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WALKING WITH POLES IMPROVES STEP LENGTH OF SIMULATED PROSTHETIC GAIT

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Abstract

Individuals with lower limb loss often demonstrate gait deviations which cause instability and asymmetry. Walking with poles has been shown to improve stride length and stability in older adult and other patient populations. The enhanced arm involvement in bipedal walking with poles has the potential to facilitate interlimb coordination among arms and legs, and therefore is likely to improve prosthetic gait. The purpose of this study was to examine the effect of walking poles on temporal gait parameters and the center of mass (COM) displacement during prosthetic and simulated prosthetic gait. It was hypothesized that walking with poles would improve gait parameters, symmetry index and COM displacement while walking. One transfemoral amputee and seven non-amputee individuals participated. The iWalk, a hands-free crutch device, was used to simulate the gait pattern of individuals with above-the-knee amputation. All participants walked at a self-selected comfortable walking speed on a treadmill under two conditions: hands free walk (W) and pole walk (PW). Gait parameters including cadence, stance time of intact and iWalk/prosthetic legs as well as the vertical displacement of COM were analyzed. Symmetry index was calculated using iWalk/prosthetic stance phase divided by intact stance phase. Our results showed that participants exhibited significantly lower cadence and corresponding longer stance phase of the intact leg and iWalk/prosthetic leg in the PW condition compared to W at the same walking speed. In addition, there was a significantly larger vertical displacement of COM during the stance phase of the iWalk/prosthetic leg in PW compared to W. The lower cadence and greater stance time of both the intact and iWalk/prosthetic limb on the treadmill represented longer step length in PW compared to W. The longer step length also reflected in the observed larger vertical displacement of COM in PW. The longer step length may indicate better confidence level and stability during PW compared to W.

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Introduction

The number of people expected to be living with limb loss in the United States will rise from 1.6 million in 2005 to 3.6 million in 2050. Of those individuals, 65% will have undergone an amputation to the lower extremity, most commonly due to diabetes or peripheral neuropathy (Lo et al., 2020; Ziegler-Graham et al., 2008). For the year 2019, the total projected healthcare costs for individuals after lower extremity amputation was estimated to be $878,927. Not only is the cost associated with limb amputation high, but there are also significant effects to social and physical status after an amputation. Reports of individuals with lower extremity amputation (LEA) found that 42% were unable to work 7 years after the amputation and 53.9% remained non-ambulatory for 1 year after their amputation (Lo et al., 2020). While the return of ambulation is critical to functional recovery after LEA, proper, symmetrical form during gait is also important to reduce incidence of secondary musculoskeletal injuries.

Research has shown that individuals with LEA have asymmetrical gait, demonstrated by shorter times in stance and longer swing phases with the prosthetic limb, as well as 13% greater force asymmetry between limbs during walking than able-bodied controls (Nolan et al., 2003). These movement patterns could be due to the amputee relying more heavily on the intact limb in order to avoid producing larger forces through the amputated limb. This reduces the demands on the musculature of the amputated limb and can lead to asymmetrical strength between limbs overtime (Lloyd et al., 2010). Side bending of the trunk towards the intact limb combined with axial rotation away from the intact limb during double-limb support phases of gait has also been found in individuals with transtibial amputation likely as a means to compensate for weak muscular control. This has the potential to lead to asymmetry of active and passive components of the individuals' dynamic stability (Yoder et al., 2015).

With regards to the physical effects of LEA, there are many factors that can impact an amputee's gait, such as a painful stump, how well the prosthetic is fitted, and if the individual is in good health (Isakov et al., 2000). Altered biomechanics caused by these and other factors increase the risk of developing musculoskeletal pathologies including chronic and degenerative changes in weight bearing joints and tissues (Lloyd et al., 2010; Yoder et al., 2015). As the amount of weight distributed through the intact limb increases, the risk of developing osteoarthritis in the amputated limb and the intact knee increases as well. This was seen in 64% of veterans with LEA (Gailey et al., 2008). Individuals with LEA have increased rates of low back pain that has been found to be related to social experiences, level of amputation, leg length discrepancies, postural dysfunction and deviations in the kinematics of the spine and pelvis (Highsmith et al., 2019). Improving the gait symmetry of individuals with LEA could have a positive impact on health outcomes with regards to back pain, joint degeneration and osteoporosis in both the intact and amputated limb. Reducing the secondary complications of LEA would increase this population's ability to participate in everyday tasks and improve their overall quality of life (Lloyd et al., 2010).

Current strategies being implemented to improve the aforementioned gait deviations include gait training and muscle strengthening. Exercise programs consisting of supervised walking, muscle strengthening, balance training, part-to-whole gait training, and functional gait and activity training are strategies more commonly used, although there is conflicting evidence regarding improved gait with these methods (Wong et al., 2016). An ideal method of prosthetic gait training is difficult to determine due to heterogeneity between studies and conflicting evidence. In order to maximize an individual's ability to walk effectively after an amputation, new strategies to improve gait training should be considered.

Given that anatomical and physiological deficits due to limb loss are non-modifiable, rehabilitation of the prosthetic limb could benefit from utilizing the concept of neural coupling among the four limbs. Neural coupling has been defined as "flexible, task-specific, physiologically meaningful linkage of limbs during complex movements." Neural coupling explains the relationship between arm swing and leg movement as a reflex that may be influenced with pole walking (Dietz & Schrafl-Altermatt, 2016). It has been found that the active arm involvement in locomotion-like movement contributes to the modulation of lower extremity muscular activities due to neural coupling (Ferris et al., 2006; Kawashima et al., 2006). The concept of neural coupling can potentially be utilized to modulate muscle activation of the amputated limb, possibly improving symmetry between the lower limbs. Studies show that the movement of the arms and legs are in fact related and have varying degrees of influence on the other. Utilizing poles while walking may affect leg movement, facilitate more upper extremity use, influence trunk lean, and improve energy expenditure (Balter & Zehr, 2007, Pellegrini et al., 2015, 2017). In addition, the somatosensory inputs from the hands holding the poles could provide a light touch cue which has been shown to improve center of mass stability during walking (Boonsinsukh et al., 2009; Dickstein & Laufer, 2004). Despite the potential advantage from neural coupling being evident for pole walking, it is not clear whether individuals with LEA will improve their gait with this method (Pellegrini et al., 2017).

Currently, no study has investigated the benefit of pole walking on individuals with LEA, or the use of the iWalk to simulate prosthetic gait. The iWalk is a new form of crutch that allows the wearer to be hands-free without placing any of their body weight through an affected foot or ankle and may be used to simulate the gait pattern of individuals with LEA. Currently, there have been no studies comparing iWalk gait to gait with a lower extremity prosthesis, but it has the potential to simulate gait abnormalities as seen with LEA. The iWalk could potentially be utilized when recruiting participants with LEA is not possible. Based on the available research, we propose that a study investigating pole walking in individuals walking with simulated prosthetic gait through the use of an iWalk may demonstrate improvements in gait deviations. The findings of this study will inform the development of rehabilitation programs that target improvement of gait symmetry and stability in persons with LEA by implementing facilitation of neural coupling techniques through the use of pole walking.

The aims of this study were to 1) simulate prosthetic gait utilizing the iWalk crutch and 2) examine the effects of pole use on gait mechanics (i.e.gait symmetry and gait stability) and muscle activation during the simulated prosthetic gait. We hypothesized that 1) utilizing an iWalk device during walking will induce greater gait instability as evidenced by greater trunk displacement; 2a) using poles during the simulated prosthetic gait will exhibit greater gait symmetry and greater gait stability (identified by greater symmetry of the bilateral stance phases and smaller the displacements of the upper and lower trunk, respectively) and 2b) there will be less activation of the gluteus medius when utilizing poles than without during the simulated prosthetic gait as evidenced by a lower EMG reading.

Methods

Participants

One transfemoral amputee and seven non-amputee individuals participated. The iWalk, a hands-free crutch device, was used to simulate the gait pattern of individuals with above-the-knee amputation and was implemented in the non-amputee participants. Participants using the iWalk were considered eligible for the study if they were: 1) over the age of 18 and 2) had no LE comorbidities. The participant with a LEA was considered eligible for the study if they met the following inclusion criteria: 1) unilateral lower extremity amputation, 2) capable of walking independently without the use of an assistive device for at least 5 minutes, 3) use of the same prosthesis for at least 3 months and 4) over the age of 18. The following were predetermined exclusionary factors: 1) previous neurological impairments affecting walking ability and 2) having used Nordic walking poles within the past 3 months. We recruited eight non-amputee participants (4 males and 4 females) with an age range of 24 to 28 years. We also recruited one participant with transfemoral amputation as a reference (male, 34 years old). All participants were required to sign an informed consent form issued by the UNLV Department of Physical Therapy.

Procedure

The walking poles were fitted to each participant by allowing them to hold the handles with their elbows flexed to 90 degrees, then reducing the height of the poles by approximately 2 inches (5.08 cm). Participants were given a demonstration on how to properly walk with the poles, including cues to use the pole to make contact with the ground in the middle of their stride, swing the opposite arms and legs and maintain an upright posture. Participants were then given time to practice walking with the poles in the hallway of the facility, until they reported feeling comfortable with the technique.

The participants without LEA were then fitted to the iWalk by measuring the distance from the ground to the tibial tuberosity on their dominant leg and adjusting the iWalk to that height. Leg dominance was determined by asking the participant which leg they would use to kick a ball. The iWalk was then attached to the participant and time was allowed for the participant to familiarize themselves with the orthotic device. Once the participant reported feeling comfortable with the device, additional time was given to practice walking with the poles while wearing the iWalk device.

Next, the participants were brought to the split belt treadmill and fitted with a safety harness. The participant's preferred walking speed was determined with the participant wearing the iWalk. The speed of the treadmill was slowly increased until the preferred speed was reached, as reported by the participant. This procedure was then repeated without the iWalk. Subjects were blinded from the treadmill speed display.

Reflective markers were then placed on the following bony landmarks of each participant: bilateral anterior-superior iliac spine, posterior-superior iliac spine, iliac crest, greater trochanter, medial and lateral femoral epicondyle, medial and lateral malleoli, 1st and 5th metatarsal head, 2nd toe, acromioclavicular joint, medial and lateral humeral epicondyle, ulnar head, styloid process of the radius, spinous process of C7, spinous process of L5, superior/inferior aspects of shoulder girdle, acromioclavicular joints, and anterior and posterior aspects of the right and left side of the cranium. Tracking markers were a set of 3-4 markers which were attached to the lateral thigh, lateral shank, postero-lateral heel, lateral upper arm and forearm segments. The iWalk had tracking markers at the lateral shank level. The walking poles had tracking markers on the distal and proximal ends. The Trigno® Wireless Biofeedback System was used to record gluteus medius activation with Trigno AvantiTM Sensors placed halfway between the iliac crest and the greater trochanter bilaterally. Static calibration was performed to define joint centers, then the landmark markers were removed.

The outcome measures were assessed under the following conditions: normal walking with hands free (NW), walking with iWalk with hands free (IW), and walking with iWalk plus walking poles (IWP). The participant with the LEA was assessed under the prosthetic walking with and without poles. Participants first walked without the use of poles at their preferred walking speed for 2 minutes to obtain four 30 second data recordings. Following this, participants were asked to use walking poles in the second condition, while walking an additional 2 minutes at the same speed for another four 30 second data recordings. The same two step procedure was used for the iWalk conditions. Marker trajectories were recorded using the 12-camera Vicon system.

In order to compare EMG data, a maximum voluntary isometric contraction (MVIC) trial was recorded at the end of the aforementioned sessions. The participant was asked to lay on their side with the lower leg bent at a 90˚ angle while keeping the upper leg extended. A research assistant instructed the participant to activate their gluteus medius by abducting their lower extremity while the assistant applied an inferior force to the distal lower extremity.

Data Processing

Once data collection was complete, .mdh model templates were made for each condition (NW, NWP, IW, and IWP). This was necessary to account for the difference in number and placement of markers to fit the equipment used in the different conditions. Each marker for each participant was then labeled to correspond to the appropriate model template. Next, biomechanical models were created in Visual3D (C-Motion, Inc., Rockville MD, USA) for all trials and used to check for accuracy of the labeling and orientation of the assessed structures. Utilizing the Visual3D motion analysis software, bilateral stance phase duration, C7 and L5/S1 medio-lateral and vertical displacement were recorded under all four conditions for each participant. Stance time during the trails was determined by assigning time values to the initial contact and toe-off of each limb. The stance phases were calculated and averaged for analysis. Trunk movements were calculated via C7 and L5/S1 markers moving in the frontal, sagittal and transverse directions. Displacement values for these markers were averaged over 5 consecutive steps to correspond to the axis in a given plane of motion.

Statistical Analysis

All statistical analyses were performed with SPSS 25.0 statistical software (International Business Machines Corp., Armonk, NY, USA). Paired samples t-test were used to compare the different variables (medio-lateral and vertical displacement of C7 and L5) between the NW and IW conditions. Paired samples t-test were also used to compare variables (medio-lateral and vertical displacement of C7 and L5, intact limb stance phase, iWalk stance phase, and stance phase symmetry index) between IW and IWP conditions. A priori significance was set at $p \leq 0.05$.

Results

Prosthetic Simulation

A total of eight participants (4 men and 4 women) with an age range of 24 to 34 years met the criteria to be part of the study. Within this sample, we observed a statistically significant difference in the medio-lateral displacement of C7 during the stance phase of intact limb while using the iWalk when compared to normal walking as well as during stance phase of the iWalk limb when compared to normal walking (NW = 0.0584 ± 0.0065, IW intact = 0.1398 ± 0.0137, $p < 0.001$; IW iWalk = 0.1247 ± 0.0174, $p < 0.001$). We also observed a statistically significant difference in the medio-lateral displacement of L5 during the stance phase of intact limb while using the iWalk when compared to normal walking as well as during stance phase of the iWalk limb when compared to normal walking (NW = 0.0589 ± 0.0088, IW intact = 0.1106 ± 0.0220, $p < 0.001$; IW iWalk 0.0854 ± 0.0224, $p = 0.008$). (Table 1).

Table 1. C7 and L5 displacements during stance phase of normal walk (NW) and iWalk (IW) (Group mean of standard deviation of 7 non-amputees)

	Normal Walk (NW) stance phase	iWalk (IW) intact stance phase	iWalk (IW) iWalk stance phase
C7 medio-lateral displacement	0.0584 ± 0.0065	0.1398 ± 0.0137* ($p < 0.001$)	0.1247 ± 0.0174* ($p < 0.001$)
C7 vertical displacement	0.0322 ± 0.0104	0.0265 ± 0.0067 ($p = 0.12$)	0.0253 ± 0.0064 ($p = 0.057$)
L5 medio-lateral displacement	0.0589 ± 0.0088	0.1106 ± 0.0220* ($p < 0.001$)	0.0854 ± 0.0224* ($p = 0.008$)

L5 vertical displacement	0.0320 ± 0.0094	0.0329 ± 0.0052 (*p* = 0.789)	0.0307 ± 0.0048 (*p* = 0.675)

* indicates significant difference from NW

Effects of Walking with Poles – Gait Symmetry and Stability

We observed a statistically significant difference in the stance time of the intact side when comparing hands-free walking with the iWalk to using poles while walking with the iWalk (IW = 1.13 ± 0.15, IWP = 1.31 ± 0.21, *p* = 0.003). We also observed a statistically significant difference in the stance time of the iWalk wearing side when comparing hands-free walking with the iWalk to using poles while walking with the iWalk (IW = 0.88 ± 0.12, IWP = 1.01 ± 0.22, *p* = 0.025). There was also a statistically significant difference in the amount of L5 vertical displacement during the intact limb stance phase when compare walking with the iWalk hands-free and with walking poles (IW = 0.03 ± 0.01, IWP = 0.04 ± 0.01, *p* = 0.047), as well as the amount of L5 vertical displacement during the iWalk limb stance phase (IW = 0.03 ± 0.00, IWP = 0.04 ± 0.01, *p* = 0.049). (Table 2).

Table 2. Gait parameters, C7 and L5 displacements of iWalk (IW) and iWalk with pole (IWP) of 7 non-amputees

	iWalk (IW)	iWalk with poles (IWP)	*p* value
Intact side stance phase (s)	1.13 ± 0.15	1.31 ± 0.21	*p* = 0.003*

iWalk side stance phase (s)	0.88 ± 0.12	1.01 ± 0.22	p = 0.025*
Stance phase symmetry index	0.78 ± 0.02	0.77 ± 0.05	p = 0.563
C7 mediolateral displacement during intact side stance phase (m)	0.14 ± 0.01	0.13 ± 0.02	p = 0.124
C7 vertical displacement during intact side stance phase (m)	0.03 ± 0.01	0.03 ± 0.01	p = 0.239
L5 medio-lateral displacement during intact side stance phase (m)	0.11 ± 0.02	0.11 ± 0.02	p = 0.448
L5 vertical displacement during intact side stance phase (m)	0.03 ± 0.01	0.04 ± 0.01	p = 0.047*
C7 medio-lateral displacement during iwalk side stance phase (m)	0.12 ± 0.02	0.12 ± 0.02	p = 0.642
C7 vertical displacement during iwalk side stance phase (m)	0.03 ± 0.01	0.03 ± 0.01	p = 0.433
L5 mediolateral displacement during iwalk side stance phase (m)	0.09 ± 0.02	0.10 ± 0.03	p = 0.254
L5 vertical displacement during iwalk side stance phase (m)	0.03 ± 0.00	0.04 ± 0.01	p = 0.049*

Effects of Walking with Poles – EMG Data

For the EMG data we compared three non-amputee controls (C) with individuals with transfemoral amputation (TFA). For the intact limb when walking hands-free the average MVIC of the gluteus medius was found to be TFA = 0.0845, C = 0.0626 ± 0.0152. For the intact limb when walking with poles the average MVIC of the gluteus medius was found to be TFA = 0.0490, C = 0.0559 ± 0.0214. (Figure 1). For the iWalk/amputated limb when walking hands-free the average MVIC of the gluteus medius was found to be TFA = 0.0367, C = 0.0492 ± 0.0142. Lastly, for the iWalk/amputated limb when walking with poles the average MVIC of the gluteus medius was found to be TFA = 0.0279, C = 0.0349 ± 0.0070. (Figure 2).

Figure 1. Average EMG during intact leg stance phase, iWalk and iWalk with poles conditions

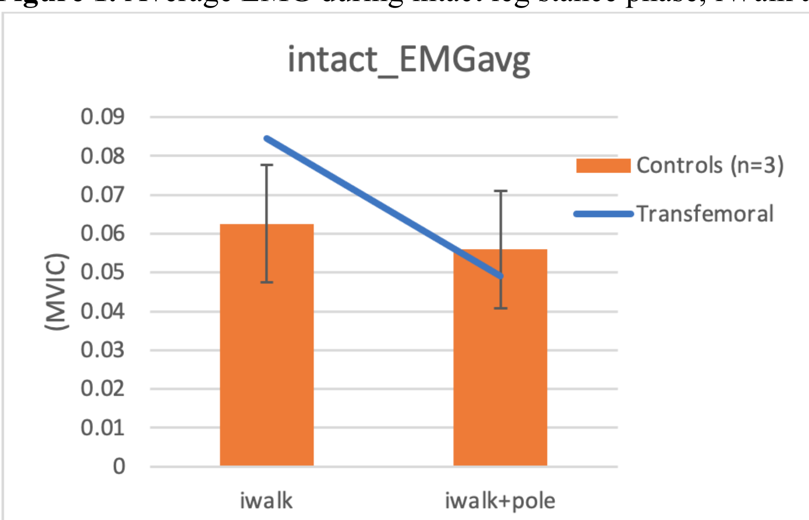
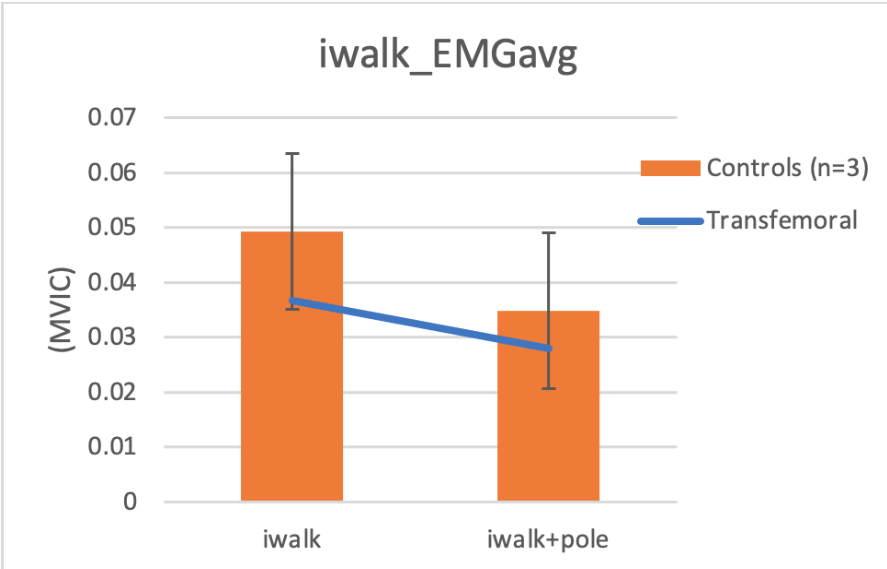


Figure 2. Average EMG during iWalk leg stance phase, iWalk and iWalk with poles conditions



Discussion

The aims of this study were 1) to examine the effect of simulated prosthetic gait through the use of the iWalk crutch in non-amputee individuals; 2) to examine the effects of pole walking on gait mechanics and muscle activation of the gluteus medius during the simulated prosthetic walking; We found that 1) with the application of iWalk, the non-amputee individuals showed larger medio-lateral displacement at C7 and L5. This aligned with our hypothesis that utilization of the iWalk would introduce greater gait instability; 2a) there were significant differences in the stance times of both the intact limb and the iWalk limb when using poles compared to hands-free walking, as well as the vertical displacement of L5 during intact and iWalk stance. However, the stance phase symmetry index was not found to be significantly different between IW and IWP. We hypothesized that the displacement of the upper and lower trunk during simulated prosthetic gait would be significantly smaller in the medio-lateral and vertical directions when walking with the poles compared to hands-free when wearing the iWalk device. This hypothesis was partially supported. While we observed significant changes in the vertical displacement of L5 during both the intact limb stance and iWalk stance, there was however, no significant difference in L5 medio-lateral displacement or C7 medio-lateral and vertical displacement. Our hypothesis was that individuals using the iWalk device will exhibit greater gait symmetry as evidenced by comparable stance phases between the two limbs when walking with the poles compared to walking hands-free was not supported. Our findings indicate that the use of the poles allowed participants to walk with a slower cadence and longer steps when walking with poles.

Our observation in the effect of the simulated prosthetic gait through the application of the iWalk device provided new findings to current literature. Lemaire et al, focused on hip and knee mechanics with a prosthetic simulator similar to the iWalk. Their design differed from our own by implementing two 45-60 minute gait training sessions before having subjects walk a 10 meter distance. The study also used canes as an assistive device instead of walking poles for comparison. In conclusion, Lemaire's experiment found that inexperienced simulator subjects produced more work at the hip to ensure foot clearance during swing phase. Duration of hip flexor activity was longer for simulator subjects compared to above knee amputation subjects (Lemaire, 2000).

In the study by Champagne (2017) prosthetic gait was also simulated using the iWalk crutch device, however their methods and outcomes were different than our study. Champagne assessed gait in barefoot participants, compared to the shod participants in our study. The fitting of the iWalk was consistent between studies, however, it is difficult to determine how much effect barefoot vs. shod would have on proper fit or gait mechanics. Champagne (2017) also fit the iWalk to alternating sides for repeated trials, whereas we determined a single side to don the iWalk device based on leg dominance. In terms of assessing gait, Champagne (2017) controlled for gait speed across all conditions at 1.03 m/s. Contrarily, we applied preferential gait speed for "normal walking" and "iWalk" conditions, respectively. Furthermore, we observed differences in stance and body segment displacement as opposed to Champagne (2017) which observed hip angle during phases of gait. With regard to the results of our study compared to Champagne (2017) we observed larger medio-lateral displacement at the upper and lower trunk during the simulated prosthetic gait, indicating decreased gait stability. Champagne (2017) found that the iWalk device caused a significant increase in hip flexion angle of individuals using the iWalk compared to walking without the device during mid support. This increased hip flexion angle is postulated to cause an increase in moment at the hip during stance phases. Although the specific deficits in gait in a person with lower limb amputation may vary, changes in hip angle at stance may occur. These deficits between sides can contribute to hip joint and low back pain, among other conditions. This may also contribute to the increased amount of work performed at the simulated LEA side. Since there is no simulated ankle or knee movement at the iWalk, more movement at the hip is required to clear the ground throughout swing phase or to propel during stance.

Additionally, we found pole-walking to have several effects on gait parameters such as increased stance phase which reflects the findings of (Wilson, 2001) and (Shim 2012). We also observed an increased L5 vertical displacement and decreased gluteus medius activation with pole-walking. The increased time in stance phase could be attributable to the greater base of support that the poles provide to their users, thus decreasing the instability that comes from weight/bearing on a narrow, non-compliant device and increasing user confidence during stance phase. It is also possible that the increased time in stride phase we observed was due to the horizontal propulsion the Nordic-style of pole-walking provides its users via its postero-inferior direction of force, thus requiring its users to take longer steps. Studies comparing gait during Nordic walking and unassisted walking have also shown that step length is significantly increased by pole walking (Shim, 2012). There remain significant differences in cadence in the Nordic pole group in gait excluding the phase between heel off to toe off. Nordic poles shorten the time from foot flat to heel off and may shorten the midstance during stance phase (Shim, 2012).

Although it was a small study with a single session design, the findings from our study can be utilized to create protocols for larger studies in the future. Walking with poles provides an economical alternative to current prosthetic gait training. Further studies are recommended to determine if the compensations seen during iWalk gait are comparable to those seen after LEA.

We also had a small sample size of eight, including a single participant with LEA. Further studies are needed in order to determine if these findings can be generalized to a larger amputee population. The one-session design of this study led to limited time for the participants to learn how to walk while wearing the iWalk and learn the technique of walking with poles. Further studies are needed to determine the effects of long-term training with pole walking on prosthetic gait.

The findings of our study are limited by several factors. The COVID-19 pandemic limited access to a larger number of individuals with LEA, therefore the majority of the data is taken from non-amputee participants utilizing an iWalk device. This study also had a small sample size of eight, including a single participant with LEA. Further studies are needed to determine if these findings can be generalized to a larger amputee population. The one-session design of this study led to time limitations for the participants to learn how to walk while wearing the iWalk as well as learning the technique of walking with poles. Further studies are needed to determine the effects of long-term training with pole walking on prosthetic gait, and potentially a dose-response relationship.

Conclusion

With this study we aimed to simulate prosthetic gait by utilizing the iWalk device and examine the effects of walking with poles on gait mechanics during simulated prosthetic gait. The findings from this study suggest that utilizing pole walking during simulated prosthetic gait allowed participants to walk with lower cadence and greater stance time on both sides. The longer step length was also reflected in the observed larger vertical displacement of L5 when utilizing the poles. The longer step length may indicate increased confidence level and stability when utilizing the poles during the simulated prosthetic walking. Future studies are needed to examine the long-term training effects of pole walking on prosthetic gait. The use of a pole walking technique during prosthetic gait training may be beneficial in order to improve the gait pattern of those with lower extremity amputation. Further research is needed to determine if the symmetry between stance phases for individuals with LEA can be positively influenced by the use of walking poles.

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