulding & Patla, 1991). When going over a fragile obstacle, toe clearance, vertical position of the hip, and hip vertical velocity increases to minimize the risk of tripping (Patla et al., 1996). Toe clearance variability is increased in the trailing limb when compared to the lead limb potentially because there is a lack of visual exteroceptive information and the trail limb has less time to adapt its trajectory after toe-off (Patla et al., 1996). This visual exteroceptive input may also be defined as obstacle memory. Obstacle memory is using obstacle size and position to modify the limb trajectory over an obstacle in subsequent trials (Heijnen et al., 2014; McVea & Pearson, 2006, 2007). When comparing limb trajectories of fragile and solid obstacle clearance, initially at toe-off they look similar, but then start to deviate: first, in response to the height and secondly, in response to the fragility in which the toe clearance is even higher than the rigid obstacle (Patla et al., 1996). Visual exproprioceptive information is used more than visual exteroceptive information to alter the limb trajectory when crossing an obstacle (Rhea & Rietdyk, 2007). Exproprioception (limb position relative to the obstacle) is compromised more for the trail limb than the lead limb because of the lack of online visual information, thus viewing the obstacle during approach is more critical for the trail limb (Heijnen et al., 2014).

After an obstacle is contacted, gait is adapted to prevent future contact with obstacles (Rhea & Rietdyk, 2011). When a limb contacts an obstacle, toe clearance and peak toe elevation are ipsilaterally increased for multiple subsequent trials (Rhea & Rietdyk, 2011) to ensure clearance over the obstacle and the risk of tripping and injury is minimized (Patla et al., 1996). When the trail limb contacts an obstacle, the subsequent trail limb clearance is increased by 75%, but the lead limb clearance does not change (Heijnen, Muir, & Rietdyk, 2012). This supports the idea that during adaptive gait, the limbs are independent from one another as the motion and/or feedback of one limb is not used to control the opposite limb (Heijnen et al., 2012; Patla et al., 1996; Rhea &

Rietdyk, 2011). Additionally, there is evidence to suggest this increased clearance during adaptive gait lasts for at least 30 subsequent obstacle crossings (Heijnen et al., 2012).

Obstacle Avoidance in Amputees

In able-bodied healthy gait, an individual should have the ability to lead with whichever foot encounters the obstacle first. In contrast, an amputee has an inherent asymmetrical gait due to missing musculature and joints and must compensate for this by typically choosing a preferred lead limb. Trans-tibial amputees prefer to cross obstacles with their prosthetic limb as the lead (Vrieling et al., 2007). With trans-tibial amputees, knee flexion decreases along with ankle dorsiflexion in the prosthetic limb during obstacle crossing (Hill et al., 1997). When compared to able-bodied individuals, amputees exhibit a reduced approach velocity, reduced foot placement distance before and after the obstacle and reduced foot clearance over it, and reduced lead-limb knee flexion during the step following crossing (Buckley, De Asha, Johnson, & Beggs, 2013). Reducing foot placement before the obstacle in order to become closer to it seems to be a strategy to guarantee that the heel of the lead limb clears the obstacle before being lowered to the ground (Buckley et al., 2013). Despite placing the heel closer to the obstacle before crossing, the lead limb was positioned closer to the obstacle after crossing as well. This may be a result of the desire to attain a limb/foot angle and orientation at the instant of landing that minimizes loads placed on the prosthetic limb as may be seen by the reduced knee flexion in the lead limb following crossing (Buckley et al., 2013). Because of the changes in foot placement, the foot is in a different part of swing when crossing the obstacle and may explain the reduced foot clearance in amputees (Buckley et al., 2013). Lead-limb toe clearance during obstacle crossing for amputees is approximately half of when able-bodied individuals exhibit (Buckley et al., 2013), regardless if they lead with the intact or the prosthetic limb (Hill et al., 1997). Trans-tibial amputees are unable to increase toe clearance by dorsi-flexing the foot during swing as dorsiflexion is limited by the prosthesis (Hill et al., 1997). Additionally, the posterior edge of the socket limits knee flexion (Hill et al., 1997).

Foot placement and toe clearance variability are not different between the intact and prosthetic limb when leading (De Asha & Buckley, 2015). Toe clearance is not different between the intact and the prosthetic limb; however, it is higher when amputees lead with their preferred limb (De Asha & Buckley, 2015). Trans-tibial amputees should be encouraged to step over obstacles with their preferred limb in contrast to leading with the prosthetic limb (Vrieling et al., 2007) or the intact limb (Barnett, Vanicek, & Polman, 2013b) as laterality may be important in successful and safe obstacle crossing (De Asha & Buckley, 2015). Laterality is as important in lower-limb amputees as it is in able-bodied individuals (De Asha & Buckley, 2015).

Effects of Obstacle Heights on Amputee Gait

Hip and knee flexion increases for the swing and stance limb as obstacle height increases. However, hip elevation only increases with the swing limb (Hill et al., 1997). The posterior shell of the prosthetic socket limits the prosthetic limb's swing knee flexion and increases the lead intact limb's dorsiflexion which leads to increased swing prosthetic foot angle and increased intact limb stance ankle plantarflexion (Hill et al., 1997). This limitation of the prosthesis may also contribute to the preference of a prosthetic limb lead as the decreased ability of knee flexion makes it less reliable as a trail limb (Hill et al., 1997). There is no overall preference between the intact or the prosthetic limb being the lead (Hill et al., 1997). However, this may be due to stump length and thus the length of the prosthesis. Those who have no preference in lead limb are the most functional and adaptable walkers and do not have to break stride to successfully clear an obstacle (Hill et al., 1997). Toe clearance is similar for the intact and the prosthetic limbs with each exhibiting similar increases with increases in obstacle height (Hill et al., 1997). Swing hip flexion and knee flexion show the largest magnitude of changes with differences in obstacle height (Hill et al., 1997; Patla & Rietdyk, 1993). There are no differences between the intact and the prosthetic limb in terms of swing hip flexion, trunk angle, hip elevation, stance limb hip and knee flexion, and toe-off distance from obstacle (Hill et al., 1997). The differences between the intact and the prosthetic limb lie within swing knee flexion (mechanically limited in the prosthetic side and thus leading to increased intact stance ankle plantarflexion), ankle flexion (decreased in the intact to compensate for reduced swing knee flexion in prosthetic), and foot angle (Hill et al., 1997). The method and effort with which an amputee crosses an obstacle depends on their strength, flexibility, and prosthesis.

The prosthetic limb suffers a loss of sensory feedback after an amputation. In order to compensate for this loss, kinesthetic information from other body segments and visual feedback must be synthesized together to control the prosthetic limb’s trajectory. This is an intersegmental dynamics strategy (Hill et al., 1999). Limb-elevation strategies are different for each limb regardless of the kinematic patterns showing major similarities

(Hill et al., 1997). The knee typically modulates limb elevation strategies as obstacle height changes. However, in amputees, the rotational work at the hip is the modulator in the prosthetic side. This change in limb-elevation strategy may be due to the aforementioned physical limitation of knee flexion by the posterior shell of the prosthetic. However, another reason may be the increased knee flexor power generation to control knee flexion may result in instability and discomfort in the residual stump (Hill et al., 1999).

Changes Throughout the Rehabilitation Process

Individuals with recent lower-limb amputation create strategies to improve obstacle crossing, gait initiation, and gait termination (Vrieling et al., 2009). In the early stages of rehabilitation, new amputees are not familiar with prosthesis use and have not yet created adjustment strategies for walking with a prosthetic device. Through rehabilitation, gait and balance training allows amputees to create these strategies to compensate for the missing musculature. With obstacle crossing, amputees increase their success rate of clearing obstacles, gait velocity, and swing knee flexion of the prosthetic limb during the rehabilitation process (Vrieling et al., 2009). Following discharge from rehabilitation, walking velocity increases and a greater reliance is placed on the intact limb (Barnett et al., 2013b). During swing phase, peak knee flexion and peak knee power absorption are increased in intact lead limb compared to the prosthetic lead limb (Barnett et al., 2013b). In contrast to gait and balance training programs, there is no previous literature that has examined the changes of the amputee's obstacle crossing behavior throughout rehabilitation while utilizing obstacle crossing within their training or rehabilitation programs. The changes due to gait and balance training programs throughout the rehabilitation process suggest a motor learning effect, which will be covered more in depth in the next section.

Motor Learning

Motor learning covers a wide spectrum of concepts in which a person learns or re-learns a skill or task. This learning cannot be directly observed, but it can be inferred via a person's performance. Performance is observable behavior and refers to the execution of a skill at a specific time and in a specific situation (Magill, 2001). Motor learning may then be defined as a set of processes associated with practice or experience leading to relatively permanent gains in the capability for a skilled performance (Schmidt & Lee, 2014). As skill learning takes place, four performance characteristics emerge: (1) Improvement – performance of the skill shows improvement over time; (2) Consistency – early in learning performance is quite variable, but then becomes increasingly more consistent; (3) Persistence – as the person progresses in learning the skill, the improved performance capability lasts over increasing periods of time, which refers to the relatively permanent improvement in performance; (4) Adaptability – the improved performance is adaptable to a variety of performance context characteristics (Magill, 2001).

Learning may be assessed by observing practice performance, conducting retention and transfer tests, and assessing coordination dynamics. Observing practice performance is typically conducted via analysis of illustrated performance curves. Unfortunately, practice performance may misrepresent the amount of learning actually achieved by over- or under-estimating the performance. Therefore, retention and transfer tests are important in assessing learning. Retention tests are used to determine the degree of permanence or persistence of the performance improvements an individual can exhibit after a period of rest. However, it can be argued that the most important assessment of learning is a transfer test as transfer tests examine the ability of an individual to apply the learning to a similar situation. Transfer tests require the participant to perform a novel skill variation by changing the novel context characteristics, such as the availability of augmented feedback, the physical environment, or the personal characteristics (Schmidt, 1982).

The transfer of learning is defined as the influence of previous experiences on performing a skill in a new context or on learning a new skill (Magill, 2001; Schmidt & Lee, 2014). Transfer of learning is one of the most universally applied principles in education and rehabilitation. In the rehabilitation clinic, this principle forms the basis for the systematic development of protocols that therapists implement with patients. This transfer principle helps to understand the underlying processes of learning and control of motor skills and provides the basis for assessing the effectiveness of practice conditions.

Learning transfer may be positive, negative, or zero (no influence). Positive transfer is said to occur when the previous experiences or practice facilitates the performance of a skill in a new context of the learning of a new skill (Magill, 2001; Schmidt & Lee, 2014). Positive transfer occurs for two reasons. The first reason is that the components of the skills performed in both the practice context and the transfer context are similar. The more similar the component parts of the two skills or performance contexts are, the greater the amount of positive transfer occurs. This leads to the Identical Elements Theory proposed by Thorndike. The Identical Elements Theory states that “elements” are general characteristics of a skill or performance context, such as the purpose of the skill or the attitude of the person performing the skill, or specific characteristics of the skill, such as the components of the skill being performed (Thorndike, 1914). In addition, the Identical Elements Theory is considered to include mental processes that share the same brain cell activity as the physical action (Magill, 2001). The second reason positive transfer occurs is because of the similarities between the amounts and typed of learning processes required. As the similarity of the cognitive processes that are required by the two skills of two performance situation increases, the amount of positive transfer increases as well. This leads to the Transfer-Appropriate Processing Theory which discusses the importance of the similarity between the learning or performance cognitive processes required by the two performance situations. In this theory, two components of positive transfer are crucial: the cognitive processing activity a person must do to be successful in performing the task, and the similarity between that activity and the activity required during the training experience (Lee, 1988). Examples of transfer-appropriate processing include when the transfer task requires a person to engage in problem-solving activity rapid decision making, application of rules, attention control, and the simultaneous performance of two or more tasks (Magill, 2001). It is important to note that the training and transfer tasks do not need to have similar movement components, but do have to each involve similar cognitive processing demands to each other.

In stark contrast to positive transfer, negative transfer occurs when the previous experiences or practice hinders the performance of a skill in a new context of the learning of a new skill (Magill, 2001; Schmidt & Lee, 2014). Negative transfer appears to be rare and temporary. However, it occurs when an old stimulus requires a new, yet similar response, such as the environmental context characteristics of two performance situations are similar, but the movement characteristics are different (Magill, 2001). Some situations in which negative transfer may arise are when a change in the spatial locations or the timing structure of the movement. Negative transfer is typically only influential in the early learning stage. This is important for the practitioner in clinical settings as the participant may become discouraged in their interest and motivation to pursue the learning of a new skill or a new way to perform a well-learned skill. The practitioner can give specific instructional attention to the aspects of the training that are most susceptible to negative transfer to help the participant overcome their discouragement and lack of motivation (Magill, 2001).

The key to enhancing the learning of a new motor skill is the feedback provided to the participant. Feedback is the information about performance or errors that the learner can use for making future decisions (Schmidt & Lee, 2014). There are two main types of feedback to enhance learning. The first is task-intrinsic feedback which is the sensory-perceptual information that is a natural part of performing a skill (Magill, 2001). Each of the sensory systems can provide this type of feedback, but the most common are visual, proprioceptive, auditory, and tactile. The second is augmented feedback which is performance-related information in addition to task-intrinsic feedback that influences how the participant directs their attention (Magill, 2001). Augmented feedback may be given while the movement is in progress, also known as concurrent augmented feedback, or after the skill has been performed, known as terminal augmented feedback. There are two categories in which augmented feedback may be divided: knowledge of results and knowledge of performance. Knowledge of results is externally presented information about the outcome of performing a skill or about achieving the goal of the performance (Schmidt & Lee, 1999). Essentially, it describes performance outcomes, such as EMG tracings and contact location for a dart throw, or it informs the participant whether or not the goal was achieved, such as a yes or no on a 30-degree knee flexion task. In contrast, knowledge of performance gives information about the movement characteristics that led to the performance outcome, such as limb position and angle relative to the body (Schmidt & Lee, 1999).

There is some debate on whether to give augmented feedback information of the errors or of the correct aspects of performance (Salmoni, Schmidt, & Walker,1984; Wulf & Shea, 2004). Some argue that error information is more effective for encouraging skill improvement as the experience in error correcting by the participant is important in skill acquisition (Annett, 1959; Lintern & Roscoe, 1980). Others argue that the information indication that the participant performed certain aspects correctly tells them they are on task in learning the skill, and therefore, encourages the participant to keep trying (Magill, 2001). In sum, it seems that both forms are of importance and it depends on the task and the participant as to which is given and when. Error-related information works to better facilitate skill acquisition, whereas information about the correct aspects motivates the participant to continue (Magill, 2001).

Regardless of the type or timing of augmented feedback, augmented feedback is important in skill acquisition. Augmented feedback helps to facilitate the achievement of the goal of the skill by providing information about the success of the skill in progress or just completed. This allows the participant to determine whether or not what they are doing is appropriate for performing the skill correctly. Augmented feedback also motivates the learner to continue to strive toward a goal by allowing them to compare their performance to the performance goal (Magill, 2001).

The motor learning principle of attentional focus has also received much attention over the past two decades. Evidence has amassed for the benefits of an external focus on the intended movement effect versus an internal focus on one’s own body movements. An external focus of attention has the most beneficial effects in acquisition, learning, retention, and transfer of a motor skill (Shea & Wulf, 1999). In numerous studies, directing attention towards an external focus results in increased movement effectiveness, such as balance and accuracy, and movement efficiency, such as muscle activity, force production, and speed and endurance (Wulf, 2013). Consequently, an internal focus of attention on an individual’s own movements may actually degrade the learning of a new skill and disrupt the execution of an automated skill (Wulf & Weigelt, 1997).

Apart from movement effectiveness and efficiency, external focus induced by instructions can affect movement coordination on a larger scale, such as whole-body coordination patterns (Wulf, 2013). This may be due to the fact that when focusing internally on body movements, semi-independent body segments are linked and the motor system becomes constrained (Ford, Hodges, Huys, & Williams, 2009). With an external focus of attention, those constraints are released and allow for functional variability (Muller & Loosch, 1999), in which the motor system automatically adjusts the various degrees of freedom to achieve the desired outcome (Miller et al., 2001). With virtual reality immersion, having an amputee focus externally on the obstacle they are crossing may produce the best benefits for improved obstacle avoidance. By focusing on the obstacle, the underlying automatic control processes may operate freely and produce a learning advantage (McNevin, Wulf, & Carlson, 2000).

Virtual Reality

Gait is an important aspect of rehabilitation following lower limb amputation. Most individuals with lower limb amputations are able to regain the ability to walk. However, their gait deviates from an able-bodied person after rehabilitation is completed. Clinicians have sought to improve the methods of rehabilitation and create new and more advanced rehabilitation techniques. Virtual reality has become a very plausible means of rehabilitation and is becoming more commonly used in conjunction with traditional rehabilitation techniques. The use of virtual reality is defined as a simulation of real world environment that is generated through computer software and is experienced by the user through a human-machine interface (Holden, 2005). It has been suggested that virtual reality systems may enhance skill acquisition and retention by providing task specificity, repetition, and external real-time feedback (Wulf, 2007). Virtual reality-based gait training has been studied and implemented with many clinical populations such as post-stroke (Jaffe, Brown, Pierson-Carey, Buckley, & Lew, 2004), Parkinson's (Mirelman et al., 2010), and multiple sclerosis (Baram & Miller, 2006). Virtual reality training has been beneficial in increasing walking speed, stride length, walking endurance, and obstacle crossing (Baram & Miller, 2006; Jaffe et al., 2004; Mirelman et al., 2010)

The key concepts of integrating virtual reality with rehabilitation use motor learning principles such as repetition, feedback, and motivation. Repetition must be linked to incremental success at a task or goal in order to produce motor learning (Holden, 2005). This may be achieved through feedback about performance success. However, an individual must be motivated for repeated practice in order to see incremental performance successes (Holden, 2005). Thus, repetition, feedback, and motivation are interlinked with one another and are each crucial to successful motor learning and improved performance in a task.

Virtual reality offers an advantage for learning motor skills because it provides participants with all three of the motor learning principles: motivation, repetition, and feedback. Virtual reality experiences can be highly engaging both physically (because it requires the user to interact with obstacles in their virtual environment) and mentally (because it feels like a game as the use is submersed within a different world). This high engagement provides crucial motivation for rehabilitative applications that require consistent, repetitive practice that may otherwise not be engaging within a medical office or rehabilitation clinic. In several studies for upper-body exercise with feedback, virtual reality has shown significant improvement in the movement, use and control, relative to baseline measurements and to other more conventional rehabilitation approaches, such as patient-guided exercise and group physical therapy (Henderson, Korner-Bitensky, & Levin, 2007; Merians et al., 2002). Additionally, virtual reality is an excellent tool for rehabilitation because it allows the researcher or clinician to record and follow minute changes and track a patient's improvement over time.

Gait training on a treadmill has many advantages including continuous data collection occurring within a small physical space. The walking is unimpaired by the environment unless created to do so, thus gait and fall risk may be assessed and gait may potentially re-trained (Darter & Wilken, 2011). Virtual reality may be integrated to provide visual cues, such as optic flow and real-time feedback. Virtual obstacles can also be negotiated. For example, Shema et al. (2014) examined the clinical experience of a challenging virtual obstacle course environment in which obstacles appeared on an outdoors pathway on either the left or right side causing the user to plan ahead and adapt their steps to avoid contacting the obstacles. Obstacles varied in frequency as well as height and size to increase step clearance and step length. Results from this study suggest improved functional mobility and increased obstacle negotiation. Crossing virtual obstacles of a variety of sizes is beneficial in terms of space and time on a treadmill compared to over ground walking. Virtual obstacles have also been examined overground. Heijnen et al. (2014) compared the ability to cross a virtual obstacle marked on the ground via masking tape after removing a real obstacle created from Masonite. Foot placement showed no change between the two conditions. In contrast, obstacle clearance was compromised when the real obstacle was removed, thus suggesting the need for height reference in combination with position reference to successfully navigate an obstacle. Gait training within a virtual reality environment is similar enough to walking overground that changes with the training should carry over from virtual reality to overground (Gates, Darter, Dingwell, & Wilken, 2012). Additionally, able-bodied and trans-tibial amputees both respond in this manner (Gates et al., 2012).

Treadmills may also be used for rehabilitation with trans-tibial amputations. Gait adaptability to obstacle crossing may be assessed using a treadmill with projected visual targets on the surface (Houdijk et al., 2012) or by dropping physical objects on the belt (Hofstad et al., 2009). The downside to dropping physical objects is the individual does not have adequate time to use the visual feedback to implement a successful obstacle avoidance strategy and adapt their stride, as the response by the amputee is bilaterally delayed and reduced (Hofstad et al., 2009). Additionally, an amputee exhibits high failure rates when under pressure to cross an obstacle because a short step strategy is used to avoid the obstacle (Hofstad et al., 2006). Thus, a reduced response time will most likely produce a failure to avoid the obstacle. Virtual reality obstacle crossing gives an amputee time to incorporate visual feedback to adjust stride and implement an obstacle avoidance strategy.

Current Gaps in the Literature with Regards to This Thesis

Previous literature has lacked any quantified normative data of able-bodied individuals in response to a virtual environment or virtual obstacle crossing. As virtual reality is quickly becoming thought of as a plausible rehabilitation technique, obtaining quantitative normative data of a virtual reality to real world transfer task is of crucial importance for comparison to clinical populations seen in rehabilitation clinics.

Additionally, previous studies have not been completed with repeated virtual obstacles on a pathway with feedback about their performance and results. Based on the motor learning literature, feedback is crucial to motor learning along with repetition, and virtual obstacle training on a treadmill with instantaneous feedback may provide more successful obstacle clearances and more rapid motor learning. Furthermore, the virtual obstacle crossing is a fun and engaging environment, which increases the motivation to repeatedly step over these obstacles to ensure learning.

Finally, determining if the learning completed within a virtual obstacle environment will transfer to real world over-ground obstacles is crucial. For virtual rehabilitation to be effective, it is important that the learned obstacle avoidance is able to transfer to everyday obstacle situations. This transfer will enhance an individual's ability to cross those obstacles and ultimately lead to a decreased risk of falling due to poor obstacle avoidance ability.

CHAPTER III

OUTLINE OF PROCEDURES

Participants

Twenty young healthy adults (22.45 ± 3.65 years) and twenty older healthy adults (55.60 ± 5.98 years) were recruited to participate. All participants met the inclusion criteria, which included normal or corrected to normal vision, no cognitive impairment, no current musculoskeletal injuries, and ability to walk 10 minutes without aid.

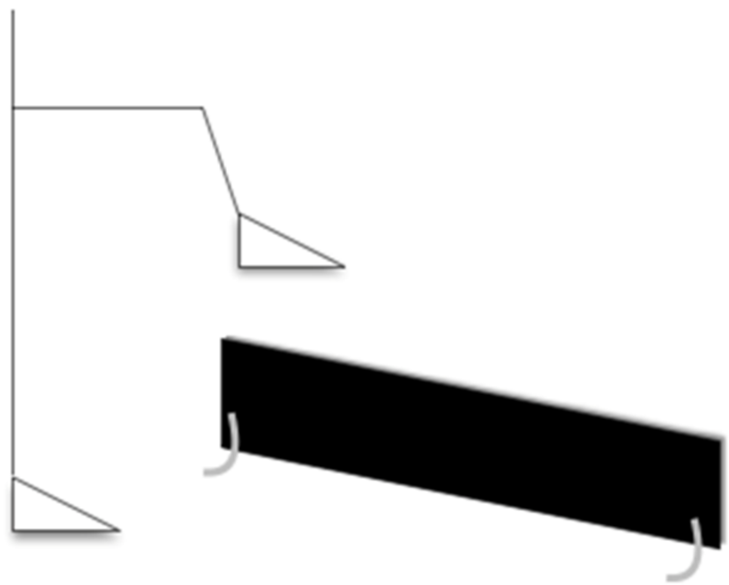
Instrumentation

Kinematic variable data were collected using Qualysis motion capture cameras (Gothenburg, Sweden) while participants walked on a Simbex Active Step treadmill (Lebanon, NH) and overground across the laboratory. A total of 36 markers were used and placed on the shoulders, anterior superior and posterior superior iliac crests, the thigh and shank segment panels, medial and lateral knee and ankles, medial first and lateral fifth metatarsals, and the most anterior superior position on the shoe toe and the calcaneus.

Obstacle Set-up

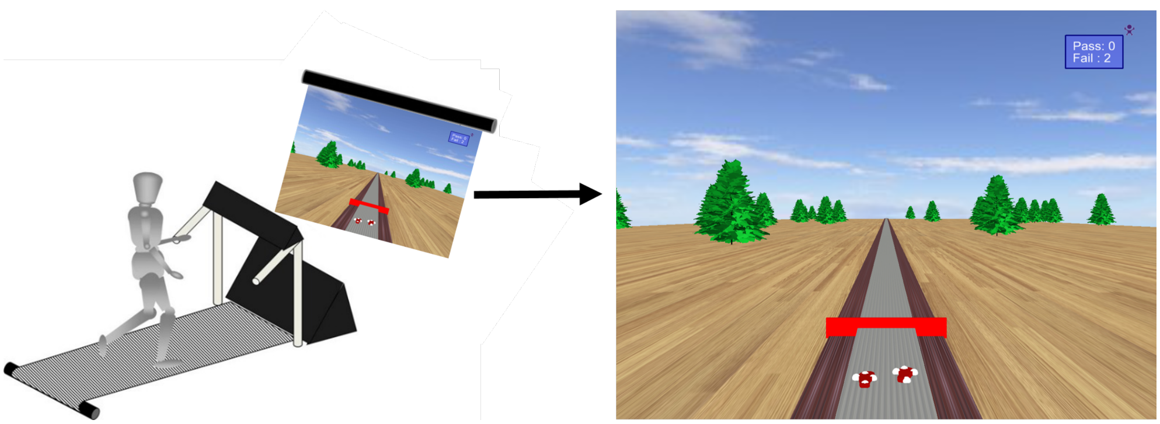
A real environment obstacle was set-up at the halfway mark (4m) along an 8m pathway across the laboratory. The obstacle was made of Masonite board, painted flat lack and designed to easily tip over if contacted. The dimensions for the obstacle were 10 cm high, 100 cm wide, and 0.5 cm deep. A height of 10 cm was selected because it is the average curb height.

Figure 2. Real Environment Obstacle



The virtual environment consisted of a grey and red walking path through a brown landscape with trees and a blue sky (Figure 3). A counter set in the top right gave feedback of obstacles hit and obstacles cleared. A virtual obstacle in the virtual environment was set to appear as 10 cm and on pace with the subject's normal walking speed with approximately 30 seconds in between obstacles to prepare for the obstacle that had to be crossed. The obstacle covered the entirety of the pathway requiring the subject to step over with both feet. The virtual environment also contained virtual feet that moved in real-time and corresponded to the participant's own foot movement by synchronizing with the motion capture system. The toe, heel, medial, and lateral metatarsal markers were white and portrayed on the feet in real-time. If the virtual obstacle was contacted, the specific foot marker that was hit turned red to indicate a collision, while the others that cleared the obstacle remained white. If the foot successfully cleared the obstacle, all markers turned green.

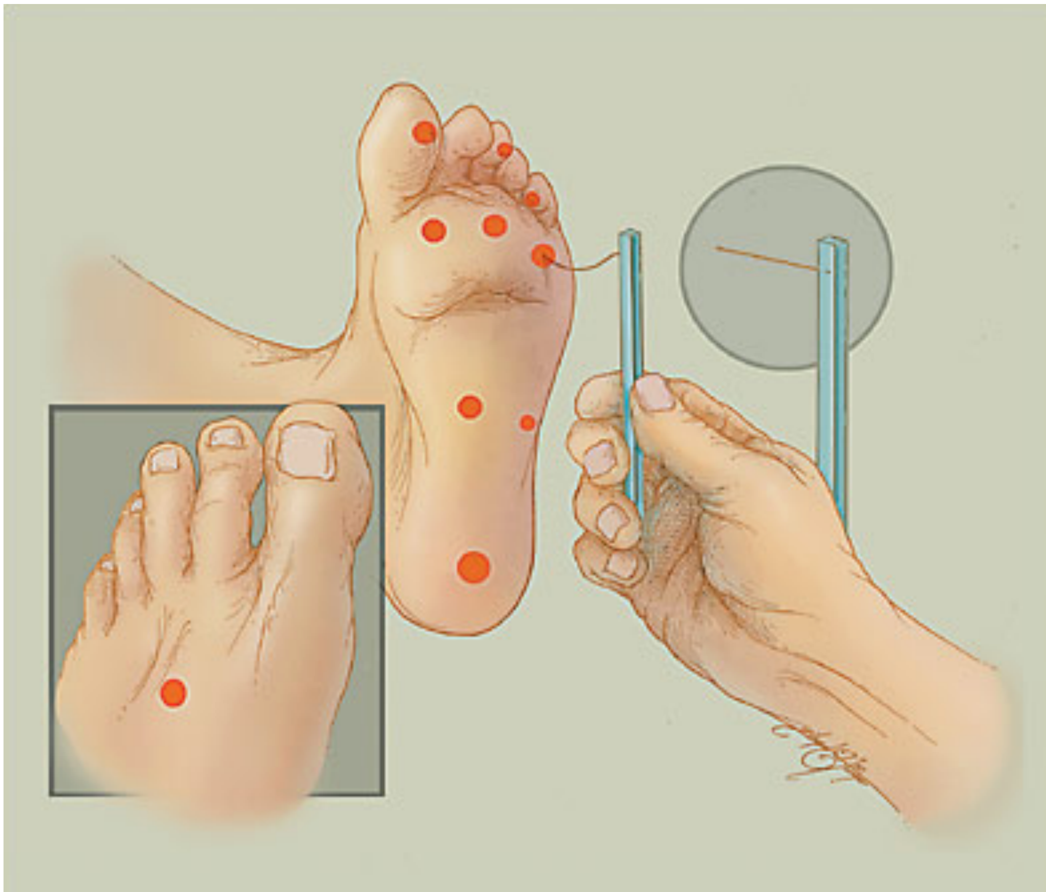
Figure 3. Virtual Obstacle Crossing Environment



Procedure

Participants provided a medical history and consent form. The 5.07 Semmes-Weinstein (10-g) monofilament test was administered for all participants to test for limb sensitivity. The 5.07 monofilament exerts 10g of force and a patient must be able to detect the force on each of the multiple areas of the feet to have full limb sensitivity.

Figure 4. The Semmes-Weinstein (10-g) Limb Sensitivity Test.
(Courtesy of Aring, Jones, & Falko, 2005)



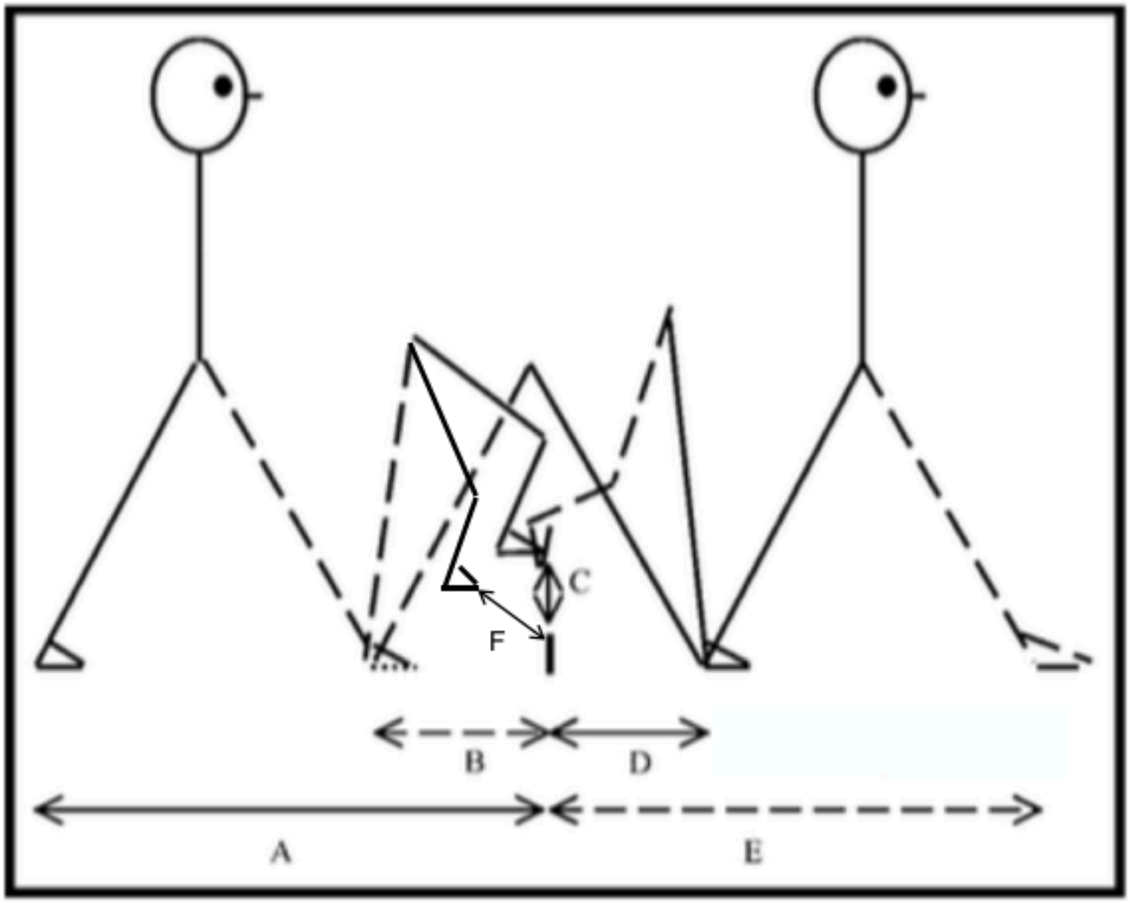
Anthropometric data was collected and participants were put in a safety body harness to be utilized during the virtual obstacle crossing sessions. Retro-reflective markers were then applied to the participants' body.

Normal walking pace was determined on the treadmill by starting at a very slow pace and incrementally increasing speed until the participant decided that speed was their normal pace. Then, the participant was sped up with a fast walking pace and incrementally slowed until they decided that speed was their normal pace. The two speeds were then averaged to determine their normal walking pace.

This study utilized a pre/post design using a real environment over-ground obstacle. Each participant walked over the real environment obstacle after being given standardized instructions, “Walk at your normal pace and safely step over the obstacle with your dominant foot without stopping until you reach the end of the walkway.” They completed 10 consecutive trials. The subject was given 5 minutes of rest and then walked on the treadmill in the virtual obstacle-crossing environment for one session of 15 obstacle crossings, the first 5 being practice trials. After the first session of virtual obstacle crossing, the subject was given a 5-minute rest period before continuing on for a second session of 10 more virtual obstacle crossings. After the virtual obstacle crossing environment was completed, the subject stepped off the treadmill and rested for 5 minutes before walking over the real environment over-ground obstacle 10 more times given the same previously stated standardized instructions. The 5-minute rest breaks were strategically placed to reduce fatigue, and if amputees were to be included in the future, allow the amputee to make any needed prosthesis and sock adjustments.

In the real environment, measures of vertical and radial toe and heel clearance, foot placement before and after the obstacle, peak toe and heel elevation, and the delta scores of the aforementioned variables between sessions were measured (Figure 5). In the virtual environment, peak toe and heel elevation, the delta scores of toe and heel elevation between sessions, and the pass/fail rate were measured. Toe clearance was measured in two ways. First, vertical toe clearance as the distance between the toe marker and the top of the obstacle at the time the toe marker is directly above the obstacle, and second, radial toe clearance as the minimal distance between the toe marker and the top of the obstacle. Similarly, heel clearance was measured in two ways. First, vertical heel clearance as the distance between the heel marker and the top of the obstacle at the time the heel marker is directly above the obstacle, and second, radial heel clearance as the minimal distance between the toe marker and the top of the obstacle. Foot placement was measured as the horizontal distance between the toe marker and the obstacle for before-obstacle foot placement and as the horizontal distance between the heel marker and the obstacle for after-obstacle foot placement. Foot placement was measured before and after the obstacle for both the lead and trail limb. Peak toe elevation and heel elevation for each obstacle was obtained and averaged per obstacle crossing session.

Figure 5. Real Environment Obstacle Foot Placement and Clearance Variables



The solid line represents the lead limb and the dashed line represents the trail limb. The variables identified are: (A) lead-limb toe obstacle distance, (B) trail limb toe-obstacle distance, (C) lead limb heel elevation and trail limb toe elevation, (D) lead limb heel-obstacle distance, (E) trail limb heel-obstacle distance, and (F) radial heel and toe clearance for both limbs. (Adapted from Said et al., 2005)

Ratings of Perceived Exertion (RPE) was determined throughout the different tasks. Participants were asked to rate their perceived exertion after walking over the real environment over-ground obstacles after trial 1 and 10 for each session. With the virtual obstacle crossing, participants were asked their RPE during the first session after trial 1, 10, and 15, and during the second session after trial 1 and 10.

Data Collection and Analysis

The motion capture data (Qualysis, Gothenburg, Sweden) was exported to Visual3D (C-Motion, Germantown, MD) to create a file to import into MATLAB (Mathworks, Natick, MA) to determine foot placement before and after the obstacle, vertical and radial heel clearance for the lead limb, vertical and radial toe clearance for the trail limb, peak toe and heel elevation. Microsoft Excel was used to calculate deltas scores for each of the aforementioned variables. SPSS (IBM, Armonk, NY) was used to run all statistical tests. Alpha level was set a priori to 0.05.

To address hypothesis 1, a 2 x 2 (age by time) repeated measures MANOVA for pass rate and peak elevation of the lead and trail limb in the virtual environment was run. To address hypothesis 2, two separate 2 x 2 (age by time) repeated measures MANOVAs were run: one for foot placement (position of the both lead and the trail limb before and after the obstacle) and one for foot clearance (peak elevation and radial clearance for both the lead and trail limbs) in the real environment. Finally, to address hypothesis 3, Pearson's correlations were run for the delta scores of each variable of foot placement and clearance in the real environment against the number of successful obstacle passes in the virtual environment.

CHAPTER IV

MANUSCRIPT

Introduction

Obstacles are a naturally occurring part of our daily environment when defined as any physical object that requires the modulation of an individual's current gait pattern. Obstacles may be stationary and allow for a slow and early adaptation to the gait pattern such as stairs, curbs, and puddles. Obstacles may also be dynamic and require a sudden adaptation, such as a ball rolling or an animal running into the gait path. The ability to avoid such obstacles is a crucial part of safe ambulation. High success rates of obstacle avoidance are due to an individual's ability to modify their limb trajectory within a step cycle (Patla et al., 1991). Healthy able-bodied individuals contact obstacles 1-2% of the time (Rhea & Rietdyk, 2007; Rietdyk & Rhea, 2006), with the trail foot as the contactor 67-100% of the time (Heijnen, Romine, Stumpf, & Rietdyk, 2014; Mohagheghi, Moraes, & Patla, 2004; Rhea & Rietdyk, 2007; Rhea & Rietdyk, 2011, 2011; Rietdyk & Rhea, 2006). Contact of the trail limb with the obstacle is higher than the lead limb due to the observation that the movement trajectories of the leading and trailing limb are different (Patla, Rietdyk, Martin, & Prentice, 1996)—as the lower limbs are independently controlled during obstacle crossing (Heijnen et al., 2014; Patla et al., 1996; Rhea & Rietdyk, 2011)—and the fact that online visual information can be used to guide the lead limb, but not the trail limb due to it being outside the visual field (Patla, 1998).

Thirty-four percent of falls in older adults occur as a result of a trip, with the majority of these falls occurring during obstacle negotiation (Berg, Alessio, Mills, & Tong, 1997). Because of this, older adults exhibit a more careful strategy of obstacle crossing. Older adults cross obstacles at a slower speed, with a shorter step length, and a shorter obstacle-heel strike distance (Chen et al., 1991; Muir et al., 2015). The shortened step length is a result of the lead toe being raised more vertically after toe-off, which causes the foot to extend beyond the landing position during swing phase, thus overshooting before retracting backwards to the final landing position (Muir et al., 2015). This lead foot overshoot becomes progressively increased with aging (Muir et al., 2015). Additionally, older adults cross the obstacle so that it is 10 percent further forward in their obstacle-crossing step (Chen et al., 1991). Understanding the typical obstacle crossing strategies adopted by older adults is useful in order to develop fall-prevention programs aimed at enhancing obstacle avoidance ability.

Obstacle avoidance training programs have been developed in the real world with some success. Basille et al. (2003) discovered that patients post-stroke increased walking velocity and distance after four weeks of training with obstacles along a walkway. Similarly, Jaffe et al. (2004) concluded that patients post-stroke with hemiplegia showed increases in gait velocity, stride length, walking endurance, and obstacle clearance capacity after two weeks of obstacle avoidance training. However, these training programs often require physical space, materials, and human resources which may not be available in clinical settings.

To this end, clinicians have begun employing new and more advanced rehabilitation techniques, one of which is virtual reality. The use of virtual reality is defined as a simulation of real world environment that is generated through computer software and is experienced by the user through a human-machine interface (Holden, 2005). From a motor learning perspective, it has been suggested that virtual reality systems may enhance skill acquisition and retention by providing task specificity, repetition, and external real-time feedback (Wulf, 2007). Virtual reality gait training on a treadmill offers the advantage of providing a visually challenging task (i.e., ambulation over multiple virtual obstacles) in a relatively small and safe environment, especially if a harness system is used. Virtual reality training has been beneficial in increasing walking speed, stride length, walking distance, and obstacle crossing (Baram & Miller, 2006; Jaffe et al., 2004; Mirelman et al., 2010). Additionally, virtual obstacles can be modified for a specific task. For example, Shema et al. (2014) examined the clinical experience of a challenging virtual obstacle course environment with obstacles of various heights and sizes, resulting in improved functional mobility and increased obstacle negotiation.

When considering virtual obstacle crossing as a potential rehabilitation technique, it is important to address the differences in sensory information provided to the participant in each the virtual and the real obstacle crossing environments. The virtual environment lacks the tactile feedback of contacting an obstacle. This lack of tactile information requires individuals to use proprioception to sense how close their foot may be to the virtual obstacle in the virtual environment. Additionally, individuals must modify their limb trajectories using the visual information about foot clearance given in the virtual environment in a feedback and feedforward manner. However, the virtual environment gives consistent visual information of the trail limb which is otherwise unavailable in the real environment. When crossing a real obstacle, visual information about the trail limb ceases with the crossing of the lead foot over the obstacle. Thus, individuals may be able to expediently use this information to modify their trail limb trajectory.

While the use of virtual obstacles during treadmill gait training appears promising, it is not yet clear how foot clearance over a virtual obstacle changes with increased training/exposures, nor is it clear whether these performance changes transfer to a real-world obstacle crossing task or how aging effects this training. Understanding these training and transfer effects are necessary prior to adopting virtual reality obstacle crossing into a clinical setting designed to decrease fall-risk. Thus, the purpose of this study was to (1) determine the biomechanical obstacle crossing behavior of an able-bodied individual within a virtual environment, (2) determine if a learning effect existed with virtual obstacle crossing, and (3) determine if the learning effect would transfer to overground obstacle crossing. Three hypotheses were tested: (1) a training effect would be observed at the end of the virtual obstacle crossing training in the form of the adoption of a safer obstacle crossing strategy in the virtual environment, (2) a safer obstacle crossing strategy in the real environment would be adopted in the post-test relative to the pre-test, and (3) the performance changes in the virtual environment would be correlated with the performance changes in the real environment, suggesting an association between motor learning in a virtual environment and transfer to a real environment task. It was also postulated that each hypothesized finding would be affected by age, with older adults showing less learning and transfer (albeit still significant) compared to the younger adults.

Methods

Participants

A total of forty healthy, able-bodied individuals participated in the study (see Table 1 for demographics). Twenty younger adults and twenty older adults were recruited from a convenience sample from the University of North Carolina at Greensboro and their colleagues within the Greensboro community. All participants had normal or corrected-to-normal vision, no current musculoskeletal injuries, no cognitive impairment, able to walk unaided for 10 minutes, and were not currently pregnant. Each participant provided an informed consent and a basic healthy history and physical activity questionnaire. All procedures for the study were approved by the University of North Carolina at Greensboro Institutional Review Board.

Instrumentation

Twenty-eight retro-reflective markers were placed bilaterally on the shoulders, anterior superior and posterior superior iliac crests, medial and lateral knee and ankles, medial first and lateral fifth metatarsals, and the most anterior superior position on the shoe toe and the calcaneus. Additionally, rigid plates of 4 markers each were placed on the shank and thigh segment of each limb. A 12-camera motion capture system (Qualisys

AB, Gothenburg, Sweden) collected movement data at 100 Hz for each obstacle crossing session.

Experimental Design

Participants signed a consent form and provided a basic health history. Height and mass were collected and participants were put in a safety body harness to be utilized during the virtual obstacle crossing sessions. The retro-reflective markers were then applied to the participants' body. Normal walking pace was determined on the treadmill by starting at 0 m/s and incrementally increasing speed until the participant verbally indicated that speed was their normal pace. Then, the participant was sped up with a fast walking pace (2.0 m/s) and incrementally slowed until they verbally indicated that speed was their normal pace. The two speeds were then averaged to determine their normal walking pace.

This study utilized a pre-test/training/post-test design. During the pre-test, participants walked overground and crossed a real environment obstacle 10 times. The obstacle was set-up at the halfway mark (4m) along an 8m pathway across the laboratory. The obstacle was made of Masonite board, painted flat lack and designed to easily tip over if contacted (Figure 2). The dimensions for the obstacle were 10 cm high, 100 cm wide, and 0.5 cm deep. A height of 10 cm was selected because it is the average curb height. Participants were asked to "Walk at your normal pace and safely step over the obstacle with your dominant foot without stopping until you reach the end of the walkway."

During the training, participants walked on a treadmill and crossed a virtual environment obstacle. The virtual environment consisted of a grey and red walking path through a brown landscape with trees and a blue sky (Figure 3). A counter set in the top right gave feedback of obstacles hit and obstacles cleared. Virtual obstacles in the virtual environment were set to appear as 10 cm and on pace with the subject's normal walking speed with approximately 30 seconds in between obstacles to prepare for the obstacle that had to be crossed. The obstacle covered the entirety of the pathway requiring the subject to step over with both feet. The virtual environment also contained virtual feet that moved in real-time and corresponded to the participant's own foot movement by synchronizing with the motion capture system. The toe, heel, medial, and lateral metatarsal markers were white and portrayed on the feet in real-time. If the virtual obstacle was contacted, the specific foot marker that was hit turned red to indicate a collision, while the others that cleared the obstacle remained white. If the foot successfully cleared the obstacle, all markers turned green. Following five practice trails, participants completed two sessions of 10 trials each (20 total virtual environment obstacle crossings).

During the post-test, participants again walked over-ground and crossed a real environment obstacle 10 times to test for transfer from the virtual to the real environment. Participants crossed an obstacle a total of 40 times—10 real environment crossings in the pre-test, 10 virtual environment crossings in training session 1, 10 virtual environment crossings in training session 2, and 10 real environment crossings in the post-test. Five minute breaks were given between each test/session to minimize fatigue.

In the real environment, measures of vertical and radial lead and trail limb clearance, foot placement before and after the obstacle, peak lead and trail limb elevation, and the delta scores of the aforementioned variables between sessions were measured (Figure 5). In the virtual environment, peak lead and trail limb elevation, the delta scores of lead and trail limb elevation between sessions, and the pass/fail rate were measured. Trail limb clearance was measured in two ways. First, vertical trail limb clearance as the distance between the toe marker and the top of the obstacle at the time the toe marker is directly above the obstacle, and second, radial trail limb clearance as the minimal distance between the toe marker and the top of the obstacle. Similarly, lead limb clearance was measured in two ways. First, vertical lead limb clearance as the distance between the heel marker and the top of the obstacle at the time the heel marker is directly above the obstacle, and second, radial lead limb clearance as the minimal distance between the heel marker and the top of the obstacle. Foot placement was measured as the horizontal distance between the toe marker and the obstacle for before-obstacle foot placement and as the horizontal distance between the heel marker and the obstacle for after-obstacle foot placement. Foot placement was measured before and after the obstacle for both the lead and trail limb. Peak lead limb elevation and trail limb elevation for each obstacle was obtained and averaged per obstacle crossing session, using the heel and toe markers respectively.

Ratings of Perceived Exertion (RPE) was determined throughout the different tasks. Participants were asked to rate their perceived exertion after walking over the real environment over-ground obstacles after trial 1 and 10 for each session. With the virtual obstacle crossing, participants were asked their RPE during the first session after trial 1, 10, and 15, and during the second session after trial 1 and 10.

Data Reduction

The data from the Qualysis system was exported to Visual3D (C-Motion, Bethesda, MD) to create a file to import into MATLAB (The MathWorks Inc., Natick, MA). Custom MATLAB scripts were written to determine foot placement before and after the obstacle, vertical and radial heel clearance and peak elevation of the heel for the lead limb, and vertical and radial toe clearance and peak elevation of the toe for the trail limb. Deltas scores were calculated for each of the aforementioned variables.

Statistical Approach

To address hypothesis 1, a 2 x 2 (age by time) repeated measures MANOVA for pass rate and peak elevation of the lead and trail limb in the virtual environment was run. To address hypothesis 2, two separate 2 x 2 (age by time) repeated measures MANOVAs were run: one for foot placement (position of the both lead and the trail limb before and after the obstacle) and one for foot clearance (peak elevation and radial clearance for both the lead and trail limbs) in the real environment. Finally, to address hypothesis 3, Pearson's correlations were run for the delta scores of each variable of foot placement and clearance in the real environment against the number of successful obstacle passes in the virtual environment. Alpha level was set a priori to 0.05

Results

Table 1. Participant Demographics by Group. Demographics [mean(SD)] for participants including age, mass, height, normal walking speed (NWS), and average Ratings of Perceived Exertion (RPE) for Real Environment crossing sessions 1 (Real 1) and 2 (Real 2) and Virtual Reality crossing sessions 1 (VR 1) and 2 (VR 2).

Group	Foot Dominance	Age (yrs)	Mass (kg)	Height (m)	NWS (m/s)	RPE Real 1	RPE Real 2	RPE VR 1	RPE VR 2
Younger Adults	20 Right	22.45 (3.65)	71.05 (13.68)	1.70 (0.084)	0.959 (0.179)	7.08 (0.62)	7.38 (0.91)	7.40 (0.60)	7.40 (0.70)
Older Adults	18 Right 2 Left	55.60 (5.98)	70.93 (13.92)	1.70 (0.094)	0.429 (0.080)	7.25 (1.03)	7.40 (1.35)	7.72 (1.50)	7.60 (1.55)

Table 2. Summary of Results. Results include lead and trail limb position before the obstacle (PBO), lead and trail limb position after the obstacle (PAO), lead and trail limb foot clearance (FC) in the radial and vertical directions, lead and trail limb peak elevation (PE), and pass rate. Shading indicates non-significance at $p<0.05$.

		Group	Session 1	Session 2	Delta Score
Real Environment	*Lead Limb PBO*	Older	0.889(0.128)	0.929(0.122)	0.043(0.104)
		Younger	0.802(0.119)	0.855(0.181)	0.053(0.137)
	Trail Limb PBO	Older	0.234(0.061)	0.272(0.075)	0.040(0.050)
		Younger	0.197(0.055)	0.227(0.086)	0.030(0.063)
	Lead Limb PAO	Older	-0.293(0.041)	-0.276(0.049)	0.018(0.034)
		Younger	-0.308(0.070)	-0.282(0.069)	0.026(0.043)
	Trail Limb PAO	Older	-1.003(0.080)	-0.997(0.084)	0.003(0.030)
		Younger	-1.009(0.132)	-0.968(0.123)	0.040(0.074)
	Lead FC (Radial)	Older	0.124(0.032)	0.134(0.031)	0.012(0.022)
		Younger	0.141(0.039)	0.151(0.050)	0.010(0.023)
	Trail FC (Radial)	Older	0.102(0.043)	0.124(0.056)	0.023(0.023)
		Younger	0.094(0.034)	0.114(0.041)	0.020(0.024)
	Lead FC (Vertical)	Older	0.138(0.040)	0.152(0.042)	0.016(0.021)
		Younger	0.163(0.050)	0.180(0.067)	0.017(0.025)
	Trail FC (Vertical)	Older	0.123(0.061)	0.145(0.069)	0.021(0.027)
		Younger	0.121(0.054)	0.145(0.055)	0.024(0.024)
	Lead Limb PE	Older	0.403(0.058)	0.420(0.064)	0.018(0.019)
		Younger	0.405(0.040)	0.421(0.048)	0.015(0.022)
	Trail Limb PE	Older	0.480(0.099)	0.497(0.096)	0.017(0.033)
		Younger	0.499(0.063)	0.525(0.062)	0.027(0.023)
Virtual Environment	*Lead Limb PE*	Older	0.466(0.095)	0.468(0.114)	0.022(0.115)
		Younger	0.460(0.143)	0.447(0.099)	-0.013(0.069)
	Trail Limb PE	Older	0.360(0.119)	0.387(0.124)	0.038(0.088)
		Younger	0.382(0.208)	0.423(0.209)	0.041(0.077)
	Pass Rate (%)	Older	39.00(23.60)	46.00(27.22)	5.25(13.71)
		Younger	33.50(22.31)	43.50(23.68)	5.00(13.38)

Hypothesis 1: A training effect would be observed at the end of the virtual obstacle crossing training in the form of the adoption of a safer obstacle crossing strategy in the virtual environment.

There was no age x time interaction ($F_{3,36}$=0.714, p=0.550, partial η^2=0.056) for the variables of pass rate and lead and trail limb clearance peak elevation in the virtual environment. A multivariate main effect for time was identified ($F_{3,36}$=5.069 p=0.005, partial η^2=0.297), but not age ($F_{3,36}$=0.557, p=0.647, partial η^2=0.044). Follow up univariate tests concluded trail limb clearance peak elevation ($F_{1,38}$=8.915, p=0.005, η^2=0.190) and pass rate ($F_{1,38}$=4.812, p=0.034, partial η^2=0.112) drives this multivariate effect for time. Follow up pairwise comparisons confirmed there is a significant difference in trail limb clearance peak elevation between session 1 and session 2, $t(39)$=3.011, p=0.005, two-tailed. Follow up pairwise comparisons also confirmed there is a significant difference in pass rate between session 1 and session 2, $t(39)$=2.218, p=0.032, two-tailed.

Hypothesis 2: A safer obstacle crossing strategy in the real environment would be adopted in the post-test relative to the pre-test.

There was no age x time interaction for the variables of lead and trail limb position before the obstacle and lead and trail limb position after the obstacle ($F_{4,34}$=1.765, p=0.159, partial η^2=0.172). Multivariate main effects were identified for time ($F_{4,34}$=4.424, p=0.005, partial η^2=0.342), but not age ($F_{4,34}$=1.053, p=0.394, partial η^2=0.110). Follow up univariate tests concluded each variable of foot placement drives this multivariate effect for time: Lead limb position before the obstacle ($F_{1,37}$=5.725, p=0.022, η^2=0.132), trail limb position before the obstacle ($F_{1,37}$=13.789, p=0.001, η^2=0.272), lead limb position after the obstacle ($F_{1,37}$=11.497, p=0.002, η^2=0.237), and trail limb position before the obstacle ($F_{1,37}$=6.499, p=0.015, η^2=0.149). Follow up pairwise comparisons confirm there is a significant difference in between session 1 and session 2, for lead limb position before the obstacle [t(38)=2.438, p=0.020, two-tailed], trail limb position before the obstacle [t(38)=3.742, p=0.001, two-tailed], lead limb position after the obstacle [t(38)=3.432, p=0.001, two-tailed], and trail limb position after the obstacle [t(38)=2.515, p=0.016, two-tailed].

No age x time interactions for the variables of radial lead and trail limb clearance and lead and trail limb peak elevation in the real environment ($F_{4,34}$=1.350, p=0.272, partial η^2=0.137) were identified. A multivariate main effect for time was identified ($F_{4,34}$=13.094, p<0.001, partial η^2=0.606), but not age ($F_{4,34}$=2.141, p=0.097, partial η^2=0.201). Follow up univariate tests concluded each variable of foot clearance drives this multivariate effect for time: Radial lead limb clearance ($F_{1,37}$=9.238, p=0.004, η^2=0.200), radial trail limb clearance ($F_{1,37}$=31.493, p<0.001, η^2=0.460), lead limb peak elevation ($F_{1,37}$=24.536, p<0.001, η^2=0.399), and trail limb peak elevation ($F_{1,37}$=22.125, p<0.001, η^2=0.374). Follow up pairwise comparisons confirm there is a significant difference in between session 1 and session 2, for radial lead limb clearance [t(38)=3.082, p=0.004, two-tailed], radial trail limb clearance [t(38)=5.666, p<0.001, two-tailed], lead limb peak elevation [t(38)=5.005, p<0.001, two-tailed], and trail limb peak elevation [t(38)=4.715, p<0.001, two-tailed].

Hypotheses 3: The performance changes in the virtual environment would be correlated with the performance changes in the real environment, suggesting an association between motor learning in a virtual environment and transfer to a real environment task.

Pearson's correlations between the pass rate in the virtual environment and delta scores of each dependent variable were conducted. The relationship between pass rate and change in radial lead limb clearance in the real environment yielded a weak positive correlation trending towards significance (r=0.300, n=39, p=0.063). When separated by age group, this relationship was driven by the older adults, as the older adults exhibited a statistically significant strong correlation (r=0.635, n=19, p=0.003) compared to the very weak correlation exhibited by the younger adults (r=0.032, n=20, p=0.892). In addition, the relationship between pass rate in the virtual environment and change in lead limb peak elevation in the real environment yielded a weak positive correlation (r=0.355, n=39, p=0.027). When separated by age group, this relationship was driven by the younger adults, as the younger adults exhibited a moderate positive correlation (r=0.458, n=20, p=0.042) compared to the weak correlation exhibited by the older adults (r=0.243, n=19, p=0.317). Lastly, the relationship between pass rate in the virtual environment and change in trail limb peak elevation in the real environment yielded a moderate positive correlation (r=0.545, n=39, p<0.001). When separated by age group, this relationship was driven by the older adults, as the older adults exhibited a strong correlation (r=0.657, n=19, p=0.002) compared to the moderate correlation exhibited by the younger adults (r=0.430, n=20, p=0.059). However, the correlation for the younger adults is approaching significance.

Figure 6. The Relationship Between the Delta Score of Radial Lead Limb Clearance and Pass Rate Separated by Age.

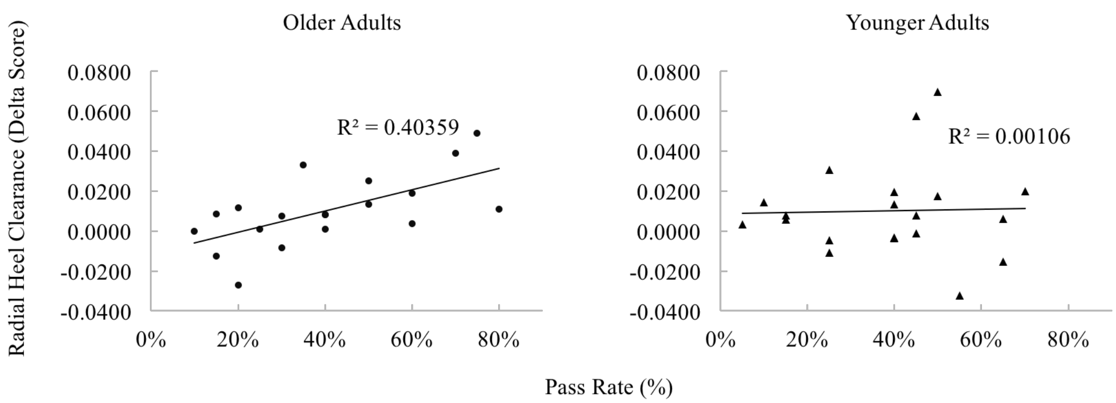


Figure 7. The Relationship Between the Delta Score of Lead Limb Peak Elevation and Pass Rate Separated by Age.

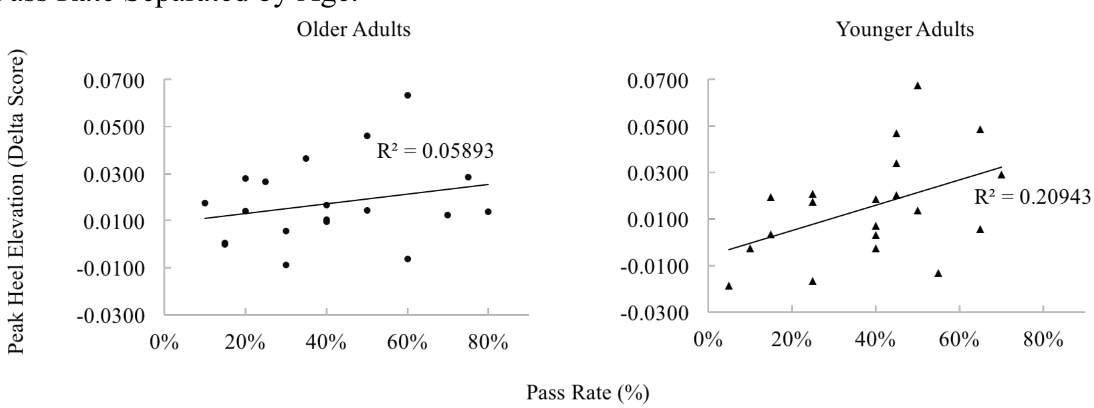
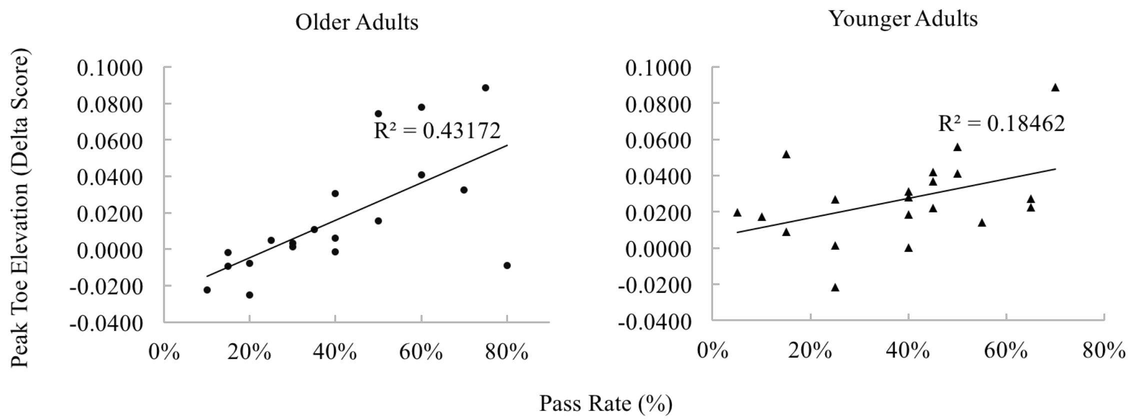


Figure 8. The Relationship Between the Delta Score of Trail Limb Peak Elevation and Pass Rate Separated by Age.



Discussion

This study examined whether changes in gait during a virtual obstacle avoidance task on a treadmill transferred to overground gait obstacle avoidance with real environment obstacles. Further, this study examined whether age influenced the learning and transfer of the obstacle crossing strategy. In general, our hypotheses were supported, as participants did perform better with increased obstacle crossing exposure in the virtual environment and that change in performance was associated with changes in the real environment obstacle crossing during the post-test. Furthermore, both age groups similarly increased their foot placement and clearance in the real environment, but employed different adaptive strategies to successfully avoid the obstacles.

Younger and older adults learned to enhance their obstacle crossing in the virtual environment via peak trail limb elevation and pass rate, evidenced by an increase in these variables from session 1 to session 2 of the virtual environment training. This suggests that participants chose an obstacle crossing strategy of increasing their trail limb's overall peak elevation after focusing on the lead limb clearing the obstacle to try to ensure that the trail limb would not contact the obstacle. These findings are in concurrence with previous work that has shown that after contact with an obstacle, clearance and peak elevation remain elevated for multiple subsequent obstacle crossings (Rhea & Rietdyk, 2011). As participants frequently contacted the obstacle within the virtual session, their toe elevation progressively increased and remained increased during the real environment session following virtual obstacle crossing. Additionally, the increases in pass rate suggest that the participants became progressively more successful at crossing the obstacles from session 1 to session 2, furthering the evidence for learning within the virtual environment.

Both younger and older adults adjusted their adaptive obstacle crossing strategies via foot position, clearance, and elevation in the real world following virtual obstacle crossing training. Previously, distance between foot placement and the obstacle has been related to foot contact with the obstacle (Chou & Draganich, 1998; Patla & Greig, 2006). A main effect for time was seen in all foot position variables in the real environment from pre- to post-test. Participants increased their foot position before and after the obstacle for both the lead limb and the trail limb. This may suggest an obstacle crossing strategy of elongating their stride to ensure proper time for increasing foot clearance and successful obstacle crossing. Additionally, a main effect for time was seen for all foot clearance and peak elevation variables in the real environment from session 1 to session 2. Participants increased the minimal distance between their foot and the obstacle, as observed by increases in radial foot clearance, which may be a function of their overall increases in foot peak elevation. Increasing foot clearance decreases the chance of contacting the obstacle, but this may come at the cost of increased metabolic energy demand and decreased biomechanical stability. There is likely a specific range of foot clearance that will optimize the ratio of increased metabolic cost to decreased obstacle contact. However, this ratio may be person and context specific. Future research may test this optimization using perturbation and physiological testing.

Lastly, younger and older adults were able to transfer the learning obtained from the virtual environment to the real environment as evidenced by the positive correlational relationships between pass rate and the changes in each radial lead limb clearance, peak lead limb elevation, and peak trail limb elevation. After separating the correlations by age, it was observed that younger and older adults each applied separate adaptive obstacle strategies which transferred to the real environment. Younger adults adapted increases in lead limb peak elevation in the real environment. This suggests that the younger adults focused on increasing the overall elevation of the lead limb in the virtual world to increase the success of obstacle crossing, while allowing the trail limb to cross without modulation. Conversely, older adults adapted increases in radial lead limb clearance and trail limb peak elevation. This may suggest that after increasing the clearance of the lead limb over the obstacle, older adults increased the overall trail limb elevation due to lack of confidence in the trail limb to successfully clear the obstacle. Due to the fact that the trail limb is the main obstacle contactor at 67-100% of the time (Heijnen, Romine, Stumpf, & Rietdyk, 2014; Mohagheghi, Moraes, & Patla, 2004; Rhea & Rietdyk, 2007;

Rhea & Rietdyk, 2011, 2011; Rietdyk & Rhea, 2006) and older adults are at an increased fall risk, adopting an overall clearance of the trail limb is a strategy for older adults to decrease contact and also fall risk. Increasing clearance and elevation may be metabolically costly, but it may be argued that the benefit of decreasing obstacle contact, and thus fall risk, outweighs the increase in metabolic cost for older adults.

The virtual environment lacked the tactile information of obstacle contact, thus participants had to rely on proprioception and visual information in both a feedforward and a feedback manner to modify limb trajectory. However, visual information of the trail limb was provided in the virtual environment which was otherwise absent in the real environment. Participants used this information advantageously to modify the trajectory of their trail limb.

This study is limited by the fact that it did not include a group which walked on the treadmill in the virtual environment without the virtual obstacles. As such, it is difficult to ascertain that the changes seen in the real environment were due solely to the virtual environment obstacles and not to the potential practice effect from crossing the real environment obstacle a total of 20 trials. Additionally, the variables of foot placement and clearance which were analyzed in the real environment were not analyzed in the virtual environment due to technical difficulties. The ability to compare all variables across conditions would strengthen the comparisons and effects of the virtual environment on the real environment and give a better understanding of the biomechanical adaptive obstacle crossing strategies used by the participants in this study.

In conclusion, virtual obstacle crossing changes an individual's obstacle crossing behavior. These changes are not limited to age, as each age group responded similarly from pre-testing to post-testing in the real environment in terms of biomechanical behavior and within the virtual obstacle crossing sessions in terms of performance in pass rate. This study sets up the foundation for employing virtual reality training with patients with pathology who may exhibit decreased gait adaptability, reduced ability for obstacle negotiation, and an increased fall risk.

CHAPTER V

DISCUSSION

The motivation behind this study was to create a virtual reality obstacle crossing program for use in increasing a trans-tibial amputee's ability to safely and successfully traverse obstacles within their gait path. The virtual reality obstacle crossing required an individual to adapt their gait patterns to appropriately respond to the obstacle while maintaining a safe environment to determine and practice an optimal obstacle crossing strategy. This study served to determine the feasibility of a virtual obstacle crossing program and the plausibility of its use in a clinical rehabilitation setting, as it established normative data based on healthy able-bodied adults.

This study examined the change in an able-bodied individual's adaptive obstacle crossing strategies in response to virtual environment obstacle crossing. Participants increased foot position, clearance, and elevation from session 1 to session 2 in the real environment as a response to the virtual obstacle crossing. Between the two sessions of virtual obstacle crossing, participants became progressively more successful in clearing the obstacle with both limbs indicating a learning over the virtual environments sessions. Participant age dictated the adaptive obstacle crossing strategy used in response to their performance in the virtual environment. Younger adults increased overall lead limb peak elevation, whereas older adults increased overall trail limb peak elevation and increased the minimal distance of the lead limb to the obstacle.

It is important to note the difference in sensory information received by the participant in each environment which may have influenced the strategy used to modify limb trajectory. The virtual environment lacked the tactile information of obstacle contact, resulting in reliance on proprioception and visual information in both a feedforward and a feedback manner to order to modify limb trajectory. However, visual information of the trail limb was provided in the virtual environment which was otherwise unavailable in the real environment. Participants were able to use this additional visual information to advantageously modify trail limb trajectory.

Overall, this study specifically contributes to the literature by showing that performance changes in obstacle crossing via the virtual environment are not confined by age. Both younger and older adults showed improvements in virtual reality, which also translated to the real environment. However, the specificity of the transfer was dictated by age—a finding that should be taken into account in future research. The clinical implications for these results are that an individual of any age may be able to benefit from virtual reality obstacle crossing training.

Some limitations to this study are as follows. First, a true control group did not exist. A group which walked on the treadmill in the virtual environment, but did not cross virtual obstacles was not included. This creates difficulties in ascertaining that the virtual reality obstacle crossing was solely responsible for the changes in the real environment. The changes in the real environment may be due to a practice effect of crossing the real environment obstacles for a total of 20 trials. This leads to the second limitation that it is possible that there were not enough virtual obstacles crossed to create a large enough learning effect to significantly affect more of the crossing variables. Third, the transfer task was conducted within day and there was no retention test. We cannot determine for sure that the effects seen will remain for the long-term without 24 hours in between sessions to determine retention. Lastly, the virtual environment lacked the variables of foot clearance and placement due to technical difficulties. Determining these variables would optimize the assessment of the biomechanical behaviors and obstacle crossing strategies within the virtual environment and would allow for direct comparisons between the virtual and real environment conditions.

Future directions for this research are to first include a control group to determine if there is a practice effect with the real environment obstacle. Second, there is a need to determine the optimization of exposure to the virtual environment. Whether the amount of trials in one session must be increased or if the amount of sessions must be increased, the optimization remains to be determined. Also, a training program over days for consecutive weeks must be created for this virtual reality program to be considered for use within rehabilitation clinics. Lastly, direct comparison of biomechanical behaviors and the inclusion of a retention test are needed to determine causal effects of the virtual environment on the real environment.

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