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Postural and movement adaptations by individuals with a unilateral below-knee amputation during gait initiation

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Abstract

The present study examined the compensatory strategies adopted by individuals with a unilateral below-knee amputation (BKA) during gait initiation. Eleven individuals with a unilateral BKA and 11 able-bodied subjects initiated gait at three step length conditions (+0, +25 and +50% of preferred step length). A lead-limb condition was also introduced, such that all participants were required to initiate gait with both their left and right limbs. For all step length and lead-limb conditions, it was found that individuals with a unilateral BKA required more time to initiate gait, as compared with the able-bodied. This increase in movement duration was attributed to the stability and movement limitations of the prosthetic limb. On the other hand, by prolonging the task duration, these individuals were also able to employ a 'horizontal impulse' strategy, whereby they could create a similar magnitude of horizontal impulse as the able-bodied without the need to apply a large magnitude of peak antero–posterior (A–P) force.

Keywords: Gait initiation; Below-knee amputee; Postural adjustment

1. Introduction

Each year, approximately 135 000 North Americans experience the loss of one or more limbs due to trauma and/or disease [1]. The majority (53%) of these cases are classified as a trans-tibial or below-knee amputation (BKA), where there is a loss of one or both legs below the level of the knee [2]. In the case of a unilateral BKA, the individual becomes structurally asymmetrical, as there is an altered sensation and a loss of musculature on the amputated side. Together, these changes present a difficult challenge in being able to continue with daily activities. Thus, understanding the adaptations that occur due to the loss of a lower limb is an important aspect in devising a successful rehabilitation program.

One common activity is upright walking, a task which requires the alternating and balanced motion of the lower limbs. It consists of three major components: gait initiation, the transition from quiet stance to steady state; steady state walking and gait termination, the period from steady state to upright stance. Of particular interest to this paper is the phase of gait initiation. Similar to continuous walking, gait initiation requires an individual to achieve forward movement while at the same time, maintain body stability. Whereas these responsibilities are shared equally and alternatively between the two limbs during steady state gait, this does not occur during gait initiation. Rather, each limb undergoes a distinct set of commands depending upon its precise role during the course of the movement.

Prior to the initiation of gait, an individual will be standing upright with both feet placed on the ground. Once the decision to initiate gait is made, postural adjustments are immediately completed in order to facilitate the upcoming stepping motion. One limb, termed the swing or lead limb, exerts a lateral force such that body's center of mass is shifted towards the contralateral limb [3]. This reduces the lead limb's need to maintain body support, so that it can focus on generating a portion (typically a peak posterior force of 7% body weight) of the initial forward thrust [4]. Concurrently, the role of the non-stepping limb, termed

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the trail or stance limb, becomes one of body stability, as it accepts the load, which was previously borne by the lead leg. Although stability and balance are the primary concerns for the trailing limb, a small burst of posterior force (a peak of 14% body weight) is applied in order to augment the forward thrust that was produced from the leading limb [4]. The distinct responsibilities for each leg remain unchanged until the initial step has been completed.

After approximately 0.54 s from the onset of gait initiation, the heel of the leading limb returns towards stance phase [4]. It is at this time that the duties of the lead and trail limbs reverse. The trail limb, initially relied upon for body support and to a lesser extent forward movement, begins to apply a greater amount of horizontal force (a peak of 22% body weight) in the posterior direction [4]. This is needed in order to achieve the required movement for the second step. The lead leg, originally responsible for creating the initial forward motion, now becomes responsible for absorbing much of the shock generated from the forthcoming heel-strike. Body support also becomes a major concern, particularly when the trailing limb undergoes its own swing phase. Once the trailing limb completes its initial step, steady state gait will be achieved [4,5].

While the unequal sharing of responsibilities associated with gait initiation may not be a concern for those who have structurally and functionally similar limbs, conflicts can certainly arise for persons with a unilateral BKA. One issue relates to the overall stability of the prosthetic limb as these individuals are known to exhibit a greater amount of body sway during upright stance [6–8]. Although they may be able to counteract this instability by placing a greater proportion of their body weight on their non-amputated limb during upright stance [9,10], it is not entirely known how these individuals compensate during gait initiation, when the duration of single limb support depends upon the choice of the leading limb.

Another concern for individuals who have experienced a unilateral BKA is the ability to generate adequate amounts of forward propulsion. Much of the forward momentum needed for walking is provided by the ankle plantarflexors [11], which are now lacking in the amputated limb. Since the prosthetic ankle is unable to generate a comparable amount of propulsion as a biological leg, compensations measured during steady state gait, such as an increased contribution from the hip joint of the prosthetic limb [12–14], as well as the hip and knee joints of the non-amputated limb [12,15], may be needed in order to successfully initiate gait. How these, or any other adaptations can fully restore the functions of a physiological ankle during gait initiation remain to be seen.

Since the problems of stability and forward movement will occur regardless of the lead-limb condition, it may be that the best strategy for gait initiation is to simply minimize these areas of concern. Factors, such as duration and overall importance of each particular limitation, may also need to be considered in order to maximize the ability to walk forward. Yet, despite these potential conflicts, little research has been conducted in this area. The first study, conducted by Nissan [16], found several kinematic and kinetic differences between the amputee and control groups, but little explanation was given as to why these adaptations had occurred. Further, the study failed to examine many of the postural components, such as the movement of the center of pressure (COP), that are strongly related to gait initiation.

The most recent experiment, conducted by Rossi et al. [17], found that the movement of the COP, an indicator of postural control, was similar for both the amputee and control groups. However, this is quite unlikely to occur since the path of the COP during gait initiation is highly dependent upon the muscle activity of the lower limbs. Two of these muscles, the tibialis anterior and gastrocnemius, are absent after a BKA and thus, these individuals would certainly need to employ a different strategy in order to appropriately move forward. This choice in strategy will further be related to the responsibilities and abilities of each limb, as gait initiation requires both stability and movement.

The present study was conducted to further distinguish the postural and movement adaptations made by persons with a unilateral BKA during gait initiation. It was hypothesized that temporal, kinematic and kinetic compensations would be undertaken in order to compensate for the limitations of the prosthetic limb. To further differentiate between the two groups, three step length requirements were introduced, where the distance of the initial step was experimentally controlled. It was hypothesized that a step length effect would be found, as the longer step length conditions would represent a greater postural and movement challenge.

2. Methods

2.1. Subjects

Two groups of subjects were recruited. The amputee group comprised of 11 individuals who had undergone a unilateral BKA (eight males, three females; 44.1 ± 14.1 years of age; Table 1). Once the amputee group was formed, 11 age- and gender-matched individuals (eight males, three females; 44.9 ± 12.3 years of age) with no history of neurological or orthopaedic impairments were recruited to form the control group (Table 1). All participants read and signed an informed consent form revealing all details of the experimental protocol, which

Table 1
Characteristics of A, the 11 individuals with a unilateral BKA and B, the control subjects

Below-knee amputees	Age (years)	Sex	Mass (kg)	Amputated limb	Reason for amputation	Foot type	Time since amputation
A							
1	24	Male	90	R	Trauma	Flex	4
2	50	Male	92	L	Vascular	Seattle	1
3	43	Male	105	R	Trauma	Flex	3
4	31	Female	69	R	Trauma	Flex	3
5	67	Male	126	R	Vascular	Seattle	7
6	48	Male	71	L	Trauma	Seattle	11
7	55	Male	101	R	Trauma	Seattle	39
8	65	Male	63	R	Trauma	Seattle	7
9	30	Female	55	L	Trauma	Seattle	4
10	36	Male	84	R	Trauma	C-Foot	13
11	41	Female	54	L	Cancer	Flex	18
Average (±S.D.)	44.1±14.1	Eight males; three females	82.7±22.5	Seven right leg; four left leg	Eight trauma; two vascular; two cancer	Six Seattle; four Flex; one C-foot	1 0.0±10.8
Non Amputees	Age (years)	Sex	Mass (kg)				
B							
1	59	Male	65				
2	41	Male	57				
3	24	Male	60				
4	52	Male	69				
5	53	Female	46				
6	39	Male	64				
7	60	Male	62				
8	37	Female	46				
9	59	Female	59				
10	38	Male	75				
11	32	Male	69				
Average (±S.D.)	44.9±12.3	Eight males; three females	61.1±9.0				

L, left; R, right.

had been approved by the University of British Columbia Ethical Review Committee.

2.2. Protocol

Each individual underwent an initial 'pre-test' assessment, which required the subject to walk, from a standing upright position, along a short walkway for a total of six strides. No instruction was given as to which limb should begin the walking process. During these five trials, the primary measurement of interest was the mean initial step length (i.e. the distance from the starting position to the heel strike of the first step), as this was used to scale the step length requirements for the experiment to each individual. Additional measurements included the preferred limb, defined as the limb that initiated gait, and the preferred stride length, defined as the distance from the heel strike of one foot to the following heel strike of the same foot.

After this initial assessment, reflective markers for the heel and fifth metatarsal were placed onto the subject's two shoes. Landmarks for the prosthetic foot were estimated from the marker positions of the non-amputated limb. Once the markers were in place, participants were brought into the measurement area and were then asked to stand with each foot on a separate force platform (Bertec model 4060) (Fig. 1). The specific angle and placement of the feet were set to the individual's preference and was established as the 'start' position. This position remained constant throughout the entire experiment by tracing the individual's footprints onto a piece of paper placed overtop the two platforms.

For each experimental trial, participants stood upright at the start position and were instructed to walk along a short (3 m) mounted walkway once the visual cue had been presented. Not only did this visual cue, a pair of lights located at the end of the walkway, indicate when to start, it also signified the stepping limb. If the light on the left side of the walkway was to light up, the subject was required to initially step with the left leg, while a right lead-limb condition was denoted by the illumination of the light on the right side. The order of the lead-limb requirement was randomly determined prior to data collection.

In addition to the lead-limb requirement, all of the trials were completed such that the first step landed on a third force platform (Kistler model 9261A) that was located in front of the participant. This platform was moved forward or backward, so that for each trial, it was at a distance of +0, +25 or +50% of the individual's preferred step length. The distances for each of the three step length conditions were determined from the initial 'pre-test' measurement. Similar to the order of the swing leg, the order of step length was presented randomly.

No practice trials were given and subjects were informed to focus their attention towards the end of the walkway rather than down towards the ground. They were also reminded that speed was not an important aspect for this study and that anticipation was strongly discouraged. In cases when the subject failed to land properly on the Kistler platform (e.g. the entire foot did not contact the platform), the trial was repeated.

In total, each individual participated in 42 trials. Fourteen trials were completed at each of the three step length conditions, with half the trials from each step length condition initiating with the left leg and the other

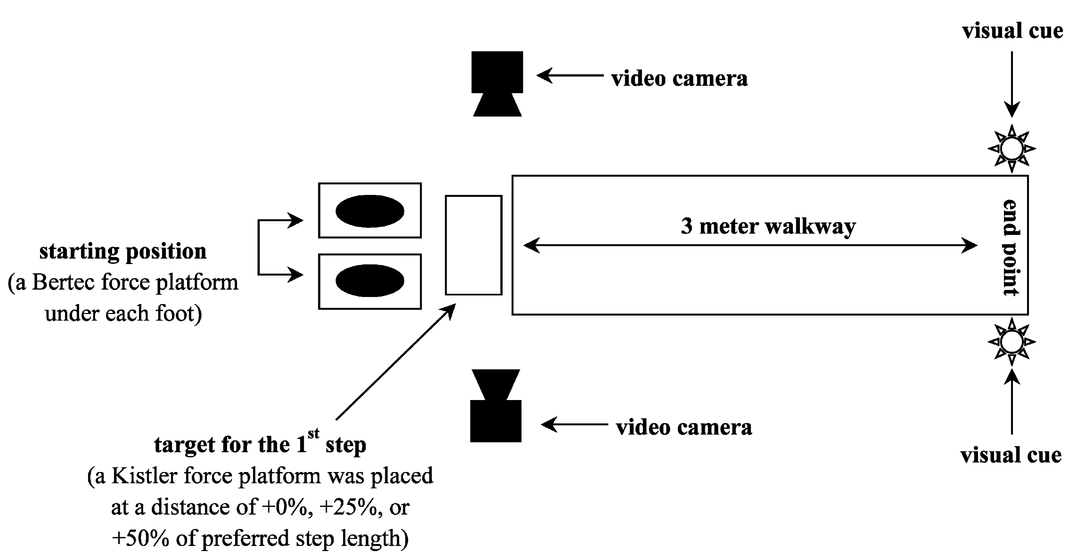

Fig. 1. Diagram of the lab set-up. Participants stood with each foot on a separate force platform. When one of the lights (i.e. the visual cue) at the end of the walkway lit up, they were required to start walking towards the end of the walkway. The first step had to land on the Kistler force platform. Two video cameras recorded the movements of the individual during the entire trial.

half initiating with the right. The trials were presented in a blocked (step length), random (lead-limb) fashion.

2.3. Data processing

During each of the 42 gait initiation trials, kinematic measurements along the walkway were obtained using two video cameras placed 4 m from the starting position, and perpendicular from the line of action. One camera recorded the movements of the right limb, the other tracked the left leg. Both cameras were sampled at a rate of 60 Hz and were synchronized in time by flashing a reference light in each camera's field of view. The video clips from each trial were then digitized into two-dimensional co-ordinate data using the Peak Motus (version 5.1.8) video analysis system. Raw co-ordinate data were filtered using a 4th order, dual pass Butterworth filter with a cut-off frequency of 5 Hz.

In addition to the kinematic data, the three force platforms collected ground reaction force and COP data. These were analog-to-digital converted (Data Translation Model 3010) using a PC compatible computer at a sampling frequency of 600 Hz. For the purposes of plotting, all of the collected data were temporally normalized to 100% of the gait initiation cycle, where 0% represented the start of sway and 100% denoted the second toe-off of the leading leg. Similar to the work by Maki and McIlroy [3], the start of sway was defined as the time at which there was a 4 mm shift from the baseline COP for a duration of at least 10 ms. Normalized data were then averaged for each lead-limb condition at each of the three stride lengths. Group averages were calculated from all participants within each of the two groups.

For the COP data that were obtained from the two Bertec force platforms, additional processing steps were undertaken. The position of the COP was then normalized to 100% foot length and 100% stance width, as foot size and placement were not controlled between subjects. Second, to quantitatively analyze the excursion of the COP, an inflection point was calculated. This was determined as the point in which the COP stopped increasing in the posterior direction for a period of at least 0.05 s.

For the calculation of the horizontal impulses, the anterior–posterior force–time curves were integrated using the Trapezoid Approximation method. From the starting position, three impulse values were calculated: one for the leading limb ($\int_{\mathrm{sway}}^{\mathrm{TO1}} Fx_{\mathrm{lead}}\,\mathrm{d}t$), another for the trailing limb ($\int_{\mathrm{sway}}^{\mathrm{TO2}} Fx_{\mathrm{trail}}\,\mathrm{d}t$), as well as the combined impulse from the leading and trailing limbs ($\Sigma(\int_{\mathrm{sway}}^{\mathrm{TO1}} Fx_{\mathrm{lead}}\,\mathrm{d}t + \Sigma + \int_{\mathrm{sway}}^{\mathrm{TO2}} Fx_{\mathrm{trial}}\,\mathrm{d}t)$. The starting frame for both the leading and trailing limbs was the start of sway. The end frame for the leading limb was the initial toe-off (TO1) while the first toe-off of the trailing limb (TO2) was used as the last frame for the trailing limb.

To calculate the total impulse for a given trial, the sum of the leading and trailing impulses was determined. For example, during a left lead-limb trial, the sum of impulses would include the impulse generated by the left lead-limb as well as the impulse generated by the right trail limb. This summed value was used to represent the total change in momentum for each individual and consequently, total change in displacement for the four (i.e. the left, right, intact and prosthetic) lead-limb conditions. Since both test groups had similar step lengths relative to their height (i.e. a step length difference of 8 cm and a leg length difference of 9 cm), it was hypothesized that the two groups would generate similar totals of horizontal impulse. Furthermore, trials at the longer step requirements were hypothesized to show greater amounts, as larger impulses would be needed to create greater changes in body displacement.

Statistical analysis was conducted using a 2 (group) $\times$ 2 (limb condition) $\times$ 3 (stride length) mixed factorial design, with repeated measures on the last two factors. Alpha was set a priori at 0.05.

3. Results

3.1. Pre-test assessment

The results from the pre-test assessment indicated that the majority of the subjects preferred to lead with their right limb. For the able-bodied, ten of the 11 subjects preferred the right leg; for the amputees, nine of the 11 subjects preferred the right leg. It also appeared that the prosthetic limb was the preferred limb for the individuals in the amputee group, as eight of the 11 subjects preferred stepping with this leg, as opposed to their non-amputated (i.e. intact) limb.

In addition, it was found that the amputee group exhibited a longer preferred step length (0.74 ± 0.09 m for the BKAs vs. 0.66 ± 0.08 m for the able-bodied; $F_{1,20} = 4.41$, $P < 0.05$) and stride length (1.48 ± 0.18 m for the BKAs vs. 1.30 ± 0.14 m for the able-bodied; $F_{1,20} = 6.72$, $P < 0.02$) than their able-bodied counterparts. However, it is likely that these small differences were due to differences in leg length (0.92 ± 0.08 m for the amputees vs. 0.83 ± 0.04 m for the able-bodied).

3.2. Task duration

Individuals with a unilateral BKA required more time to initiate gait as compared with the able-bodied ($F_{1,20} = 40.96$, $P < 0.001$) (Fig. 2). The choice of lead-limb did not have a significant effect, as a leg $\times$ group interaction did not occur ($F_{1,20} = 0.67$, $P > 0.79$).

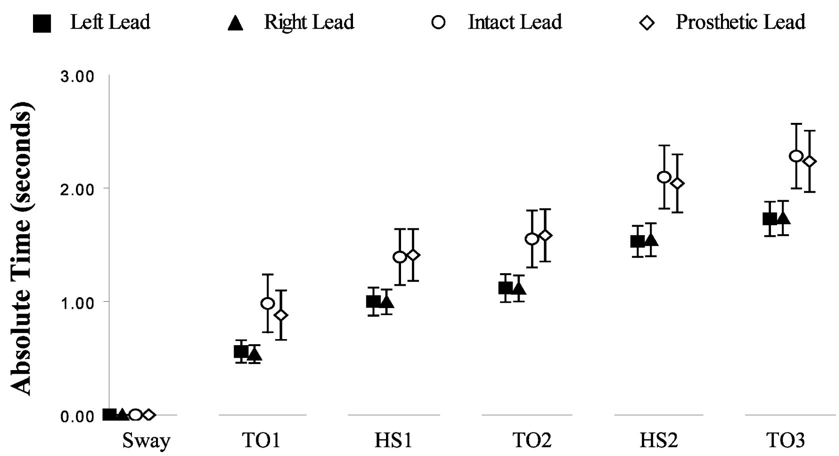

Fig. 2. Mean (±S.D.) time (in seconds) taken from sway to the third toe-off of the four lead-limb conditions during the +0% SL requirement. The data from the +25 and +50% SL conditions exhibited similar patterns.

Since the individuals with a unilateral BKA, regardless of the lead-limb condition, required more time to complete the gait initiation task, the entire movement was broken down into several components to determine whether the difference in overall duration could be explained by one or more of these events. As shown in Fig. 2, when the gait initiation task was separated into each heel-strike and toe-off event, it was found that the amputee group required more time to reach each heel-strike and toe-off event than their able-bodied counterparts. Although the temporal difference between the two groups became larger at each successive heel-strike and toe-event event, the majority of this difference was explained by the first toe-off of the leading limb. For example, during the preferred step length condition, 383 ms of additional time, or 72% of the overall difference, was needed by persons with a unilateral BKA in order to get the leading limb off of the ground and into its initial swing phase. This prolongation occurred when these individuals lead with either the intact or prosthetic limb, as there were no leg, or leg × group interaction effects.

3.3. Center of pressure

The displacement of the COP under each foot was measured from the starting position. It was found that for the able-bodied, the amount of posterior COP displacement was similar regardless of the foot's responsibility (Fig. 3 and Table 2A). On the other hand, a large degree of asymmetry was observed in the amputee group, where the posterior displacement of the COP under the intact limb was similar to the control limb conditions, while a smaller displacement was measured under the prosthetic foot ($F_{1,20} = 19.12$, $P < 0.001$ for leg main effect, $F_{1,20} = 17.60$, $P < 0.001$ for the leg × group interaction).

The net COP (i.e. the summed COP from both feet) was also measured, as this indicated the overall postural control for the individual. As shown in Fig. 4, the path of the COP was visually different between the control, intact and prosthetic lead-limb conditions. Unlike the able-bodied, who initiated gait symmetrically by moving in the typical 'J' shaped pattern, individuals with a unilateral BKA appeared to have two distinct strategies. During trials with a prosthetic lead-limb, the COP displacement looked similar to the control conditions, while an entirely different pattern emerged during trials with an intact lead-limb.

To quantitatively analyze this data, the location where the COP inflected from a M-L to an antero–posterior (A–P) direction was determined. This inflection point, as defined in the methods section, was found to be different between the leg ($F_{1,20} = 15.81$, $P = 0.001$) and group ($F_{1,20} = 17.83$, $P < 0.001$) conditions at each of the three step length requirements ($F_{2,40} = 9.62$, $P < 0.001$). Specifically, when an individual from the amputee group lead with their intact limb, the COP inflected at a more anterior position than the prosthetic, left or right lead-limb conditions. As the step length requirement increased, the inflection point occurred at a more anterior position for all leg conditions.

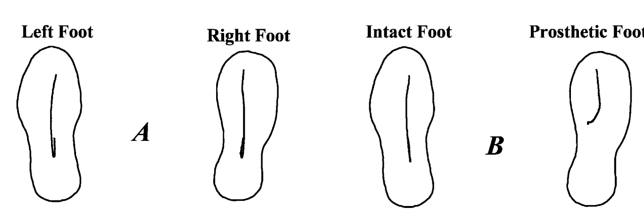

Fig. 3. Mean displacement of the COP under each foot during the +0% step length condition for A, the left lead-limb condition; and B, the intact lead-limb condition. The right lead-limb condition exhibited a similar pattern as the left lead-limb condition; the prosthetic lead-limb condition exhibited a similar pattern as the intact lead-limb condition.

Table 2
(A) The mean (±S.D.) minimum posterior displacement (expressed as percent foot length) of the COP under each foot during the +0% (SL1), +25% (SL2), and +50% (SL3) step length conditions; (B) the mean (±S.D.) peak posterior force (expressed in N per body weight) from the leading and trailing limbs during the +0% (SL1), +25% (SL2), and +50% (SL3) step length conditions; (C) mean (±S.D.) peak braking and propulsive force (expressed in N per body weight) from the landing and subsequent push-off of the initial step during the preferred step length condition; (D) peak (mean±S.D.) vertical forces (expressed in N per body weight) during weight acceptance, midstance and push-off from the landing and subsequent push-off of the initial step during the preferred step length condition

Leg	SL1	SL2	SL3
A			
When leading			
Non-amputee	−23.24 (5.81)	−24.24 (5.74)	−24.51 (6.46)
Intact	−23.74 (10.04)	−23.73 (9.76)	−25.67 (8.63)
Prosthetic	−1.41 (16.81)	−0.62 (15.76)	−2.39 (17.74)
When trailing			
Non-amputee	−24.90 (4.84)	−24.48 (4.35)	−24.51 (6.46)
Intact	−20.62 (9.31)	−22.08 (8.61)	−22.06 (8.62)
Prosthetic	−0.69 (17.01)	−4.23 (18.12)	1.01 (15.74)
B			
When leading			
Non-amputee	−0.07 (0.02)	−0.07 (0.03)	−0.08 (0.03)
Intact	−0.07 (0.03)	−0.07 (0.03)	−0.08 (0.03)
Prosthetic	−0.02 (0.01)	−0.02 (0.02)	−0.03 (0.02)
When trailing			
Non-amputee	−0.27 (0.04)	−0.30 (0.05)	−0.34 (0.06)
Intact	−0.23 (0.06)	−0.31 (0.08)	−0.34 (0.07)
Prosthetic	−0.20 (0.04)	−0.23 (0.04)	−0.25 (0.06)
Leg	Peak braking	Peak propulsion	
C			
Non-amputee	−0.16 (0.02)	0.22 (0.03)	
Intact	0.15 (0.03)	0.24 (0.06)	
Prosthetic	0.09 (0.03)	0.17 (0.03)	
Leg	Weight acceptance	Midstance	Push-off
D			
Non-amputee	1.08 (0.05)	0.80 (0.04)	1.08 (0.06)
Intact	1.17 (0.09)	0.75 (0.11)	1.13 (0.08)
Prosthetic	1.03 (0.04)	0.82 (0.07)	1.02 (0.06)

For all four tables, the 'non-amputee' condition corresponds to the average of the left and right legs from the able-bodied group; the 'intact' and 'prosthetic' leg conditions correspond to the amputee group.

3.4. Ground reaction forces

Several components of force application were measured during the gait initiation task. These include the A–P forces from the starting position, as well as the A–P and vertical forces from the end of the initial step. The vertical and medio–lateral ground reaction force components from the starting position are not described in this paper as they were found to be similar to previous studies [16,17].

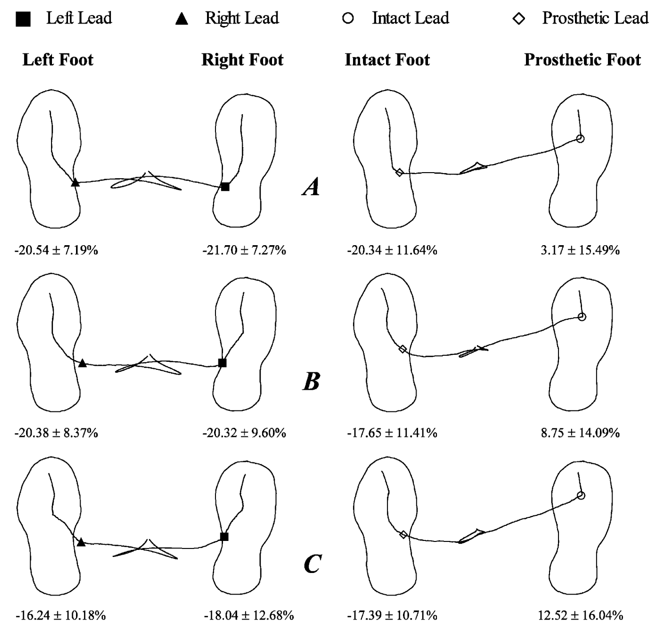

Fig. 4. Mean displacement and mean (±S.D.) inflection point of the COP during gait initiation for A, the +0% step length condition; B, the +25% step length condition; and C, the +50% step length condition.

First, differences were found in the peak force applied by the leading limb (Table 2B). Subjects from the amputee group generated a smaller peak A–P force ($F_{1,20} = 4.56$, $P < 0.05$) due to a smaller magnitude of force application with the prosthetic foot ($F_{1,20} = 26.73$, $P < 0.001$ for the leg main effect, $F_{1,20} = 13.60$, $P = 0.001$ for the leg × group interaction). As the step length requirement increased, a greater peak A–P force was applied by both groups of subjects ($F_{2,40} = 17.291$, $P < 0.001$ for the step main effect). The differences between the four limb conditions remained similar, as the step × leg × group interaction was insignificant.

Second, differences were also apparent when examining the magnitudes of peak A–P force that was applied by the trailing leg (Table 2B). Individuals with a unilateral BKA employed a significantly smaller peak magnitude of force ($F_{1,20} = 9.08$, $P = 0.007$) in the posterior direction than their able-bodied counterparts. A leg × group interaction did not occur, implying that the level of symmetry with the amputee group was similar compared with those in the control group. As the step length requirement became longer, the applied force significantly increased ($F_{2,40} = 65.59$, $P < 0.001$) by a similar amount for the two groups of subjects. The differences between the four limb conditions remained similar, as the three way interaction was insignificant.

The third force platform, located in front of the starting position, measured the ground reaction force from the landing of the initial step. Peak A–P and vertical forces from the preferred step length condition are shown in Table 2C and D, respectively. Although there were no differences between the two groups in the magnitude of braking and propulsion forces, there were significant leg × group interactions ($F_{1,20} = 17.83$, $P < 0.001$ for peak braking; $F_{1,20} = 45.985$, $P < 0.001$ for peak propulsion). The interaction for both the braking and propulsion phases occurred due to a decrease in force application by the prosthetic limb, as compared with the intact, left and right limbs. As the step length requirement became longer, the magnitude of peak braking and propulsion forces increased ($F_{2,40} = 146.95$, $P < 0.001$ for braking; $F_{2,40} = 11.39$, $P < 0.001$ for propulsion) for all four lead-limb conditions.

There were no group main effects when analyzing the peak magnitudes of the vertical ground reaction force. However, there were leg × group interaction effects for the peak force during weight acceptance ($F_{1,20} = 49.05$, $P < 0.001$), midstance absorption ($F_{1,20} = 17.33$, $P < 0.001$) and push-off ($F_{1,20} = 19.749$, $P < 0.001$). The interaction during weight acceptance and push-off occurred due to a larger magnitude of peak force during the intact lead-limb condition, combined with a decreased magnitude during the prosthetic lead-limb condition. On the other hand, the midstance absorption interaction arose since the peak minimum during the intact lead-limb condition was substantially lower as compared with the other three limb conditions.

3.5. Horizontal impulse

Horizontal impulses were calculated from both the leading and trailing limbs. For both of these limb conditions, it was found that the amputee group exhibited a greater asymmetry between their two leg conditions (i.e. the intact vs. the prosthetic limbs), as compared with the left and right limbs of the able-bodied individuals ($F_{1,20} = 12.91$, $P = 0.002$ for the lead-limb × group interaction; $F_{1,20} = 12.89$, $P = 0.002$ for the trail limb × group interaction; Fig. 5). For both the lead and trail-limb conditions, asymmetry had occurred as the impulse generated from the prosthetic limb was always less than the control conditions, while the intact limb generated a greater impulse than the controls. A step length main effect was found, such that greater impulses were applied during the longer step length requirements for both the lead ($F_{2,40} = 45.60$, $P < 0.001$) and trail ($F_{2,40} = 256.71$, $P < 0.001$) limb conditions.

The sum of the horizontal impulses was also calculated for each gait initiation trial. This involved the addition of the impulse generated by the leading limb with that applied by corresponding the trailing limb. The results indicated that despite the differences in the lead and/or trail limbs between the four leg conditions, the total impulse generated during all lead-limb conditions were not statistically different from one another. Further, as the step length requirement increased, all four leg conditions systematically increased by the same amount ($F_{2,40} = 281.38$, $P < 0.001$ for step main effect; step × group interaction was not significant).

4. Discussion

During gait initiation, the most global adaptation occurred with respect to task duration. Similar to the study conducted by Mouchnino et al. [18], when individuals with a unilateral BKA could not laterally raise their limbs as fast as the able-bodied, the present study found that the individuals from the amputee group required more time to achieve steady state gait than their able-bodied counterparts. Although temporal differences between the two groups were found during each instance of single and/or double limb support, much of the extra duration that was needed by individuals with a unilateral BKA was a consequence of the initial double limb support phase. For example, during the preferred step length condition, of the 530 ms difference that was found in the total task duration between the two groups, 72% of this difference had occurred during the time from upright stance until the initial toe off of the leading limb.

It is proposed that individuals with a unilateral BKA were required to adopt this ‘movement time’ strategy in order to counteract the stability and propulsion limitations of the prosthetic limb.

4.1. Stability

Proprioceptive information from the plantar sole plays an important role in the control of posture and gait [19–21]. Unfortunately, individuals with a unilateral BKA lose this specific form of sensory feedback from one of their legs, resulting in a change in which they maintain and regulate lower limb movements. Not only can this cause these individuals to become more cautious when creating body disequilibrium [22], it can create some uncertainty as the feedback from the residual limb may not be consistent with the sensory feedback that is actually received. The degree to which this may influence the task of gait initiation appeared to be dependent upon the choice of lead limb.

When individuals with a unilateral BKA initiated gait with the intact limb, they had to completely rely on the prosthesis during the initial instance of single limb support. Since re-adjustments to postural errors are difficult to complete without the usual sensory feedback from the foot, these individuals would have needed to be certain that their postural shift was accurate before actually lifting the intact limb off of the ground. Thus during the intact lead-limb condition, the initial double limb support phase was found to significantly longer than either of the two control conditions and slightly (i.e. 200 ms), though not statistically significant, than

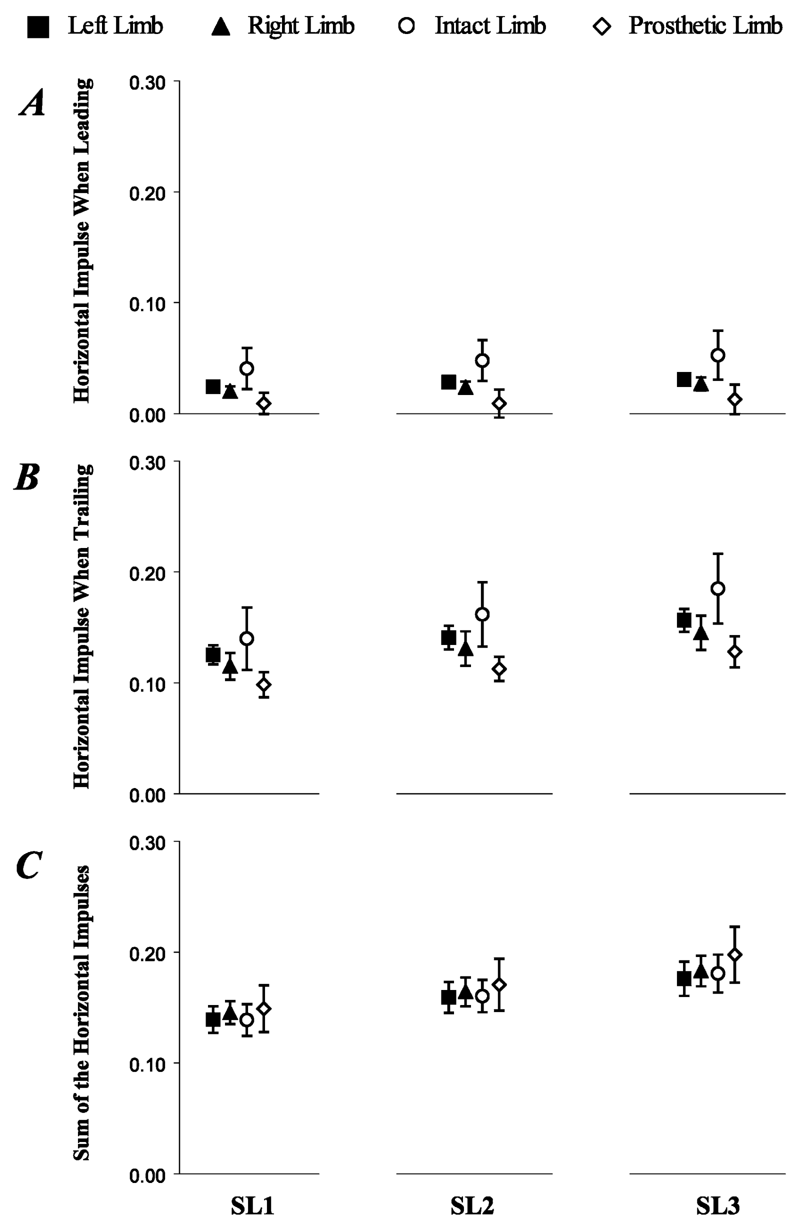

Fig. 5. Mean (±S.D.) horizontal impulse, expressed as Ns/BW, from A, the leading limb; and B, the trailing limb. The mean (±S.D.) sum of the horizontal impulses (i.e. the impulse from the leading leg plus the trailing leg) is shown in C.

the prosthetic lead-limb condition. Conversely, when the prosthetic limb was used for the initial step, the immediate body shift towards the prosthetic side was still required. But during this situation, they were able to rely on the intact limb for slight adjustments as both feet were still positioned on the ground. Since maintaining balance was not as difficult to achieve during this lead-limb condition, the duration of double-limb support was found to only increase a small amount as compared with the control conditions.

While a statistically significant difference was not found in the duration of the initial double limb support phase between the intact and prosthetic lead-limb conditions, other studies have similar temporal asymmetries. For example, in the study by Michel and Do [23], individuals with a unilateral above knee amputation were required to initiate gait as fast as possible. Due to the loss of both the thigh and shank, these amputees required 0.32 s longer to reach the first toe-off stage when leading with their intact, as opposed to their prosthetic limb. Also, many studies examining the gait patterns of individuals with a unilateral BKA during steady-state gait have found significantly longer stance durations on the intact side [13,14]. Clearly, the idea of instability of the prosthesis, and its magnitude of effect, warrants further study.

The idea of instability during the initial double limb support phase also explained the compensatory strategy found with respect to the movement of the COP during gait initiation. Regardless of the lead-limb requirement, individuals with a unilateral BKA exhibited little posterior displacement on the prosthetic side, while the COP under the foot of the intact limb exhibited the typical range of movement. Furthermore, changes to the net COP profile (i.e. the sum of the COP under each foot) were found to be of a greater magnitude during the intact lead-limb condition. When these subjects led with their prosthetic limb, there was little change to the COP profile since the maintenance of stability was being controlled by the intact limb. However, when leading with the intact limb, the COP was found to move in an anterior, as opposed to posterior, direction. As a result, the inflection point, considered the time when forward movement became an important factor, consistently occurred near the fore part of the prosthetic foot. This was distinctly different from the two control foot conditions, as the inflection point occurred towards the heel.

Despite the obvious differences between the four lead-limb conditions, these results are in disagreement with those found by Rossi et al. [17], who found that regardless of the leading limb, the path of the COP from an individual with a unilateral BKA did not differ from the able-bodied. There are several reasons to explain this discrepancy. First, it is likely that the lack of posterior displacement was one method to limit the body's disequilibrium. By moving the COP backwards, forward movement and consequently, displacement of the center of gravity occurs [24]. By limiting this particular displacement, persons with a BKA would have experienced less disequilibrium and hence, diminished the chance of falling or stumbling during gait initiation. However, a more probable reason for the change in the movement of the COP may simply have been due to the prosthetic foot itself. Depending on the particular model and type of prosthetic foot that was used by each of the amputees, the prosthetic ankle could have limited the posterior displacement of the COP. Since these devices are fairly rigid, individuals with a unilateral BKA may not have been able to rock back on their prosthetic heel without having the toes to be lifted off of the ground. Since this would certainly be undesirable, it was likely that they chose to rock back by only using their intact limb.

Interestingly, for three of the subjects in the amputee group, the prosthetic heel was actually not on the platform. It was not clear if this arose from poor fitting of the prosthesis or deformation from the keel of the prosthetic foot, but it did result in their knee being a slightly flexed position, thereby lifting the heel off of the ground. Thus for these particular individuals, having the COP move towards the heel would not have been possible unless the heel dropped back down. This of course, would be redundant as it would not only increase the already prolonged task duration, but would also increase the loading at the base of the residual leg.

4.2. Forward propulsion

In agreement with the results obtained by Nissan [16] and Rossi et al. [17], during the initial push-off phase, very little force was applied in the posterior direction with the prosthetic limb. This was regardless of its responsibility, as only 0.02 N/BW of peak propulsive force was applied when used as the leading limb, and 0.20 N/BW of peak force when utilized as the trailing limb. Both were considerably less than the leading (0.07 N/BW) and trailing (0.27 N/BW) limbs from a control condition.

Consequently, individuals with a BKA relied more heavily on the intact limb for forward propulsion. However, the peak A–P force did not statistically differ from either of the control limbs when leading or trailing. Since this was not large enough to completely make up for the deficiencies of the prosthetic limb, it would appear that they were not able to create enough force for forward movement. However, this was not the case, as task success, as determined by the landing of the lead limb onto the third platform, by these individuals was close to 100%.

To explain this apparent contradiction, the horizontal impulse generated by each of the limbs needs to be considered. Impulse, measured as the amount of applied force over a given period, causes an object to change velocity and consequently, will lead to changes in displacement.

The results obtained from the integrated force–time curve indicate that the prosthetic limb showed very little impulse generation regardless of whether it was acting as the leading or trailing limb. The contralateral intact limb did, however, compensate for this deficit, by increasing its own magnitude of impulse generation. When used as the lead limb, the intact limb applied two times a greater impulse than either of the control conditions; when acting as the trailing limb, it applied a 1.15 times greater impulse than normal.

When the total impulse during each of the four lead-limb conditions were compared, the outcome was very intriguing. The total impulse during all limb conditions exhibited similar magnitudes, where the greatest difference between any two conditions was only 0.01 Ns/N. In addition, it was found that as the step length requirement lengthened, the total impulses for all lead leg conditions systematically increased at very similar rates for all leg conditions. These results indicate that for all three step length conditions, individuals with a unilateral BKA were successful in generating large enough impulses with the intact limb to completely compensate for the limitations of the prosthesis. Since large impulses were not completed using larger peak forces, it was concluded that this was accomplished through an increase in the duration of force application. Hence, by employing a longer movement duration, individuals with a unilateral BKA were able to generate adequate amounts of propulsive force, which otherwise may not have been possible with the passively-acting prosthesis.

4.3. Ground reaction forces after the initial step

Unlike the two previous studies examining gait initiation in individuals with a unilateral BKA, the present study incorporated the use of a third force platform to measure the forces from the end of the first step. This was valuable as further compensations that support the notions of instability and decreased propulsive ability of the prosthesis were observed.

First, there was a significant decrease in the amount of fore–aft force application by individuals with a unilateral BKA when stepping with the prosthetic limb. This is not too surprising given the lack of actively contracting muscles below the level of the knee. These results are also comparable to those found in studies examining unilateral below-knee amputees during steady state walking [13,25].

However, a more noticeable difference occurred in the vertical component of the ground reaction force during the intact lead-limb condition. It was found that the individuals with a unilateral BKA applied a significantly greater load during weight acceptance and push-off, as well as a significantly increased amount of unweighting during mid-stance, as compared with the control and/or prosthetic lead-limb conditions. Although the general shape of the force time curve was similar to those found in studies examining steady state gait, the differences in magnitude between the limb conditions were intriguing.

When leading with the intact limb, these individuals would have been relying on the prosthetic limb for the initial body support. Unlike the other three trailing limb conditions, the prosthetic limb lacks the ankle musculature to actively control ankle dorsiflexion and thus, they would have had much difficulty in controlling the rate of downward acceleration. Hence, as the initial heel strike of the leading limb occurred, a significant increase, in the magnitude of 0.08 N/BW, in vertical force was observed. Whether this difference in impact loading is clinically relevant remains unknown. If it is, it may be possible that this phenomenon is strongly linked to the greater incidence of lower back pain found in the amputee population [26].

As these individuals moved forward and approached mid-stance, the intact limb became responsible for body support whereas the prosthesis was undergoing push-off. For this lead-limb condition, there was a substantial increase in the amount of midstance absorption, which again is not observed during steady-state gait. It is reasonable not to expect the prosthetic limb to undergo such a large absorption, as this would facilitate the compressive forces at the residual limb–socket interface. However, why the intact limb absorbs more weight than the controls remain unknown. Despite the results found with the horizontal impulses, it is possible that the preceding push-off of the contralateral limb was not large enough for these individuals. Lowering the body's center of mass may have lead to eccentric loading, and thus the storage of energy, within the muscles of the intact limb. This, in turn, could be used to generate more force and, therefore, more forward movement, for its upcoming push-off.

As they began to push-off with the intact limb, the individual created a larger vertical force than the other limb conditions, presumably as compensation for the lack of push-off from the previous step. This is not too surprising, as this effect has been found during steady state walking [13].

5. Conclusions

The purpose of this study was to investigate the postural and movement strategies adopted by persons with a unilateral BKA during gait initiation. It was hypothesized that these individuals would employ two unique strategies, depending upon the responsibilities and capabilities of their two structurally asymmetrical limbs. Longer step length requirements were hypothesized to produce greater magnitudes of change to the gait initiation strategy.

The results indicated that individuals with a unilateral BKA required more time to complete the gait initiation task. This increase in movement duration was primarily attributed to one interval: the initial instance of double-limb support, where the individual had to, from an upright standing position, lift the stepping foot off the ground. The increase in stance duration allowed a greater time for force to be applied in the posterior direction. Not only was this important as the prosthetic foot could not generate as much of a peak force, it allowed the intact limb to compensate, by applying a similar amount of force, but for a longer period of time. Consequently, these individuals were able to apply a similar 'sum of horizontal impulses' as compared with their able-bodied counterparts. This enabled all of the individuals with a unilateral BKA to successfully step far enough to satisfy the three step length requirements.

The decision by these individuals to use this 'horizontal impulse strategy' warrants further investigation. It would be of interest to determine whether this strategy transpires for other locomotor tasks such as gait termination and/or immediate stepping. Furthermore, it may be of interest to examine other populations who experience stability and locomotor difficulties, as this particular strategy may in fact be employed regardless of the specific disability.

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