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# Assessment of Stabilization and Muscle Recruitment in Amputees following Sit-To-Stand Transfer

An Honors Thesis Submitted in Partial Fulfilment of Requirements for Honors Study in  
Kinesiology

By

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## Section 1: Introduction

Standing from a seated position is a task of daily living that is essential for independence relying heavily on balance and lower-extremity muscle strength and coordination (Whitney et al., 2005). With age, many once healthy individuals lose this ability and are institutionalized or find themselves reliant on others for care due to the development of balance disorders or loss of muscle mass. The majority of research present today on transitional movements affected by balance disorders, such as the sit-to-stand transition, has been focused on the elderly population. A correlation has been established between the Sit-To-Stand Test and factors such as postural control, fall risk, lower-extremity strength, proprioception, and degree of disability (Whitney et al., 2005). Among these, the risk of falling and the act of falling has also been shown to have negative effects summarized as “post-fall syndrome” which is characterized by loss of self-efficacy in routine daily tasks resulting in withdrawal, depression, or confusion, as well as, self-imposed restrictions on activity and independence (O’Laughlin, Robitaille, Boivin, & Suissa, 1993). More so, falls resulting in an injury are among the top causes of death in elderly of the United States and costs the nation directly and indirectly around \$85 billion per year (Pauley, Devlin, & Heslin, 2006).

In addition to the elderly, another population that is naturally considered to have a higher risk for falls and post-fall syndrome include persons with amputations. This increase in risk is due to altered balance, strength, and gait patterns and puts amputees at danger for injury as well as inactivity as a result of fear of falling (Miller, Speechley, & Deathe, 2001a). Miller, Speechley, and Deathe (2001a) found as much as 52.4% of their sample of amputees to have fallen within the last year, and 49.2% experience an active fear of falling. The sit-to-stand transition is often the limiting factor, even above walking, in mobility of persons with

amputations; the Sit-To-Stand Test has been considered an evaluative tool for functional mobility and risk of falling. Even with the high risk of falling, the very limited research on assessing stability and muscle recruitment derivations among amputee populations has controversial results. Therefore, this study will qualitatively assess functional mobility by its two main components of stability and coordinated muscle recruitment through a Sit-to-Stand Test with measurements of time to stabilization and degree of muscle group activation. In particular, this study will investigate the interactions between lower-extremity muscle coordination and time to stabilization in relation to sit-to-stand performance through cinematography analysis software, as well as muscle activation through the use of electromyography (EMG). Very little research has been completed on the comparison of amputees' sound and un-sound limb, therefore, this qualitative study will serve to strengthen the few experiments completed and may potentially contribute to training procedures to improve and preserve the sit-to-stand function in new and aging amputees in the clinical setting.

## Section 2: Review of Literature

The occurrence and consequences of falls have been the focus of many studies for the last 30 years. Some studies report the incidence of falls among community-dwelling elderly to be as high as 44.1 falls per 1,000 person-months, and of these reported falls 46.2% resulted in an injury (O’Laughlin, Robitaille, Boivin, & Suissa, 1993). The number of falls that go unreported are bound to make this statistic conservative. Although most falls do not result in serious injury, those that result in fractured or broken bones often prove to be fatal in elderly individuals. The majority of falls occur during transitional movements, including the transfer from a seated to a standing position or Sit-to-Stand (STS) transfer (National Patient Safety Agency, 2007). The STS test is a functional and physical performance assessment that has been used as an objective, predictive tool for the risk of falling by measuring balance and strength. Failure of a STS test has been linked to subject-related factors including: a longer than average time to execute the movement; muscle weakness; lack of coordination; and impaired balance (Riley, Krebs, & Popat, 1997; Lindemann et al., 2007; Janssen, Bussmann, & Stam, 2002).

Sit-to-Stand transfer is defined as the transition from a seated to standing position without losing balance as the body’s center of mass is moved upward and forward (Janssen, Bussmann, & Stam, 2002). During this movement, the center of mass (COM) moves away from the base of support (BOS) to a less stable position while the weight and momentum is shifted forward and upward. To complete a successful STS, the momentum, balance, and posture must be stabilized by returning the COM to rest over the BOS by the end of the task (Akram & Macllroy, 2011). Individuals that complete a successful STS transfer have high postural stability and short times to stabilization, as well as normal weight bearing symmetry between their legs. Time to stabilization (TTS) is the time it takes for a person to reach a stable postural control after

performing a task. This time begins with the attainment of an upright stance and ends when motion becomes stabilized.

Due to decreased strength and power that occurs naturally with age, elderly tend to have difficulty with the STS transitory movement (Dehail et al., 2007). Therefore, the elderly have been the general focus of stability and fall studies through STS assessment. In comparison to young and healthy individuals, the elderly show a longer completion time, a longer TTS, and increased postural sway (Gross, Stevenson, Charette, Pyka, & Marcus, 1998; Wolfson et al., 1992). Altered strategies to compensate for less power involved a greater dependence on trunk flexion for momentum when standing at faster rates, but no significant differences were found in the order of muscle activation or muscle groups used (Dehail et al., 2007). These factors decrease their overall stability and place elderly at an increased risk of falling (Hasan, Lichtensein, & Shiavi, 1990). Other potential risk factors for falling have been reported as age, gait and balance instability, decreased vision, alcohol consumption, acute and chronic health problems, dizziness, and medications (O'Laughlin et al., 1993; Miller, Speechley, & Deathe, 2001a).

Though these factors are often most prevalent in elderly populations, they also have strong correlation to the amputee who uses a prosthesis (Miller, Speechley, & Deathe, 2001a). Miller et al. discovered the prevalence of falls among these individuals to be as high as 52% for falls within the last year and 75% of these individuals having fallen more than once. Nearly half (49%) of the subjects maintained a fear of falling whether or not they had experienced a fall and 76% avoid activity due to their fear of falling (Miller, Speechley, Deathe, & Koval, 2001b). This increased risk for and fear of falling has serious implications on the health, independence, and quality of life that clinicians strive to improve. Amputees have been shown to have standing

difficulties and longer completion times much like the average elderly person, and it is generally assumed that amputees are at an increased risk for falling due to asymmetry of weight distribution, a decrease in proprioception, and a decrease in balance (Isakov, Mizrahi, Ring, Susak, & Hakim, 1992).

Asymmetry of weight bearing between sound and unsound limbs has been studied at length even among the amputee population. In the general healthy population, a certain degree of asymmetry (5-7%) has been accepted as normal and is contributed to favoring the dominant leg during the sit-to-stand transition (Janssen, Bussmann, & Stam, 2002). Among amputees this degree of asymmetry while transferring has been reported as high as 16% with excessive loading of the sound limb so that 42% of body weight is on the amputated limb and the remaining 58% is overloaded onto the sound limb (Buger, Kuzelicki, & Marincek, 2005; Ozyurek, Demirbuken, & Angin, 2013). This asymmetry, due to overloading the sound limb, adds further wear on remaining joints and muscles and could initiate or accelerate degeneration of the sound side. Though studies (Gao, Zhang, & Haung, 2011; Highsmith et al., 2011; Jassen et al, 2002; Buger et al., 2005; Ozyurek et al., 2013) have shown a degree of asymmetry is typical of amputees, the data from research to back the assumptions of decreased balance and increased risk of falls do not exist.

With loss of limb, there is a loss of proprioceptive sensors. These sensors, found in the muscle fibers and joints of the body, provide information on the body's position in space. These specialized cells include spindle fibers and stretch receptors, and allow for location of limbs even without visual input. In amputees, losing proprioceptive sensors by losing muscle and joints increases the reliance of visual input to locate body parts and can result in the phenomenon known as phantom limb (Isakov et al., 1992). Though this loss of information inflow initially



causes impaired balance, research finds mixed results on long term effects. Isakov et al. (1992) found that even as soon as four weeks from starting rehabilitation, amputees showed improvement in balance and adjustment to loss of proprioceptive sensors.

Most studies assessing balance in amputees have measured standing postural sway, but these limited studies have contradictory results. Isakov et al. (1992) found an increase in postural sway in amputees using the Romberg test and two force plates; the increase was attributed to decreased proprioception. These results contradicted other studies (Fernie, & Holling, 1978) that did not find significant differences in postural sway of an amputee versus a healthy individual when standing with eyes open. The difference was reasoned to be from the visual inflow that was needed to contribute the information input that was previously attributed by spindle fibers. All the aforementioned research studies failed to consider if these symptoms were due to comorbid diseases or an adjustment period after initial use of their prosthesis as opposed to the actual loss of limb. Diabetes alone has been correlated to decreased stability in the general population (Kalyani et al., 2013). Additionally, the risk factors for falls in amputee inpatient rehabilitation facilities were found to be strikingly similar to those of other rehabilitative facilities for other impairments (Pauley, Devlin, & Heslin, 2006). Comorbid diseases, age, cognitive impairment, and medications have to be taken into consideration as contribution to imbalance and falls in this population; these factors may even be the main reason for falls and functional limitation over amputation.

This study was the first to this researcher's knowledge to simultaneously assess TTS and muscle group activity following a sit-to-stand transfer. More so, this study was one of the first to assess both below-knee (BK) amputees that do not have diabetes or vascular disease and those that do have a disease in addition to a healthy individual. This provided a comprehensive subject

sample through the inclusion of both diseased and healthy BK amputees that also resulted in assessment of complicating factors that affect balance. TTS was used to assess balance, and electromyography (EMG) was used to qualitatively assess asymmetry between sound and un-sound limbs of amputees. This study was used to determine if individuals that have had time to adjust to their prosthesis (by excluding participants that have not used their prosthesis for more than six months) differ from a healthy non-amputee in risk for falls through a STS test. This, in turn, added to data that could discriminate risks associated with prosthetic use from complications of comorbid diseases, adjustment time, and sedentary life styles, and to test the causal relationship assumed about limb amputation and balance impairment.

### Section 3: Methods

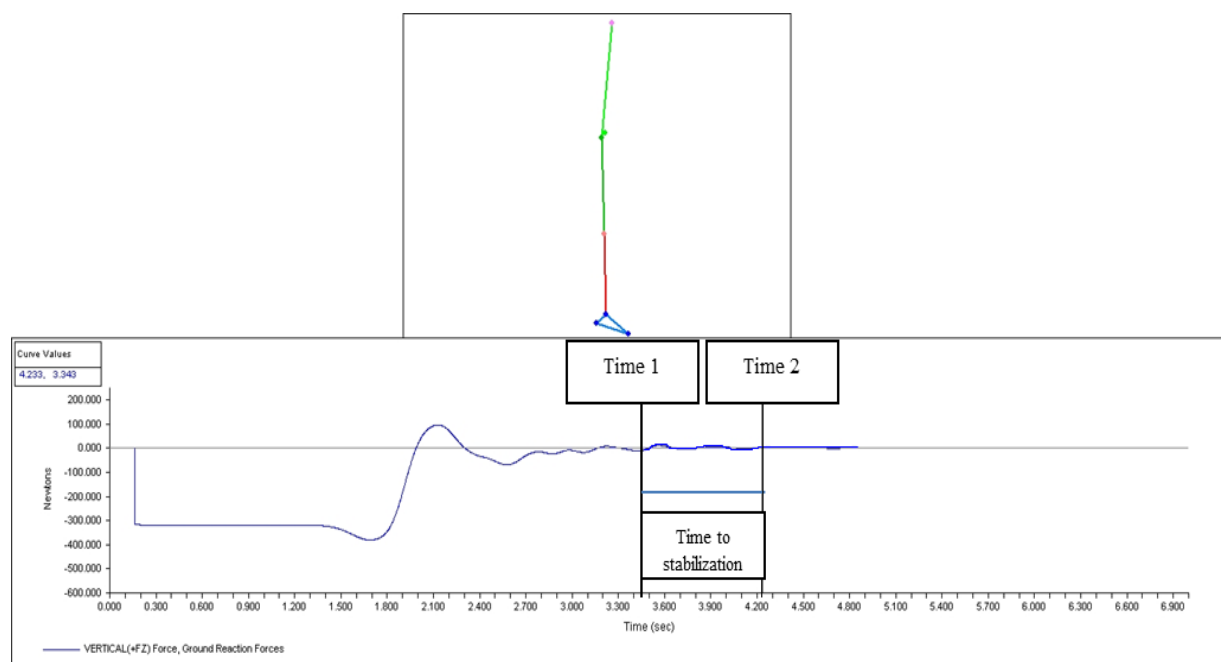
Adults over 18 years of age were recruited to participate in the current investigation. Four male trans-tibial amputees from reasons including trauma, infection, or diabetes and peripheral artery disease, were recruited, as well as one healthy male non-amputee. Upon arrival to the Human Performance Laboratory, each participant completed a health-history questionnaire, IPAQ physical activity questionnaire for the past six months, and demographic information (age, height, weight, body composition via DXA). All participants performed a chair-stand test five times with a 60-second rest period between trials.

Chair-stand muscle recruitment and time to stabilization were assessed during and following a sit-to-stand task. The subject was seated in a standard backed chair (seat height ~0.43-m) with both feet placed hip-width apart on a force plate and asked to stand as quickly as possible with hands placed across their chest. During each trial, the participant had reflective markers attached (via adhesive tape) to their acromion process (shoulder), greater trochanter (hip), lateral epicondyle of femur (knee), lateral malleolus (ankle), posterior aspect of calcaneus (heel), and the base of the fifth metatarsal (pinky-toe) of amputated side for amputees and the preferred side of non-amputees. The subject was filmed from their amputated side for amputees and right side on the non-amputees using a high speed camera (Bosch, Farmington Hills, MI) and 2D motion analysis capture system (60 Hz; Vicon Peak Motus, v. 9.0, Centennial, CO). In addition, participants were asked to wear an electromyographic (EMG) (Delsys, Natick, MA) sensor over the quadriceps (vastus lateralis & medialis), hamstrings (biceps femoris), hip (gluteus maximus), calf (medial & lateral gastrocnemius), and shin (tibialis anterior) to assess muscular activation of each muscle. All of the aforementioned muscles were recorded on the sound side of amputees and of both limbs of non-amputees. The sites used on the un-sound limb

of amputees were a subset of the list based on the muscles present and were determined on an individual basis. EMG signals were assessed for 5 seconds from initiation of the movement during each of the chair-stand trials and provided the forces of muscle groups.

The time to stabilization (TTS) was calculated from the force plate ground reaction forces with a sample rate of 600 Hz (Kistler, Novi, MI), and provided a measure of stability to compare amputees and non-amputees for significant differences. Time to stabilization was defined as the time it takes for a person to reach a controlled stance after a task, in this case a sit-to-stand transfer. The force plate provided a measurement of ground reaction forces exerted during the standing transition and immediately after and was recorded for 5 seconds from the initiation of the movement. Along with the force plate, a high-speed video camera (Bosch, Farmington Hills, MI) and 2D motion analysis system (60 Hz; Vicon Peak Motus, v. 9.0, Centennial, CO) was synchronized to provide a measurement of the phases of standing and stabilization in the sagittal plane. A calibration frame was captured and digitalized before each subject's trial. To create the 2D model that was used by the motion analysis system, adhesive reflective markers were placed on the amputated side of amputees and the right side of non-amputees on the following anatomical sites: acromion process, greater trochanter, lateral joint line of the knee, lateral malleolus, posterior aspect of the heel outside of the shoe, and the base of the fifth metatarsal on the outside of the shoe. Two time measurements were recorded during standing and stabilization; Time 1 was the moment of maximal knee extension, and Time 2 was captured as the end of oscillation with the GRF line stabilized and returned to body weight. Time to stabilization was recorded as the difference between Time 2 and Time 1 (Figure 1).

Figure 1



All assessments were completed in one visit to the Human Performance Lab and took no longer than 2 hours.

### Statistical Analysis

Subjects were placed into one of two groups (independent variable): below-the-knee amputee, or able-bodied control. Dependent variables included TTS, and EMG amplitude for: vastus lateralis, vastus medialis, biceps femoris, gluteus maximus, medial gastrocnemius, lateral gastrocnemius, and tibialis anterior. Descriptive characteristics were compared between groups for TTS and EMG activation as well as within subjects (sound versus amputated limb) for EMG activation, and maximal trunk flexion. Values were reported as means  $\pm$  *sd*.

## Section 4: Results

A total of five individuals participated in this study, four male, transtibial amputees and one male, non-amputee. Descriptive data for participants are summarized in Table 1. Individuals were recruited from local prosthetic clinics, and through flyers and school newspaper clips. Individuals that chose to participate, reported to the testing facility and completed an informed consent, health history questionnaire, physical activity questionnaire, basic anthropometric measurements, and a body composition scan through a DXA prior to the study. It should also be noted that all participants but participant A 4 were able to stand from a seated position without the use of their hands to push off the seat.

Table 1.

<i>Participant Anthropometric and Health History Data</i>										
Measure	Height (in)	Age (yrs)	Weight (kg)	BMI	% Fat Mass	Years since Amp	Amputated Side	Etiology	Presence of Disease	PA Level
A 1	68.0	29	84.5	28.3	25.6	5	L	Trauma	None	High
A 2	69.5	58	65.0	20.9	17.8	31	L	Trauma	None	High
A 3	70.0	68	94.5	29.9	36.1	9	L	Trauma	Diabetes	High
A 4	67.7	66	97.5	33.0	28.3	17	L	Trauma	Heart Disease, Diabetes	Moderate
N 5	66.0	28	86.5	30.7	38.8	—	—	—	—	High

Table 1 - Note: A: amputee participant, N: non-amputee participant

The aim of this study was to gather descriptive data on muscle group activity and time to stabilization in amputees while transferring from sitting to standing. Although the sample size was too small for statistically significant differences, the data were analyzed for patterns and

comparisons among different groups. All of these comparisons were analyzed with a t-test and are summarized in the tables that follow.

### **Hypothesis 1**

One of the main hypotheses of this study was that during sit-to-stand transitions in amputees, muscle groups from the sound limb would create greater EMG amplitudes in comparison to the corresponding muscles of the amputated limb. Table 2 compares EMG magnitude of sound verses amputated limb in the following pairs: vastus medialis, vastus lateralis, biceps femoris, and semitendinosus. The data are presented as pairs of muscles with pair 1 being vastus medialis, pair 2 being the vastus laterals, pair 3 being the biceps femoris, and pair 4 being the semitendinosus muscles. Although the sample is too small for significant differences, for all pairs the muscles of the amputated side contracted harder than the sound side.

Table 2

<i>EMG Muscle Group Amplitudes by Pairs of Muscles</i>				
Muscle Pair	Mean (mV)	Std. Deviation	% Difference	n
Pair 1				
EMG_VM_A	1.798	2.318	62.6	3
EMG_VM_S	0.673	0.593		3
Pair 2				
EMG_VL_A	2.775	2.251	4.86	3
EMG_VL_S	2.640	1.993		3
Pair 3				
EMG_BF_A	1.709	2.745	37.7	3
EMG_BF_S	1.064	1.547		3
Pair 4				
EMG_Semi_A	2.518	3.302	93.6	2
EMG_Semi_S	0.161	0.067		2

Note. Mean represents the average amplitude of the muscle contraction in millivolts. The sample number (n) varied by the muscle groups present as well as equipment malfunction resulting in failure to retrieve data for some individuals. EMG\_VM\_A is the EMG of the vastus medialis on the amputated side and EMG\_VM\_S is the EMG of the vastus medialis of the sound side. Likewise, EMG\_VL is the vastus lateralis of the amputated and sound side respectively. EMG\_BF\_A and EMG\_BF\_S are the EMG of the biceps femoris of amputated and sound side, and EMG\_Semi\_A and EMG\_Semi\_S is the EMG of the semitendinosus of the sound and amputated side respectively.

## Hypothesis 2

The second hypothesis was the comparison of amputees' to non-amputees' time to stabilization, expecting amputees to take longer to stabilize. These data are reported in Table 3. Amputees took 0.072 seconds on average longer to stabilize than the non-amputee subject, a difference of 13.6%.

## Hypothesis 3

The third hypothesis of this study was to compare the time it took individuals to reach a stable stance based on different factors. Three comparisons were performed within the amputee group alone and were comparing activity level, presence of disease, and time since amputation



separately in relation to TTS. It was hypothesized that higher physical activity levels, absence of disease, and greater years since amputation would all be associated with shorter times to stabilization. These data as well as the percentage difference among the groups were also calculated and reported in Table 3.

Table 3

<i>Time to Stabilization Comparisons</i>				
Comparison Group	TTS Mean (s)	Std. Deviation	% Difference	n
Pair 1				
Amputee	0.531	0.068	13.6	4
Non-amputee	0.459	—		1
Pair 2				
Amp High PA	0.497	0.008	21.5	3
Amp Moderate PA	0.633	—		1
Pair 3				
Amp Diseased	0.565	0.097	12.0	2
Amp No Disease	0.497	0.011		2
Pair 4				
Amp > 10yrs	0.561	0.102	10.7	2
Amp < 10yrs	0.501	0.006		2

Note. TTS Mean is the average time to stabilization for the group in seconds, Amp High PA is the group classified as high physical activity by the IPAQ and the Amp Moderate PA is the group classified as moderate physical activity. Amp Diseased is comprised of individuals diagnosed with a disease, while Amp No Disease are individuals with no known disease. Amp > 10yrs are individuals having lost their limb more than ten years ago, and Amp < 10yrs are individuals that have lost a limb within the last ten years. The number of individuals in each category is the variable n.

#### **Hypothesis 4**

Lastly, as is commonly seen in the elderly, it was hypothesized that amputees would compensate for lower muscular strength by using upper body momentum to propel themselves off the chair. This would be done through greater trunk flexion to move the center of gravity forward. Therefore, maximal degree of trunk flexion was also gathered for each individual from the motion analysis software, and t-test comparisons were performed based on physical activity

level, disease state, and years since amputation. These data are summarized in Table 4. As shown in the table, maximal flexion was 27.8% less in non-amputees than in amputees, 3.1% greater in moderately active amputees than highly active amputees, 15.5% greater in subjects with a disease compared to subjects without any diseases, and the greatest difference among the amputee comparison groups of 18.8% greater flexion in subjects that had been an amputee for less than 10 years when compared to amputees of greater than 10 years.

Table 4

<i>Maximal Degree of Trunk Flexion Comparisons</i>				
Comparison Group	Max Flexion (degrees)	Std. Deviation	% Difference	n
<b>Pair 1</b>				
Amputee	52.876	8.587	27.8	4
Non-amputee	73.202	—		1
<b>Pair 2</b>				
Amp High PA	52.460	10.467	3.1	3
Amp Moderate PA	54.123	—		1
<b>Pair 3</b>				
Amp Diseased	57.326	4.529	15.5	2
Amp No Disease	48.426	11.022		2
<b>Pair 4</b>				
Amp > 10yrs	47.378	9.540	18.8	2
Amp < 10yrs	58.374	3.047		2

Note. Then number of individuals in each categories is the variable n.

## Section 5: Discussion

The objective of this study was to gather descriptive data on muscle activity, time to stabilization, and maximal trunk flexion components of the sit-to-stand transfer in amputees. This was done through motion capture and EMG analysis. Results of this study provide data that contradict hypothesis one and a portion of hypothesis two, but also support sections of hypothesis two and hypothesis three. The most important factor in time to stabilization appeared to be physical activity levels, and time since amputation in maximal degree of trunk flexion. The muscle groups of the amputated limb appeared to contract at higher voltage than the corresponding muscles of the sound leg.

When developing the methodology of this study, the intention had been to include both AK and BK amputees and exclude those with Peripheral Artery Disease or Diabetes. Due to the difficulty in recruiting with these restrictions and the small population, the study was opened to include those with Diabetes and/or PAD. By chance, all participants were BK amputees. Also, the thickness and bulk of the EMG sensors was not factored into the methodology. For all participants, the sensor was too thick to fit under the socket to analyze the gastrocnemius and tibialis anterior. Additionally because measurements were being compared contralaterally between sound and unsound limbs, measurements of the gastrocnemius and tibialis anterior of the sound limb were not assessed. Lastly, the liner sleeves that develop suction in the socket came higher on the thigh than anticipated, and in some cases resulted placing the sensor higher on the muscle group instead of directly over the belly of the muscle. Although this variable was controlled by adapting the placement of the sensors, this could have resulted in less accurate and more noise in the EMG readings. The muscles that were compared were: biceps femoris, semitendinosus, vastus lateralis, and vastus medialis.

## **Hypothesis 1**

The hypothesis that muscles in the sound limb would produce greater signal amplitude than compared to the muscles in the amputated limb of the same individual was not supported by the results of this study. In fact, for all four pairs of muscles the average of maximal current generated during the movement was greater on the amputated side. This was counter-intuitive, and since there are no other studies that used EMG to measure muscle activity in amputees during sit-to-stand transitions, more extensive research on a larger sample would be needed to draw any conclusions.

Several studies have been conducted and concluded that amputees do not load weight symmetrically during the seat-off phase of standing. Agrawal et al. (2011) found that transtibial amputees shifted weight to load the sound limb with 64% and the unsound limb 36% of the total body weight. It was also found that amputees without chronic illness added an additional 27% of the weight to the sound limb (Agrawal et al., 2011). This study went on to suggest that during the initiation of the movement, the excessive demand on the compromised musculoskeletal system of the amputated limb could trigger the weight shift (Agrawal et al., 2011). Similar studies have established that the sound and un-sound limb of amputees do not perform equally during sit-to-stand transfers (Burger, Kuzelicki, & Marincek, 2005; Gao, Zhang, & Haung, 2011; Highsmith et al., 2011). Only one study tracked muscle activity with EMG analysis, and this was during quiet standing (Wakasa & Fukuda, 2013). This study only tracked the gastrocnemius of the sound limb, and found that by the end of a 4-week rehabilitation program it contracted less to establish standing equilibrium than was initially needed after amputation (Wakasa & Fukuda, 2013). Lastly, a study assessing muscle strength and the biomechanics used to complete the sit-to-stand task by elderly, found that elderly had greater relative muscle activity for all muscle

groups tested than the young sample (Gross, Stevenson, Charette, Pyka, & Marcus, 1998). They attributed this to the elderly having to demand more of their muscles to perform the task, basically that the activity was a higher intensity than for the younger sample (Gross et al., 1998). This study tested three of the same muscles as the present study, namely the tibialis anterior, vastus medialis, and biceps femoris with the addition of lateral gastrocnemius, soleus, rectus femoris, and gluteus maximus (Gross et al., 1998). Combining the results of previous and the current study, the greater EMG amplitudes seen in the amputated limb may be the result of maximal effort by the compromised muscles to contribute to and stabilize the movement prior to the body weight shifting to excessively load the sound limb. In theory, if a certain level of symmetry was established, these muscles could even strengthen in response to the overload and surpass the strength of the sound limb if it was not experiencing the same level of demands. Essentially, the muscles of the amputated limb would be performing the same movement at the same workload without the contribution of primary and accessory muscles.

For the analysis of the data, the average amplitude was taken of all four individuals, and it was found that the average amplitude for the amputated limb was greater than the average amplitude of the sound limb for all pairs of muscles. This pattern is not as strict when comparing the muscle pairs within each individual. For each muscle pairing, one or more individuals experienced a disparate measurement for muscle contraction in their amputated limb. These readings might be outliers in a larger sample, but they may also suggest that each individual develops their own altered strategy for the sit-to-stand movement that may require different muscles to play a greater part in stabilizing the center of mass. The muscles used could be affected by socket type and fit, stump length, or even joint pain which can all cause the weight to be shifted differently. Since this sample size of this study is small and there are no other studies

comparing EMG of sound and un-sound limbs of amputees, these are all theories that suggest further research.

## **Hypothesis 2**

The second hypothesis, that amputees would have longer stabilization times than their non-amputee counterparts, was supported by the results of this study. Amputees took 0.072 seconds longer on average than the non-amputee to reach stabilization. This was a difference of 13.6%. Two types of studies have been conducted on the amputee population to assess stability; postural sway during quiet standing, and time to completion of the sit-to-stand task. The majority of studies measuring postural sway have found amputees to be less stable, exhibiting greater postural sway, than the healthy control samples (Isakov, Mizrahi, Ring, Susak, & Hakim, 1992; Ozyurek, Demirbuken, & Angin, 2013). Yet some studies have found contradictory results and suggest that amputees stand as soundly as non-amputees (Vittas, Larsen, & Jansen, 1986). Studies measuring time to completion in both elderly and amputees theorize that longer times to completions are correlated to greater instability and a greater risk for falling (Akram, & McIlroy, 2011; Miller, Speechley, & Deathe, 2001a). The study by Miller, Speechly, and Deathe (2001a) found that amputees exhibited completion times more similar to the elderly group than the healthy non-amputee group, suggesting compromised stability. Therefore, the findings of the current study are a novel measurement of stability in amputees, in that the measurement was taken after performing the sit-to-stand transition, and adds to the data suggesting that amputees are less stable than non-amputees. These differences are most likely due to a decrease in muscles that assist in stabilizing the body such as the tibialis anterior and gastrocnemius, as well as, a decrease in sensory input and proprioception with the loss of spindle fibers and stretch receptors embedded in the amputated muscles.

### **Hypothesis 3**

The third hypothesis was that physical activity, absence of disease, and longer time from amputation would all improve time to stabilization. The results of this study support that physical activity and the absence of disease have a beneficial effect on stability. No studies have assessed these specific variables in amputees, but studies have shown that muscle strength and power are important factors in sit-to-stand transfers and stability in the general population (Akram & McIlroy, 2011; Dehail et al., 2007; Jansenn, Bussmann, & Stam, 2002; Riley, Krebs, & Popat, 1997). Both of these factors are improved through exercise; therefore, it is reasonable to expect improvements in sit-to-stand performance in amputees with greater muscle strength and balance training, whereas, diseased states alone, such as diabetes, have been linked to decreased stability and may be partially due to effects of medications (Kalyani et al., 2013). Therefore, both of these results support the current literature, but have not been studied in amputees specifically.

Greater differences were expected to be seen based on time since amputation with expectations of individuals having the amputation greater than 10 years ago having better stability. The opposite was seen in the results of this study. Again there is no longitudinal study that follows time to stabilization in amputees, but research suggests that the natural musculoskeletal changes that occur with age result in a decrease in stability (Akram & McIlroy, 2011; Dehail et al., 2007; Gross, Stevenson, Charette, Pyka, & Marcus, 1998; Janssen, Bussmann, & Stam, 2002). It should be noted that one of the individuals in the greater than 10 years category was also the only individual to be classified as moderate physical activity level and also had to push off the chair during standing. This may suggest that level of physical activity plays a greater role in functional level and stability than does years of experience as an

amputee. Isakov, Mizrahi, Ring, Susak, and Hakim (1992) found that their sample of amputees had decreased standing sway and improved weight distribution after just four weeks of rehabilitation. This may also support the theory that the amount of physical training and therapy may bring about improvements more so than simply years since amputation.

#### **Hypothesis 4**

The fourth hypothesis was that amputees would rise with a greater degree of trunk flexion than non-amputees. It was also hypothesized that amputees with greater levels of physical activity, greater years since amputation, or no diseases would have less trunk flexion (closer to the average non-amputee measurements) than amputees with any of these factors. The results of this study suggest that amputees do exhibit a greater degree of trunk flexion than non-amputees, but they do not support that amputees without a disease, that are highly active, or that have been an amputee for a greater number of years have less trunk flexion than amputees with diseases, lower activity level, or more recent amputation. This was in opposition of what was hypothesized.

This hypothesis was based on the findings of two studies. Bugar, Kuzelicki, and Marincek (2005) found that trans-femoral amputees use a different technique for standing. This technique is characterized by moving the buttocks forward, flexing significantly more at the hips than healthy counterparts, and placing the feet backwards in order to shorten or eliminated the period of instability (Burger, Kuzelicki, and Marincek, 2005). Although this study's population was above-the-knee amputees, as opposed to below-the-knee amputees, it was hypothesized that similar methods would be used to compensate for muscle loss. Another study by Gross, Stevenson, Charette, Pyka, and Marcus (1998), compared trunk flexion among elderly and young women. This study found that the elderly also rely on greater trunk flexion to compensate for



lower levels of muscular strength due to natural sarcopenia (Gross et al., 1998). Trunk flexion was also found to be significantly correlated to hip extensor strength, with greater strength being related to less trunk flexion (Gross et al., 1998). It was from these studies, that the hypothesis of greater activity levels, the absence of disease, and more years since amputation (in general a healthier and more adjusted individual) would have stronger muscles and more stability, thus able to stand with less trunk flexion. This was not found in the results of our study. In contrary, the t-test comparisons of moderate and high physical activity groups found a 3.1% difference with moderately active individuals producing less trunk flexion compared to highly active individuals. Similarly, individuals with diseases and individuals with less than 10 years since amputation, were found to have less hip flexion by 15.5% and 18.8%, respectively when compared to individuals without diseases and greater than 10 years since amputation. This may suggest that over time and with greater strength and health, amputees actually adopt the method of greater trunk flexion to cause more momentum, to in turn use less power from the compensated muscles. This method requires more stability to control and stop forward momentum once they reach full knee extension, which may be a skill that develops over time. The greatest difference was seen in the comparison of individuals having amputation less than ten years and greater than ten years prior. Presence of diseases (the second greatest difference) may hinder the development of this strategy or medications may effect balance as was suggested by MacGilchrist et al. (2010) and Kalyani et al. (2013). One of the greatest supports for this theory is a study done by Hermodson, Ekdahl, Persson, and Roxendal (1994) which found that vascular amputees had significantly worse standing balance than trauma amputees. If this pattern continued to hold through a larger sample, it may suggest that training on this method of movement could prove helpful in a clinical setting with individuals of a certain health profile.

## **Specific Case**

One participant showed particularly good results of all aspects of the sit-to-stand task, participant A2. This participant was extremely active, had no diagnosed diseases, and had been an amputee for 31 years. He was neither the youngest nor oldest participant at 58 years old. Of the four amputees, he had the fastest time to stabilization and greatest degree of trunk flexion. His time to stabilization (0.489 seconds) was only 0.099 seconds slower than the average time to stabilization for young adults (0.340 seconds). Although this is only one individual, this does support that overall health and activity level is correlated to functional performance.

## **Conclusions**

The sit-to-stand transition is a crucial component of ADLs required for independence. Amputees are one of many groups of people that struggle with this movement, and very little data has been collected to understand the cause of sit-to-stand issues and failures within this group. This study is limited by the small subject number and variances of characteristics among the subjects. Nevertheless, this study's findings suggest that health and activity level are important factors in the sit-to-stand movement. These hypotheses should be studied individually with larger sample sizes to obtain statistically significant data. Understanding the source of limitations with this transitional movement whether it be muscle imbalance, low activity levels, co-morbid diseases, adjustment time since amputation, or simply the amputation itself, would improve the strategies used in the rehabilitation setting to train these individuals. These hypotheses also suggest that amputees should not be considered as an entity, but considered on a case-by-case basis when developing treatment and rehabilitation plans.

**Limitations**

Due to the small sample size of this study, none of the findings are of statistical significance. This fact and the lack of supporting or refuting data from research banks suggest that further research should be conducted on these hypotheses. If any of these theories should be supported through further research, they would all have great contribution to the clinical setting. They could provide information for developing treatment and therapy plans, as well as, highlight the important factors associated with improving and preserving the sit-to-stand movement.

## Section 6: References

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