**1. Introduction**

Walking is a basic daily human activity. To be efficient and safe, it needs to be adaptable to changing environments and activity demands. In the last twenty years, walking adaptation has been studied to understand how the nervous system changes locomotion patterns based on differing constraints. A variety of perturbations have been used; for example, a split belt treadmill [[1](#_ENREF_1), [2](#_ENREF_2)], a curved trajectory [[3](#_ENREF_3), [4](#_ENREF_4)], robotic orthoses [[5](#_ENREF_5), [6](#_ENREF_6)] or a swing phase resistance [[7](#_ENREF_7)]. The motivation of these studies has been to use these paradigms to improve post-stroke gait asymmetry [[8](#_ENREF_8), [9](#_ENREF_9)].

In young non-disabled adults, locomotor adaptation paradigms consistently induce an initial gait asymmetry that gradually decreases with practice. When the perturbation is removed, the gait pattern immediately becomes asymmetrical in the opposite direction (a negative aftereffect) before returning to its initial state [[1](#_ENREF_1), [7](#_ENREF_7), [10](#_ENREF_10)]. This is taken as evidence of the nervous system’s capacity to alter gait through feedforward mechanisms in response to a perturbation and temporarily store these changes, which can be thought of as a form of short-term motor learning [[10](#_ENREF_10)].

Only a few studies have looked at unilateral weighting, an inexpensive, easily available way to alter limb dynamics and perturb gait, in young, non-disabled adults. One study showed that after applying a weight to one leg during treadmill walking, step length, stance and swing times developed an asymmetrical pattern [[11](#_ENREF_11)]. When the weight was removed, stance time immediately returned to a symmetrical pattern. However, step length showed an aftereffect, indicating the nervous system stored the new weight-induced step length pattern. In a different paradigm, a weight was used to investigate the effects of inertial manipulation during overground walking. At a self-selected walking speed, the weight promoted asymmetry of stance [[12](#_ENREF_12)] and step time [[13](#_ENREF_13)] but aftereffects were not investigated. Therefore, the existence of a short-term learning effect after overground training is not known.

The current study’s purpose was to investigate the ability of a unilaterally applied ankle weight to drive locomotor adaption during overground and treadmill walking and to compare the overground negative aftereffects of each in young, non-disabled participants. Based on Noble and Prentice [[11](#_ENREF_11)], we hypothesized that adapting to an ankle weight during treadmill walking would result in step length negative aftereffects only during subsequent overground walking. Given the attentional demands of walking overground, we hypothesized that step length symmetry negative aftereffect magnitude would be greater for those who adapted gait overground versus on a treadmill.

**2. Methods**

**Participants**

Eighteen young non-disabled participants (12 female) aged 18 to 25 (21.39±0.46) years were recruited from the XXXXXXXXX student community. See Table 1 for participant characteristics. The inclusion criterion was age between 18 and 30 years. Exclusion criteria were: (1) a self-reported inability to walk for 30 minutes without stopping and, (2) any self-reported neurological, orthopedic or other deficit that could compromise gait. All participants reported having previous treadmill experience and provided written informed consent before testing. The experimental protocol was approved by the XXXXXXXXXXX Institutional Review Board.

**Paradigm**

The adaptation paradigm consisted of three conditions: unweighted baseline, adaptation with a weight and de-adaptation with the weight removed. During adaptation, participants walked for 10 minutes with a weight equal to 5% of their body weight secured with Velcro straps on their right or left ankle. We chose 10 minutes for the adaptation period, 5% weight, and the ankle based on previous literature [[7](#_ENREF_7), [9](#_ENREF_9)] and pilot work. Additionally, given the leg’s long lever arm, the ankle allowed the greatest perturbation with the least weight.

**Data Collection**

To measure gait parameters, participants walked on an eight-meter-long gait mat (GAITRite, CIR Systems, Inc., Sparta, NJ, USA) with their arms free to swing, wearing walking shoes. The GAITRite mat is a portable walkway containing a grid of sensors that records footfalls and calculates spatial/temporal gait parameters [[14](#_ENREF_14)]. Walking the length of the GAITRite mat equaled one trial. Gait parameters were measured on five occasions: (1) *Baseline (BL),* the average of the last two of three trials*,* (2) *Early Adaptation (EA),* the average of the first trial performed after adding weight to the participant’s ankle, (3) *Late Adaptation (LA)*, the average of one trial performed at the end of the adaptation walking session, (4) *Early Deadaptation (ED)*, the average of the first trial performed after the weight was removed and (5) *Late Deadaptation (LD)*, the average of the last two trials (Figure 1).

After the BL condition, step length (SL) data were reviewed and the weight was placed on the leg with the shorter average BL SL. We chose the leg with the shorter BL SL to intentionally exaggerate the participant’s SL asymmetry so that negative aftereffects, if present, would result in the perturbed leg having a longer SL during the deadaptation condition [[1](#_ENREF_1)]. We chose to analyze SL and single limb support (SLS) time because these parameters are often asymmetrical in clinical populations[[15-18](#_ENREF_15)]. Participants were randomized into three groups based on the context in which adaptation took place: an overground group (OG), a treadmill group (TG) and a control group (CG). The CG was not weighted and was included to verify that walking for 10 minutes would not alter gait symmetry. Controls were split into two equal sub-groups (OG and TG).

The OG adapted gait in a corridor (1.6 m wide, 64.0 m long) adjacent to the gait lab. Participants were instructed to walk continuously at a comfortable speed without stopping or talking to others. At the end of the adaptation condition, participants returned to the lab and walked over the gait mat (Late Adaptation). The TG adapted their gait on a treadmill (Woodway, Waukesha, WI) in an adjacent room. The treadmill speed was set to the participant’s average baseline overground walking speed. Participants were instructed to look ahead, not talk, and let their arms swing freely. For participant safety, a research assistant supervised the participant at all times while they were on the treadmill. To ensure that adaptation to the weight occurred only while on the treadmill, a wheel chair was used to transport participants from the gait mat to the treadmill and to take them back to the gait mat. Thus, all adaptation steps (except those occurring during EA and LA conditions) occurred on the treadmill.

The participants in the CG, which did not have a weight placed on their leg, performed their adaptation walking session overground or on the treadmill following the same procedures as the OG and TG. Adaptation conditions were counterbalanced so that equal numbers of participants adapted their gait overground and on the treadmill. To ensure all steps occurred on the gait mat during deadaptation, a chair was placed at each mat end. Participants were instructed to stop at the mat end and sit in the chair until the start of the next trial.

**Data Reduction**

Primary outcomes were SL symmetry, the forward distance between consecutive heels at initial contact, SLS time symmetry, the time the contralateral leg was in swing, and gait speed. The weighted leg was named the “perturbed leg”.

Data were collected and processed with GAITRite specific software. To remove acceleration and deceleration effects, the first and last steps performed on the mat were not analyzed. All GAITRite ASCII files were exported to Microsoft Excel spreadsheets. Symmetry was quantified by a symmetry index (SI):

where Xu is the value for the unperturbed and Xp is the value for the perturbed leg. Perfect symmetry will result in an SI value of zero. Positive values indicate the unperturbed leg had a longer SL/SLS time and negative values indicate the perturbed leg had a longer SL/SLS time [[19](#_ENREF_19), [20](#_ENREF_20)]. All variables were calculated for individual steps in each participant, then individual averages were calculated for each experimental condition. Finally, a group average was determined for each measuring period.

**Statistical Analysis**

We ran a Kruskall Wallis test comparing outcomes of the CG participants who walked overground to those who walked on the treadmill. There were no between-group differences for any outcome measures. Therefore, CG group data were collapsed into one group (CG). Data were normally distributed as indicated by a Shapiro-Wilk test. A repeated measures two-way analysis of variance (ANOVA) with group (TG, OG and CG) and experimental condition (BL, EA, LA, ED and LD) was used with the following dependent variables: SL symmetry, SLS symmetry and gait speed. When an ANOVA was significant, Fisher post-hoc tests were employed. Group average limb weights were compared with a two-tailed T-Test. Within-group effect sizes (Cohen’s d: difference in means divided by pooled standard deviation) were calculated for each pair of significant comparisons. An alpha level of 0.05 was adopted for all statistical tests, which were conducted using SPSS software version 22.0 and Statistica (Statsoft/Dell).

**3. Results**

See Table 2 for the average number of strides in each key period and Table 3 for SL values for each leg. The ANOVA for SL symmetry revealed a group by time interaction (F8,60= 3.62, p=0.001) and time main effect (F4,60= 11.62, p<0.001). (See Figure 2A). Post hoc testing indicated no within group differences at early adaptation (EA) compared to baseline (BL) for any group (p>0.30). At late adaptation (LA), the TG, but not OG, had greater asymmetry with a longer SL on the perturbed leg compared to BL (p=0.003, *d*=1.24). At early deadaptation (ED), both TG and OG demonstrated a negative aftereffect, with an asymmetry associated with a longer SL on the perturbed leg (p<0.001, *d*=1.97 and 2.41 respectively) without differences between groups (p=0.10). By late deadaptation (LD), all groups returned to their BL symmetry pattern.

For SLS time symmetry, the ANOVA revealed a group by time interaction (F8,60= 16.46, p<0.001), a time main effect (F4,60= 31.87, p<0.001) and a group main effect (F2,15= 31.87, p<0.001) (see Figure 2B). See Table 4 for individual SLS values. Post hoc testing indicated that at EA, both OG and TG SLS time became more asymmetric compared to their respective BLs with the unperturbed leg spending a longer time in SLS (p<0.001, *d*=3.67 and 3.17 respectively). This difference was maintained for both groups through LA. During ED, there was a significant difference in SLS time symmetry between the OG and TG (p<0.001, *d*=1.82). The TG returned to BL values while the OG was unchanged from LA (p=0.51). By LD, all groups had returned to their BL symmetry.

Figure 2C shows gait speed changes across the key time points of interest. The ANOVA revealed a main effect of time only (F4,60= 9.48, p<0.001). Post hoc testing revealed that all participants decreased their gait speed at EA compared to BL (p<0.01, *d*=0.46). At LA, gait speed increased but not significantly. At ED, gait speed increased significantly compared with BL (p<0.001, *d*=0.26) and further increased at LD (p=0.003, *d*=0.51).

**4. Discussion**

The present study compared the ability of a unilaterally applied ankle weight to drive locomotor adaptation during overground versus treadmill walking in young, non-disabled participants. During EA, participants maintained their baseline step length (SL) symmetry but became asymmetrical in their single limb support (SLS) while decreasing their speed. After 10 minutes of walking with a weight, participants on the treadmill increased their SL asymmetry. When the weight was removed, participants from both groups had a negative aftereffect in SL symmetry and increased their gait speed; TG participants also became symmetrical (baseline) in SLS. With continued walking, all variables eventually returned to baseline values.

The main finding from this study is that both training paradigms produced equal negative aftereffects for spatial (SL symmetry) but not temporal outcomes (SLS symmetry). Our data indicate that participants temporarily stored a changed SL symmetry pattern when walking was perturbed either on a treadmill or overground. These results are similar to split-belt treadmill perturbations of Morton and Bastian [[21](#_ENREF_21)] and Reisman et al. [[1](#_ENREF_1)], but not the unilateral swing leg perturbation of Savin et al. [[7](#_ENREF_7)] or unilateral weighting on a treadmill [[11](#_ENREF_11)], which showed a negative aftereffect for spatial and temporal outcomes in non-disabled young individuals.

One unexpected observation was that immediately after adding a weight, no difference was observed in SL symmetry. Kodesh et al. [[13](#_ENREF_13)] also found no change in SL symmetry when adding a weight to one leg during overground walking at a self-selected speed. We speculate that we could not measure a change because participants compensated for the extra weight by slowing their gait velocity, thereby reducing the rotational energy required to advance the weighted leg and reducing the perturbation. (See Figure 2C) This compensation would not be seen in other experiments where the perturbation was applied in a constant velocity condition on the treadmill. Since our experimental paradigm required overground testing for all groups, we did not see the typical initial symmetry decrease. Participants adjusted unweighted leg SL to match that of the weighted leg even though they were asked to maintain their earlier preferred walking speed. Had we assessed the TG on the treadmill, we would likely have seen the same decreased symmetry results as Nobel and Prentice [[11](#_ENREF_11)] or Savin et al. [[7](#_ENREF_7)]. Encouragingly, even though the OG were not constrained to keep a constant velocity, they clearly adapted to the differential weight of the leg as demonstrated by a negative aftereffect similar to the TG.

Despite similarities in kinematic parameters described during overground vs. treadmill walking [[22](#_ENREF_22), [23](#_ENREF_23)], differences have been observed during walking in these contexts in speed [[24](#_ENREF_24)], kinetic [[22](#_ENREF_22), [23](#_ENREF_23)], electromyographic parameters [[22](#_ENREF_22)] and contextual cues (visual and proprioceptive) [[24](#_ENREF_24), [25](#_ENREF_25)]. These differences can affect learning and adaptation to a perturbation [[26](#_ENREF_26), [27](#_ENREF_27)] and may explain differences between our results and those of Noble and Prentice [[11](#_ENREF_11)].

Regarding SLS symmetry, no differences were observed between weighted groups until the weights were removed. At EA, both groups immediately became less temporally symmetrical by increasing unweighted leg support time. Both maintained this asymmetry, with a slight trend toward symmetry, through LA. However, the OG produced an adaptive effect by continuing to maintain the same increased asymmetry at ED while the TG immediately returned to baseline. This suggests that the TG altered this gait parameter through a feedback/reactive mechanism, allowing SLS symmetry to revert to baseline values once the perturbation was removed [[8-11](#_ENREF_8), [19](#_ENREF_19)]. On the other hand, the OG showed maintenance of temporal asymmetry after weight removal, implying the use of feedforward mechanisms (albeit a positive aftereffect). Although somewhat speculative, this positive aftereffect could be due to the relatively unconstrained nature of overground walking compared to treadmill walking. Without a moving treadmill belt to drive stance leg hip extension, thereby limiting SLS time, the nervous system would be freer to alter its output in a feedforward manner. The initial SLS time asymmetry increase at EA for both TG and OG is due to increased weighted leg swing time. The continued asymmetry at ED for OG could be attributed to feedforward changes occurring during adaptation. The difference between SLS and SL symmetry aftereffects may also reflect distinct neural control strategies underlying spatial and temporal adaptation [[26](#_ENREF_26), [28](#_ENREF_28)].

Aftereffects may be explained by use-dependent learning mechanisms, when repetitive movement performed in a particular condition can influence future similar movements [[29](#_ENREF_29)]. This learning mechanism that changed the movement in response to the perturbation appears to be task dependent [[30](#_ENREF_30)] and, according to our results, context dependent. Additionally, the immediate change of SLS time by the TG could be due to an inability to transfer one’s gait pattern from a more constrained condition (treadmill walking) to a less constrained condition (overground walking), again suggesting context plays a role during locomotor adaptation. Alternatively, SLS time differences at ED could be due to the OG group adapting to a greater average weight (see Table 1). This, however, would not explain the finding of no between-group differences in SL symmetry at ED.

The unilateral weight paradigm proposed here differs from previous paradigms designed to provide perturbations and induce adaptation in three key ways. First, the perturbation is applied throughout the gait cycle, not just during the swing [[7](#_ENREF_7), [31](#_ENREF_31)] or stance phases [[6](#_ENREF_6)]. Second, the ankle weight allows gait perturbation to occur on different surfaces i.e., overground and treadmill. Third, it could be easily transferred to the clinic because it is a cheaper, more accessible, more easily implemented way to induce locomotor adaptation than other previously described methods and can flexibly match individual patient training protocol characteristics (surface, perturbed limb, weight load, training schedule). With no between group SL symmetry aftereffect differences, overground or treadmill adaptation could be used depending on patient needs/safety. After stroke, our data support targeting SL asymmetries by weighting the leg with the shorter step length with aftereffects increasing step length.

Study limitations include a small sample size, although effect sizes were moderate to large indicating meaningful differences, a lack of gait speed control overground, different average perturbing weights between TG and OG, and not assessing the TG in the same context in which they were trained. Because of the latter, we cannot directly compare our results to the one previous study that used weights on a treadmill. However, our paradigm has the advantage of being more clinically relevant since it is overground walking we are trying to ultimately improve through perturbation training. To our knowledge this is the first study to use a weight to promote locomotor adaptation during overground walking and therefore is a first step to understanding the ability of this paradigm to bring about locomotor adaptation during functional walking.

**Conclusion**

Using a short-term unilateral weighting protocol, we found that participants adapted their step length symmetry using feedforward mechanisms in either an overground or treadmill context. For single limb support, it appeared that feedback mechanisms were used on the treadmill but overground training involved feedforward mechanisms. To understand the potential effects of this paradigm and to help clinical decision making, future studies should investigate short and long-term training in participants with stroke as well as kinematic, kinetic and electromyographic parameters associated with the changes in gait asymmetry observed in the present study.

**Conflict of interest statement**

The authors report no conflicts of interest, financial or otherwise.

**References**

[1] D.S. Reisman, H.J. Block, A.J. Bastian, Interlimb coordination during locomotion: what can be adapted and stored?, J. Neurophysiol. 94 (2005) 2403-15.

[2] F. Mawase, et al., Kinetic adaptation during locomotion on a split-belt treadmill, J. Neurophysiol. 109 (2013) 2216-27.

[3] C.R. Gordon, et al., Adaptive plasticity in the control of locomotor trajectory, Exp. Brain Res. 102 (1995) 540-5.

[4] K.D. Weber, et al., Motor learning in the "podokinetic" system and its role in spatial orientation during locomotion, Exp. Brain. Res. 120 (1998) 377-85.

[5] T. Lam, M. Anderschitz, V. Dietz, Contribution of feedback and feedforward strategies to locomotor adaptations, J. Neurophysiol. 95 (2006) 766-73.

[6] M. Noel, K. Fortin, L.J. Bouyer, Using an electrohydraulic ankle foot orthosis to study modifications in feedforward control during locomotor adaptation to force fields applied in stance, J. Neuroeng. Rehabil. 6 (2009) 1-11.

[7] D.N. Savin, S.C. Tseng, S.M. Morton, Bilateral adaptation during locomotion following a unilaterally applied resistance to swing in nondisabled adults, J. Neurophysiol. 104 (2010) 3600-11.

[8] D.S. Reisman, et al., Locomotor adaptation on a split-belt treadmill can improve walking symmetry post-stroke, Brain 130 (2007) 1861-72.

[9] D.N. Savin, et al., Poststroke hemiparesis impairs the rate but not magnitude of adaptation of spatial and temporal locomotor features, Neurorehabil. Neural Repair 27 (2013) 24-34.

[10] D.S. Reisman, A.J. Bastian, S.M. Morton, Neurophysiologic and rehabilitation insights from the split-belt and other locomotor adaptation paradigms, Phys. Ther. 90 (2010) 187-95.

[11] J.W. Noble, S.D. Prentice, Adaptation to unilateral change in lower limb mechanical properties during human walking, Exp. Brain Res. 169 (2006) 482-95.

[12] J.D. Smith, P.E. Martin, Walking patterns change rapidly following asymmetrical lower extremity loading, Hum. Mov. Sci. 26 (2007) 412-25.

[13] E. Kodesh, et al., Walking speed, unilateral leg loading, and step symmetry in young adults, Gait Posture 35 (2012) 66-9.

[14] S. Nadeau, M. Betschart, F. Bethoux, Gait analysis for poststroke rehabilitation: the relevance of biomechanical analysis and the impact of gait speed, Phys. Med. Rehabil. Clin. N. Am. 24 (2013) 265-76.

[15] J.L. Allen, S.A. Kautz, R.R. Neptune, Step length asymmetry is representative of compensatory mechanisms used in post-stroke hemiparetic walking, Gait Posture 33 (2011) 538-43.

[16] K.K. Patterson, et al., Gait asymmetry in community-ambulating stroke survivors, Arch. Phys. Med. Rehabil. 89 (2008) 304-10.

[17] J.H. Yang, et al., Asymmetrical gait in adolescents with idiopathic scoliosis, Eur. Spine J. 22 (2013) 2407-13.

[18] G. Yogev, et al., Gait asymmetry in patients with Parkinson's disease and elderly fallers: when does the bilateral coordination of gait require attention?, Exp. Brain Res. 177 (2007) 336-46.

[19] J.T. Choi, et al., Walking flexibility after hemispherectomy: split-belt treadmill adaptation and feedback control, Brain 132 (2009) 722-33.

[20] L.A. Malone, E.V. Vasudevan, A.J. Bastian, Motor adaptation training for faster relearning, J. Neurosci. 31 (2011) 15136-43.

[21] S.M. Morton, A.J. Bastian, Cerebellar contributions to locomotor adaptations during splitbelt treadmill walking, J. Neurosci. 26 (2006) 9107-16.

[22] M.P. Murray, et al., Treadmill vs. floor walking: kinematics, electromyogram, and heart rate, J. Appl. Physiol. 59 (1985) 87-91.

[23] P.O. Riley, et al., A kinematic and kinetic comparison of overground and treadmill walking in healthy subjects, Gait Posture 26 (2007) 17-24.

[24] R. Bayat, H. Barbeau, A. Lamontagne, Speed and temporal-distance adaptations during treadmill and overground walking following stroke, Neurorehabil. Neural Repair 19 (2005) 115-24.

[25] D.S. Reisman, et al., Split-belt treadmill adaptation transfers to overground walking in persons poststroke, Neurorehabil. Neural Repair 23 (2009) 735-44.

[26] G. Torres-Oviedo, A.J. Bastian, Natural error patterns enable transfer of motor learning to novel contexts, J. Neurophysiol. 107 (2012) 346-56.

[27] G. Torres-Oviedo, A.J. Bastian, Seeing is believing: effects of visual contextual cues on learning and transfer of locomotor adaptation, J. Neurosci. 30 (2010) 17015-22.

[28] L.A. Malone, A.J. Bastian, G. Torres-Oviedo, How does the motor system correct for errors in time and space during locomotor adaptation?, J. Neurophysiol. 108 (2012) 672-83.

[29] K.V. Huynh, et al., Comparing aftereffects after split-belt treadmill walking and unilateral stepping, Med. Sci. Sports Exerc. 46 (2014) 1392-9.

[30] J. Diedrichsen, et al., Use-dependent and error-based learning of motor behaviors, J. Neurosci. 30 (2010) 5159-66.

[31] S.C. Yen, B.D. Schmit, M. Wu, Using swing resistance and assistance to improve gait symmetry in individuals post-stroke, Hum. Mov. Sci. 42 (2015) 212-24.

**Figure Captions:**

Figure 1. Time course of the experimental conditions and testing periods. BL = Baseline, EA = Early Adaptation, LA = Late Adaptation, ED = Early Deadaptation, LE = Early Deadaptation, OG = Overground group, TG = Treadmill group, CG = Control group.

Figure 2A. Group average step length symmetry across all testing periods and participant groups. 2B. Group average single limb support symmetry across all testing periods and participant groups. For 2A and 2B, \* = within group difference, † = between weighted groups difference. 2C. Group averages for gait speed across all testing periods and participant groups. \* = different from Baseline. Error bars for all represent ± 1SD. BL = Baseline, EA = Early Adaptation, LA = Late Adaptation, ED = Early Deadaptation, LE = Early Deadaptation, OG = Overground group, TG = Treadmill group, CG = Control group.