Longitudinal changes in transtibial amputee gait characteristics when negotiating a change in surface height during continuous gait

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1. Introduction

The negotiation of a change in surface height during ongoing gait, such as stepping onto or from a pavement when crossing a road, is an important activity of daily living (ADL) that individuals are required to perform regularly (Begg and Sparrow, 2000, Buckley et al., 2008, Buckley et al., 2011). When stepping down from a raised surface, the lead limb must control the downward momentum of the whole body centre of mass (COM) via eccentric muscle actions and conversely, when stepping up to a raised surface, it must perform positive work via concentric muscle actions, in order to raise the COM (Buckley et al., 2008, Buckley et al., 2011, van Dieen et al., 2007, van Dieen et al., 2008). In both scenarios, the lead limb must be able to safely support bodyweight whilst providing propulsion in the context of ongoing gait and avoiding contact with the step.

Although stepping gait may be executed by young able-bodied individuals without apparent difficulty (Barbieri et al., 2013, Begg and Sparrow, 2000, Buckley et al., 2011, van Dieen et al., 2007, van Dieen et al., 2008), it is more mechanically challenging compared to level gait (Nadeau, McFadyen, & Malouin, 2003). To the authors' knowledge, no data have been reported previously on the development of lower limb amputee (LLA) stepping gait. However, investigations into LLA function during challenging motor tasks similar to stepping gait, such as stair negotiation and obstacle crossing, have outlined specific biomechanical adaptations which may also be adopted during LLA stepping gait. For example, during stair descent, transtibial amputees (TTA) maintain the affected lead limb in an extended position in an attempt to reduce the demands on the knee extensor musculature, avoiding potential limb buckling, whilst during stair ascent intact trail limb ankle plantarflexion and knee extension during stance aids the elevation of the COM in preparation for affected limb stance (Aldridge, Sturdy, & Wilken, 2012, Alimusaj et al., 2009, Jones et al., 2006, Powers et al., 1997, Ramstrand and Nilsson, 2009, Schmalz, Blumentritt, & Marx, 2007, Vanicek et al., 2010, Winter and Sienko, 1988). When negotiating obstacles, recent TTAs also display an inter-limb asymmetry in joint

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kinematics and kinetics, preferring to lead with the intact limb shortly following discharge from rehabilitation (Barnett, Polman, & Vanicek, 2013).

With these results in mind, it could be suggested that stepping gait may also present recent TTAs with a challenging task given that movement strategies are still being established. Subsequently, this may increase the potential for falling and fall-related injury, which are worldwide major public health concerns. Lower limb amputees have been shown to fall more frequently than age-matched controls (Miller, Speechley, & Deathe, 2001), indicating that the impact of falls may be exacerbated in this population.

Previous research has documented significant long-term biomechanical adaptations in recent LLA level gait, obstacle negotiation, and balance activities, (Barnett et al., 2009, Barnett, Polman, & Vanicek, 2013, Barnett, Vanicek, & Polman, 2013, Jones, Bashford, & Bliokas, 2001, Vrieling et al., 2009). Understanding how TTAs develop strategies for the successful completion of ADLs following formal rehabilitation is important as it establishes an objective evidence base from which further potential therapeutic or prosthetic interventions can be designed. Specifically, recent TTAs are likely to continue adapting their stepping gait strategies following discharge from rehabilitation. Therefore, understanding how this process occurs longitudinally with a view to optimising targeted clinical interventions is pertinent given that physical function in recent TTAs has been linked to quality of life and fear of falling (Barnett, Vanicek, & Polman, 2013).

Therefore, the aim of the current study was to investigate biomechanical changes that occur when stepping onto and from a raised surface, in recent TTAs, during the six-month period following discharge from rehabilitation. Previous research has shown long-term adaptation to ADL during this time period following discharge from rehabilitation (Barnett, Polman, & Vanicek, 2013, Barnett, Vanicek, & Polman, 2013). It was predicted that walking speed would increase over time, reflecting an improvement in overall task performance. In addition, it was

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predicted that self-selected lead limb preference (LLP) would change over time reflecting changes in participants' preferred movement strategies, as previously reported in obstacle crossing (Barnett, Polman, & Vanicek, 2013). Finally, it was predicted that improvements in task completion and changes to LLP would be underpinned by increased intact limb joint mobility (peak joint angles and ranges of motion) and power bursts (peak joint powers), as seen previously during obstacle crossing (Barnett, Polman, & Vanicek, 2013).

2. Methods

2.1 Participants

Having completed rehabilitation within a national healthcare physiotherapy department, A consecutive sample of unilateral TTAs were recruited and gave informed consent to participate in the current study. Participants were excluded if they experienced pain or discomfort whilst using their prostheses, had any current musculoskeletal injuries or cognitive deficits. Participants were included if they were at least 18 years of age, were able to use their prosthesis to complete a number of functional tasks without the use of a walking aid, including walking a distance of five metres and stepping onto/from a pavement. The study was approved by a local national healthcare service research ethics committee (08/H1304/10).

2.2 Experimental Set-up

In order to assess the biomechanical adaptations in stepping gait, a custom raised-surface walkway (5m length, 1.5m width) was constructed with a step height that replicated a standard roadside kerb (7.5cm) and placed within a 10m walkway (Buckley et al., 2005b, Buckley, Jones, & Johnson, 2010, Jones et al., 2005, Jones et al., 2006) (Figure 1). A ten-camera motion capture system (Qualisys, Gothenburg, SE) and two force platforms (Kistler, Model No: 9281B, Kistler, Winterthur, CH) sampled synchronous kinematic (100Hz) and ground reaction force (GRF) (1000Hz) data via Qualisys Track Manager software v2.8 (Qualisys, Gothenburg, SE).

2.3 Experimental Design and Protocol

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A longitudinal repeated measures design was employed with participants attending standardised data collection sessions at one, three and six months following discharge from their rehabilitation programme. Participants wore their own comfortable, flat footwear and were able to fit and re-adjust their own prostheses prior to data collection. Segmental six degree-of-freedom kinematics of the lower limbs were recorded by attaching reflective markers (14mm) bilaterally to the posterior aspect of calcaneus, dorsum of the 2nd metatarsal, medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanter, superior aspect of iliac crest, anterior-superior iliac spines, posterior-superior iliac spines in accordance with the six degrees-of-freedom marker set (Buczek et al., 2010, Cappozzo et al., 1995, Collins et al., 2009). Four-marker rigid clusters were securely attached to the thigh and shank segments. Marker placement on the affected limb was estimated from anatomical landmarks on the intact limb (Barnett et al., 2009, Powers, Rao, & Perry, 1998). A static calibration was performed by collecting kinematic data of each participant standing in the anatomical neutral position. Following several practice trials to ascertain a self-selected starting position, participants walked towards and stepped onto the walkway, continued to walk, turned 180° and then returned along the walkway before stepping off, at a self-selected pace. This allowed for the capture of continuous gait while stepping onto and from a new level with a minimum of five and a maximum of ten trials being recorded for each task across multiple time periods.

****Insert Figure 1 here****

2. 4 Data Analysis

Raw kinematic and GRF data were exported to Visual 3D (C-Motion, Inc, Germantown, USA), interpolated using a cubic spline algorithm and filtered using a fourth-order low pass Butterworth filter with cut-off frequencies of 6Hz and 30Hz respectively. Medial and lateral landmarks defined anatomical frames from which segment co-ordinate systems were defined following the right hand rule (Cappozzo et al., 1995). An XYZ Cardan sequence was used to

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define the order of rotations to calculate joint kinematics. For stepping trials, data from the transition step, as participants stepped onto/from the raised-surface, were analysed (Figure 1). The lead limb was defined as the first limb to approach/lead from the elevated walkway; the contralateral limb was designated as the trail limb. Self-selected LLPs were noted during the performance of each stepping trial using the motion capture video playback and calculated as percentages for both the intact and affected limbs (Figure 1). Gait events were identified using GRF data in order to normalise data to the gait cycle as defined in Figure 1.

Walking speed ($m.s^{-1}$) and stance duration (% gait cycle) were calculated along with joint angle data for the ankle, knee and hip (º). Kinetic data were recorded following stepping for the lead limb and prior to stepping for the trail limb (Figure 1). Peak ground reaction forces in the vertical (Fz) and anterior-posterior (Fy) directions were normalised to body weight (BW). Normalised peak joint power (W/kg) data were calculated for the ankle, knee and hip joints using standard inverse dynamics procedures.

In addition to the reporting of standard gait biomechanics data, task specific variables were selected based upon their relevance to the role of a particular limb during stepping gait (Barnett, Polman, & Vanicek, 2013). Therefore, during stepping down, lead limb variables that related to the controlled lowering of the COM during stance (e.g. load rate, peak joint angles during loading response and knee power burst K1) and to trail limb support of body weight during lead limb swing (e.g. joint ranges of motion (ROM) during single limb support and peak knee and hip power bursts K1 and H2 during mid-stance) were analysed. Similarly, during stepping up, lead limb variables that related to the raising and progression of the COM (e.g. peak joint power generation bursts throughout stance phase, A2, K2 and H3) were selected whilst variables related to trail limb progression and clearance were analysed (e.g. peak knee and hip flexion during swing).

2.5 Statistical Analysis

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Group mean data were analysed using a linear mixed model, Limb (Affected, Intact) * Time (One, Three and Six Months) with repeated measures on the last factor allowing for analyses of changes in multiple gait variables (Brown and Prescott, 2006). Each feature of the design (Time and Limb) was modelled as a fixed effect with the appropriate covariance structure being selected according to the lowest value for Hurvich and Tsai's Criterion, indicating improved model fit (Bias Corrected Akaike Information Criteria). Underlying assumptions were checked using conventional graphical methods and were deemed plausible unless stated otherwise. In the instance of a significant result, post-hoc comparisons were conducted using a Sidak adjustment in IBM SPSS v19.0 (IBM, Portsmouth, UK). The alpha level of statistical significance was set at $p \leq 0.05$.

3. Results

Participant details are presented in Table 1.

****Insert Table 1 here****

3.1 Stepping Down Temporal-Spatial

Walking speed increased between one and six months post-discharge (p=0.04) with both an affected (36%) and intact (24%) LLP (Table 2). The affected LLP diminished between one month (90.8%) and six months (52.6%) post-discharge (Table 2). Intact trail limb stance duration was greater than affected trail limb stance duration (p=0.01) with trail limb stance durations decreasing between one and three (p=0.04) and one and six months (p=0.01) post-discharge (Table 2), although no significant interaction effect was present.

****Insert Table 2 here****

3.2 Stepping Down Joint Kinematics

Lead limb peak ankle plantarflexion (p=0.01) and peak knee flexion (p=0.01) during loading response were greater with an intact LLP compared to an affected LLP (Figure 2). Ankle ROM

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during stance (p<0.01) and knee ROM during single limb support (p=0.05) were both greater with an intact trail limb compared to an affected trail limb (Figure 2).

3.3 Stepping Down GRF and Joint Kinetics

During early stance, intact limb load rate (p=0.02), initial peak vertical GRF (Fz1) (p=0.05) and peak posterior GRF (Fy1) (p<0.01) were significantly higher compared to the affected limb (Figure 2). A significant increase in lead limb peak anterior GRF (Fy2) (p=0.02) was observed between one and six months post-discharge (Figure 2). A significant interaction effect was reported for trail limb peak posterior (Fy1) GRF (p=0.01) as this was generally greater in the intact limb (Figure 2). Peak anterior GRF (Fy2) (p=0.01) was significantly greater with an intact trail limb compared to an affected trail limb (Figure 2).

****Insert Figure 2 here****

Peak lead limb knee power absorption during swing (K4) was greater in the intact vs. affected limb (p=0.01) (Figure 2). Peak ankle power absorption (A1) (p=0.01) and generation (A2) (p=0.04) and peak knee power generation during stance (K2) (p=0.05) were increased with an intact trail limb compared an affected trail limb (Figure 2). Peak knee power absorption during swing (K4) reduced over time with an affected trail limb with variable changes in the intact trail limb, resulting in a significant interaction effect (p=0.03) (Figure 2). Peak power absorption during stance (H2) increased significantly between one and three months post-discharge (p=0.04). A significant time main effect was also reported for peak hip power absorption in pre-swing H3 (p=0.05), although post-hoc analysis did not reveal the time points between which the significant increases occurred.

3.4 Stepping Up Temporal-Spatial

Walking speed was comparable at six months post-discharge irrespective of LLP (Table 2). The predominately intact LLP at one month post-discharge (70.0%) decreased at six months post-

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discharge (54.6%) (Table 2). Intact limb stance duration was significantly greater when acting as both the lead ($p=0.02$) and trail limb ($p=0.05$) (Table 2).

3.5 Stepping Up Joint Kinematics

Lead limb ankle ROM during stance ($p=0.02$) and peak knee flexion during loading response ($p<0.01$) were significantly greater with an intact LLP compared to an affected LLP (Figure 3). Peak plantarflexion during swing was greater when trailing with the intact limb compared to the affected limb ($p=0.01$).

****Insert Figure 3 here****

3.6 Stepping Up GRF and Joint Kinetics

Intact lead limb peak posterior GRF (Fy1) was significantly greater when compared to the affected limb ($p=0.01$) (Figure 3). Both load rate and peak posterior GRF (Fy1) were greater with an intact trail limb vs. and affected trail limb at one month post-discharge and converged six months post-discharge, resulting in significant interaction effects ($p=0.03$ and $p=0.05$, respectively) (Figure 3).

Peak ankle power generation (A2) ($p=0.02$), peak knee power generation during stance (K2) ($p<0.01$) and peak knee power absorption during swing (K4) ($p<0.01$) were significantly greater with an intact LLP compared to an affected LLP (Figure 3). Peak knee power absorption during late stance (K3) increased over time and was greater with an intact LLP resulting in a significant interaction effect ($p=0.01$) (Figure 3).

Peak ankle power generation (A2) ($p=0.02$), peak knee power absorption during loading response (K1) ($p=0.05$) and peak knee power generation during stance (K2) ($p<0.01$) were greater with an intact vs. affected trail limb (Figure 3). An initial increase followed by a subsequent decrease in peak knee power absorption during late stance (K3) resulted in a significant time main effect between three and six months post-discharge ($p=0.02$) (Figure 3).

4. Discussion

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The current study investigated biomechanical changes that occur when stepping onto and from a raised surface, in recent TTAs during the six-month period following discharge from rehabilitation.

4.1 Stepping Down

As predicted, there was an overall improvement in task performance as represented by a significant increase in walking speed. Participants initially preferred to lead with the affected limb, although at six-months post-discharge, this LLP had all but ceased.

As indicated previously, research has sought to explain LLA stair descent ability by describing the function of the affected limb (Aldridge, Sturdy, & Wilken, 2012, Alimusaj et al., 2009, Jones et al., 2006, Schmalz, Blumentritt, & Marx, 2007, van Dieen et al., 2007). However, the results from the current study suggest that the initial affected LLP was based upon participants' preference to exploit the capacity of the intact trail limb during stance.

Participants had greater stance duration, displayed greater ankle and knee mobility and ankle, knee and hip power absorption bursts during intact vs. affected trail limb stance. These results indicated that participants initially preferred to exploit the capabilities of the intact limb to safely control the lowering of the whole body COM during trail limb stance and potentially an initial cautionary approach to stepping down, which has been reported in perturbed stepping down in older adults (Buckley et al., 2005a, Buckley et al., 2005b).

Another factor that may have contributed to the initial affected LLP was the observation of a greater propulsive mechanism in the intact trail limb, reflected by higher ankle and knee power generation bursts (A2, K2 and K3) and propulsive GRFs (Fy2) in stance when compared to the affected limb. These results suggested that participants preferred to propel the intact limb forwards, while in single limb support on a relatively 'rigid' affected lead limb. These results are unsurprising given that for many TTAs, it is reasonable to assume that intact limb function is more readily utilised thus likely to adopt a more dominant role (Barnett, Polman, & Vanicek,

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2013). In addition, the current participant group were encouraged to lead with their 'weaker' limbs when descending stairs and steps during rehabilitation, which is likely to have influenced this LLP at one month post-discharge.

However, the reduction of the affected LLP at six months post-discharge reflected the underlying shift in the strategies used by participants during stepping down gait which occurred alongside improvements in overall task performance, characterised by increased walking speed. Results suggested that adaptations did occur in affected trail limb function resulting in an improved controlled lowering mechanism and, although these adaptations did not result in repeatedly significant interaction effects, this may have reflected participants' increased confidence in utilising this strategy. In addition, results from the current study suggested that task performance at six months post-discharge was also underpinned by the increased exploitation of intact limb vs. affected limb capacity, which had not changed significantly over time. The lack of dorsiflexion possible in the trail limb prosthetic ankle joint during single limb support is likely to have necessitated the increased lead limb intact ankle plantarflexion in late swing, as has been reported previously in LLA stair descent (Alimusaj et al., 2009, Schmalz, Blumentritt, & Marx, 2007). This mechanism would have allowed participants to probe the ground before 'falling' onto the intact lead limb in weight acceptance (Buckley et al., 2008, Schmalz, Blumentritt, & Marx, 2007). In addition, it could be suggested foot contact occurs earlier and more energy is absorbed by the lead limb when utilising a toe first contact when stepping down onto the intact limb, compared to a heel first contact, with a dorsiflexed prosthetic ankle, with the affected limb (van Dieen et al., 2008). Furthermore, increased intact lead limb loading during touchdown, as reflected by GRF data, and greater observable but not statistically significant peak joint powers bursts compared to the affected limb, suggested that the intact lead limb knee extensor and ankle plantarflexor musculature were more capable of

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lowering the body in a controlled fashion, corroborating the mechanisms underpinning an intact LLP.

4.2 Stepping Up

Overall task performance, as indicated by walking speed, was consistent over time with an affected LLP and, although improvements in task performance over time were noted with an intact LLP, these effects were not statistically significant.

Initially, participants utilised an intact LLP strategy. However, while stance duration did not change over time, it was greater in intact limb compared to the affected limb, regardless of role (lead or trail limb) which may have reflected a reluctance to transfer weight onto the affected limb (Powers et al., 1997). In the current study, an explanation for the initial intact LLP were related to the observations of greater intact limb ankle and knee joint mobility demands and power bursts during stance, as reflected by ankle and knee joint kinematic and peak joint power burst data, respectively. During stance, participants preferred to exploit the capacity of the intact lead limb in order to manage weight acceptance following foot contact and then do positive work in order to raise the COM and maintain progression in preparation for swing. Thus, as predicted, the higher utilisation of intact limb capacity initially led to its preferential use as the lead limb one month following discharge. It must also be stated that, conversely to stepping down gait, participants were encouraged to utilise an intact LLP during rehabilitation when stepping up stairs and steps. Therefore, it is probable that this effect persisted into the timeframe of the current study.

A shift from an initial intact LLP to more balanced LLP strategies at six months post-discharge occurred in stepping up gait, with comparable walking velocities observed throughout. This suggested that participants were more flexible in their strategy selection when performing the task. Participants spent more time in intact trail limb stance with an affected LLP and during this period, the intact limb experienced greater loading, as reflected by increased GRFs. In

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addition, increased peak joint power generation and absorption bursts were associated with the intact limb indicating that it aided the control of whole body momentum in preparation for stepping up during early stance with continued progression prior to swing. These results corroborated previous research highlighting the role of the intact trail limb in the elevation of the COM in more experienced LLAs (Schmalz, Blumentritt, & Marx, 2007). Seemingly, the participants in the current study who adapted to using the affected limb as their lead limb, increased their flexibility of strategy selection. While these individuals may have been better equipped to deal with unpredictable configurations of the physical environment, these adaptations in strategy selection occurred despite a persistent disparity between the capacity of the intact vs affected limbs.

Summary

To the authors' knowledge, the current longitudinal study is the first to investigate the biomechanical changes present in the stepping gait of recent TTAs. Following discharge from rehabilitation, participants' overall performance of stepping down from and stepping up to a raised surface displayed trends towards improvement. Moreover, participants' willingness to deviate from an initial preferred strategy could be interpreted as a positive increase in plasticity when completing this motor task. Participants preference to exploit intact limb function may be beneficial initially, although potential problems may arise in the future when a situation does not allow for the self-selection of a particular LLP and thus, necessitates a strategy requiring increased utilisation of affected limb function. An example of such a situation would be the presentation of an unexpected change in surface height where it could be assumed TTA stepping performance would be reduced or even become hazardous given that TTAs have been shown to perform worse under increasing time pressure during an obstacle avoidance task (Hofstad et al., 2006). Therefore, it is important that TTAs are adaptable in terms of LLP selection and do so according to the task requirements rather than a preference to utilise the capacity of a

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particular limb. Results from the current study have implications for TTAs rehabilitation as they suggest further functional utilisation of the affected limb is required, as the disparity in utilisation was evident at one month and persisted at six-months post-discharge. Interventions aimed at encouraging the use and exploration of different strategies in safe, controlled but challenging environments may address this disparity. In addition, interventions targeting the eccentric lowering mechanism and concentric raising mechanism of the knee extensors within the affected limb would benefit stepping down and stepping up gait respectively, particularly in the early stages following discharge. Such training may in turn reduce TTAs falls risk by increasing adaptability when performing stepping gait. It is possible that these changes may be achieved through affected limb resistance and flexibility training aimed at improving knee extensor strength and joint mobility. Also, the prescription of advanced prosthetic components and improved prosthetic design aimed at increasing ankle mobility may also aid TTAs functional performance, thus investigation into the effects of these interventions are warranted.

Limitations

Although the results from the current study were obtained over a six-month period in recent TTAs, it is not possible to elucidate what the long-term health effects are arising from the apparent adaptations in stepping performance. Research has shown that asymmetries in LLA mechanics may be linked to bone health, although further causal relationships must be established (Sherk, Bemben, & Bemben, 2008). Given the small sample size of this study, variation in participants' cause of amputation may have limited statistical power. The assessment of one step height representing a street kerb may not have induced the biomechanical adaptations associated with a more challenging step height. Finally, variation in prosthetic componentry may have increased the variation in some biomechanical variables reported.

5. Conclusion

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Following discharge from rehabilitation, trends towards improvement in task performance occur in stepping gait. Although LLPs changed over time, reflecting an increased flexibility in strategy selection, TTAs continued to exploit intact limb function to a greater extent when compared to the affected limb, regardless of the role being performed. The novel data presented provide an objective basis on which an understanding of how TTAs learn to perform this important ADL can be structured, thus informing future therapeutic and prosthetic interventions.

6. Declaration of Interest

The authors report no conflicts of interest. The authors alone are responsible for the content and writing of the paper.

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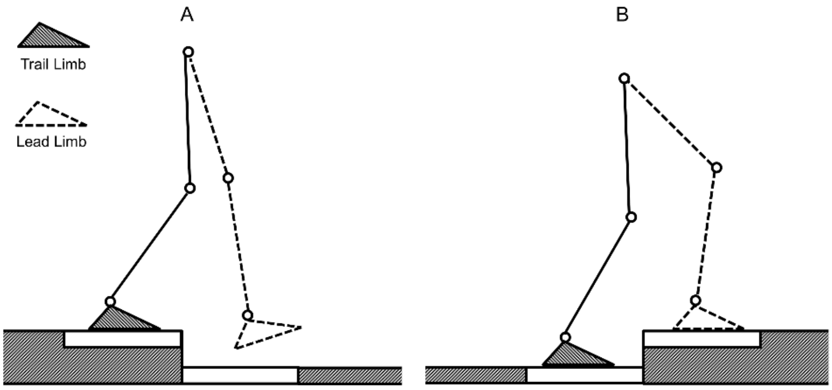

Figure 1. Schematic diagram depicting stepping down (A) and stepping up (B) during ongoing gait with force platform locations (white blocks) indicated. The lead limb is defined as the first limb to approach the ledge of the elevated walkway. For stepping gait trials, the lead limb gait cycle was defined from toe-off to subsequent toe-off, with the trail limb trials' gait cycles being defined from foot contact to subsequent foot contact.

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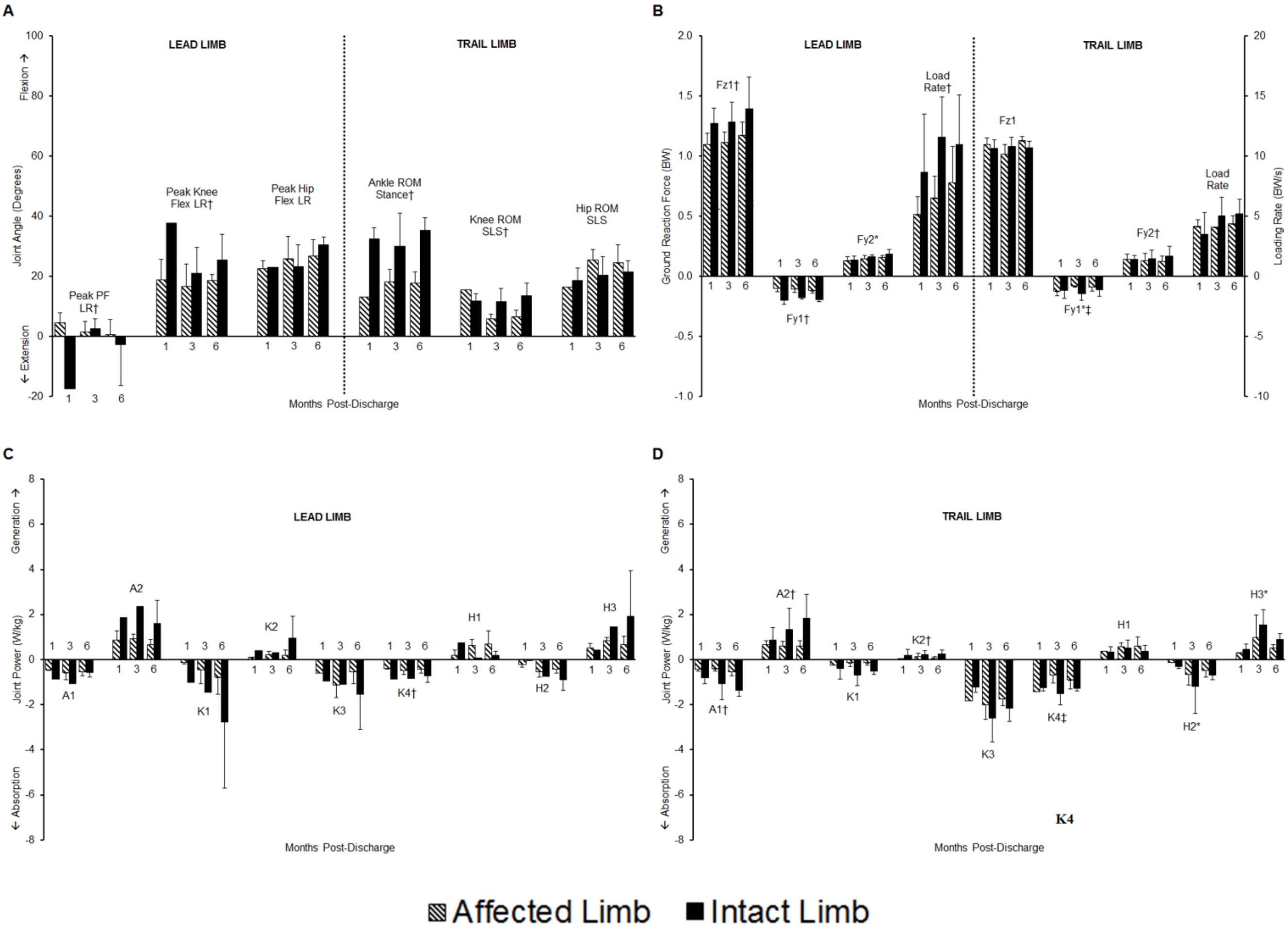

Figure 2. Group mean (SD) joint kinematics (A), loading rates and peak ground reaction forces (B) and lead limb (C) and trail limb (D) peak joint powers during stepping down. Symbols denote significant □time, †limb and ‡interaction effects ($p \leq 0.05$). Peak joint power burst definitions are as follows: Ankle power absorption during stance (A1); Ankle power generation during pre-swing (A2); Knee power absorption during loading response (K1); Knee power generation during mid-stance (K2); Knee power absorption during pre-swing (K3); Knee power absorption during terminal swing (K4); Hip Power generation during loading response (H1); Hip power absorption during stance (H2); Hip power generation during preswing (H3).

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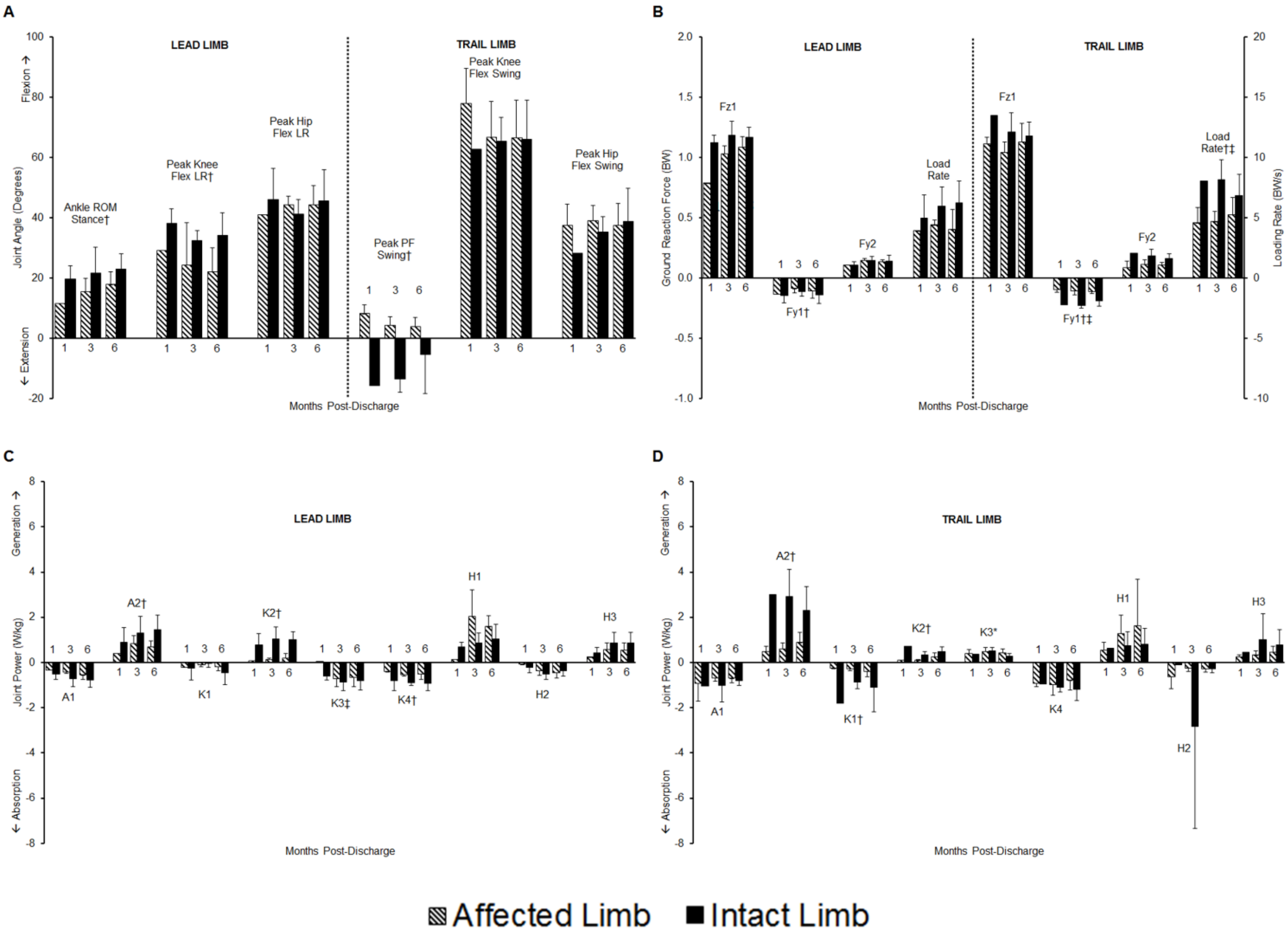

Figure 3. Group mean (SD) joint kinematics (A), loading rates and peak ground reaction forces (B) and lead limb (C) and trail limb (D) peak joint powers during stepping up. Symbols denote significant * time, †limb and ‡interaction effects ($p \leq 0.05$). Peak joint power burst definitions are as follows: Ankle power absorption during stance (A1); Ankle power generation during pre-swing (A2); Knee power absorption during loading response (K1); Knee power generation during mid-stance (K2); Knee power absorption during pre-swing (K3); Knee power absorption during terminal swing (K4); Hip Power generation during loading response (H1); Hip power absorption during stance (H2); Hip power generation during pre-swing (H3).

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