

Reactive stepping strategies following lateral surface translations in individuals  
with a unilateral lower limb amputation

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## **Abstract**

Following a loss of balance, individuals may utilize specific stepping strategies to prevent themselves from falling. The stepping strategies that are used by lower limb amputees, which likely consist of unique properties because of the limitations when using a prosthesis, have not been thoroughly identified. Therefore, this study examined the lateral reactive stepping strategies used eight amputees and ten non-amputees. They experienced four support-surface translations in both the leftward and rightward direction. Results indicated that amputees use unique reactive stepping strategies, particularly with the unloaded leg and when the direction of the perturbation causes unloading of the prosthesis. Amputee stepping strategies were characterized by fewer steps, lower quality of balance recovery, and wider variety compared to non-amputees. This study's findings highlight the reliance on the hip strategy in amputee reactive balance, and future studies should explore how amputees use their hip and trunk while executing reactive stepping strategies.

*Keywords: lower limb amputation, reactive stepping strategy, surface translation, balance control, prosthetic leg*

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**Abbreviations:**

BK (below-knee)

AK (above-knee)

AP (anteroposterior)

ML (mediolateral)

COM (centre of mass)

COP (centre of pressure)

BOS (base of support)

LLS (loaded leg step)

COS (crossover/crossunder step)

MSS (medial side step)

ULS (unloaded leg step)

**Terminology:**

Residual leg: refers to the leg above the site of amputation (this term will be used when addressing properties associated with the stump)

Prosthetic leg: refers to the residual leg with an attached prosthesis (this term will be used when addressing balance control)

Intact leg: refers to the leg contralateral to the amputation.



## **1. Introduction**

Falls are a significant issue among individuals with a unilateral lower limb amputation, with over half of this population reporting a fall each year (Kulkarni, Wright, Toole, Morris, & Hirons, 1996; Miller, Speechley, & Deathe, 2001). Approximately 40% of these falls result in an injury, of which ~20% require medical attention. Additionally, falling often leads to fear of future falls, which commonly leads to more sedentary behaviour and consequently, a lower quality of life (Miller et al., 2001).

An unexpected perturbation, such as a slip or trip, is often the cause of a loss of balance. The reactive balance strategies used by an individual after experiencing such balance loss can be indicative of their level of stability for future fall events. Therefore, gathering data regarding the manner in which these individuals attempt to recover their balance may lead to changes and improvements to current rehabilitation practices.

In order to effectively examine the reactive balance of individuals with a unilateral lower limb amputation, balance control in individuals without amputation must first be explored to provide a groundwork of typical balance control seen in humans.

### **1.1. Balance Control in Humans**

The effective maintenance of balance is achieved through proactive and reactive balance strategies. Proactive balance strategies involve postural adjustments

that act to minimize the de-stabilizing effects of upcoming movements, such as during gait initiation (Maki, Edmondstone, & McIlroy, 2000; Vrieling, van Keeken, Schoppen, Otten, Halbertsma, et al., 2008). Conversely, reactive balance strategies, which will be defined in more detail, occur in response to an unexpected loss of balance.

## **1.2. Reactive Balance Control**

Reactive balance refers to the generation of rapid postural reactions in response to an unexpected loss of balance. Following a perturbation to an individual's centre of mass (COM) the balance control system integrates relevant sensory input from the visual, vestibular, and somatosensory systems, leading to the execution of postural reactions via motor outputs to maintain the COM within the base of support (BOS). The COM-BOS relationship is a critical component of the postural control system as it is indicative of an individual's level of stability (Horak, 2006).

During quiet standing, movement of the COM within the BOS is controlled by the centre of pressure (COP), the point at which the ground reaction forces from the standing surface summate to act on the feet (Winter, Prince, Frank, Powell & Zabjek, 1995). The COP moves in phase with the COM during postural sway, attempting to 'push' the COM back toward the centre of the BOS (Massion, 1992; Winter et al., 1995). Movement of the COP, much like movement of the COM, is used to infer an individual's level of stability, as evidenced by smaller and less variable COP excursions during standing balance in young adults compared to less stable individuals, such as older adults (Massion, 1992; Winter et al., 1995).

Research studies have incorporated various methods of perturbation to elicit a sudden loss of balance, such as waist-pulls (Mille, Johnson, Martinez, & Rogers, 2005; Young, Whittall, Bair, & Rogers, 2013) and surface translations (King & Horak, 2008; Maki et al., 2000). When the support surface is translated beneath a stationary individual's feet, the resulting postural response involving the lower limbs can be classified as either a fixed-support recovery strategy (e.g., ankle strategy or hip strategy) or a change-in-support recovery strategy (e.g., stepping strategy) (Maki et al., 2000; Massion, 1992). Fixed-support balance strategies are effective for smaller perturbations to the COM. Muscles are typically activated in a distal-to-proximal pattern where the furthest muscles that cross the ankle joint are activated approximately 100 ms post-perturbation, closely followed by the activation of more proximal muscles. This creates torques about the ankle and knee (i.e., ankle strategy) to control the body's COM (Nashner, 1977). As the task becomes more difficult (e.g., greater surface translation acceleration), the ankle strategy becomes insufficient to maintain balance. Instead, it is necessary to activate the hip and trunk muscles (i.e., hip strategy) in addition to the ankle strategy to compensate for the larger excursions of the body's COM (Horak, Henry, & Shumway-Cook, 1997).

A hip strategy is also more prevalent when individuals are perturbed in the lateral direction due to differences in the biomechanical requirements of balance recovery associated with these perturbations. A loss of balance in the lateral direction results in the COM travelling laterally over the BOS, thereby loading one leg with the majority of the individual's weight. Conversely, the leg opposite the direction of the loss of balance is unloaded, causing asymmetrical distribution of the individual's

weight (Gray, Yang, McCombe Waller, & Rogers, 2017; Maki, McIlroy & Perry, 1996). Thus, torques generated by the hip adductors and abductors are initially relied upon for generating the correctional torques (Day, Steiger, Thompson & Marsden, 1993). This is reflected by the early activation of the tensor fascia latae of the loaded leg at approximately 100 ms after lateral translation onset, followed 20 ms later by the activation of distal muscles that cross the ankle joint (Henry, Fung, & Horak, 1998).

When a perturbation is too large for an individual to maintain their COM within the BOS, a stepping strategy is observed. Various classification systems of stepping strategies have been proposed. For example, a complex classification of 11 stepping strategies that include contributions from the upper body and trunk was identified by a recent study (Batcir et al., 2018). However, since the aim of this thesis is to identify the lower limb balance response, this thesis will use a simplified classification system similar to Maki et al. (1996), which includes: (i) medial side step, (ii) crossover step, (iii) loaded leg step, and (iv) unloaded leg step (Figure 1). These four strategies are summarized in the following sections, and for ease of understanding, described according to a loss of balance resulting in a rightward shift of the COM. In the event of a surface translation that causes a lateral loss of balance to the right side, the right leg is passively loaded, while the left leg is passively unloaded.

### **1.3. Reactive Stepping Strategies**

#### **(i) *Medial Side Step (MSS)***

The initial step of the MSS strategy is completed by the passively unloaded (left) leg, where the left leg takes a small, quick step toward the right leg, but does

not cross in front or behind the loaded right leg. The initial medial step results in a narrower BOS, resulting in a follow up lateral step with the loaded right leg, (Gray et al., 2017; Mille et al., 2005; Maki et al., 1996). While the MSS strategy has been identified as a distinct strategy, some studies have grouped it with stepping responses that consist of more than one step, a response known as a multiple step strategy. Multiple step strategies are often observed in populations that are at risk for falling, such as older adults and individuals with stroke (Young et al., 2013; Gray et al., 2017).

**(ii) Crossover Step (COS)**

The COS strategy is characterized by the passively unloaded (left) leg crossing either in front of or behind the passively loaded (right) leg. Once the left foot is planted on the ground, another step is typically initiated with the right leg in order to return to the original BOS. The initiation of the COS is facilitated by the nature of the fall since the stepping leg is already being unloaded, thus allowing quick movement of the unloaded leg. Adequate strength and flexibility of the unloaded leg is necessary to effectively perform the crossover without incidental contact with the loaded leg during the crossover, and the loaded leg must be able to support the weight of the individual (Gray et al., 2017; Mille et al., 2005; Maki et al., 1996).

**(iii) Loaded Leg Step (LLS)**

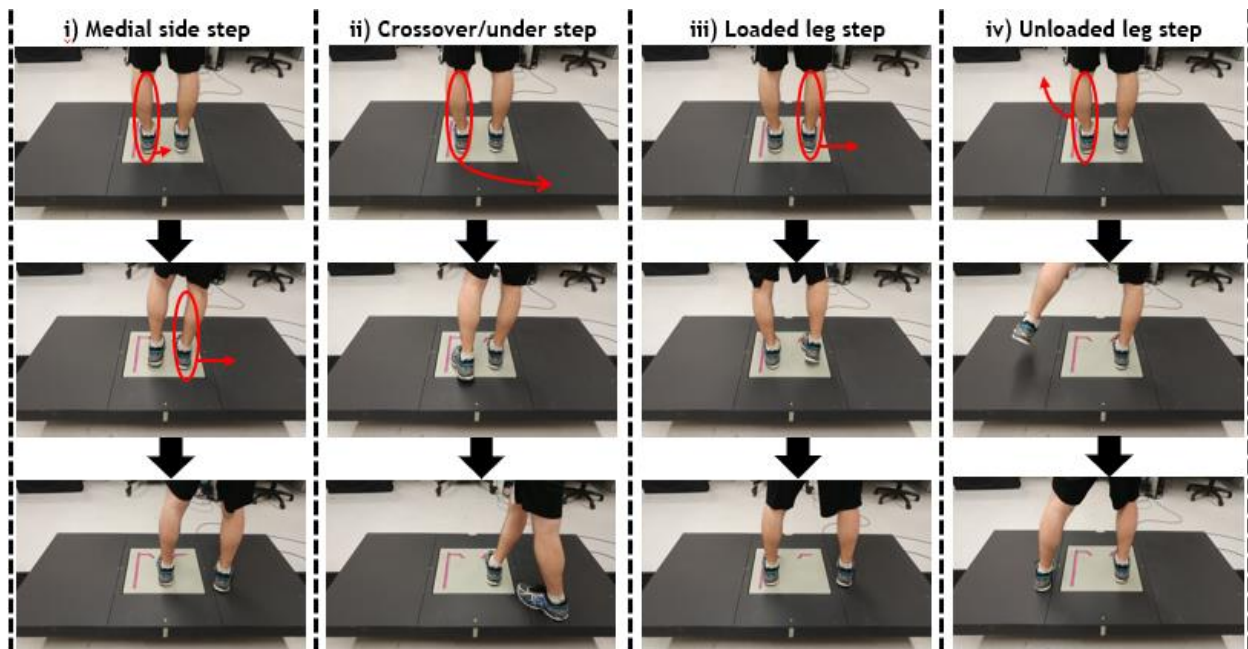
The LLS strategy is characterized by a step taken with the passively loaded (right) leg. In order for this to occur, the right leg must be actively unloaded and the left leg must be actively loaded. This stepping strategy has the benefit of a reduced

risk of limb-to-limb contact (an issue apparent with the COS) and less demand on the stepping leg, as it does not require a large range of motion. However, the unloading and loading of the right and left legs, respectively, work against the natural weight transfer of the lateral fall. Typically, a LLS is sufficient to recover balance without taking another step, and the BOS is wider once the step is complete, offering more support for the individual (Gray et al., 2017; Mille et al., 2005; Maki et al., 1996).

***(iv) Unloaded Leg Step (ULS)***

An unloaded leg step refers to a step taken with the passively unloaded (left) leg. For the purpose of this thesis, the ULS strategy includes all steps that involve only the unloaded leg, without any additional steps with the loaded leg. This can include a side step (directed away from the intended fall direction), a medial step (directed toward the loaded leg), or a leg lift (an excessive abduction of the unloaded leg).

The specific stepping strategy used in response to a lateral perturbation is affected by several external factors, including the method of perturbation, the perturbation waveform characteristics, and task instructions, as well as internal factors such as the structural and physiological constraints of the individual being perturbed. The following sections outline specifically how these factors affect the postural response.



**Figure 1:** Lateral stepping strategies often observed following a lateral perturbation to standing balance. (i) Medial side step: unloaded left leg steps medially toward the loaded right leg, followed by a lateral side step with the loaded right leg. (ii) Crossover/under step: unloaded left leg crosses in front/behind the loaded right leg. (iii) Loaded leg step: loaded right leg steps laterally in the direction of the loss of balance. (iv) Unloaded leg step: characterized by an excessive abduction of the unloaded leg, or a lateral step in the opposite direction to the loss of balance.

### 1.3.1. Influence of Perturbation Characteristics on Reactive Stepping Strategies

Perturbation characteristics (e.g., perturbation mechanism and perturbation intensity) can influence the type of reactive stepping strategy used. Two of the most common mechanisms for eliciting a lateral loss of balance are waist-pulls and support-surface translations. Both of these perturbation mechanisms cause a lateral shift of the body's COM, however there are differences in the mechanics, as well as the sensory information elicited, by each perturbation mechanism (Mansfield & Maki, 2009). Waist-pulls are conducted by attaching a motor driven belt to an individual's waist, subjecting them to potential pulling forces directed at the pelvis. Surface

translations are also motor driven, however the applied force is directed across the foot soles, as opposed to the pelvis (Mansfield & Maki, 2009).

Mansfield & Maki (2009) exposed older adults to both surface translations and waist-pulls and showed differences in the balance recovery strategies between these perturbation mechanisms. The number of lateral perturbation trials that elicited a COS was not different between perturbation method (surface translation: 30.7% of trials; waist-pull: 32.2% of trials). However, surface translations caused a greater occurrence of foot collisions during the reactive stepping response (surface translation: 55.6% of trials; waist-pull: 5.7% of trials) (Mansfield & Maki, 2009). This was attributed to the more de-stabilizing perturbations produced by surface translations, evidenced by a more rapid increase in ankle torques and larger COM excursions compared to waist-pulls (Mansfield & Maki, 2009). The increased occurrence of foot collisions following a surface translation was perhaps because the rapid movement of the support surface beneath the feet causes the stance leg to move toward the foot that is lifted, thereby increasing the chance of collision. Following a waist-pull, there is no such passive movement of the stance leg since the support surface during a waist-pull is stationary, making it less likely for a collision to occur (Mansfield & Maki, 2009). Thus, while the older adult sample collected by Mansfield & Maki (2009) displayed similar stepping strategies between surface translations and waist-pulls, the surface translation method appears to be more challenging due to the rapid movement of the support surface.

Aside from perturbation mechanism, the intensity of the perturbation (e.g., peak acceleration, velocity, displacement, etc.) has a significant influence on the



reactive balance response. It is generally accepted that the initial acceleration of the support surface translation elicits instability, and that the magnitude of the postural response is scaled to the perturbation amplitude and velocity (Diener, Horak, & Nashner, 1988). The perturbation deceleration and the acceleration-deceleration interval also impact balance recovery. When the deceleration phase occurs too soon after the initial acceleration, the individual being perturbed can use the deceleration force to stabilize themselves since the deceleration forces oppose the acceleration forces (Carpenter, Thorstensson, & Cresswell, 2005; Tokuno, Cresswell, Thorstensson, & Carpenter, 2010). Therefore, to accurately examine the stepping response following a support surface translation, the acceleration-deceleration interval must be long enough to allow the individual to perform their stepping strategy without the influence of the deceleration phase.

### **1.3.2. Effects of Aging on Reactive Stepping Strategies**

Age-related deficits have been found within the somatosensory, vestibular, and visual systems. Aging is also associated with declines in muscle strength, flexibility, joint range of motion, orthopaedic impairments, and cognitive decline. Together, these sensory and musculoskeletal impairments lead to an increased risk of falls in the older adult population (Horak, 2006).

Aging appears to impair older adults' ability to recover balance in the lateral direction (Rogers & Mille, 2003). This is evidenced by reduced electromyographic amplitudes and delayed electromyographic onset latencies of 10-20 ms when compared to young adults (Lin & Woollacott, 2002). These changes, along with

reduced somatosensation at the ankle and foot (Woollacott & Shumway-Cook, 1990), and a reduced capacity to generate torque and reduced trunk control, contribute to instability in the reactive balance of older adults (Rogers & Mille, 2003). Following lateral waist-pull perturbations, older adults require multiple steps to recover balance in 65% of trials, and use the riskier COS strategy in 59% of trials. This contrasts with young adults, who only rely on multiple steps and the COS strategy for 13% and 37% of trials, respectively (Mille et al., 2005).

#### **1.4. Lower Limb Amputation**

The following sections will introduce and describe the population of interest (unilateral lower limb amputees), and review the current literature that relates to reactive balance control within this population.

##### **1.4.1. Overview of Lower Limb Amputation**

In the USA, there are an estimated 1.6 million individuals with a lower limb amputation, and this number is estimated to increase to 3.6 million by 2050 (Ziegler-Graham, MacKenzie, Ephraim, Trivison, & Brookmeyer, 2008). In Canada, approximately 7,500 amputations to a lower limb are performed each year, and the lower limb amputation rate has recently been calculated to be 22.9 amputations per 100,000 people (Imam, Miller, Finlayson, Eng, & Jarus, 2017).

Lower limb amputations encompass all amputations that occur at or below the hip joint. Amputations that occur along the thigh are labeled as above-knee (AK), while those that occur below the knee are labeled as below-knee (BK). BK

amputations typically refer to an amputation somewhere along the tibia, however by definition it can refer to a foot amputation as well. The most prevalent level of amputation is BK (58.4%), followed by AK (23.9%) (Imam et al., 2017). For this thesis, only amputations that occur above the ankle joint and below the hip joint will be considered.

In Canada, the majority of lower limb amputations are caused by peripheral vascular disease and diabetic neuropathy. Vascular-related and diabetes-related cases of lower limb amputation are often grouped together for research purposes since vascular issues are often present with diabetes, however they can be separate issues altogether. For example, in a sample of 10,834 individuals with a lower limb amputation from Quebec, over 50% suffered from peripheral vascular disease and diabetes, 28% suffered only from peripheral vascular disease, and 4% suffered only from diabetes (Dawes, Iqbal, Steinmetz, & Mayo, 2010). Furthermore, a study that tracked amputation rates in hospitals across Canada found that of the 44,430 lower limb amputations performed in Canada between 2006 and 2011, 65.4% were related to diabetes, while 25.6% of lower limb amputations resulted from standalone vascular diseases/infections (without coinciding diabetes) (Imam et al., 2017). Trauma, cancer, and congenital amputations account for the remaining 9% of lower limb amputations in Canada (Imam et al., 2017).

The amputation of a lower limb typically occurs in people aged 50 years and over. High rates of diabetes in Canada (9.3% of population) and the increased prevalence of diabetic neuropathy and peripheral vascular disease among older adults (Canadian Diabetes Association, 2016) contributes to a mean age of 65.7 years at the

time of amputation (Imam et al., 2017). Many individuals with a lower limb amputation are 50-74 years old (45 amputations per 100,000 people), and there is an increase in amputation prevalence after the age of 75 years (107 amputations per 100,000 people). In contrast, only 4.7 amputations per 100,000 occur in individuals 50 years of age and under (Imam et al., 2017).

There is currently minimal data with reference to unilateral versus bilateral amputation rates in Canada, however general trends can be inferred based on data from other countries. For instance, during a three-year span at a Danish hospital a total of 257 individuals sustained an amputation to at least one leg. Of these individuals, 79% sustained a unilateral amputation, while 21% sustained a bilateral amputation, indicating that the rate of unilateral amputations greatly exceeds that of bilateral amputations (Helm, Engel, Holm, Kristiansen, & Rosendahl, 1986). Additionally, it has been estimated that 55% of individuals with vascular-related unilateral lower limb amputation will require an amputation to their remaining intact leg within three years. In other cases, such as traumatic lower limb amputations, the chance of having a bilateral amputation is much smaller, affecting only 4.6% of traumatic lower limb amputees (Pasquina et al., 2014).

For the purpose of this thesis, the remainder of the document will be focused on the most prevalent type of amputation, which is a unilateral lower limb amputation. As this study will be examining reactive balance following lateral surface translations, unilateral balance deficits will be more likely to result in unique direction-dependent stepping strategies during fall recovery due to the asymmetrical structure and abilities of the lower limbs in individuals with a unilateral amputation.

This is based on compensations demonstrated by other populations with lower limb neuromuscular asymmetries, such as stroke (Gray et al., 2017).

#### **1.4.2. Musculoskeletal Changes Associated with Unilateral Lower Limb Amputation**

The ability to generate torques around the ankle and knee joints of the residual limb is compromised due to the loss of an active ankle joint and the associated musculature. Consequently, individuals with a unilateral lower limb BK amputation must rely on a passive prosthetic ankle, which is controlled largely through the movements of the knee and hip of the residual leg. However, individuals with a unilateral lower limb amputation also experience muscle atrophy of the residual leg. This is evident through isometric strength measurements of the intact and the residual leg, where a 51-57% decrease in residual leg knee extension torques and a 42-51% decrease in residual leg knee flexion torques have been observed (Lloyd, Stanhope, Davis, & Royer, 2010; Isakov, Burger, Gregoric, & Marincek, 1996).

Due to the reduced muscle strength at the knee there is compromised control of the prosthetic ankle joint, and much of the torque generation is provided by the hip joint on the prosthetic side (Shell, Segal, Klute, & Neptune, 2017). This reliance on hip muscle activity is exacerbated in AK amputees, as there is no active knee joint. This puts greater stress on the hip joint compared to BK amputees. Although this higher reliance of the hip joint minimizes strength differences between the intact and residual hips (<5% difference between hip abduction torques) (Lloyd et al., 2010; Isakov et al., 1996), the increased requirements of the hip joint often leads to a higher likelihood of experiencing osteoarthritis and low back pain (Burke, Roman, &

Wright, 1978). Joint problems also arise in the intact leg of individuals with a unilateral lower limb amputation, as it is relied upon for the majority of movements. Compared to the residual leg, the intact leg is subjected to greater loading and therefore greater torque generation and the ankle, knee, and hip, often leading to osteoarthritis and contributing to low back pain (Royer & Wasilewski, 2006).

AK amputations have a greater affect on an individual's physical capabilities as there is a lack of an active knee joint following an AK amputation. This leads to disadvantages during walking, where an intact femur and knee joint with BK amputations can allow for greater loading forces and force transfer to the prosthesis. This has been associated with less energy consumed during gait, and higher walking velocities in BK amputees compared to AK (Hagburg, Berlin & Renstrom, 1992). Therefore, BK amputees are typically regarded as higher functioning amputees compared to AK.

#### **1.4.3. Sensory Changes Associated with Unilateral Lower Limb Amputation**

Individuals with a unilateral lower limb amputation lack somatosensory information from the distal portion of their residual leg, and thus lack a critical component of the postural control system. Not only is the postural control system missing important afferent information from the lower limb, the remaining afferent nerves located at the stump of the residual leg are often damaged, potentially leading to inappropriate depolarization of the neurons and incorrect sensory feedback (Geurts & Mulder, 1992).

Sensation at the amputated site is highly dependent on the reinnervation of the skin, with the number and type of axons that successfully grow back and connect with the skin determining the quality of sensation at the stump (Braune & Schady, 1993). It is critical to facilitate skin reinnervation early on during post-amputation rehabilitation since cortical reorganization occurs soon after amputation. Since a portion of the lower limb is no longer sending afferent nerve impulses to its represented area in the somatosensory cortex, the cortical representation of other body parts, specifically those areas represented in close proximity to the 'unused' area in the somatosensory cortex, begin to take over the unused lower limb cortical area (Taub, Uswatte, & Mark, 2014).

The early stages of rehabilitation for individuals with a lower limb amputation focus heavily on the recovery of the skin at the site of amputation and acclimatizing the individual to the unfamiliar sensations present following their operation (Cole, 2003). When provided with the proper care, individuals with a unilateral lower limb amputation can regain normal sensation in the residual leg. A study that examined light touch, deep pressure, vibration, and pinprick sensation in individuals who suffered a traumatic unilateral lower limb amputation showed no sensory impairments in the residual leg that were due to amputation, however they did find deficits in pinprick and vibration sensibility that were associated with aging (Kosasih & Silver-Thorn, 1998).

Research that examined skin sensation in individuals with a unilateral lower limb amputation due to traumatic (n=34) or vascular reasons (n=14) found that the touch-pressure threshold was higher in the residual compared to the intact leg for

both groups (vascular related amputation group:  $3.90 \pm 0.33$  intact leg;  $4.09 \pm 0.46$  residual leg; trauma-related amputation group:  $3.70 \pm 0.44$  intact leg;  $4.08 \pm 0.68$  residual leg) (Kavounoudias, Tremblay, Gravel, Iancu, & Forget, 2005). This indicates that the amputation of a leg leads to worse tactile sensation in the residual leg, as similar effects were observed in both the trauma-related and vascular-related amputation groups.

The assessment of movement detection threshold in the Kavounoudias et al. (2005) study produced interesting results. As expected, the vascular-related amputation group had impaired movement detection in the residual leg, with a 55% greater detection threshold in the residual leg compared to the intact leg. However, in the vascular-related amputation group there was no difference in movement detection threshold between legs. The authors attributed this lack of difference between legs to central re-organization, stating that central sensory adaptation may occur as an attempt to make both legs 'equal', thereby minimizing the amount of sensory conflict that may arise from differing sensory information from each leg (Kavounoudias et al., 2005). However, the functional significance of these findings has yet to be determined.

#### **1.4.4. Asymmetrical Standing Posture**

Rehabilitation practices commonly focus on improving standing balance during the initial stages of rehabilitation in individuals with a unilateral lower limb amputation by stressing the importance of weight bearing over the intact leg. Weight bearing asymmetry during standing balance is common among individuals with



unilateral lower limb amputation even after post-acute rehabilitation. These individuals typically shift their weight toward the intact leg, with multiple studies reporting that >56% of an individual's weight is borne over the intact leg (Bolger, Ting, & Sawers, 2014; Nadollek, Brauer, & Isles, 2002; Nederhand, Van Asseldonk, der Kooij, & Rietman, 2012; Quai, Brauer & Nitz, 2005). This weight bearing asymmetry causes the mean COP position to be shifted toward the intact leg by  $7\pm 2$  mm when standing with eyes open. When the standing balance task is made more difficult by removing vision (standing with eyes closed), there is a larger shift of mean COP position ( $11\pm 3$  mm) toward the intact leg (Duclos, Roll, Kavounoudias, Mongeau, Roll & Forget, 2009).

In addition to increased weight bearing over the intact leg, individuals with a unilateral BK amputation have a greater COP excursion under the intact leg. Nadollek et al. (2002) found greater point-to-point displacement of the COP in the AP direction under the intact ( $8.40\pm 4.16$  mm) compared to the prosthetic leg ( $4.34\pm 2.39$  mm). Similarly, another study observed greater root mean square distance of the COP in the AP direction (intact leg:  $5.75\pm 3.3$  mm; prosthetic leg:  $2.82\pm 2.3$  mm) (Jayakaran, Johnson & Sullivan, 2015). Again when the task becomes more difficult through the removal of vision, there is a more pronounced effect, with both studies (Jayakaran et al., 2015; Nadollek et al., 2002) observing a doubling of COP excursions in the AP direction.

The magnitude of difference in AP COP excursions between the prosthetic and intact leg does not extend to the ML direction. However, when compared to individuals without amputation, individuals with a unilateral BK amputation show

greater ML sway. When standing with eyes open, individuals with a unilateral BK amputation demonstrate greater mean ML sway amplitude ( $5.7 \pm 1.6$  mm) compared to controls ( $4.7 \pm 1.4$  mm), with a stronger effect when standing with the eyes closed (BK amputees:  $7.2 \pm 2.3$  mm; control:  $6.0 \pm 2.3$  mm) (Hermodsson, Ekdahl, Persson, & Roxendal, 1994). Other findings indicate that the ML COP excursion range is also larger in individuals with a unilateral BK amputation compared to those without amputation (BK amputees: range of ~18 mm; control: ~10 mm range) (Buckley, Driscoll, & Bennett, 2002).

When the standing balance task incorporates movements of the support surface, similar effects of weight bearing asymmetry and COP excursions are observed. The degree of weight bearing asymmetry is greater when standing balance is perturbed (e.g., with sinusoidal platform movements or waist-pulls), ranging from 55%-69% of weight bearing over the intact limb (Curtze, Hof, Postema, & Otten, 2012; Nederhand et al., 2012; Vrieling, van Keeken, Schoppen, Otten, Hof, et al., 2008). Also, sinusoidal platform movements in the AP direction elicit a 2.5x larger COP displacement in the intact leg compared to the prosthetic leg (Vrieling et al., 2008), closely resembling results found during static standing balance (Hermodsson et al., 1994). When participants were blindfolded, there was a further increase in COP displacement under the intact but not the prosthetic leg relative to the eyes open condition (Vrieling et al., 2008). This may indicate that as the task becomes more challenging, individuals with a unilateral BK amputation rely more heavily on the intact leg for body stability.

Research conducted by Curtze and colleagues (2012) further highlights the reliance on the intact leg for stability. Curtze et al. (2012) used a waist-pull method to induce a loss of balance in the AP and ML directions, and found that a backward waist-pull resulted in a 30% larger ankle plantarflexion moment in the intact compared to the prosthetic leg. This suggests that there is a greater contribution of the intact leg for balance recovery. The discrepancy in joint contribution is even greater after a forward waist-pull, where an 85% larger ankle plantarflexion moment was produced by the intact compared to the prosthetic leg (Curtze et al., 2012). This contrasts with the able-bodied control group, who displayed relatively symmetrical ankle plantarflexion moments between the left and right legs (8-14% difference) for both forward and backward waist-pulls (Curtze et al., 2012). For these AP waist-pulls, the ankle moments are of interest due to the predominance of the ankle strategy for balance recovery in these directions. The perturbation magnitude was not large, and therefore did not require a hip strategy, as demonstrated by the small hip moments produced by both the unilateral BK amputee and able-bodied groups.

In contrast to AP waist-pulls, ML waist-pulls resulted in direction-dependent hip moments. For waist-pulls inducing a loss of balance toward the intact limb, there was a 39% larger hip abduction moment on the intact compared to the prosthetic leg. Conversely, when the induced loss of balance was directed toward the prosthetic leg, there was a 30% larger hip abduction moment produced at the hip of the prosthetic compared to the intact leg, but this difference did not reach statistical significance (Curtze et al., 2012). Conversely, able-bodied controls exhibited 41% greater hip adduction moments at the hip joint contralateral to the direction of balance loss.

Together, these data highlight the presence of asymmetrical feet-in-place balance recovery strategies in individuals with a unilateral BK amputation. Much of the research examining balance control in individuals with a unilateral BK amputation has provided evidence of asymmetrical balance strategies, but the precise adaptations in stepping strategies following lateral perturbations are still unknown.

#### **1.4.5. Asymmetrical Reactive Stepping in the Sagittal Plane**

Curtze and colleagues (2010) used a tether-and-release method to induce a forward fall and describe the stepping strategy used by individuals with a unilateral BK amputation. The prosthetic leg was found to be the predominant stepping leg across participants, even when the intact leg was identified as an individual's preferred stepping leg. Those who preferred to step with their intact leg were able to recover their balance using one step (as opposed to multiple steps) more frequently when stepping with their prosthetic leg compared to their intact leg (78% of trials resulting in one step (prosthetic step) compared to 65% of trials resulting in one step (intact step)). A similar trend was seen for those individuals who preferred stepping with their prosthetic leg compared to their intact leg [53.6% of trials resulting in one step (prosthetic step); 42.9% of trials resulting in one step (intact step)]. Also, recovery steps performed with the prosthetic leg were characterized by a faster step initiation time and a larger step length (Curtze, Hof, Otten, & Postema, 2010).

The findings of this study may have important implications for reactive stepping in the lateral direction. For example, given that asymmetrical stepping responses

were observed in response to symmetrical perturbations in the AP direction, this may suggest that further asymmetry will be observed following a lateral perturbation.

## **2. Rationale, Purposes, Research Questions, and Hypotheses**

### **2.1. Rationale**

Although there is plenty of evidence that individuals with a unilateral lower limb amputation are falling, there is a lack of information detailing how and why they are falling. One risk factor for falls is the inability to step rapidly in different directions (Dite, Connor, & Curtis, 2007). While much research has investigated how individuals with a unilateral amputation recover their balance in the AP direction, few have examined how individuals with a unilateral amputation rely on stepping strategies in response to a loss of balance in the lateral direction.

Balance recovery in the lateral direction is unique because of the passive loading of weight onto one leg and the passive unloading of the other. This asymmetrical response leads to direction-dependent recovery strategies in people with poor balance control. For example, recent work with older adult and stroke populations has shown that these individuals require multiple steps to recover their balance in 60-70% of perturbation trials, compared to only 12.5% of trials in healthy young adults (Gray et al., 2017; Mille et al., 2005; Young et al., 2013). Also, the stepping strategies used by these at-risk populations is often different than in able-bodied controls. Many older adults respond to a lateral perturbation with a COS strategy, which is inherently riskier than other stepping strategies due to the high chance of limb-to-limb contact during the COS which could lead to a fall (Mille et al.,

2005; Young et al., 2013), while individuals with stroke tend to avoid stepping with their paretic leg first regardless of perturbation direction, and use a COS strategy frequently when falling toward the paretic side (Gray et al., 2017). Thus, these findings, along with the obvious unilateral somatosensory and musculoskeletal changes experienced with a unilateral amputation and the associated alterations to balance control, provide reason to investigate the stepping strategies used by individuals with a unilateral amputation following lateral perturbations.

## **2.2. Purposes**

The purposes of this thesis are (i) to identify the stepping strategies used by individuals with a unilateral lower limb amputation when perturbed in the lateral direction, and to compare them to individuals without amputation, and (ii) to identify whether the direction of perturbation changes the type of stepping strategy used, and (iii) explore the relationship between collected measures and variables associated with lower limb amputation, such as level of amputation and time since amputation.

## **2.3. Research Questions**

- 1) Do individuals with a lower limb amputation use the same reactive stepping strategies as those without amputation?
- 2) Do individuals with a unilateral lower limb amputation exhibit direction-dependent reactive stepping strategies following a lateral surface translation?

3) Do individuals with a unilateral lower limb amputation exhibit faster first step onset latencies when stepping with the prosthesis? How does this compare to individuals without amputation?

4) Is there an association with the stepping strategies utilized by individuals with a unilateral lower limb amputation and their level of amputation or experience with a prosthesis?

#### **2.4. Hypotheses**

1) It is hypothesized that individuals with a lower limb amputation will require more steps to recover their balance compared to individuals without amputation.

2) It is hypothesized that individuals without amputation will exhibit the same stepping strategy regardless of perturbation direction (Young, 2013; Mille, 2005), while individuals with a unilateral amputation will exhibit a COS strategy following perturbations toward the intact leg, and a higher frequency of multiple step trials following perturbations toward the prosthesis.

3) It is hypothesized that a step taken with the prosthetic leg will be characterized by a faster step initiation time (Curtze et al., 2010) compared to a step taken with the intact leg.

4) It is hypothesized that AK amputees and less experienced amputees will have fewer successful balance recovery trials, and will require more steps to recover balance.

### 3. Methods

#### 3.1. Participants and Recruitment

Eight individuals with a unilateral lower limb amputation (AK=2; BK=6) ( $63.38 \pm 13.95$  years of age, range 38-81;  $167.64 \pm 6.25$  cm in height;  $79.16 \pm 18.56$  kg in weight) and ten adults without amputation ( $66.5 \pm 8.64$  years of age, range 55-79;  $168.65 \pm 9.6$  cm in height;  $71.25 \pm 14.35$  kg in weight) participated in this study (Table 1). All participants had to be able to stand for 60 seconds and walk for 10 metres without assistance. Participants were recruited from the Niagara and Hamilton regions via advertisements posted at Niagara Prosthetics & Orthotics, Niagara Amputee Association, Hamilton Orthotics & Prosthetics, and their associated websites. All participants provided informed consent prior to participating in this study, and ethics clearance was granted by the Bioscience Research Ethics Board at Brock University.

<i>ID</i>	<i>Sex</i>	<i>Height (cm)</i>	<i>Weight (kg)</i>	<i>Age (yrs)</i>	<i>Years Since Amputation (yrs)</i>	<i>Cause of Amputation</i>	<i>Level of Amputation</i>	<i>Falls (1 yr)</i>
<i>A1</i>	M	173	70	65	25	Traumatic	AK	1
<i>A2</i>	F	167	58	53	6	Traumatic	AK	1
<i>B1</i>	M	175	94	77	41	Traumatic	BK	0
<i>B2</i>	M	173	109	67	3	Diabetes	BK	1
<i>B3</i>	M	170	65	56	8	Diabetes	BK	0
<i>B4</i>	F	162	97	38	37	Congenital	BK	1
<i>B5</i>	F	157	63	81	43	Cancer	BK	1
<i>B6</i>	F	162	74	70	61	Congenital	BK	0

**Table 1:** Individual amputee participant characteristics.



### 3.2. Questionnaires and Functional Assessment

Participants provided demographic information in the form of their age, sex, height, and weight. Participants were also asked which leg they prefer to kick with in order to determine leg dominance (van Melick, Meddeler, Hoogeboom, Nijhuis-van der Sanden, & van Cingel, 2017). Additionally, participants with a unilateral lower limb amputation were asked which leg they preferred kicking with prior to their amputation. The preferred kicking leg was denoted as the dominant leg, whereas the non-kicking leg was denoted as the non-dominant leg. Next, participants were asked to recall the number of falls they have experienced over the past year. A fall was defined as "an unexpected event in which an individual comes to rest on the ground, floor, or lower level" (W. C. Miller et al., 2001).

Next, all participants completed the Activities Specific Balance Confidence (ABC) Scale to determine their balance confidence when performing a variety of daily activities (W. C. Miller, Deathe, & Speechley, 2003) and the Fall Risk Questionnaire (FRQ) to determine their risk of experiencing a future fall (Rubenstein, Vivrette, Harker, Stevens, & Kramer, 2011). Participants with a unilateral lower limb amputation also completed the Locomotor Capabilities Index (LCI), a 14-item interview designed to assess one's ability to perform daily activities with their prosthesis (Dite et al., 2007).

To assess each participant's mobility, the Timed Up and Go (TUG) test and the TUG Cognitive (TUG<sub>cog</sub>) were administered. The TUG is a valid, reliable, and clinically appropriate mobility measure among older adults (Shumway-Cook, Brauer, &

Woollacott, 2000) and individuals with a unilateral BK amputation (Schoppen, Boonstra, Groothoff, de Vries, Goeken & Eisma, 1999). Younger individuals with a unilateral BK amputation have shown ceiling effects in the TUG, therefore the TUG<sub>cog</sub> was implemented as a more challenging variation of the test. For the TUG, participants stood up from a seated position, walked forward 3 m, turned around 180°, walked back 3 m to the chair and sat down. The TUG<sub>cog</sub> is the same as the TUG except that the participant counted backward by seven's or three's from a randomly selected number between 20 and 100 while performing the mobility task (Shumway-Cook et al., 2000). Performance for both variations of the TUG was measured by the time it takes to complete the task, starting when the participant stands up, and finishing when they sit back down. For both TUG variations, two trials were collected and the average time from the two trials was used as the performance measure.

Next, cutaneous sensitivity of the foot sole and residual limb was assessed using a set of Semmes-Weinstein monofilaments (Baseline Tactile Semmes-Weinstein Monofilament Sensory Evaluator Set, Fabrication Enterprises Inc., White Plains, NY, USA). Amputees were assessed at the skin over the first metatarsal of the intact foot, at the medial aspect of the distal 1/3<sup>rd</sup> of the residual leg, and at an equivalent location on the intact leg. Non-amputee controls were assessed at the skin over the first metatarsal of the right foot, and at the medial aspect of the right knee, in-line with the tibial tuberosity. Participants sat in a chair with their eyes closed and their feet propped up. Beginning with the thinnest monofilament, the monofilaments were applied perpendicular to the location of interest with enough force to bend the monofilament for one second. Monofilaments were applied with increasing thickness

until the tactile threshold, determined as the monofilament thickness that was correctly identified for three out of five trials at a particular location, was observed.

### **3.3. Participant Set-up**

After completing the questionnaires and functional assessments, participants were fitted with a safety harness that was attached to an overhead track. The harness eliminated the possibility of participants falling to the ground during the surface translation trials. Then, infrared markers were placed at the great toe and heel of each foot. Movement of these markers were monitored using a three-dimensional motion analysis system (3D Investigator system, Northern Digital Inc., Waterloo, ON, Canada) to record the first step onset latency (time from surface translation onset to toe-off).

### **3.4. Quiet Standing Trials**

Participants stood with a self-selected stance width of <40 cm from heel to heel on a forceplate (AMTI, OR6-7-2000, Watertown, MA, USA) that was located at the centre of a 1.8 m x 0.9 m wooden platform. They were asked to "stand quietly" while keeping their arms at their sides and looking straight ahead at an 'X' on the wall in front of them. Participants stood for two trials of 60 s. During these trials, COP data from the force platform was sampled at 1000 Hz (micro1401, Cambridge Electronics Design, UK). AP and ML COP data obtained during each quiet standing trial were low-pass Butterworth filtered at 5 Hz before determining the standard deviation during each 60 s trial. The COP data was also differentiated to obtain AP and ML velocities. The collection of quiet standing COP data was to provide additional descriptive data

regarding the postural control of the participants. This data was used to infer the level of active control an individual exhibited during a static balance task.

Independent to the quiet standing measures, COP data was collected during the 5 s period preceding each surface translation. This COP data was collected at the same frequency as in the quiet standing trials in order to determine the ML COP position during the period prior to each surface translation and to ensure that participants were not anticipating the upcoming surface translation. Each Optotrak foot marker was located in order to determine the location of the centre of each individual's stance. This was then referenced to the centre of the platform, which is used as the reference point for data output in Spike2. By subtracting the location of the centre of the platform by the location of the centre of stance, the ML COP position was determined. Positive and negative ML COP position corresponds to a shift of the COP toward and away from the prosthesis (or non-dominant leg), respectively.

### **3.5. Surface Translations**

Participants stood on the same forceplate as the quiet standing trials with their feet placed in the same self-selected position, their arms at their sides, and their gaze directed in front of them. At a time unknown to the participant, the platform in which the forceplate was embedded, rapidly translated in the leftward or rightward direction. This was achieved using a motor-driven 4.3 m linear stage (H2W Technologies Inc., Valencia, CA, USA). Each surface translation had a 300 ms acceleration phase, followed by a 500 ms constant velocity phase (peak velocity of 0.8 m/s), and finished with a 300 ms deceleration phase (peak deceleration of 0.5

m/s<sup>2</sup>). The peak acceleration depended on the individual's step threshold (see below). The total platform displacement was 50 cm. In response to each translation, participants were instructed to do whatever they needed to do in order to keep themselves from falling over. Prior to the experimental trials, participants were shown an example surface translation, and then experienced two practice trials (one leftward and one rightward) at a peak acceleration of 0.25 m/s<sup>2</sup>. For both the practice and experimental trials, two research assistants acted as spotters in case the participant was unable to recover their balance.

For the remainder of the document, surface translations will be referred to as 'Away' or 'Toward' surface translations, indicating in which direction the COM is passively shifted. For the amputee group, the direction of the loss of balance is referenced to the prosthesis (i.e., Away from the prosthesis or Toward the prosthesis). For example, the Away direction would refer to a surface translation that passively unloads the prosthesis. For the non-amputee group, the direction is referenced to the non-dominant leg (i.e., Away from the non-dominant leg or Toward the non-dominant leg).

In addition to the surface translation balance task, participants were engaged with a secondary task to divide their attention. The purpose for including a secondary cognitive task was to simulate an authentic reactive balance strategy. Participants were required to occupy their attentional resources with the cognitive task in addition to the balance control task, thereby reducing their ability to prepare themselves for the impending postural perturbation. Participants were presented with a sequence of nine letters, or nine numbers, spoken aloud by an automated voice

generation software ('Balabolka' for Windows). While standing on the forceplate in preparation for the surface translation onset, participants were asked to listen for a specific letter (or number), and to keep track of the amount of times this specific letter (or number) was mentioned in the nine-character sequence. The timing of the character sequence onset was offset relative to the surface translation; the surface translation onset was coded to occur at a randomized time between 5-10 seconds after the command was sent to the program. Therefore, the character sequence did not prompt the participant for the onset of the surface translation.

### **3.5.1. Determination of Single Step Threshold**

Single step threshold was defined as the minimum surface translation acceleration that elicited a stepping response for two consecutive trials. To determine each individual's single step threshold, the initial surface translation acceleration was  $0.25 \text{ m/s}^2$ . For each trial that did not elicit a step, the acceleration was increased by  $0.25 \text{ m/s}^2$ . When a stepping response was observed at a given acceleration, a second trial was completed at the same acceleration. This progression continued until a stepping response was elicited for two consecutive trials at the same acceleration. Participants were unaware of both the timing and direction of the surface translations, both of which were randomized. They were informed beforehand that the surface translation acceleration would gradually increase, however they were not informed each time the acceleration changed. Single step threshold was determined for each direction; thus, two step thresholds were determined (Away step threshold and Toward step threshold).

### 3.5.2. Determination of Stepping Strategies

Once the single step threshold was determined for each direction, participants experienced eight surface translations. Two surface translation accelerations were used based on step threshold: 120% of Away step threshold, and 120% of Toward step threshold. Four trials were completed in each direction, for a total of eight trials. Surface translation intensity was relative to each direction, meaning that Away translations always had an acceleration relative to Away step threshold, and Toward translations always had an acceleration relative to Toward step threshold. The timing, direction, and intensity of each translation was randomized.

In addition to the eight surface translations at 120% of step threshold, each participant experienced eight trials at 100% of step threshold. However, due to inconsistencies in stepping responses at 100% of step threshold, these trials were not included in the analyses.

All trials were video recorded for retrospective confirmation of stepping strategies. A Go Pro video camera recorded each surface translation trial, and the video was recorded from the mid-section of the participants body to the ground, so as to not include the participants face in the video. Each surface translation trial was analyzed by the primary researcher and the assistant researcher to determine the stepping strategy classification for each trial. A stepping sequence that involved an initial medial step that did not cross the sagittal plane of the loaded leg, followed by a lateral step with the loaded leg, was considered a MSS. For trials that involved a step with the unloaded leg that crossed the sagittal plane of the loaded leg was

considered a COS. Any trial that involved a step with the unloaded leg, but not the loaded leg, was considered an ULS. Additionally, the number of balance recovery steps were determined for each trial using the video recordings.

Through video observation, the quality of each balance recovery trial was determined using a three-item scale. Each trial was viewed and rated on two separate occasions by the primary investigator. The quality of balance recovery scale consisted of the following three items:

- i) Stable: A 'stable' trial was characterized by effective balance recovery with no external assistance.
- ii) Unstable: An 'unstable' trial was characterized by effective prevention of a fall with noticeable instability following the execution of a reactive stepping strategy. For example, if the participant was left in a vulnerable position following the stepping strategy, where an additional perturbation would have resulted in a fall, that trial was rated as unstable.
- iii) Fail: A 'failed' trial was characterized by clear external assistance from the spotters or safety harness. These trials would have hypothetically resulted in a fall without any external assistance.

### **3.6. Statistical Analysis**

Independent samples t-tests were used to compare scores from the ABC scale and the fall risk questionnaire between the amputee and non-amputee groups. To compare the walking times from the TUG and TUG<sub>cog</sub> tests between groups,



independent samples t-tests were used, while paired samples t-tests were used to compare walking times within groups.

Step initiation time of amputees and non-amputees was compared using independent samples t-tests.

Stepping strategies were assessed using a three variable Loglinear Analysis (Group x Direction x Strategy). Significant interactions were followed up with chi-square analyses at each level of the interacting variables.

All statistical tests were performed with SPSS software (SPSS, Chicago, IL, USA). For all statistical analyses, significance will be set at  $p \leq 0.05$ .

## **4. Results**

### **4.1. Non-Amputee Group and Amputee Group**

#### **4.1.1. TUG Tests**

Paired samples t-tests were completed separately for each group and showed that both the amputee group ( $t_7 = -4.11, p = 0.005$ ) and the non-amputee group ( $t_9 = -4.87, p = 0.001$ ) took more time to complete the TUG<sub>cog</sub> than the TUG (Table 2). Independent samples t-tests showed no difference in TUG ( $t_{16} = -2.08, p = 0.054$ ) or the TUG<sub>cog</sub> ( $t_{16} = -0.039, p = 0.96$ ) times between groups (Table 2).

	<i>Non-Amputee</i>	<i>Amputee</i>	<i>p-value</i>
<i>Sex</i>	4 M, 6 F	4 M, 4 F	
<i>Age (years)</i>	66.5 ± 8.6; range 55-79	63.4 ± 13.9; range 38-81	0.56
<i>Height (cm)</i>	168.6 ± 9.6	167.6 ± 6.2	0.80
<i>Weight (kg)</i>	71.2 ± 14.3	79.1 ± 18.5	0.32
<i>TUG (s)</i>	7.5 ± 1.7; range 5.7-11.6	8.9 ± 1.1; range 7.7-10.7	0.54
<i>TUG<sub>cog</sub> (s)</i>	9.9 ± 3.1; range 6.4-17.3	10.0 ± 1.4; range 7.9-11.7	0.96
<i>ABC (%)</i>	95.8 ± 4.8; range 86-100	91.8 ± 5.7; range 80-100	0.12
<i>FRQ (/14)</i>	1.6 ± 2.2; range 0-6	2.0 ± 1.3; range 0-4	0.65
<i>LCI (/56)</i>	n/a	53.6 ± 2.7; range 49-56	
<i>Fallers</i>	2/10 (20%)	5/8 (63%)	

**Table 2:** Participant characteristics and group means ± 1 SD [TUG= Timed Up and Go; TUG<sub>cog</sub>= Timed Up and Go (Cognitive); ABC= Activities Specific Balance Confidence Scale; FRQ= Fall Risk Questionnaire; LCI= Locomotor Capabilities Index].

#### 4.1.2. ABC-Scale and FRQ

No difference was found for the ABC-scale scores between amputees (91.80 ± 5.74%) and non-amputees (95.84 ± 4.83%) ( $t_{16} = 1.62$ ,  $p = 0.12$ ) (Table 2). Also, FRQ scores were not different between amputees (2.0 ± 1.31) and non-amputees (1.6 ± 2.22) ( $t_{16} = -0.44$ ,  $p = 0.65$ ) (Table 2).

#### 4.1.3. Monofilament Tactile Threshold

No differences were observed in the monofilament tactile threshold (Table 3) between the intact foot of amputees and the right foot of non-amputees at the first metatarsal head ( $t_{16} = -1.80$ ,  $p = 0.09$ ), or the heel ( $t_{16} = -1.76$ ,  $p = 0.09$ ). No difference was found in the tactile threshold of the intact leg of amputees ( $t_{16} = -1.68$ ,  $p = 0.113$ ), or the distal 1/3<sup>rd</sup> of the residual leg of amputees ( $t_{16} = -0.753$ ,  $p = 0.462$ ), compared to the right leg of non-amputees. Furthermore, the tactile threshold was not

significantly different between the residual leg and the intact leg in amputees ( $t_7 = -0.247$ ,  $p = 0.812$ ).

	<i>Non-Amputee</i>	<i>Amputee</i>	<i>P-Value</i>
<i>1<sup>st</sup> Metatarsal</i>	1.90 ± 0.41 g	22.90 ± 13.13 g	0.09
<i>Heel</i>	3.24 ± 0.49 g	21.60 ± 11.77 g	0.09
<i>Intact Leg</i>		1.25 ± 0.22 g	0.11
<i>Residual Leg</i>	0.84 ± 0.13 g	1.17 ± 0.47 g	0.46

**Table 3:** Mean ± 1 SE monofilament tactile threshold of amputees and non-amputees. All units are expressed as grams of force required to bend the monofilament.

#### 4.1.4. COP During Quiet Standing

The SD of position and velocity COP data obtained during 60 second quiet standing trials are shown in Table 4. Differences were observed between amputees and non-amputees in the SD of the ML velocity ( $t_{16} = -2.569$ ,  $p = 0.021$ ) and the SD of the AP velocity ( $t_{16} = -2.581$ ,  $p = 0.020$ ), where amputees experience greater fluctuations of the COP during quiet standing than non-amputees. No differences were found in the SD of the ML position ( $t_{16} = -1.960$ ,  $p = 0.068$ ) or the AP position ( $t_{16} = 0.128$ ,  $p = 0.900$ ) between groups.

	<i>ML COP</i>			<i>AP COP</i>		
	Non-Amputee	Amputee	P-value	Non-Amputee	Amputee	P-value
<i>SD of COP Position (mm)</i>	2.57 ± 0.26	4.14 ± 0.84	0.07	6.02 ± 0.62	5.90 ± 0.63	0.90
<i>SD of COP Velocity (mm/s)</i>	3.69 ± 0.25	8.33 ± 2.01	0.02*	8.84 ± 0.78	14.68 ± 2.35	0.02*

**Table 4:** SD of the medio-lateral (ML) and antero-posterior (AP) COP position and velocity (mean ± 1 SE).

#### 4.1.5. Surface Translations

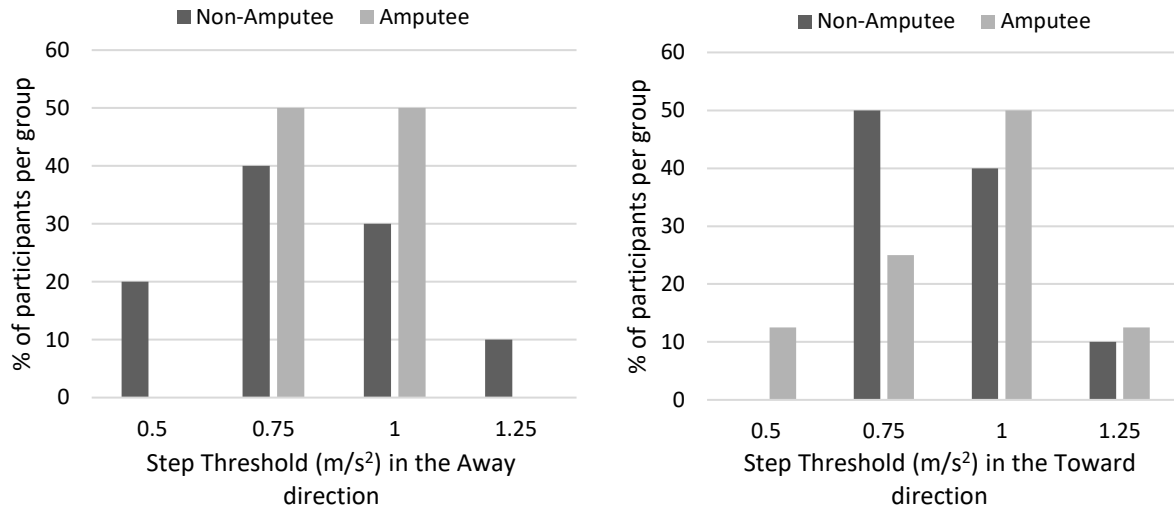
The following sections will examine the measures related to the surface translations trials, including: step threshold, first step onset latency, COP position prior to perturbation, and reactive stepping strategies (preferred strategy, quality of balance recovery, and recovery step count). Individual data for each of these measures can be found in Table 5. Each of these measures will first be statistically compared between non-amputees and amputees. Next, exploratory analyses will be performed within the amputee group, where the influence of the level of amputation (AK or BK) and the time since amputation will be examined for each of the collected measures. These analyses are exploratory due to the small size and large variability within the amputee group, which deteriorates the strength of statistical analysis. In addition, the effect of age will be explored in each section.

ID	Preferred Strategy		Step Threshold (m/s <sup>2</sup> )		Stable Trials (/4)		Avg. # of Recovery Steps per Trial		1 <sup>st</sup> Step Onset Latency (ms)		SD of COP Position During QS (mm)		SD of COP Velocity During QS (mm)		ML COP Position Prior to ST (mm)	Avg. Stance Width (cm)
	Dir.	A	T	A	T	A	T	A	T	A	T	ML	AP	ML	AP	
A1	MSS	ULS	1.0	1.0	3	2	2	1.25	322	272	5.9	7.0	15.7	21.0	11.5	32.1
A2	MSS	MSS	0.75	0.5	0	3	3.75	3	320	547	8.7	7.9	18.4	27.7	-18.7	24.1
B1	MSS	MSS	1.0	1.0	4	4	2	2	277	285	1.4	6.7	3.5	10.8	-1.5	27.6
B2	ULS	MSS	1.0	0.75	3	4	1.25	2	360	305	3.5	7.7	7.9	14.7	11.6	35.4
B3	ULS	ULS	0.75	1.0	3	1	1	1	310	397	4.6	5.1	6.8	13.8	-12.9	28.6
B4	COS	COS/ ULS	0.75	1.25	1	3	2.75	1	337	290	4.2	5.5	6.4	11.2	-22.9	28.4
B5	COS	COS	1.0	1.0	0	0	1.75	2	342	292	2.1	4.4	4.3	11.0	9.9	25.9
B6	ULS	MSS	0.75	0.75	4	4	1.5	1.75	250	262	2.5	2.7	3.3	7.1	-31.4	26.0
C1	MSS	COS	0.5	0.75	4	4	2	3	337	400	2.9	4.5	2.9	6.5	27.1	25.1
C2	ULS	MSS	0.5	1.0	4	4	2	1	342	307	3.5	5.6	4.6	12.1	21.1	29.8
C3	MSS	MSS	1.0	1.25	4	4	2	2	256	263	1.9	5.2	3.4	7.4	-6.2	27.9
C4	COS	COS	1.0	1.0	4	4	1.75	2	292	285	1.1	3.5	3.0	5.4	40.9	25.0
C5	MSS	MSS	1.0	1.0	4	4	2	2	257	232	3.2	10.3	5.3	9.5	10.6	24.7
C6	MSS	MSS	0.75	0.75	4	4	2.5	3	290	307	1.8	7.9	3.1	8.9	1.7	23.6
C7	MSS	MSS	0.75	0.75	2	0	3.33	4	273	266	1.9	5.6	4.0	9.2	-6.3	25.5
					(/3)	(/3)										
C8	MSS	COS	0.75	0.75	4	4	2	2	295	290	3.2	4.9	3.2	6.7	-1.4	22.4
C9	MSS	COS	0.75	0.75	4	4	2	2	262	267	2.2	5.0	3.1	9.1	15.6	23.4
C10	MSS	MSS	1.25	1.0	4	4	2	2	207	207	3.5	7.2	4.0	13.2	-16.8	28.2

**Table 5:** Individual data for preferred stepping strategy, step threshold, number of stable balance recovery trials, average number of recovery steps per trial, first step onset latency for Away (A) and Toward (T) trials, SD of COP position and SD of COP velocity in ML and AP direction during quiet standing, average ML COP position during 5 s prior to surface translation (positive values indicate COP position toward the prosthesis/non-dominant leg, negative values indicate COP position away from the prosthesis/non-dominant leg) and average stance width. Participant ID's are represented with a subject number and an 'A' indicating AK amputee, 'B' indicating BK amputee, or C indicating non-amputee control.

#### 4.1.5.1. Single Step Threshold (Non-Amputee and Amputee)

No differences were found for the single step threshold in the Away ( $t_{16} = -0.531$ ,  $p = 0.603$ ) or the Toward direction ( $t_{16} = -0.066$ ,  $p = 0.948$ ) between groups (Figure 2). No differences were observed in single step threshold between directions within the amputee ( $t_7 = -0.35$ ,  $p = 0.73$ ) or non-amputee group ( $t_9 = -1.15$ ,  $p = 0.27$ ). These results are corroborated by the number of participants who exhibited identical step thresholds in both directions. Four of eight amputees (or 50%) and six of ten non-amputees (or 60%) demonstrated the same step threshold in each direction.



**Figure 2:** Single step threshold for the non-amputee (dark-grey) and amputee (light-grey) groups in the Away (left) and Toward (right) directions.

#### 4.1.5.2. First Step Onset Latency (Non-Amputee and Amputee)

The first step onset latency was not different between directions in the amputee [ $315.0 \pm 12.7$  ms (Away),  $331.6 \pm 34.2$  ms (Toward);  $t_7 = -0.499$ ,  $p = 0.648$ ] or non-amputee group [ $281.5 \pm 12.6$  ms (Away),  $282.8 \pm 16.4$  ms (Toward);  $t_9 = 0.119$ ,  $p = 0.883$ ]. Due to the lack of difference between directions in each group, the combined average of each direction was calculated (amputees:  $323.28 \pm 17.17$  ms; non-amputees:  $282.13 \pm 7.82$  ms), which revealed a difference between amputees and non-amputees, where amputees took longer to initiate the first balance recovery step ( $t_{16} = -1.839$ ,  $p = 0.041$ ).

#### 4.1.5.3. ML COP Prior to Perturbation (Non-Amputee and Amputee)

Independent samples t-test revealed no difference in the mean ML COP position prior to perturbation ( $t_{16} = -0.221$ ,  $p = 0.828$ ) between amputees ( $6.79 \pm 6.01$  mm

toward the intact leg) and non-amputees ( $8.62 \pm 5.60$  mm toward the non-dominant leg).

#### 4.1.5.4. Reactive Stepping Strategy Frequencies (Non-Amputee and Amputee)

A total of 142 perturbation trials at 120% of stepping threshold were analyzed (non-amputees= 78; amputees= 64). Analyses focused on exploring the relationship between stepping strategy (COS, MSS, ULS) with group (non-amputee and amputee) and direction (Away and Toward). Stepping strategy frequencies can be found in Table 6.

The following section will examine the difference between amputees and non-amputees for each measure. Additionally, the influence of age was examined for certain measures.

	<i>Non-Amputees</i>		<i>Amputees</i>	
	# of Away trials (%)	# of Toward trials (%)	# of Away trials (%)	# of Toward trials (%)
<i>MSS</i>	31 (79.5)	23 (58.9)	11 (34.4)	14 (43.8)
<i>COS</i>	4 (10.2)	15 (38.5)	9 (28.1)	8 (25.0)
<i>ULS</i>	4 (10.2)	1 (2.6)	12 (37.5)	10 (31.2)
<i>Total</i>	39 trials	39 trials	32 trials	32 trials

**Table 6:** Reactive stepping strategy frequency for the non-amputee and amputee groups.

The preferred strategy (i.e., the most frequently used strategy for each direction) was identified for each participant. In the non-amputee group, six of ten

(60%) participants had the same preferred strategy in each direction, while four of eight (50%) amputees had the same preferred strategy.

#### 4.1.5.5. Quality of Balance Recovery (Non-Amputee and Amputee)

Each surface translation trial was rated as ‘Stable’ (effective balance recovery with no external assistance), ‘Unstable’ (effective prevention of a fall but with noticeable instability following the reactive stepping strategy), or ‘Fail’ (clear external assistance by spotter/harness).

Non-amputees were stable in their balance recovery in 95% of trials (74 of 78 trials rated as stable). Only four trials required assistance from the spotters, and each of those four trials were attributed to one participant (C7).

The amputee group had fewer stable trials compared to non-amputees, with only 53% and 66% of Away and Toward trials, respectively, rated as stable. One participant (B5) did not record a single stable trial, therefore contributing four unstable or failed trials in both directions (Table 7).

	<i>NON-AMPUTEE</i>			<i>AMPUTEE</i>					
	Combined Away & Toward Trials			Away Trials			Toward Trials		
	<i>Stable</i>	<i>Unstable</i>	<i>Fail</i>	<i>Stable</i>	<i>Unstable</i>	<i>Fail</i>	<i>Stable</i>	<i>Unstable</i>	<i>Fail</i>
<b>MSS</b>	50	1	3	7	2	2	13	1	0
<b>COS</b>	19	0	0	2	3	4	2	3	3
<b>ULS</b>	5	0	0	8	2	2	6	3	1
<b>TOTAL</b>	74	1	3	17	7	8	21	7	4

**Table 7:** Quality of balance recovery per stepping strategy in non-amputees and amputees in the Away and Toward conditions.



#### 4.1.5.6. Recovery Steps Taken (Non-Amputee and Amputee)

The number of recovery steps taken during balance recovery was recorded for each trial. Non-amputees used multiple steps for 97% and 90% of Away and Toward trials. The MSS is comprised of a minimum of two steps, and the non-amputee group used the MSS strategy for 54 of 78 (69%) of all trials. In contrast, the amputee group relied on multiple steps for fewer trials (Away= 60%; Toward= 62%) than the non-amputee group (Table 8). This may be attributed to differences in strategies between groups, where the amputee group frequently used the ULS strategy. The ULS strategy involves excessive abduction of the unloaded leg and trunk tilt in the direction of perturbation, a strategy that often requires only one step.

	<i>Non-Amputee</i>		<i>Amputee</i>	
	<i>Away</i>	<i>Toward</i>	<i>Away</i>	<i>Toward</i>
<i>Single Step Trials</i>	1 (3%)	4 (10%)	10 (31%)	12 (38%)
<i>Multiple Step Trials</i>	38 (97%)	35 (90%)	22 (69%)	20 (62%)

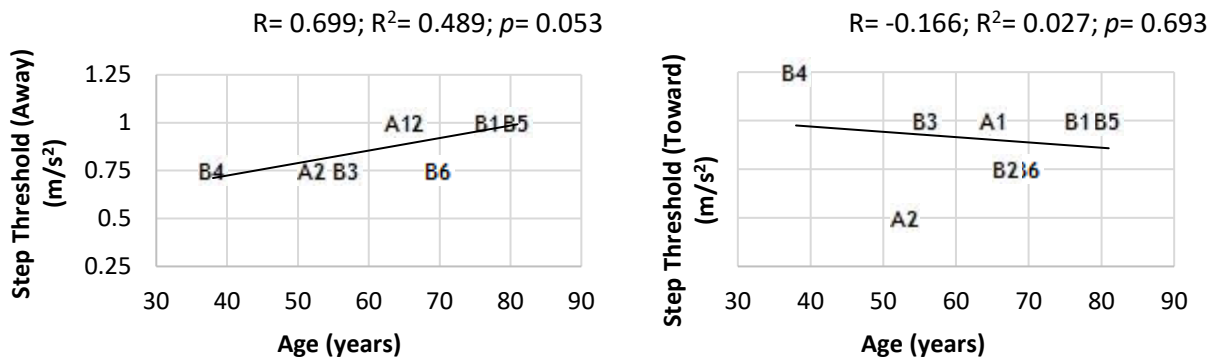
**Table 8:** Number of trials involving a single step or multiple steps in non-amputees and amputees.

#### 4.2. Exploratory Analyses (Amputee Group)

The following sections are focused on potential relationships within the amputee group. Factors such as amputation level (AK or BK), age, and time since amputation are explored.

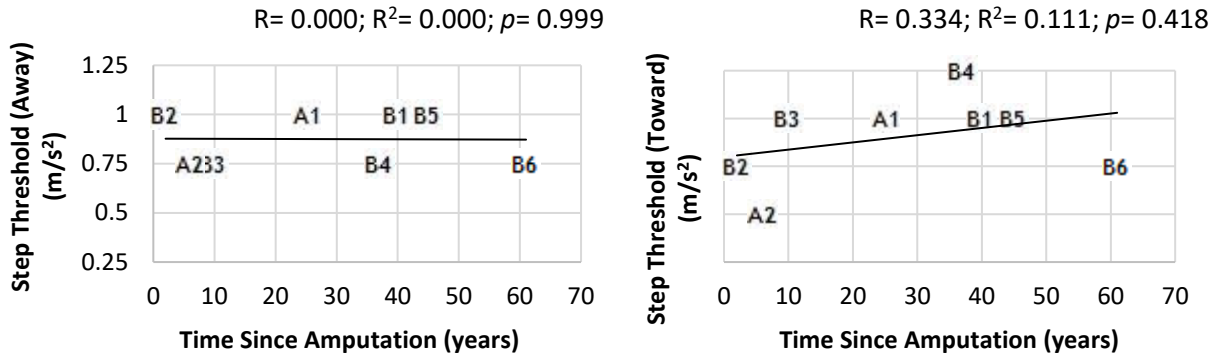
### 4.2.1. Single Step Threshold (Amputee Group)

There was no clear relationship between age and single step threshold within the amputee group (Figure 3). However, there was a near-significant association between age and step threshold for the Away trials ( $p= 0.054$ ). As noted in section 4.1.5.1., four of eight amputees had identical step thresholds for each direction (participant: A1, B1, B5, B6). For those amputees who exhibited different step thresholds in each direction, two participants (A2 and B2) had a higher step threshold Away from the prosthesis, while the remaining two amputee participants (B3 and B4) had a higher step threshold Toward the prosthesis.



**Figure 3:** Single step threshold for Away (left) and Toward (right) trials as a function of age.

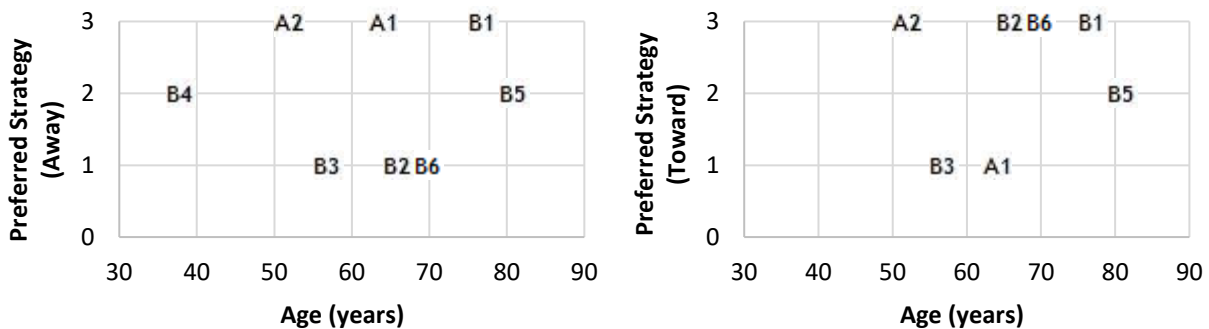
There appeared to be a relationship between the time since amputation and single step threshold (Figure 4). Each amputee participant with less than 10 years of experience with their prosthesis (participant: A2, B2, B3) exhibited different step thresholds for each direction.



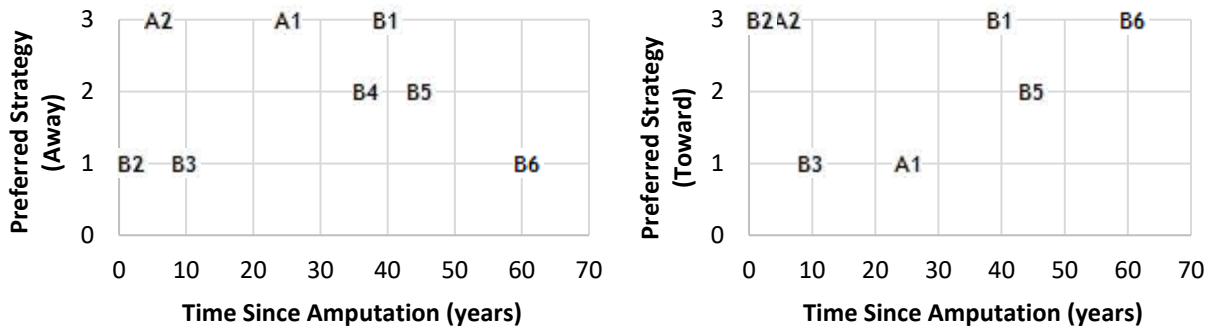
**Figure 4:** Single step threshold for Away (left) and Toward (right) trials as a function of time since amputation.

#### 4.2.2. Preferred Reactive Stepping Strategy (Amputee Group)

There did not appear to be a relationship between the type of stepping strategy and the age of the participant (Figure 5). Participants A1, B2, B4, and B6 all exhibited a different preferred strategy between directions. Of these four participants, three (excluding B4) used the MSS in one direction, and the ULS in the other. Time since amputation also did not influence the preferred stepping strategy (Figure 6).



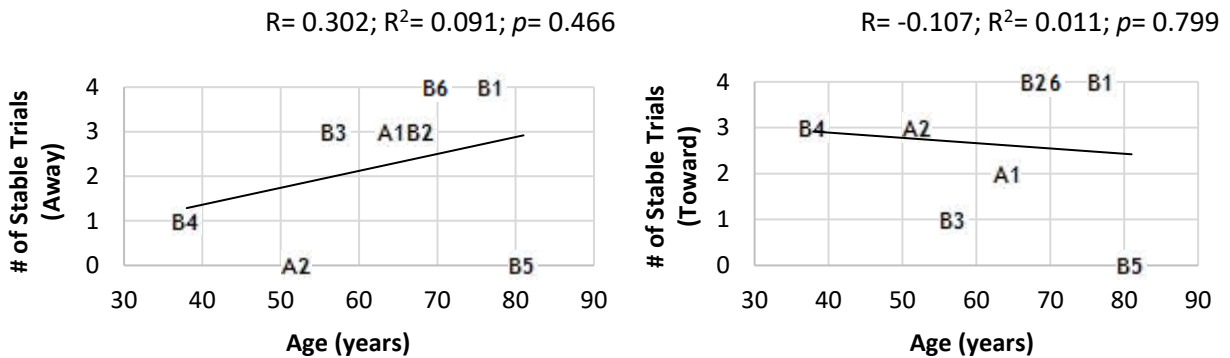
**Figure 5:** Preferred strategy for Away (left) and Toward (right) trials as a function of age. (3- MSS; 2- COS; 1- ULS). Note: participant B4 did not have a preferred strategy Toward.



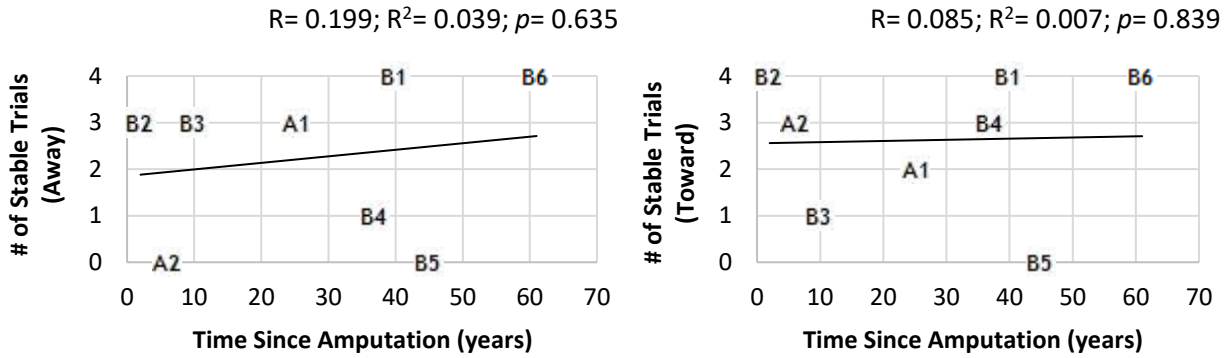
**Figure 6:** Preferred strategy for Away (left) and Toward (right) trials as a function of time since amputation. (3- MSS; 2- COS; 1- ULS). Note: participant B4 did not have a preferred strategy Toward.

#### 4.2.3. Quality of Balance Recovery (Amputee Group)

The quality of balance recovery was weakly influenced by age (Figure 7) or time since amputation (Figure 8). However, after omitting participant B5 (81 years of age; 43 years since amputation; zero stable trials) from the Away trials, age was associated with a greater number of stable trials in that direction ( $R^2= 0.648$ ,  $p= 0.029$ ).



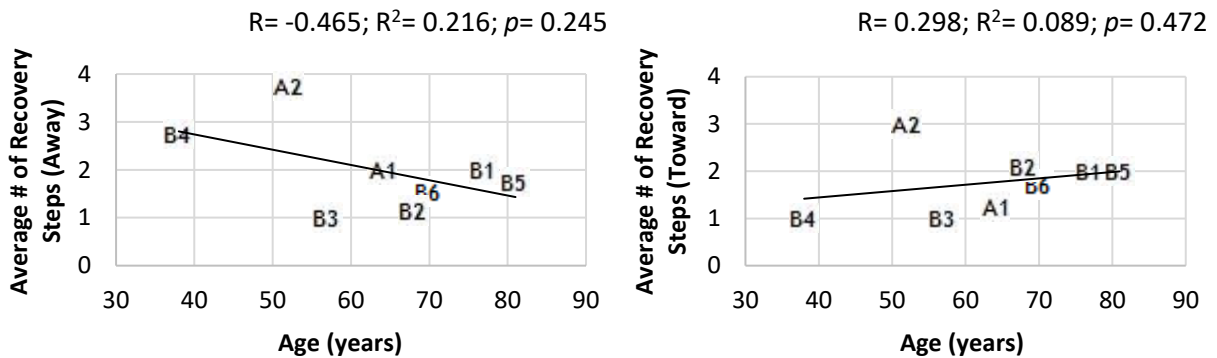
**Figure 7:** Number of stable trials for Away (left) and Toward (right) trials as a function of age.



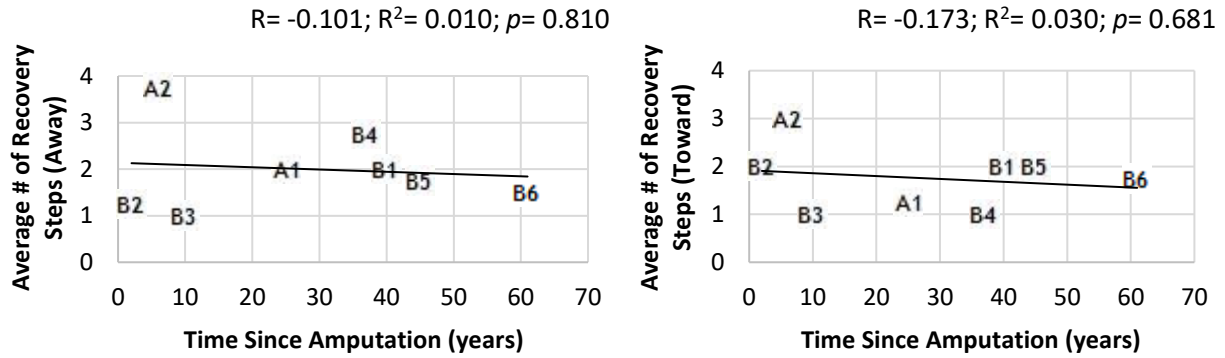
**Figure 8:** Number of stable trials for Away (left) and Toward (right) trials as a function of time since amputation.

#### 4.2.4. Recovery Steps Taken (Amputee Group)

There was no clear relationship between the number of recovery steps taken with age (Figure 9) or time since amputation (Figure 10). However, age appears to be related to fewer recovery steps for trials away from the prosthesis.



**Figure 9:** Average number of recovery steps per trial for Away (left) and Toward (right) trials as a function of age.



**Figure 10:** Average number of recovery steps per trial for Away (left) and Toward (right) trials as a function of time since amputation.

## 5. Discussion

The purposes of this thesis were (i) to identify the reactive stepping strategies used by lower limb amputees in response to a lateral surface translation, (ii) to examine the influence of perturbation direction on these reactive stepping strategies, and (iii) to explore the influence of age, level of amputation, and time since amputation on reactive balance measures. Contrary to the hypotheses, amputees required fewer steps to recover their balance compared to non-amputees. Furthermore, the frequency of each reactive stepping strategy was minimally influenced by direction, however the non-amputee group showed a greater difference in stepping strategy occurrence in each direction. No apparent relationships between age, level of amputation, or time since amputation were observed with respect to the collected reactive balance measures.

### 5.1. Unique Reactive Balance Control Among Amputees

Previous research examining the reactive balance control of lower limb amputees has highlighted differences between amputees and non-amputees.

Following medial-lateral treadmill perturbations to the leading foot during forward walking, amputees required an average of three steps more than non-amputees to prevent a fall (Segal, Orendurff, Czerniecki, Schoen & Klute, 2014). In the current study, amputees took fewer steps per trial to recover balance than non-amputees, with 34% of amputee trials resulting in a single step, compared to just 6% of non-amputee trials. However, amputees also recorded fewer successful balance trials, where only 59% of trials (compared to 95% of trials among non-amputees) did not require external assistance (i.e., spotter assistance). This may indicate that multiple steps are necessary to successfully recover one's balance following a lateral surface translation at a perturbation intensity greater than one's respective step threshold.

Also notable was the discrepancy in first step onset latency between groups. The average first step onset latency was 41 ms more delayed in the amputee group compared to non-amputees. This supports findings that have shown delayed balance recovery responses in individuals with unilateral stroke, where lower limb muscle balance response was delayed by 19 ms compared older adult controls (de Kam, Roelofs, Bruijnes, Geurts & Weerdesteyn, 2017). Furthermore, intact foot sole sensitivity was notably worse compared non-amputee controls (Table 3). The reduced tactile threshold among amputees may have contributed to the greater delay in first step onset, as there is evidence that reduced foot sole sensitivity can delay the onset of lower limb muscle responses following (Inglis, Horak, Shupert & Jones-Rycewicz, 1994; Ferguson, Polskaia & Tokuno, 2017). This is further exacerbated by the lack of foot or ankle sensory afference from the residual limb, and so a delay in first step onset should be expected from the amputee group.

A major difference between the non-amputee and amputee groups was the occurrence of the ULS strategy. The amputee group frequently exhibited the ULS strategy (22 of 64 trials; 34%), with five participants preferring to use an ULS in at least one direction. This was in contrast to the non-amputee group, where the ULS strategy was observed in only 5 of 78 trials (6%). Through video observation of the ULS trials, the ULS appeared to be an extension of the hip strategy, such that for the majority of ULS trials there was lateral trunk movement in the direction of the induced loss of balance. This would support research that has observed lower limb amputees to have an increased reliance on hip musculature for balance control. For example, when Segal et al. (2014) subjected the leading foot of lower limb amputees to medial-lateral treadmill perturbations during forward walking, they found that the ML inclination angle (i.e., the angle created by the vertical projection of the COM to the floor and the line through the COM and heel marker) decreased by 17-49% more than non-amputees following perturbations to both the prosthesis and intact foot (Segal et al., 2014). This indicates that amputees shift their weight laterally in the direction of the induced loss of balance, similar to what was observed in the current study, where the COP position of amputees was shifted  $6.79 \pm 6.01$  mm toward the intact leg. Similarly, a recent study found that amputees increase the amount of hip muscle work following medial treadmill perturbations (Miller, Segal, Klute, & Neptune, 2018). Findings from these studies, in conjunction with the current study, highlight the importance of the hip strategy for balance recovery in the lower limb amputee population.



## 5.2. Influence of Perturbation Direction on Reactive Stepping Strategies

Based on the study's classification system, the frequency of each strategy was similar in both directions. The MSS strategy was used in 34.4% and 43.8% (9.4% difference) of Away and Toward trials, respectively. The COS strategy was used in 28.1% and 25% (3.1% difference) of Away and Toward trials, respectively. Lastly, the ULS strategy was used in 37.5% and 31.2% (6.3% difference) of Away and Toward trials, respectively. In fact, contrary to hypotheses, the non-amputee group exhibited a greater effect of direction on the frequency of reactive stepping strategies (20.6% difference for MSS, 28.3% difference for COS, 7.6% difference for ULS).

Furthermore, perturbation direction did not influence the average number of steps required for balance recovery. Amputees required multiple steps in 69% and 62% of Away and Toward trials, respectively. This contrasts with the findings of Segal et al. (2014) where amputees required three additional steps to recover balance when the prosthesis was perturbed instead of the intact leg. The current study's findings also contrast with the research of Gray et al. (2017), who found that individuals with stroke demonstrated different stepping strategies depending on whether the perturbation was toward or away from the paretic leg. Specifically, stroke patients used a lateral step for 19% compared to 45% of trials toward and away from the paretic leg. However, the stroke patients in Gray et al. (2017) and the amputees in the current study appeared to avoid stepping with their affected leg.

The first step onset latency was not different between directions in either group. It should be noted however, that there was a 16 ms discrepancy between directions in the amputee group, where perturbations toward the prosthesis resulted

in a greater delay in the first step onset. This discrepancy is largely the result of one participant (A2), who exhibited excessive delays in step initiation for Toward trials. When participant A2 is omitted from the Toward trials, the average first step onset latency for the amputee group Toward trials changes from 331.56 ms to 300.71 ms. With this new group average, there is still a 14 ms discrepancy between directions, however the discrepancy is in the opposite direction, where perturbations away from the prosthesis result in a greater delay in first step onset latency. The lack of difference between directions does not support previous findings, where amputees were observed to initiate a recovery step 24-30 ms quicker with the prosthesis compared to the intact leg after a forward loss of balance (Curtze et al., 2010). However as noted elsewhere, the perturbation mechanism used by Curtze et al. (2010) allowed for perturbations in only the forward direction, thereby reducing the degree of unpredictability. This difference in findings could also be attributed to the fact that Curtze et al. (2010) used a forward lean perturbation, whereas the current study used lateral surface translations.

There was a relationship between age and the quality of balance recovery for perturbations away from the prosthesis, however only when participant B5 is omitted. Age was associated with a greater number of stable trials. Interestingly, the average number of recovery steps also appeared weakly correlated with age for trials away from the prosthesis, however this did not reach significance. But the trend suggests that age is associated with fewer balance recovery steps. Together, these relationships may suggest that multiple step balance recoveries when the prosthesis is unloaded as a result of external perturbation could be contributing to falls.

The frequency of COS trials in older adult non-amputees (24% of trials) corroborates the findings of previous studies using lateral surface translations to elicit reactive stepping in older adults. When perturbed from a standing position, older adults exhibited the COS in 30-31% of trials (King et al., 2008; Maki et al., 2000). Similarly, when perturbed while walking in place, the COS was used in 24% of trials (Maki et al., 2000). When older adults experienced either a waist-pull or a surface translation, older adults relied on a COS strategy in 30.7-32.2% of trials (Mansfield & Maki, 2008). However, some studies that used a waist-pull perturbation to elicit lateral stepping have reported higher frequencies of the COS, where older adults exhibited a COS in 51-59% of trials from a standing position (Mille et al., 2005; Young et al., 2013). This variability in results suggests that more research is required to compare the influence of perturbation type (i.e., waist-pull or surface translation) on the type of stepping strategy elicited, or that the frequency of certain stepping strategies is largely sample-dependent.

While the frequency of the COS strategy supports the findings of previous studies, the frequency of the MSS in the current study deviates from past findings. In the current study the MSS was the most frequently used strategy (69% of all trials by non-amputee older adults). In previous studies using surface translations to elicit a lateral loss of balance, a MSS strategy was either not observed (King et al., 2008), or observed in 56% of trials (Maki et al., 2000). It should be noted that King et al. (2008) required participants to stand with a narrow stance (4.5 cm stance width), which may have contributed to the lack of MSS responses, as there would not be a sufficient stepping area for the initial medial step to occur, whereas Maki et al. (2000) used a

stance width of 11% of body height (19 cm stance width for a 175 cm tall individual), which is closer to the stance width used in the current study (natural stance width: range of 22-35 cm stance width). Studies using a waist-pull perturbation to elicit a loss of balance have observed fewer MSS trials, with a 5-8% trial frequency (Young et al., 2013; Mille et al., 2005). Instead, the lateral side step strategy (a lateral step with the passively loaded leg), which was not seen in the current study, was used in 37% of trials (Young et al., 2013) and 33% of trials (Mille et al., 2005) following a waist-pull.

The stepping strategy classification system used in the current study did not yield any apparent differences in the frequency of each strategy between perturbation directions among amputees. The purpose in using the three-strategy classification was for ease of presentation, to simplify the statistical analyses, and to apply a similar classification system to previous research studies (Maki et al., 1996; Maki et al., 2000; Young et al., 2013; Mille et al., 2005). However, under different circumstances (e.g., with a larger population sample) a more complex classification system involving a wider variety of measures (e.g., trunk angle) may have highlighted differences between the reactive stepping strategies Away and Toward.

In the current sample of amputees, unique stepping strategies were observed when the perturbation caused a shift of the COM away from the prosthesis. For 28.1% of Away trials (compared to 3.1% Toward), a combination of multiple strategies was observed. These strategies include: (i) a MSS followed by abduction of the unloaded leg, indicating that a MSS was insufficient on its own, and (ii) initial abduction of the unloaded leg, followed by a shuffle step or inversion of the ankle joint (ankle roll)

with the loaded leg. These combination strategies were observed only within the amputee group, and primarily when perturbed away from the prosthesis.

Interestingly, the two AK amputees accounted for 60% of the combination strategies observed. These strategies may highlight the unwillingness of amputees to step with the prosthesis. Referring to Table 8, it is evident that the amputee group took fewer steps, on average, than the non-amputee group, where amputees used a single step for 34% of all trials, compared to 6% of trials resulting in a single step in non-amputees.

The argument that amputees are hesitant to step with their prosthesis contradicts the findings of Curtze et al. (2010), where amputees were reported to prefer stepping with the prosthesis. This difference may be attributed to the direction and predictability of perturbation, where Curtze et al. (2010) induced a forward fall using a tether-and-release method, thereby allowing the participants to prepare themselves for a forward fall. Amputation of a lower limb results in a loss of crucial sensory nerve afferents, thereby increasing an individual's reliance on other forms of sensory feedback, such as vision, to compensate (Vrieling et al., 2008). A semi-predictable forward fall paradigm, as seen in Curtze et al. (2010), provides participants with the opportunity to visualize where they will step, as their gaze is directed in line with the fall direction. This is not the case in the current study, where the direction of perturbation was randomized to occur either away or toward the prosthesis, causing a loss of balance in the lateral direction. The nature of these perturbations made it difficult for the participants to visualize their steps, as their gaze was directed forward instead of toward the direction of the induced fall. The

observed hesitation to step with the prosthesis therefore may be due to the lack of visual feedback of the support surface upon which they could place the stepping foot. This would support the findings of Gray et al. (2017) where stroke patients, who experience similar unilateral deficits to amputees, avoided stepping laterally with the paretic leg.

### **5.3. Implications**

The findings in the current study may indicate that rehabilitation practices should focus on executing balance recovery steps with the prosthesis. Many trials resulted in an ULS strategy, likely due to the pre-determined avoidance of amputees to step with their prosthesis. If small medial steps, such as those observed during a MSS strategy, are practiced during amputee rehabilitation, this may provide amputees with more confidence to make steps with their prosthesis. This would potentially increase the number of MSS trials observed in the amputee population, which appears to be a safe and effective balance recovery strategy.

Also, this study may indicate that the assessment of reactive balance control in amputees should be treated independently from non-amputees. Factors that have been identified to increase fall-risk in non-amputees do not translate to the amputee population, at least in the present sample of amputees. Specifically, multiple step balance responses, which are associated with increased fall-risk in older non-amputees, do not occur as frequently in the amputee population, even though the present amputee sample experienced a greater occurrence of unstable balance recovery trials.

Previous studies have implemented a reactive balance training program for lower limb amputees and shown that reactive stepping practice can lead to improvements in unexpected balance recovery. Lower limb amputees are able to complete a recovery step quicker and with less trunk flexion following six balance recovery training sessions (Crenshaw, Kaufman & Grabiner, 2013). The training sessions were specific to the postural perturbations that were experienced during the experimental trials (forward fall), which may suggest that rehabilitation practices should implement real-life fall scenarios when training for balance recovery. Also, the finding that six training sessions can lead to reduced trunk flexion during balance recovery is particularly important considering the number of trials in the current study that resulted in excessive trunk movement.

#### **5.4. Limitations**

The current study had limitations, primarily in the form of insufficient sample size. The current sample of amputees (n=8) is similar in size to many studies examining biomechanical factors in amputees, however the difference with the current study is the heterogeneity within the amputee group. The current sample of amputees varied widely in age (range 38-81 years), type of amputation (3 traumatic; 2 diabetic, 2 congenital, 1 cancer), and time since amputation (3-61 years). Also, the physical abilities of each amputee varied widely, however the measures used in the current study (i.e., ABC-scale, FRQ, TUG) may not have been sensitive enough to reveal these differences. Initially, this study was going to implement the L-test, a more challenging version of the TUG that involves walking a greater distance and turns in both directions. Given more time and space for data collection, the L-test

may have more clearly revealed differences in physical abilities. Furthermore, it became clear that physical activity levels varied greatly within the amputee group, through conversation with the participants. Additional questionnaires that can deduct how active an individual is, such as the CHAMPS Physical Activity Questionnaire or the Yale Physical Activity Survey, may have contributed some worthy insight.

In retrospect, the current study could have used a more sensitive protocol to determine single step threshold. With the current protocol, participants were subjected to surface translation accelerations that increased by  $0.25 \text{ m/s}^2$ , which was similar to the protocol used by a previous study (de Kam, Kamphuis, Weerdesteyn, & Geurts, 2017). If a smaller incremental increase in acceleration was used, perhaps a larger discrepancy in single step threshold could have been revealed.

Another limitation was omitting the analysis of middle-upper body balance responses. While the current study was focused on identifying the lower limb stepping strategies in response to lateral perturbations, it would have been interesting to highlight the upper body and trunk characteristics associated with certain stepping strategies due to amputees' reliance on trunk movement and hip strategies to control the COM (Miller et al., 2018).

## **5.5. Conclusion**

In conclusion, this study found that reactive stepping strategies in non-amputees and amputees are minimally affected by perturbation direction. This may suggest that the reactive stepping strategies observed in individuals without prior exposure to the current perturbation mechanism may induce a pre-determined



reaction to a lateral loss of balance, regardless of direction. However, differences were observed between the non-amputee and amputee groups with respect to the ULS strategy, where amputees relied heavily on a hip strategy to maintain their balance in as few steps as possible, in contrast to the non-amputee group where multiple steps were used in the majority of trials. Future research should aim to determine how amputees use the hip and trunk during lateral balance recovery, which could highlight further differences among amputees and non-amputees.

## References

- Batcir, S., Sharon, H., Shani, G., Levitsky, N., Gimmon, Y., Kurz, I., ... Melzer, I. (2018). The inter-observer reliability and agreement of lateral balance recovery responses in older and younger adults. *Journal of Electromyography and Kinesiology*, *40*, 39-47.
- Bolger, D., Ting, L. H., & Sawers, A. (2014). Individuals with transtibial limb loss use interlimb force asymmetries to maintain multi-directional reactive balance control. *Clinical Biomechanics*, *29*(9), 1039-1047.
- Braune, S., & Schady, W. (1993). Changes in sensation after nerve injury or amputation: the role of central factors. *Journal of Neurology, Neurosurgery & Psychiatry*, *56*(4), 393-399.
- Buckley, J. G., O'Driscoll, D., & Bennett, S. J. (2002). Postural Sway and Active Balance Performance in Highly Active Lower-Limb Amputees: *American Journal of Physical Medicine & Rehabilitation*, *81*(1), 13-20.
- Burke, M. J., Roman, V., & Wright, V. (1978). Bone and joint changes in lower limb amputees. *Annals of the Rheumatic Diseases*, *37*(3), 252-254.
- Canadian Diabetes Association. Diabetes Charter for Canada. October (2016). Retrieved from [diabetes.ca/charter](http://diabetes.ca/charter).
- Carpenter, M. G., Thorstensson, A., & Cresswell, A. G. (2005). Deceleration affects anticipatory and reactive components of triggered postural responses. *Experimental Brain Research*, *167*(3), 433-445.
- Cole, E. S. (2003). Training Elders With Transfemoral Amputations: *Topics in Geriatric Rehabilitation*, *19*(3), 183-190.
- Crenshaw, JR., Kaufman, KR., Grabiner, MD. (2013). Compensatory-step training of healthy, mobile people with unilateral, transfemoral, or knee disarticulation amputations: A potential intervention for trip-related falls. *Gait & Posture*, *38*, 500-506.
- Curtze, C., Hof, A. L., Otten, B., & Postema, K. (2010). Balance recovery after an evoked forward fall in unilateral transtibial amputees. *Gait & Posture*, *32*(3), 336-341.
- Curtze, C., Hof, A. L., Postema, K., & Otten, B. (2012). The relative contributions of the prosthetic and sound limb to balance control in unilateral transtibial amputees. *Gait & Posture*, *36*(2), 276-281.

- Dawes, D., Iqbal, S., Steinmetz, O. K., & Mayo, N. (2010). The Evolution of Amputation in the Province of Quebec. *Canadian Journal of Diabetes*, 34(1), 58-66.
- Day, B. L., Steiger, M. J., Thompson, P. D., & Marsden, C. D. (1993). Effect of vision and stance width on human body motion when standing: implications for afferent control of lateral sway. *The Journal of Physiology*, 469(1), 479-499.
- de Kam, D., Kamphuis, J. F., Weerdesteyn, V., & Geurts, A. C. H. (2017). The effect of weight-bearing asymmetry on dynamic postural stability in people with chronic stroke. *Gait & Posture*, 53, 5-10.
- de Kam, D., Roelofs, J. M. B., Bruijnes, A. K. B. D., Geurts, A. C. H., & Weerdesteyn, V. (2017). The Next Step in Understanding Impaired Reactive Balance Control in People With Stroke: The Role of Defective Early Automatic Postural Responses. *Neurorehabilitation and Neural Repair*, 31(8), 708-716.
- Diener, H. C., Horak, F. B., & Nashner, L. M. (1988). Influence of stimulus parameters on human postural responses. *Journal of Neurophysiology*, 59(6), 1888-1905.
- Dite, W., Connor, H. J., & Curtis, H. C. (2007). Clinical Identification of Multiple Fall Risk Early After Unilateral Transtibial Amputation. *Archives of Physical Medicine and Rehabilitation*, 88(1), 109-114.
- Duclos, C., Roll, R., Kavounoudias, A., Mongeau, J.-P., Roll, J.-P., & Forget, R. (2009). Postural changes after sustained neck muscle contraction in persons with a lower leg amputation. *Journal of Electromyography and Kinesiology*, 19(4), e214-e222.
- Ferguson, O. W., Polskaia, N., & Tokuno, C. D. (2017). The effects of foot cooling on postural muscle responses to an unexpected loss of balance. *Human Movement Science*, 54, 240-247.
- Geurts, A. C. H., & Mulder, W. (n.d.). *Reorganisation of postural control following lower limb amputation: Theoretical considerations and implications for rehabilitation*. 13.
- Gray, V. L., Yang, C., McCombe Waller, S., & Rogers, M. W. (2017). Lateral Perturbation-Induced Stepping: Strategies and Predictors in Persons Poststroke. *Journal of Neurologic Physical Therapy*, 41(4), 222-228.
- Hagberg, E., Berlin, K., & Renstrom, P. (n.d.). *e Function after through-knee compared with below-knee and above-knee amputation*. 6.
- Helm, P., Engel, T., Holm, A., Kristiansen, V. B., & Rosendahl, S. (1986). Function

- after lower limb amputation. *Acta Orthopaedica Scandinavica*, 57(2), 154-157.
- Henry, S. M., Fung, J., & Horak, F. B. (2001). Effect of Stance Width on Multidirectional Postural Responses. *Journal of Neurophysiology*, 85(2), 559-570.
- Hermodsson, Y., Ekdahl, C., Person, B. M., & Roxendal, G. (n.d.). *Standing balance in trans-tibia1amputees following vascular disease or trauma: a comparative study with healthy subjects*. 9.
- Horak, F. B. (2006). Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls? *Age and Ageing*, 35(suppl\_2), ii7-ii11.
- Horak, F. B., Henry, S. M., & Shumway-Cook, A. (1997). Postural Perturbations: New Insights for Treatment of Balance Disorders. *Physical Therapy*, 77(5), 517-533.
- Imam, B., Miller, W. C., Finlayson, H. C., Eng, J. J., & Jarus, T. (2017). Incidence of lower limb amputation in Canada. *Canadian Journal of Public Health*, 108(4), 374-380.
- Inglis, J. T., Horak, F., Shupert, C., & Jones-Rycewicz, C. (1994). The importance of somatosensory information in triggering and scaling automatic postural responses in humans. *Experimental Brain Research*, 101(1).
- Isakov, E., Burger, H., Gregoric, M., Marincek, C. (1996). Isokinetic and isometric strength of the thigh muscles in below-knee amputees. *Clinical Biomechanics*, 11 (4), 233-235.
- Jayakaran, P., Johnson, G. M., & Sullivan, S. J. (2015). Postural Control in Response to Altered Sensory Conditions in Persons With Dysvascular and Traumatic Transtibial Amputation. *Archives of Physical Medicine and Rehabilitation*, 96(2), 331-339.
- Kavounoudias, A., Tremblay, C., Gravel, D., Iancu, A., & Forget, R. (2005). Bilateral changes in somatosensory sensibility after unilateral below-knee amputation. *Archives of Physical Medicine and Rehabilitation*, 86(4), 633-640.
- King, L. A., & Horak, F. B. (2008). Lateral Stepping for Postural Correction in Parkinson's Disease. *Archives of Physical Medicine and Rehabilitation*, 89(3), 492-499.
- Kosasih, J. B. (n.d.). *Sensory Changes in Adults with Unilateral Transtibial*. 7.
- Kulkarni, J., Wright, S., Toole, C., Morris, J., & Hiron, R. (1996). Falls in Patients with Lower Limb Amputations: Prevalence and Contributing Factors. *Physiotherapy*, 82(2), 130-136.

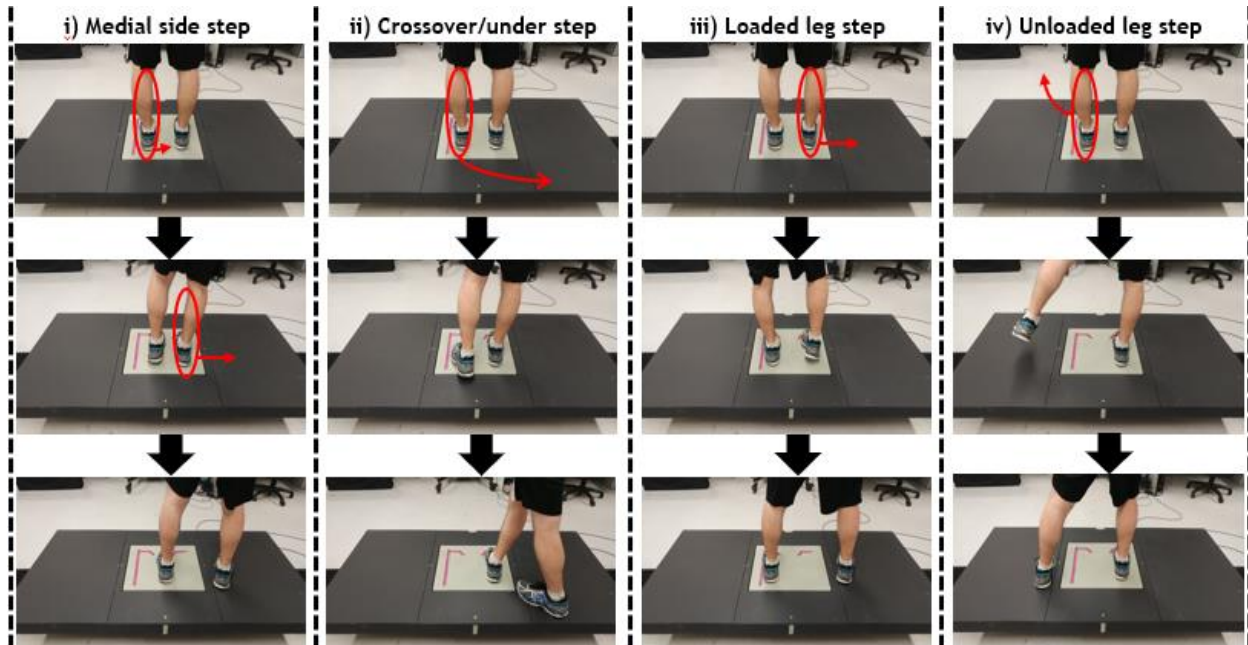
- Lin, S.-I., & Woollacott, M. H. (2002). Postural Muscle Responses Following Changing Balance Threats in Young, Stable Older, and Unstable Older Adults. *Journal of Motor Behavior*, 34(1), 37-44.
- Lloyd, C. H., Stanhope, S. J., Davis, I. S., & Royer, T. D. (2010). Strength asymmetry and osteoarthritis risk factors in unilateral trans-tibial, amputee gait. *Gait & Posture*, 32(3), 296-300.
- Maki, B. E., Edmondstone, M. A., & McIlroy, W. E. (2000). Age-Related Differences in Laterally Directed Compensatory Stepping Behavior. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, 55(5), M270-M277.
- Maki, B. E., McIlroy, W. E., & Perry, S. D. (1996). Influence of lateral destabilization on compensatory stepping responses. *Journal of Biomechanics*, 29(3), 343-353.
- Mansfield, A., & Maki, B. E. (2009). Are age-related impairments in change-in-support balance reactions dependent on the method of balance perturbation? *Journal of Biomechanics*, 42(8), 1023-1031.
- Massion, J. (1992). Movement, posture and equilibrium: Interaction and coordination. *Progress in Neurobiology*, 38(1), 35-56.
- Mille, M.-L., Johnson, M. E., Martinez, K. M., & Rogers, M. W. (2005). Age-dependent differences in lateral balance recovery through protective stepping. *Clinical Biomechanics*, 20(6), 607-616.
- Miller, S. E., Segal, A. D., Klute, G. K., & Neptune, R. R. (2018). Hip recovery strategy used by below-knee amputees following mediolateral foot perturbations. *Journal of Biomechanics*, 76, 61-67.
- Miller, W. C., Deathe, A. B., & Speechley, M. (2003). Psychometric properties of the activities-specific balance confidence scale among individuals with a lower-limb amputation. *Archives of Physical Medicine and Rehabilitation*, 84(5), 656-661.
- Miller, W. C., Speechley, M., & Deathe, B. (2001). The prevalence and risk factors of falling and fear of falling among lower extremity amputees. *Archives of Physical Medicine and Rehabilitation*, 82(8), 1031-1037.
- Nadollek, H., Brauer, S., & Isles, R. (2002). Outcomes after trans-tibial amputation: the relationship between quiet stance ability, strength of hip abductor muscles and gait. *Physiotherapy Research International*, 7(4), 203-214.
- Nashner, L. M. (1977). Fixed patterns of rapid postural responses among leg muscles

- during stance. *Experimental Brain Research*, 30(1).
- Nederhand, M. J., Van Asseldonk, E. H. F., der Kooij, H. van, & Rietman, H. S. (2012). Dynamic Balance Control (DBC) in lower leg amputee subjects; contribution of the regulatory activity of the prosthesis side. *Clinical Biomechanics*, 27(1), 40-45.
- Pasquina, P. F., Miller, M., Carvalho, A. J., Corcoran, M., Vandersea, J., Johnson, E., & Chen, Y.-T. (2014). Special Considerations for Multiple Limb Amputation. *Current Physical Medicine and Rehabilitation Reports*, 2(4), 273-289.
- Quai, T. M., Brauer, S. G., Nitz, J. C. (2005). Somatosensation, circulation and stance balance in elderly dysvascular transtibial amputees. *Clinical Rehabilitation*, 19, 668-676.
- Rogers, M. W., & Mille, M.-L. (2003). Lateral Stability and Falls in Older People: *Exercise and Sport Sciences Reviews*, 31(4), 182-187.
- Royer, T. D., & Wasilewski, C. A. (2006). Hip and knee frontal plane moments in persons with unilateral, trans-tibial amputation. *Gait & Posture*, 23(3), 303-306.
- Rubenstein, L. Z., Vivrette, R., Harker, J. O., Stevens, J. A., & Kramer, B. J. (2011). Validating an evidence-based, self-rated fall risk questionnaire (FRQ) for older adults. *Journal of Safety Research*, 42(6), 493-499.
- Schoppen, T., Boonstra, A., Groothoff, J. W., de Vries, J., Göeken, L. N. H., & Eisma, W. H. (1999). The timed “up and go” test: Reliability and validity in persons with unilateral lower limb amputation. *Archives of Physical Medicine and Rehabilitation*, 80(7), 825-828.
- Segal, A., & Klute, G. (2014). Lower-limb amputee recovery response to an imposed error in mediolateral foot placement. *Journal of Biomechanics*, 47(12), 2911-2918.
- Shell, C. E., Segal, A. D., Klute, G. K., & Neptune, R. R. (2017). The effects of prosthetic foot stiffness on transtibial amputee walking mechanics and balance control during turning. *Clinical Biomechanics*, 49, 56-63.
- Shumway-Cook, A., Brauer, S., & Woollacott, M. (2000). Predicting the probability for falls in community-dwelling older adults using the Timed Up & Go Test. *Physical Therapy*, 80(9), 896-903.
- Taub, E., Uswatte, G., & Mark, V. W. (2014). The functional significance of cortical reorganization and the parallel development of CI therapy. *Frontiers in Human Neuroscience*, 8.

- Tokuno, C. D., Cresswell, A. G., Thorstensson, A., & Carpenter, M. G. (2010). Age-related changes in postural responses revealed by support-surface translations with a long acceleration-deceleration interval. *Clinical Neurophysiology*, 121(1), 109-117.
- van Melick, N., Meddeler, B. M., Hoozeboom, T. J., Nijhuis-van der Sanden, M. W. G., & van Cingel, R. E. H. (2017). How to determine leg dominance: The agreement between self-reported and observed performance in healthy adults. *PLOS ONE*, 12(12), e0189876.
- Vrieling, A. H., van Keeken, H. G., Schoppen, T., Otten, E., Halbertsma, J. P. K., Hof, A. L., & Postema, K. (2008). Gait termination in lower limb amputees. *Gait & Posture*, 27(1), 82-90.
- Vrieling, A. H., van Keeken, H. G., Schoppen, T., Otten, E., Hof, A. L., Halbertsma, J. P. K., & Postema, K. (2008). Balance control on a moving platform in unilateral lower limb amputees. *Gait & Posture*, 28(2), 222-228.
- Winter, D. A., Prince, F., Frank, J. S., Powell, C., & Zabjek, K. F. (1996). Unified theory regarding A/P and M/L balance in quiet stance. *Journal of Neurophysiology*, 75(6), 2334-2343.
- Woollacott, M. H., & Shumway-Cook, A. (1990). Changes in posture control across the life span—a systems approach. *Physical Therapy*, 70(12), 799-807.
- Young, P. M., Whittall, J., Bair, W.-N., & Rogers, M. W. (2013). Leg preference associated with protective stepping responses in older adults. *Clinical Biomechanics*, 28(8), 927-932.
- Ziegler-Graham, K., MacKenzie, E. J., Ephraim, P. L., Trivison, T. G., & Brookmeyer, R. (2008). Estimating the Prevalence of Limb Loss in the United States: 2005 to 2050. *Archives of Physical Medicine and Rehabilitation*, 89(3), 422-429.

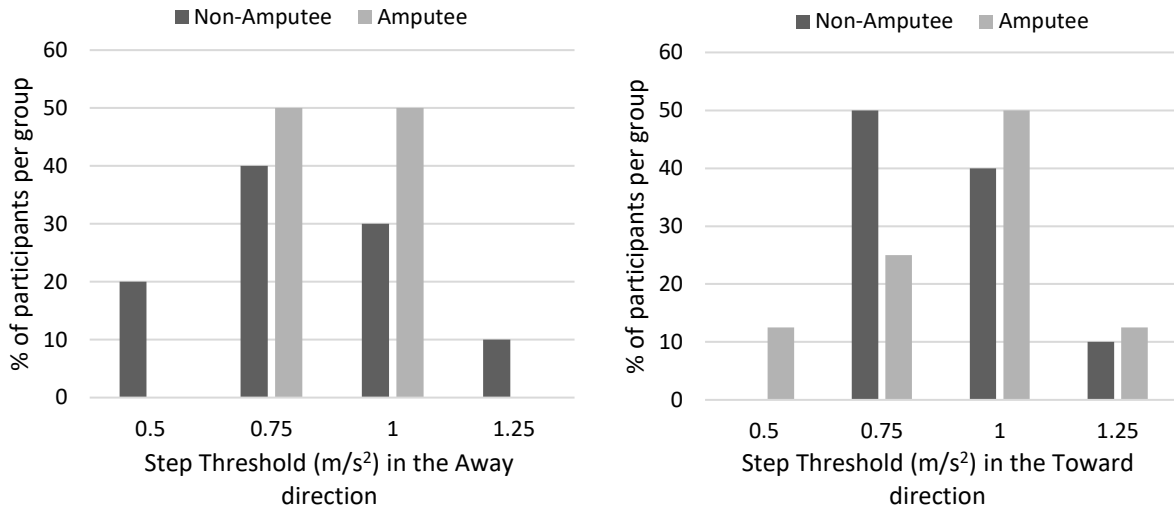
## Appendices

### Appendix I: List of Figures

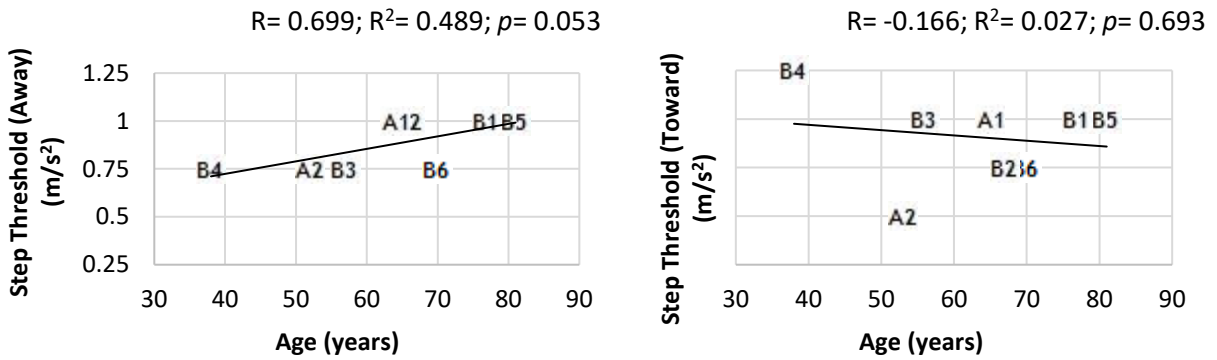


**Figure 1:** Lateral stepping strategies often observed following a lateral perturbation to standing balance. (i) Medial side step: unloaded left leg steps medially toward the loaded right leg, followed by a lateral side step with the loaded right leg. (ii) Crossover/under step: unloaded left leg crosses in front/behind the loaded right leg. (iii) Loaded leg step: loaded right leg steps laterally in the direction of the loss of balance. (iv) Unloaded leg step: characterized by an excessive abduction of the unloaded leg, or a lateral step in the opposite direction to the loss of balance.

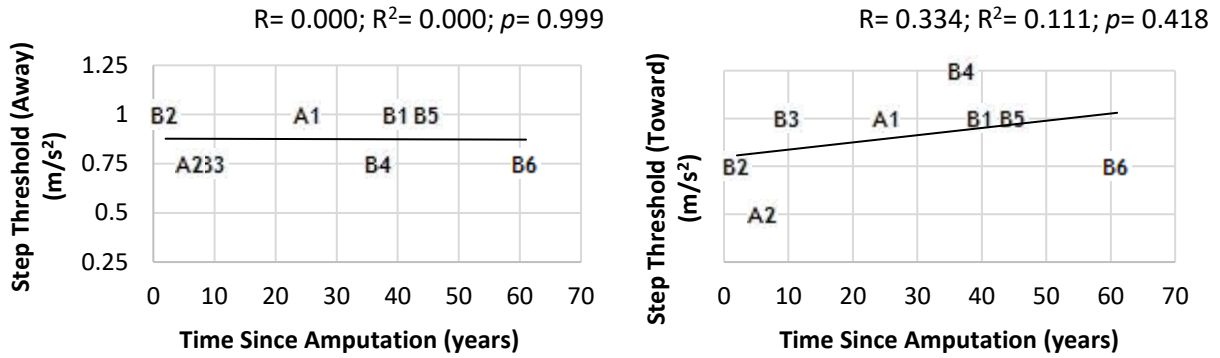




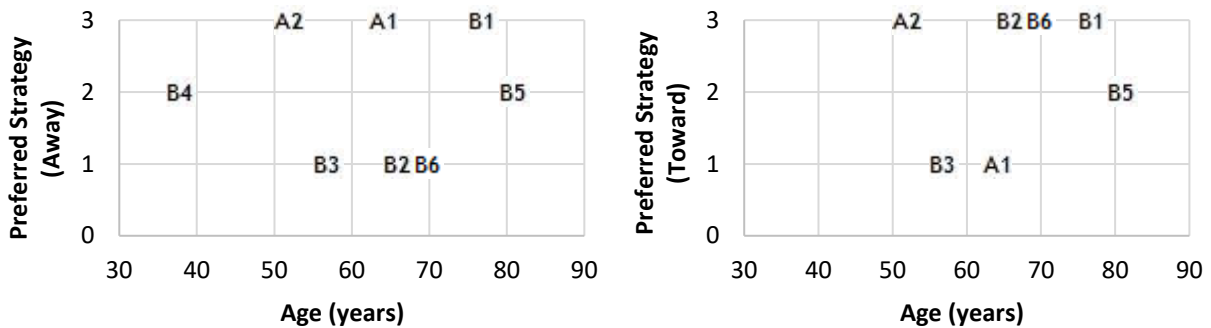
**Figure 2:** Single step threshold for the non-amputee (dark-grey) and amputee (light-grey) groups in the Away (left) and Toward (right) directions.



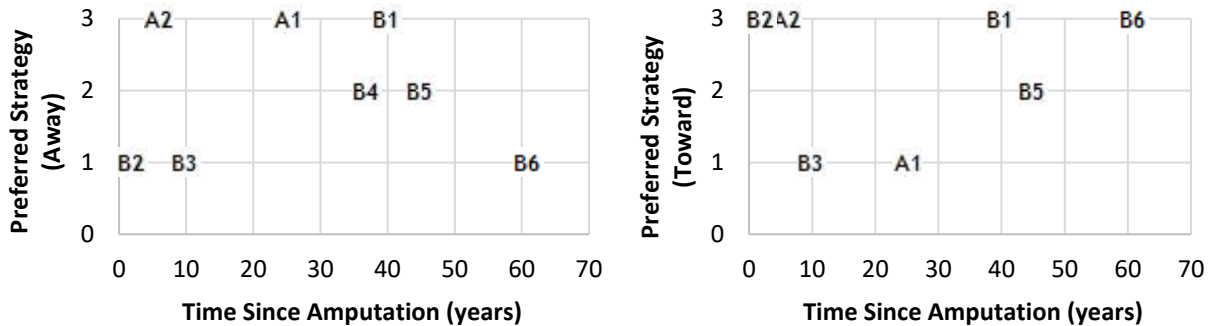
**Figure 3:** Single step threshold for Away (left) and Toward (right) trials as a function of age.



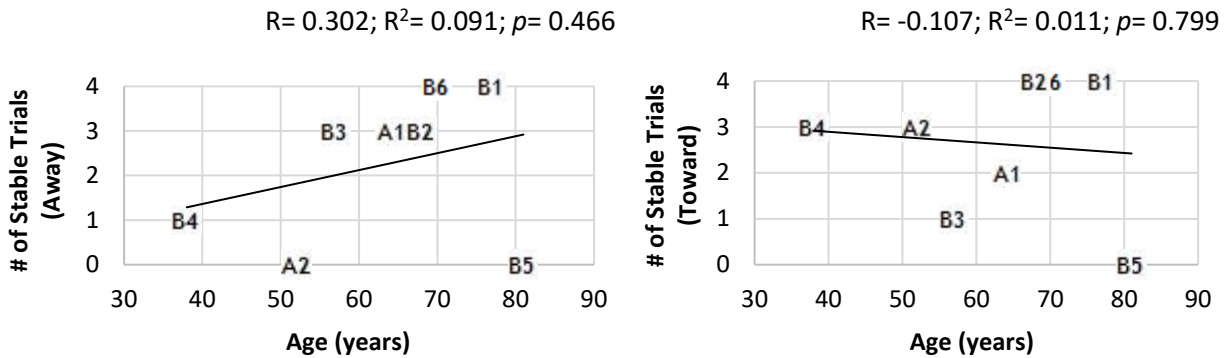
**Figure 4:** Single step threshold for Away (left) and Toward (right) trials as a function of time since amputation.



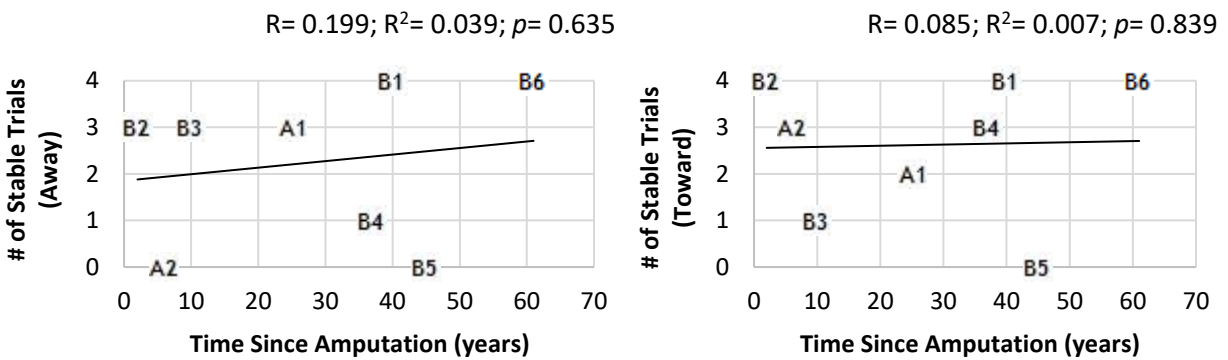
**Figure 5:** Preferred strategy for Away (left) and Toward (right) trials as a function of age. (3- MSS; 2- COS; 1- ULS). Note: participant B4 did not have a preferred strategy Toward.



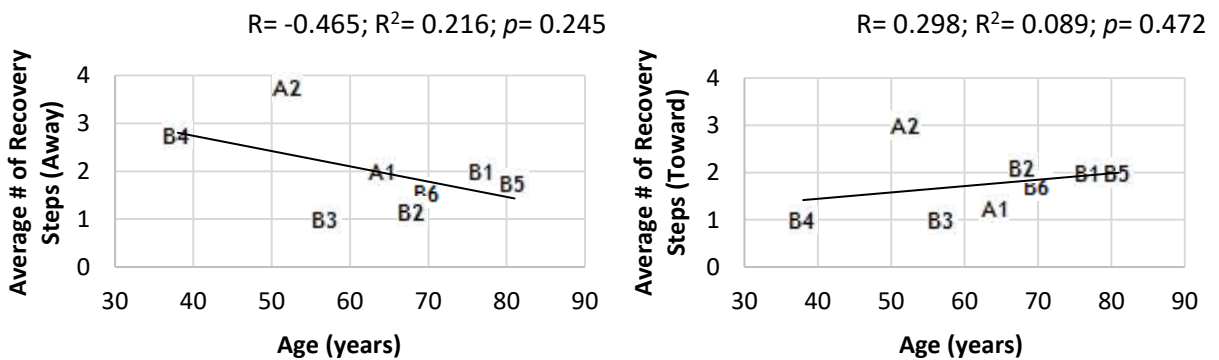
**Figure 6:** Preferred strategy for Away (left) and Toward (right) trials as a function of time since amputation. (3- MSS; 2- COS; 1- ULS). Note: participant B4 did not have a preferred strategy Toward.



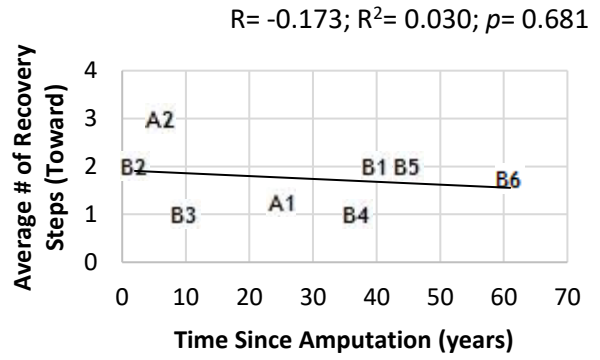
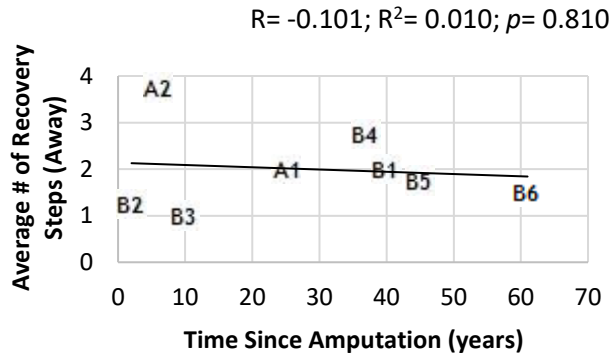
**Figure 7:** Number of stable trials for Away (left) and Toward (right) trials as a function of age.



**Figure 8:** Number of stable trials for Away (left) and Toward (right) trials as a function of time since amputation.



**Figure 9:** Average number of recovery steps per trial for Away (left) and Toward (right) trials as a function of age.



**Figure 10:** Average number of recovery steps per trial for Away (left) and Toward (right) trials as a function of time since amputation.

**Appendix II: List of Tables**

<i>ID</i>	<i>Sex</i>	<i>Height (cm)</i>	<i>Weight (kg)</i>	<i>Age (yrs)</i>	<i>Years Since Amputation (yrs)</i>	<i>Cause of Amputation</i>	<i>Level of Amputation</i>	<i>Falls (1 yr)</i>
<i>A1</i>	M	173	70	65	25	Traumatic	AK	1
<i>A2</i>	F	167	58	53	6	Traumatic	AK	1
<i>B1</i>	M	175	94	77	41	Traumatic	BK	0
<i>B2</i>	M	173	109	67	3	Diabetes	BK	1
<i>B3</i>	M	170	65	56	8	Diabetes	BK	0
<i>B4</i>	F	162	97	38	37	Congenital	BK	1
<i>B5</i>	F	157	63	81	43	Cancer	BK	1
<i>B6</i>	F	162	74	70	61	Congenital	BK	0

**Table 1:** Individual amputee participant characteristics.

	<i>Non-Amputee</i>	<i>Amputee</i>	<i>p-value</i>
<i>Sex</i>	4 M, 6 F	4 M, 4 F	
<i>Age (years)</i>	66.5 ± 8.6; range 55-79	63.4 ± 13.9; range 38-81	0.56
<i>Height (cm)</i>	168.6 ± 9.6	167.6 ± 6.2	0.80
<i>Weight (kg)</i>	71.2 ± 14.3	79.1 ± 18.5	0.32
<i>TUG (s)</i>	7.5 ± 1.7; range 5.7-11.6	8.9 ± 1.1; range 7.7-10.7	0.54
<i>TUG<sub>cog</sub> (s)</i>	9.9 ± 3.1; range 6.4-17.3	10.0 ± 1.4; range 7.9-11.7	0.96
<i>ABC (%)</i>	95.8 ± 4.8; range 86-100	91.8 ± 5.7; range 80-100	0.12
<i>FRQ (/ 14)</i>	1.6 ± 2.2; range 0-6	2.0 ± 1.3; range 0-4	0.65
<i>LCI (/56)</i>	n/a	53.6 ± 2.7; range 49-56	
<i>Fallers</i>	2/10 (20%)	5/8 (63%)	

**Table 2:** Participant characteristics and group means ± 1 SD [TUG= Timed Up and Go; TUG<sub>cog</sub>= Timed Up and Go (Cognitive); ABC= Activites Specific Balance Confidence Scale; FRQ= Fall Risk Questionnaire; LCI= Locomotor Capabilities Index].

	<i>Non-Amputee</i>	<i>Amputee</i>	<i>P-Value</i>
<i>1<sup>st</sup> Metatarsal</i>	1.90 ± 0.41 g	22.90 ± 13.13 g	0.09
<i>Heel</i>	3.24 ± 0.49 g	21.60 ± 11.77 g	0.09
<i>Intact Leg</i>	0.84 ± 0.13 g	1.25 ± 0.22 g	0.11
<i>Residual Leg</i>		1.17 ± 0.47 g	0.46

**Table 3:** Mean ± 1 SE monofilament tactile threshold of amputees and non-amputees. All units are expressed as grams of force required to bend the monofilament.

	<i>ML COP</i>			<i>AP COP</i>		
	Non-Amputee	Amputee	P-value	Non-Amputee	Amputee	P-value
<i>SD of COP Position (mm)</i>	2.57 ± 0.26	4.14 ± 0.84	0.07	6.02 ± 0.62	5.90 ± 0.63	0.90
<i>SD of COP Velocity (mm/s)</i>	3.69 ± 0.25	8.33 ± 2.01	0.02*	8.84 ± 0.78	14.68 ± 2.35	0.02*

**Table 4:** SD of the medio-lateral (ML) and antero-posterior (AP) COP position and velocity (mean ± 1 SE).

ID	Preferred Strategy		Step Threshold (m/s <sup>2</sup> )		Stable Trials (/4)		Avg. # of Recovery Steps per Trial		1 <sup>st</sup> Step Onset Latency (ms)		SD of COP Position During QS (mm)		SD of COP Velocity During QS (mm)		ML COP Position Prior to ST (mm)	Avg. Stance Width (cm)
	Dir.	A	T	A	T	A	T	A	T	A	T	ML	AP	ML		
A1	MSS	ULS	1.0	1.0	3	2	2	1.25	322	272	5.9	7.0	15.7	21.0	11.5	32.1
A2	MSS	MSS	0.75	0.5	0	3	3.75	3	320	547	8.7	7.9	18.4	27.7	-18.7	24.1
B1	MSS	MSS	1.0	1.0	4	4	2	2	277	285	1.4	6.7	3.5	10.8	-1.5	27.6
B2	ULS	MSS	1.0	0.75	3	4	1.25	2	360	305	3.5	7.7	7.9	14.7	11.6	35.4
B3	ULS	ULS	0.75	1.0	3	1	1	1	310	397	4.6	5.1	6.8	13.8	-12.9	28.6
B4	COS	COS/ ULS	0.75	1.25	1	3	2.75	1	337	290	4.2	5.5	6.4	11.2	-22.9	28.4
B5	COS	COS	1.0	1.0	0	0	1.75	2	342	292	2.1	4.4	4.3	11.0	9.9	25.9
B6	ULS	MSS	0.75	0.75	4	4	1.5	1.75	250	262	2.5	2.7	3.3	7.1	-31.4	26.0
C1	MSS	COS	0.5	0.75	4	4	2	3	337	400	2.9	4.5	2.9	6.5	27.1	25.1
C2	ULS	MSS	0.5	1.0	4	4	2	1	342	307	3.5	5.6	4.6	12.1	21.1	29.8
C3	MSS	MSS	1.0	1.25	4	4	2	2	256	263	1.9	5.2	3.4	7.4	-6.2	27.9
C4	COS	COS	1.0	1.0	4	4	1.75	2	292	285	1.1	3.5	3.0	5.4	40.9	25.0
C5	MSS	MSS	1.0	1.0	4	4	2	2	257	232	3.2	10.3	5.3	9.5	10.6	24.7
C6	MSS	MSS	0.75	0.75	4	4	2.5	3	290	307	1.8	7.9	3.1	8.9	1.7	23.6
C7	MSS	MSS	0.75	0.75	2 (/3)	0 (/3)	3.33	4	273	266	1.9	5.6	4.0	9.2	-6.3	25.5
C8	MSS	COS	0.75	0.75	4	4	2	2	295	290	3.2	4.9	3.2	6.7	-1.4	22.4
C9	MSS	COS	0.75	0.75	4	4	2	2	262	267	2.2	5.0	3.1	9.1	15.6	23.4
C10	MSS	MSS	1.25	1.0	4	4	2	2	207	207	3.5	7.2	4.0	13.2	-16.8	28.2

**Table 5:** Individual data for preferred stepping strategy, step threshold, number of stable balance recovery trials, average number of recovery steps per trial, first step onset latency for Away (A) and Toward (T) trials, SD of COP position and SD of COP velocity in ML and AP direction during quiet standing, average ML COP position during 5 s prior to surface translation (positive values indicate COP position toward the prosthesis/non-dominant leg, negative values indicate COP position away from the prosthesis/non-dominant leg) and average stance width. Participant ID's are represented with a subject number and an 'A' indicating AK amputee, 'B' indicating BK amputee, or C indicating non-amputee control.

	Non-Amputees		Amputees	
	# of Away trials (%)	# of Toward trials (%)	# of Away trials (%)	# of Toward trials (%)
MSS	31 (79.5)	23 (58.9)	11 (34.4)	14 (43.8)
COS	4 (10.2)	15 (38.5)	9 (28.1)	8 (25.0)
ULS	4 (10.2)	1 (2.6)	12 (37.5)	10 (31.2)
Total	39 trials	39 trials	32 trials	32 trials

**Table 6:** Reactive stepping strategy frequency for the non-amputee and amputee groups.

	<i>NON-AMPUTEE</i>			<i>AMPUTEE</i>					
	<b>Combined Away &amp; Toward Trials</b>			<b>Away Trials</b>			<b>Toward Trials</b>		
	<i>Stable</i>	<i>Unstable</i>	<i>Fail</i>	<i>Stable</i>	<i>Unstable</i>	<i>Fail</i>	<i>Stable</i>	<i>Unstable</i>	<i>Fail</i>
<b>MSS</b>	50	1	3	7	2	2	13	1	0
<b>COS</b>	19	0	0	2	3	4	2	3	3
<b>ULS</b>	5	0	0	8	2	2	6	3	1
<b>TOTAL</b>	74	1	3	17	7	8	21	7	4

**Table 7:** Quality of balance recovery per stepping strategy in non-amputees and amputees in the Away and Toward conditions.

	<i>Non-Amputee</i>		<i>Amputee</i>	
	<i>Away</i>	<i>Toward</i>	<i>Away</i>	<i>Toward</i>
<b><i>Single Step Trials</i></b>	1 (3%)	4 (10%)	10 (31%)	12 (38%)
<b><i>Multiple Step Trials</i></b>	38 (97%)	35 (90%)	22 (69%)	20 (62%)

**Table 8:** Number of trials involving a single step or multiple steps in non-amputees and amputees.