POSTURAL AND MOVEMENT ADAPTATIONS BY INDIVIDUALS WITH A UNILATERAL BELOW-KNEE AMPUTATION DURING GAIT INITIATION

by

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BHK, University of British Columbia, 1999

A THESIS SUBMITTED IN PARTIAL FULFILLMENT
OF THE REQUIREMENTS FOR THE DEGREE OF

MASTER OF SCIENCE

in

THE FACULTY OF GRADUATE STUDIES

(School of Human Kinetics)

We accept this thesis as conforming to the required standard

THE UNIVERSITY OF BRITISH COLUMBIA

March, 2002

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Date March 7, 2002

DE-6 (2/88)

ABSTRACT

Gait initiation, the transition from upright stance to steady state gait, requires the asymmetrical use of the two lower limbs. The initial stepping limb, called the lead leg, is mainly used to generate forward thrust, while the non-stepping, trailing limb is initially responsible for the generation of forward propulsion and the maintenance of body stability. For individuals with a unilateral below-knee amputation (BKA), this unequal sharing of responsibilities poses a potential conflict, as the prosthetic limb is known to have limitations in both stability and propulsion. Because the magnitude of this effect may differ depending upon the roles of each limb, the study hypothesized that individuals with a unilateral below-knee amputation would undergo unique postural and movement adaptations depending on the choice of the leading limb.

Eleven individuals with a unilateral BKA and eleven control subjects were recruited for this study. From a standing position, each individual initiated gait at three step length conditions (+0%, +25%, and +50% of preferred step length). Half of the trials were initiated with the right limb, while the other half were initiated with the left.

It was found that the amputees underwent postural and movement adaptations due to the presence of the prosthetic limb. The amputees required more time to initiate gait, applied a smaller magnitude of peak propulsive force, and exhibited a smaller displacement of the center of pressure. The magnitude of these changes was, however, found to be dependent upon the choice of the leading limb, as greater compensations were observed during the prosthetic trail limb condition.

Three conclusions were made from this study. First, the prolonged task duration allowed the amputees to apply a larger horizontal impulse, such that they were able to fully compensate for the decreased propulsive ability of the prosthetic limb. Second, the slower rate of initiation

and the decrease in displacement of the center of pressure allowed the amputees to remain more stable throughout the entire task. Finally, it appeared that the role of the trailing limb had a greater impact during gait initiation and thus, leading with the prosthetic limb resulted in fewer postural and movement adaptations.

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ACKNOWLEDGEMENTS

The time I have spent working on this thesis would not have been as successful without the help of several individuals, to whom I am forever grateful.

First and foremost, I would like to thank my thesis advisor, Dr. David J. Sanderson, for supporting and guiding me throughout my graduate studies. The valuable experiences I have gained not only during the composition of this thesis, but also during HKIN 363 and the B.C.M.S.F. project, will always be cherished. I also wish to thank the other committee members, Dr. J. Timothy Inglis and Dr. Romeo Chua, for their support and direction throughout the creation completion of this thesis. Their immense knowledge, especially outside the area of biomechanics, has enabled me to gain a much broader perspective on human locomotion.

The difficulty in finding individuals with a unilateral below-knee amputation was made much easier with the help of Ms. Linda McLaren, a physiotherapist at the G.F. Strong Rehab Centre. Without her enthusiasm and cooperation towards my project, none of this would have been possible.

Finally, I would like to thank my family and friends for supporting me, not only during the last few years, but also throughout my entire life. The willingness to sacrifice their time and provide much needed advice and assistance (even as control subjects) was greatly appreciated.

I will forever remember the time spent working at the U.B.C. Biomechanics Lab.

CHAPTER 1: INTRODUCTION

Each year, approximately 135,000 North Americans experience the loss of one or more limbs due to trauma and/or disease (Alberta Amputee Sport and Recreation Association, 2001). The majority (53%) of these cases are classified as a trans-tibial or *below-knee amputation*, where there is a loss of one or both legs below the level of the knee (Northwestern University Prosthetics-Orthotics Center, 2001). In the case of a unilateral below-knee amputation, 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, 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 thesis 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 gait initiation, 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 the body's center of mass is shifted towards the contralateral limb (Maki and McIlroy, 1997). This reduces the lead limb's need to maintain body support, so that it can focus on generating a portion (a peak posterior force of 7% body weight) of the initial forward thrust (Nissan and Whittle, 1990). Concurrently, the role of the non-stepping limb (termed 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 quantity 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 (Nissan and Whittle, 1990). The distinct responsibilities for both legs remain unchanged until the initial step has been completed.

After approximately 0.54 seconds from the onset of gait initiation, the heel of the leading limb will be returning towards stance phase (Nissan and Whittle, 1990). It is at this time that the duties of the lead and trail limbs become reversed. The trail limb, initially relied upon for body support and to a lesser extent forward movement, begins to apply much more of a horizontal force (a peak of 22% body weight) (Nissan and Whittle, 1990). This is needed in order to achieve the second step. The lead leg, originally responsible for creating the initial forward motion, now becomes responsible for absorbing the shock generated from the forthcoming heel-strike. Body support also becomes a concern, particularly when the trailing limb undergoes its own swing phase. Once the trailing limb finishes its initial step, approximately 0.61 seconds after the heel strike of the leading limb, steady state gait will be achieved (Nissan and Whittle, 1990; Ledebt et al., 1998).

While the unequal sharing of responsibilities associated with gait initiation may not be a concern for those who have structurally similar limbs, conflicts can certainly arise for persons with a unilateral BKA. One issue relates to the overall stability of the prosthetic limb as amputees are known to exhibit a greater amount of body sway during upright stance (Fernie and

Holliday, 1978, Geurts et al., 1992; Isakov et al., 1992), a smaller excursion of the center of pressure during forward leaning (Summers et al., 1988), and a more vertically aligned limb position during continuous walking (Sanderson and Martin, 1997). Many of these findings have been attributed to a variety of factors, such as improper prosthesis fitting (Friberg, 1984), lack of confidence in the prosthesis (Summers et al., 1988) or poor design of the socket (Engsberg et al., 1991). One of the ways unilateral below-knee amputees counteract this instability is by placing a greater proportion of their body weight on their non-amputated limb. This allows the amputee to avoid high loading forces on the prosthetic limb (Lord and Smith, 1984; Engsberg et al., 1989). Since there will be instances during gait initiation when one leg is entirely responsible for providing body support, it is of great interest to determine whether and how this compensatory strategy will be affected.

Another concern for individuals with a unilateral below-knee amputation is the generation of forward propulsion. Much of the forward moment needed for walking is provided by the ankle plantarflexors (Winters, 1991), which are now lacking in the amputated limb. Since the prosthetic ankle is unable to generate a comparable amount of propulsion as actively contracting musculature from a biological leg, compensations measured during steady state gait, may be needed in order to successfully initiate gait. These include adaptations such as an increased contribution from the musculature of hip joint from the prosthetic limb (Czerniecki and Gitter, 1992; Sanderson and Martin, 1997; Sadeghi et al., 2001), as well as the hip and knee muscles from the non-amputated limb (Czerniecki and Gitter, 1992; Nolan and Lees, 2000), How these, or any other adaptations can fully restore the functions of a physiological ankle during gait initiation remains to be seen.

Despite these potential conflicts, little research has been conducted in this area. The first study, conducted by Nissan (1991), found several kinematic and kinetic differences between the amputee and control groups, but no 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, that are strongly related to gait initiation.

The most recent experiment, conducted by Rossi et al. (1995), found that the movement of the center of pressure was similar for both the amputee and control groups. However, this is quite unlikely to occur since the path of the center of pressure during gait initiation is highly dependent upon the muscle activity of the lower limbs. Several of these muscles, such as the tibialis anterior and gastrocnemius, are absent following a below-knee amputation and therefore, these individuals would likely 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.

Due to the lack of knowledge in this area, the present study was conducted to further differentiate the strategies adopted by persons with a unilateral BKA during gait initiation as compared to their able-bodied counterparts. Postural adaptations that can influence forward movement, such as body weight distribution, as well as the position and displacement of the center of pressure, were examined to determine the amount of stability during upright stance. As each individual moved toward steady state gait, variables such as joint angles, joint moments, activation of the lower limb musculature and force application were examined to elucidate the mechanisms associated with forward propulsion. *It was hypothesized that individuals with a unilateral BKA would undergo kinematic, kinetic and electromyographic changes in order to compensate for the stability and propulsion deficiencies of the prosthetic limb.* The study also introduced three different step length requirements, where the distance of the initial step was randomly controlled. *It was hypothesized that a step length effect would be found, since the longer step length conditions would represent a greater postural and movement challenge.*

CHAPTER 2: METHODS

2.1 Overview:

Two groups (11 individuals with a unilateral below-knee amputation and 11 healthy age- and gender-matched controls) were recruited for this study. Each participant underwent a single test session that consisted of four components of data collection:

1. *A written questionnaire that assessed each individual's level of physical activity.*
2. *Five "pre-test" trials, which required each individual to take five strides from a standing position. This determined each subject's preferred limb, preferred step length and preferred stride length.*
3. *A single 1-minute standing trial that measured the individual's center of pressure and magnitude of body sway.*
4. *Forty-two gait initiation trials, which required each participant to start walking, from a standing upright position, along a three-meter walkway. Fourteen trials at each of the three step length conditions (preferred step length, +25% of preferred and +50% of preferred) were completed by each individual. Half of the trials were initiated with the left leg and the other half were initiated with the right. Kinematic (joint angles, temporal events), kinetic (ground reaction forces, linear impulse, joint moments, center of pressure) and electromyographic data were collected during each of these trials.*

2.2 Subjects:

Two groups of subjects were recruited. The unilateral below-knee amputee group comprised of 11 individuals between the ages of 24-65 years (Table 2.1). Although inclusion criteria such as reason for amputation, prosthesis type, and the time since limb loss were initially set, adequate numbers of volunteers could not be found. Consequently, the criteria were broadened such that any individual with a unilateral below-knee amputation who had the ability to walk independently was allowed to participate.

Once the BKA group was formed, 11 individuals, who were age- and gender-matched to the BKA group, were recruited to form the control group (Table 2.2). Those with a history of any neurological conditions were excluded from the study.

All participants read and signed an informed consent form (Appendix C) revealing all details of the experimental protocol, which had been approved by the University of British

Below-Knee Amputees	Age (yrs)	Sex	Mass (kg)	Amputated Limb	Reason for Amputation	Foot Type	Experience (yrs)
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	45
10	36	Male	84	R	Trauma	C-Foot	13
11	41	Female	54	L	Cancer	Flex	18
Avg. (± SD):	44.1 ± 14.1	8 males; 3 females	82.7 ± 22.5	7 right leg; 4 left leg	8 trauma; 2 vascular; 1 cancer	6 Seattle; 4 Flex; 1 C-Foot	13.7 ± 14.9

Table 2.1: Characteristics of the eleven individuals with a unilateral below-knee amputation. L = left, R = right.

Non-Amputees	Age (yrs)	Sex	Mass (kg)
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
Avg. ± SD:	44.9 ± 12.3	8 males; 3 females	61.1 ± 9.0

Table 2.2: Characteristics of the eleven age- and gender-matched controls.

Columbia Ethical Review Committee.

2.3 Data Collection and Analysis:

Each subject underwent a single 2-hour testing session that consisted of five major components.

2.3.1 Questionnaire Assessment

The first portion of the study involved the completion of a modified Godin Leisure Time Exercise Questionnaire (Appendix D). The questionnaire consisted of four questions that inquired about the regularity and rigorousness of the individual's exercise program. To quantitatively analyze this data, a weighted score was calculated using the first three questions. Each session of an extremely strenuous activity was given a raw score of 5, each moderately strenuous activity was given a raw score of 3, and the least vigorous activities were each given a raw score of 1. The sum of the scores for each individual was calculated and determined as the "level of physical activity".

2.3.2 Pre-test

Following the completion of the questionnaire, a "pre-test" was conducted so that the step length requirements could be scaled for each individual. This test 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, four measures were obtained. First, the individual's preferred limb was determined as the limb that took the first step. Two subjects did not have a consistent choice of lead limb (BKA #6, who lead with the left leg in four trials, and AB #7, who lead with the left leg in three trials), and thus, the preferred limb was designated as the one that was used for the majority of the trials. Second, the initial step length was measured as the distance from the start position to the heel strike of the first step. Third, the average stride length was calculated as the total distance of the six strides, less the first step, divided by five. Finally, a mean first step length:mean stride length ratio was calculated, as this indicated the proportion of the first step length in relation to the individual's normal stride length.

2.3.3 Subject Preparation

After these initial measurements were determined, reflective markers were placed over eleven bony landmarks: the neck (T1 vertebrae), the greater trochanter (bilaterally), the lateral condyle of the knee (bilaterally), the lateral malleolus (bilaterally), the heel (bilaterally), and the fifth metatarsal head (bilaterally). Once the reflective markers were in place, the skin of selected sites was prepared for the placement of surface electrodes (0.5 mm bipolar silver silver-chloride, Therapeutic Unlimited). Skin preparation involved the shaving of hair, the rubbing of alcohol and a slight abrasing of the skin. These standard techniques were necessary to limit the amount of impedance and to optimize the distance between the muscle and the electrode. Once this was complete, the electrodes were placed bilaterally on the gluteus medius, vastus lateralis, semitendinosus, soleus, and tibialis anterior.

When all markers and electrodes were positioned, participants were brought into the measurement area (Figure 2.1). They were then asked to stand with each foot on a separate force platform (Bertec model 4060). The specific angle and placement of the feet were set to the individual's preference. This position was established as the "start" position and remained constant throughout the entire experiment by tracing footprints onto a piece of paper placed overtop the two platforms.

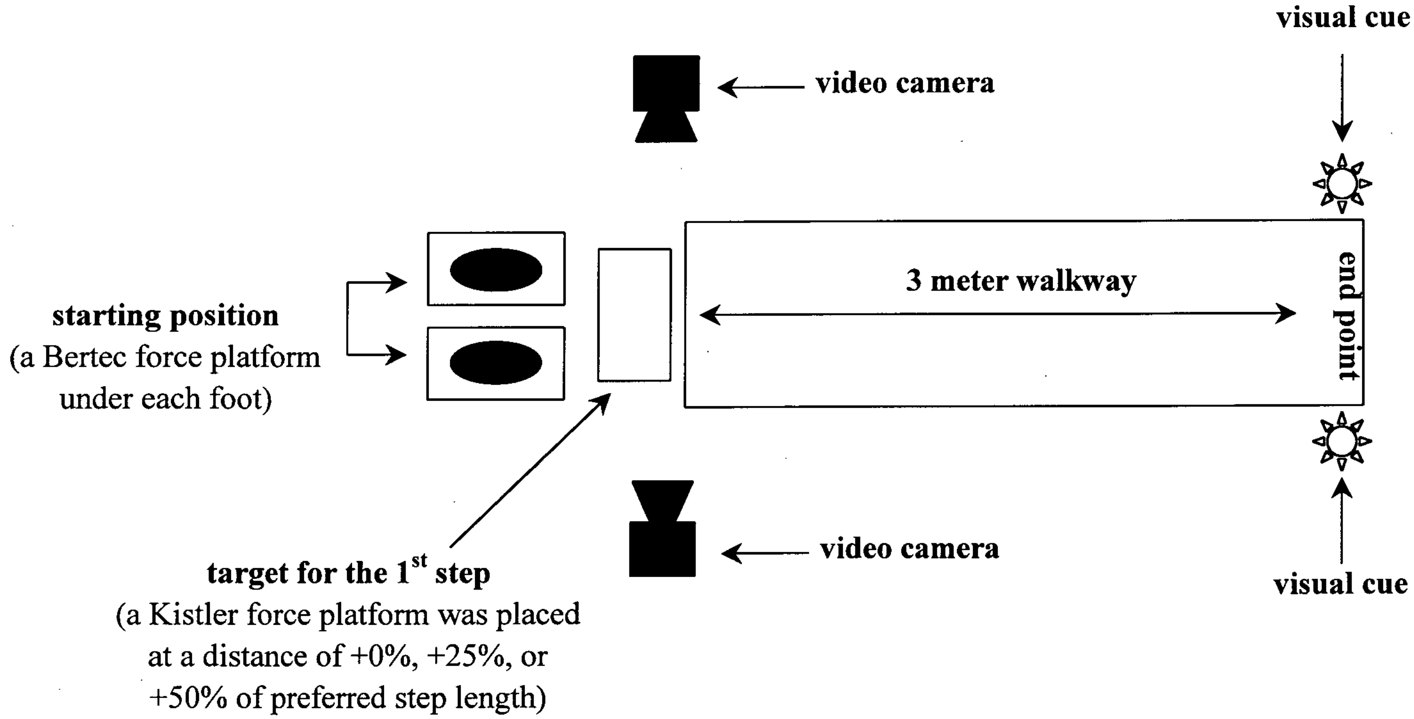


Figure 2.1: Diagram of the lab set-up. Participants stood with each foot on a separate force platform. When one of the lights (i.e. the visual cue) at the end of the walkway lit up, they were required to start walking towards the end of the walkway. The first step had to land on the Kistler force platform. Two video cameras recorded the movements of the individual during the entire trial.

2.3.4 The One-Minute Standing Trial

For the third component of the test session, subjects were instructed to stand at the "start" position, with their arms crossed in front, head focused straight ahead, while remaining as stationary as possible. Force and moment data from the two Bertec platforms were collected for each individual during a single 1-minute standing trial. These data were analog-to-digital converted (Data Translation Model 3010) using a PC compatible computer at a sampling frequency of 50 Hz. The raw data were then processed before any subsequent analysis. The first and last ten seconds of each trial were neglected to allow for measurement and subject stabilization. To account for differences in the orientation and size of the feet, foot lengths and widths were normalized as shown in Figure 2.2. Furthermore, since four of the eleven participants with a below-knee amputation had lost their left limb, their data was reversed such that the data obtained from the prosthetic side were now labelled as data from the right leg. Data from the intact, or right side, was also switched such that it corresponded to the left leg. This reversal was necessary to facilitate the averaging across the eleven amputee individuals.

There were four variables of interest during this component of the study: the relative loading of each limb, the net position of the center of pressure from both feet, the position of the center of pressure under each foot, and the standard deviation of the center of pressure position.

Since this portion of the protocol was not implemented from the beginning, one amputee subject (BKA #4) did not undergo this standing trial.

2.3.5 The Gait Initiation Trials

The last component of the study involved gait initiation. This required the subject, after receiving a visual cue, to walk from the "start" position along a short (3 meter) mounted walkway. Not only did this visual cue, a pair of lights located at the end of the walkway,

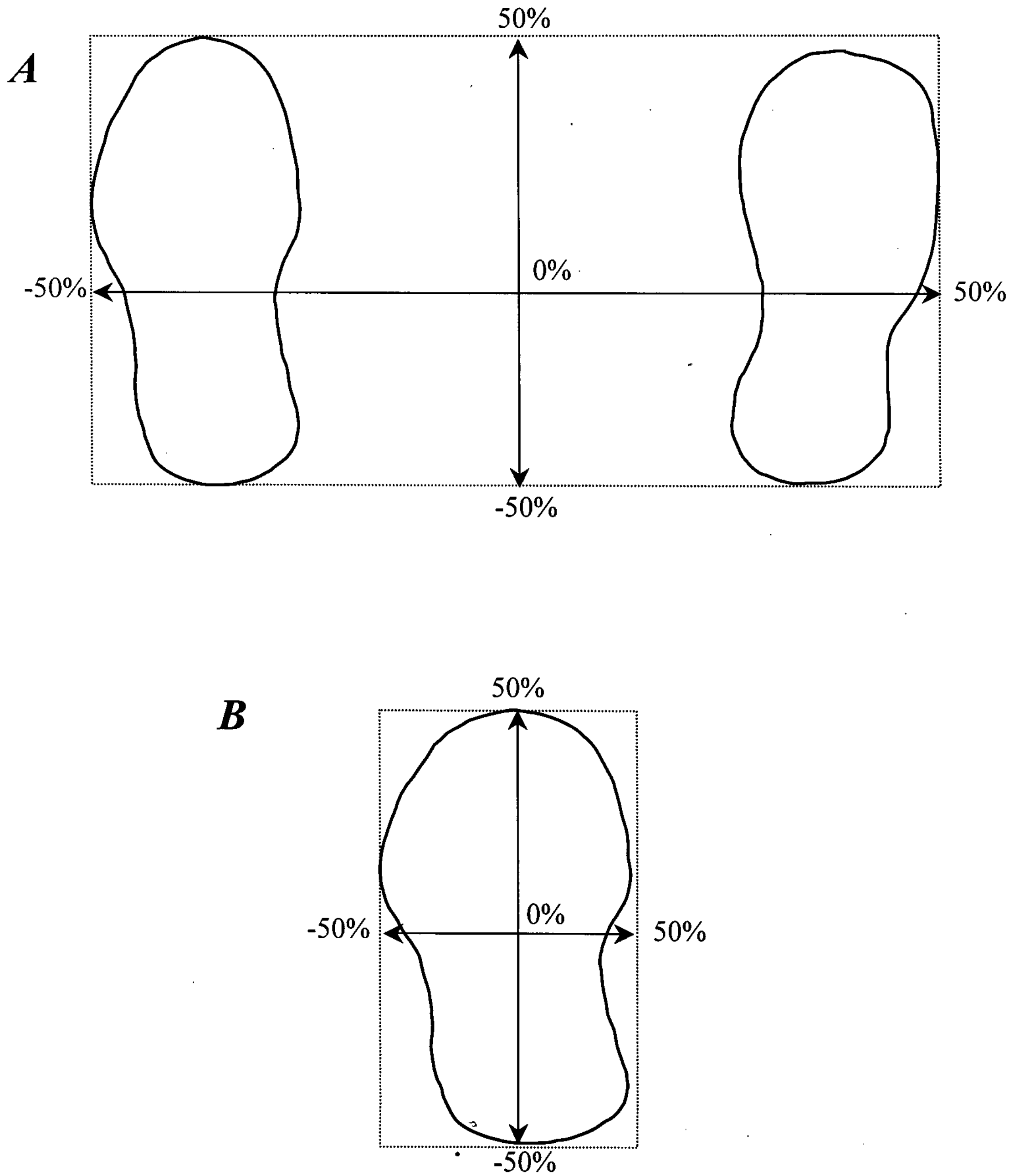


Figure 2.2: Normalized foot length and width for *A*, two-foot stance and *B*, one foot stance.

indicate the "start" signal, 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 was denoted by the illumination of the light placed 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 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 randomly 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, as determined from the initial "pre-test". 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. If the subject had failed to land properly on the Kistler platform, the trial was repeated. This did not happen often (on average, less than 2 trials per subject), and was primarily due to the cables of the EMG system getting caught on one of the force platform edges.

In total, each individual participated in 42 gait initiation trials. Fourteen trials were completed at each of the three step lengths, with half the trials from each step length condition starting with the left leg and the other half leading with the right.

During these 42 trials, kinematic measurements along the walkway were obtained using two video cameras placed four meters 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 limb. 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, Peak Performance Technologies, Colorado Springs, CO) video analysis system. Raw co-ordinate data were filtered using a 4^{th} order, dual pass Butterworth filter with a cut-off frequency of 5 Hz.

In addition to the kinematic measures, the three force platforms and the eleven surface electrodes collected various kinetic and EMG data. These were analog-to-digital converted (Data Translation Model 3010) using a PC compatible computer at a sampling frequency of 600 Hz. EMG data were full-wave rectified and filtered at 5 Hz using a Butterworth filter before continuing with any further analysis.

All of the collected data were 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. Maki and McIlroy (1997) recently defined the onset of a stepping response as a 4 mm shift in the baseline COP. Since the gait initiation task consists of a stepping response, a similar definition was used to define the start of sway. However, to reduce some of the variability that may occur with anticipation, the 4 mm shift was required to be of a duration of at least 10 ms. Normalized data were then averaged for each lead limb condition at each of the three step lengths. Group averages were finally calculated from all participants within each of the two groups.

For the kinematic data, it was found that the amputees had adopted a variety of different postures. This was not too surprising considering the results of Engsberg et al. (1991), who found that amputee children tend to exhibit greater knee flexion on the prosthetic limb during standing. However, for three of these amputees, this strategy was so excessive that it caused the prosthetic heel to be lifted off the ground. As a result, this drastically affected the group mean, and thus the kinematic data were normalized to each individual's starting position. Consequently, the kinematic results only indicated a change in joint angle, rather than the actual angle itself.

For the COP data, obtained from both Bertec force platforms, additional processing steps were taken. Similar to the processing methods during the standing trials, the position, excursion and/or area of the COP was normalized to 100% foot length and 100% stance width. 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 seconds.

For the calculation of the horizontal impulses, the anterior-posterior force-time curves were integrated using the Trapezoid Approximation method. The three impulses that were applied from the starting position were calculated: one for the leading limb, another for the trailing limb, as well as the combined impulse from the leading and trailing limbs (Figure 2.3). 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 for the trailing limb. The total impulse was calculated as the sum of the two impulses. 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.

A $$\int_{sway}^{TO1} Fx_{lead}\, dt$$

B $$\int_{sway}^{TO2} Fx_{trail}\, dt$$

C $$\sum\left(\int_{sway}^{TO1} Fx_{lead}\, dt + \int_{sway}^{TO2} Fx_{trail}\, dt \right)$$

Figure 2.3: The equations used to calculate the horizontal impulses from A, the leading limb; B, the trailing limb; and C, the sum of the leading and trailing limbs. Fx represents the A-P force.

2.4 Statistical Analysis:

When examining measures between four limb conditions (i.e. left, right, intact and prosthetic limbs), analysis was conducted using a 2 (group) X 2 (limb condition) X 3 (stride length) mixed factorial design, with repeated measures on the last two factors. It is important to note that although this analysis assesses numerous main and interaction effects, it does not allow for a direct comparison between an intact (or prosthetic) limb condition with that of a control. The leg x group interaction simply compares the level of "symmetry" between the two limb conditions from the control group versus the two limb conditions from the amputee group. Alpha was set a priori at 0.05.

CHAPTER 3: RESULTS

3.1 Overview:

The results are presented in the order in which the variables were collected. The questionnaire and pre-test results are mentioned first, followed by the one-minute standing trial and lastly, the gait initiation trials. Within the gait initiation section, the data are separated into three main categories: movement time, center of pressure, and ground reaction forces. Movement time includes both the overall task duration, as well as the time interval for each kinematic event (toe-offs and heel-strikes). The center of pressure data includes the net displacement of the COP, as well as the COP profile from each individual foot condition. Finally, the ground reaction force data contains the results obtained from the two Bertec force platforms (i.e. the forces and impulses applied from sway until toe-off) and the single Kistler force platform (the forces applied at the end of the initial step).

Measurements, such as the joint kinematics, joint moments and electromyographic activity were also collected and analysed but these results are not presented in this thesis. The joint moment and EMG measures were found to be highly variable between trials, presumably due to the experimental protocol where subjects were unaware of the lead limb requirement prior to each trial. The kinematics of the lower limb were consistent but have already been examined in detail by Nissan and Whittle (1991).

3.2 Physical Activity Questionnaire and Pre-Test Measurements:

At the beginning of the test session, each subject completed a physical activity questionnaire and participated in a short "pre-test". The questionnaire was used to gather descriptive information regarding the level of physical activity for each of the participants. The "pre-test" was conducted to determine the scaled distances for the three step length requirements.

The physical activity questionnaire did not reveal differences in the level of physical activity between the two groups (37.1 ± 29.3 for the amputees vs. 31.8 ± 19.0 for the controls; $F_{1,20}$=42.94, p>0.62), suggesting that the two groups were equally active in daily life. This however, may have been due to the large range of scores observed in both groups.

Results from the "pre-test" assessment showed a similar number of right-hand and right-limb dominant individuals between the two groups (Table 3.1). All able-bodied individuals were right-handed with 10 of the 11 subjects preferring to lead with their hand-dominant leg. There were ten right-handed and one left-handed amputee subjects, with 9 of the 11 amputees preferring to step with their hand-dominant leg. Of these nine individuals, six were stepping with their prosthetic limb. For the two who initiated gait with their non-dominant limb, both were using the prosthetic limb.

Below-Knee Amputees	**Hand Preference**	**Limb Preference**	**Step Length (m)**	**Stride Length (m)**	**Step Length : Stride Length Proportion**
1	Right	Dominant (P)	0.83	1.51	0.55
2	Right	Non-Dominant (P)	0.65	1.36	0.48
3	Right	Dominant (P)	0.93	1.79	0.52
4	Right	Dominant (P)	0.70	1.38	0.51
5	Right	Dominant (P)	0.69	1.44	0.48
6	Right	Non-Dominant (P)	0.70	1.57	0.45
7	Right	Dominant (P)	0.80	1.64	0.49
8	Left	Dominant (I)	0.81	1.63	0.50
9	Right	Dominant (I)	0.62	1.41	0.44
10	Right	Dominant (P)	0.74	1.45	0.51
11	Right	Dominant (I)	0.69	1.11	0.62
Avg. ± SD:	10 of 11 right handed	9 of 11 dominant hand 8 of 11 prosthetic limb	0.74 ± 0.09	1.48 ± 0.18	0.50 ± 0.05

Non-Amputees	**Hand Preference**	**Limb Preference**	**Step Length (m)**	**Stride Length (m)**	**Step Length : Stride Length Proportion**
1	Right	Dominant	0.74	1.40	0.52
2	Right	Dominant	0.57	1.25	0.46
3	Right	Dominant	0.67	1.34	0.50
4	Right	Dominant	0.50	1.02	0.49
5	Right	Dominant	0.65	1.41	0.46
6	Right	Dominant	0.62	1.19	0.52
7	Right	Non-Dominant	0.68	1.46	0.46
8	Right	Dominant	0.73	1.39	0.52
9	Right	Dominant	0.61	1.14	0.53
10	Right	Dominant	0.72	1.36	0.53
11	Right	Dominant	0.80	1.39	0.58
Avg. ± SD:	11 of 11 right handed	10 of 11 dominant hand	0.66 ± 0.08	1.30 ± 0.14	0.51 ± 0.04

Table 3.1: Pre-test measures for each subject. (P) refers to the prosthetic limb, (I) refers to the intact limb.

Although individuals with a unilateral below-knee amputation showed a slightly greater step length ($F_{1,20}$=4.41, $p<0.05$) and stride length ($F_{1,20}$=6.72, $p<0.02$), the step length to stride length proportion for both groups were quite similar ($F_{1,20}$=0.049, p>0.83) to one another.

3.3 One-minute Standing:

During the one-minute standing trial, several variables were measured. First, the position of the net COP (i.e. the average position of the COP, calculated from both feet, over the 1 minute sampling interval), in both the M-L and A-P directions, was found not to be significantly different between the two groups of subjects (Figure 3.1 and Table 3.2).

This was also the case when the mean position of the individual COPs was examined. All four (left, right, intact and prosthetic) feet conditions exhibited a similar positioning in both the M-L and A-P directions.

The level of limb load asymmetry between the two groups was also examined from this standing trial. Similar to the mean position of the COP, this measurement was not significantly different between the two groups of subjects.

			Position of the COP (%)		Standard Deviation of the COP (mm)	
	LEG	LIMB LOAD (% BW)	M-L	A-P	M-L	A-P
	Left	50.75 (2.96)	6.25 (5.04)	-6.48 (4.64)	1.17 (0.81)	6.92 (4.41)
AB	Right	49.25 (2.96)	-3.40 (6.62)	-7.31 (3.81)	1.11 (0.55)	5.23 (3.09)
	Net	-	0.00 (2.49)	-7.03 (3.85)	2.96 (1.77)	5.84 (2.16)
	Intact	50.40 (6.02)	6.14 (8.41)	-8.45 (11.00)	2.14 (1.26)	6.79 (5.12)
BKA	Prosthetic	49.60 (6.02)	18.69 (13.82)	0.70 (13.81)	1.03 (0.31)	2.30 (1.33)
	Net	-	-2.40 (4.78)	-4.20 (7.89)	3.54 (2.12)	6.45 (2.00)

Table 3.2: Mean (± SD) loading of the limbs (in % BW); mean (± SD) position of the individual and net COP; mean (± SD) standard deviation for the individual and net COP.

The last variable of interest during this portion of the study was the standard deviation of the COP. Although the standard deviation of the net COP was not statistically different between the two groups of subjects, the results indicated that there were adaptations within the amputee's two leg conditions. Individuals with a BKA exhibited a larger standard deviation in the M-L

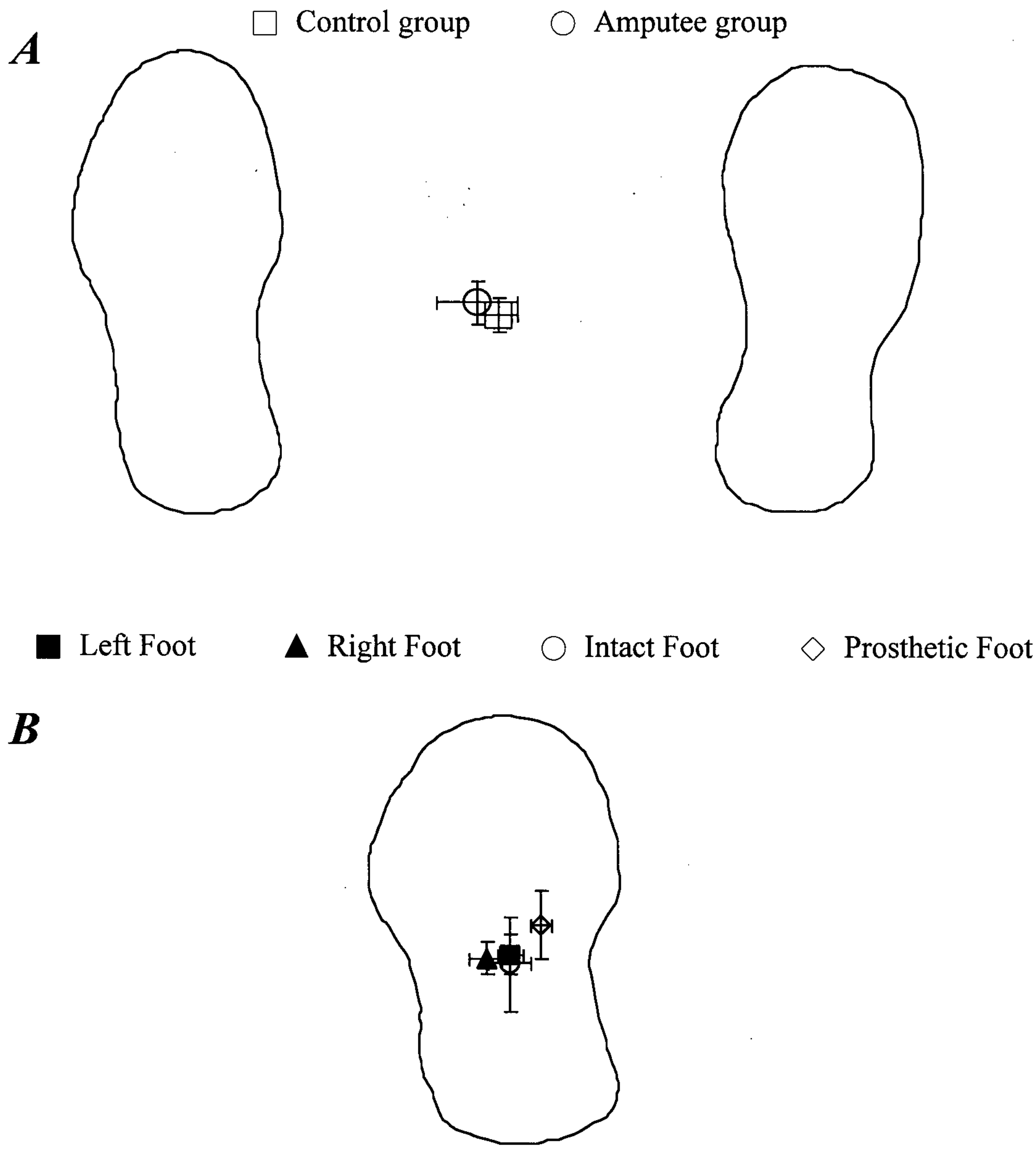


Figure 3.1: Mean (± SD) position of the COP during the one-minute standing trial for *A*, the control and amputee groups; and *B*, the left, right, intact and prosthetic limb conditions.

direction on their intact limb, and perhaps a slightly decreased variability on their prosthetic limb, as there were leg ($F_{1,19}$=6.19, p<0.03) and leg x group ($F_{1,19}$=4.86, p<0.05) interactions. No statistical differences were found in the A-P direction.

3.4 Gait Initiation Trials:

3.4.1 Task Duration:

Overall, the amputees took a longer time to complete the gait cycle (as defined as the time from the start of sway to the second toe-off of the leading limb), as compared to the able-bodied (2.26 ± 0.28 sec for the amputees, 1.73 ± 0.15 sec for the controls; $F_{1,20}$=40.96, p<0.001; Figure 3.2). Since the leg x group interaction was insignificant ($F_{1,20}$=0.67, p>0.79), this established that task duration for the amputees was not any more asymmetrical between the two lead limb conditions of the amputee group.

Due to the significantly prolonged duration found in individuals with a unilateral BKA, the entire gait cycle was broken down into each double or single limb support phase to determine whether the difference in overall duration could be explained by one or more these intervals (Figure 3.3 for the schematic explanation of events; Table 3.3 for results).

First, it was found that the amputee group spent a significantly longer duration ($F_{1,20}$=38.66, p<0.001) during the initial double limb support phase (DS 1), indicating that more time was needed to lift the lead limb off of the ground. There were no leg, or leg x group interaction effects.

During the swing phase of the lead limb (SS 1), a group main effect did not appear. However, a greater asymmetry was exhibited by the individuals with a unilateral BKA, as there were leg ($F_{1,20}$=29.52, p<0.001) and leg x group ($F_{1,20}$=23.59, p<0.001) effects.

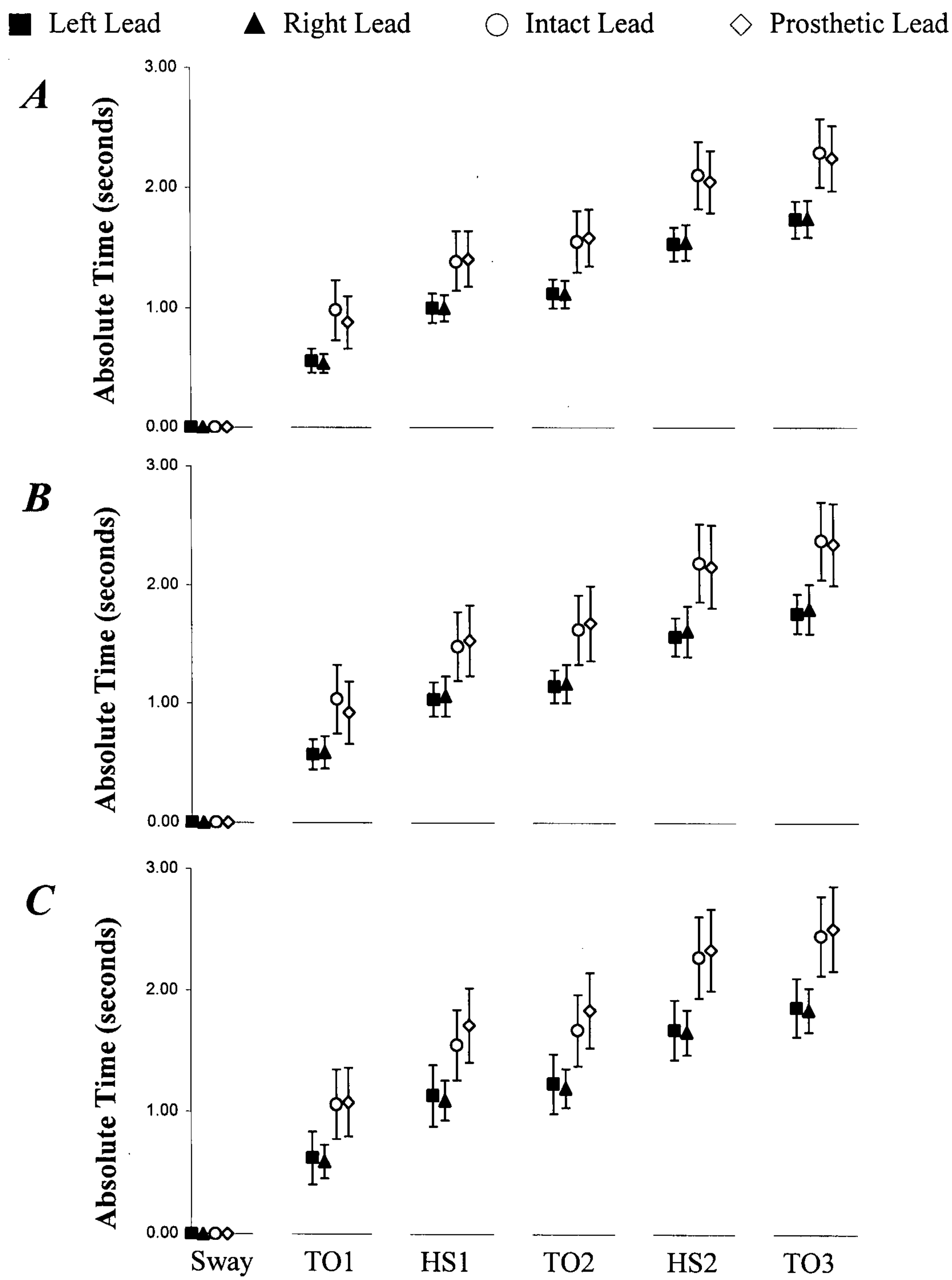


Figure 3.2: Mean (± SD) absolute time (in seconds) taken from sway to the third toe-off of the four lead limb conditions during *A*, the +0% SL requirement; *B*, the +25% SL requirement; *C*, the +50% SL requirement. There were significant group and step differences when examining the total time (i.e. from sway to TO3).

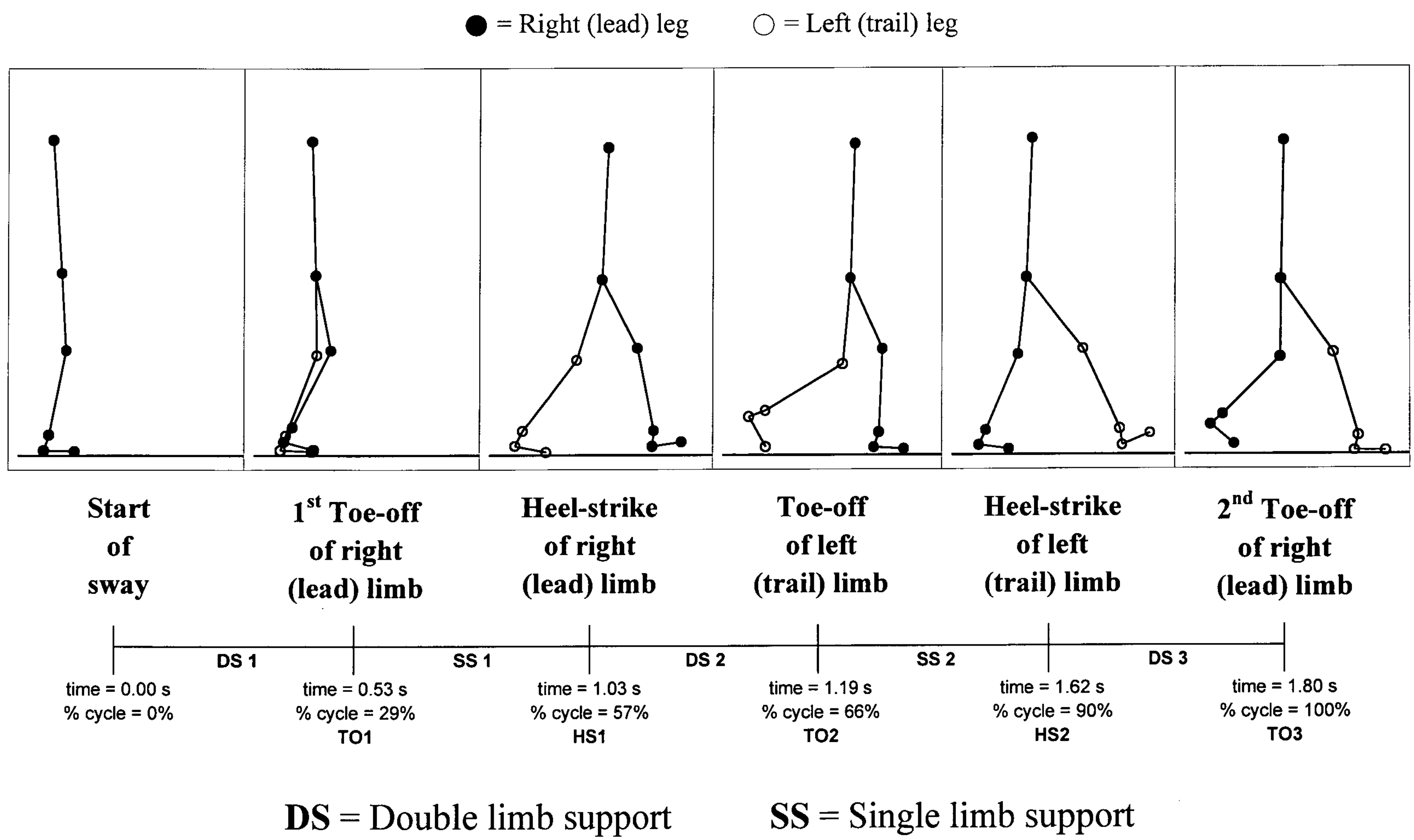


Figure 3.3: Events and timings of the gait initiation cycle. Times (in seconds) and % cycles are from a single trial (right lead limb condition) by subject AB #1.

Specifically, when amputees lead with the intact limb, single limb support phase was similar in duration to that of the able-bodied; however, when stepping with the prosthetic limb, the support phase became considerably longer.

		Leading Leg			
		Left	**Right**	**Intact**	**Prosthetic**
SL1	**DS 1**	0.56 (0.10)	0.54 (0.08)	0.98 (0.25)	0.88 (0.22)
	SS 1	0.44 (0.05)	0.46 (0.05)	0.41 (0.06)	0.53 (0.06)
	DS 2	0.12 (0.02)	0.12 (0.02)	0.16 (0.03)	0.17 (0.04)
	SS 2	0.41 (0.02)	0.43 (0.04)	0.55 (0.07)	0.46 (0.05)
	DS 3	0.20 (0.04)	0.19 (0.03)	0.19 (0.03)	0.19 (0.03)
SL2	**DS 1**	0.57 (0.13)	0.59 (0.14)	1.04 (0.29)	0.92 (0.26)
	SS 1	0.46 (0.07)	0.47 (0.07)	0.45 (0.06)	0.61 (0.11)
	DS 2	0.11 (0.02)	0.11 (0.02)	0.14 (0.02)	0.15 (0.04)
	SS 2	0.42 (0.04)	0.44 (0.06)	0.56 (0.10)	0.48 (0.06)
	DS 3	0.20 (0.04)	0.19 (0.03)	0.19 (0.04)	0.19 (0.03)
SL3	**DS 1**	0.62 (0.22)	0.59 (0.14)	1.06 (0.28)	1.07 (0.28)
	SS 1	0.51 (0.08)	0.50 (0.06)	0.49 (0.08)	0.64 (0.10)
	DS 2	0.10 (0.02)	0.10 (0.02)	0.12 (0.02)	0.12 (0.04)
	SS 2	0.45 (0.04)	0.46 (0.06)	0.61 (0.10)	0.50 (0.07)
	DS 3	0.19 (0.03)	0.19 (0.03)	0.18 (0.03)	0.17 (0.02)

Table 3.3: Mean (± SD) time (in seconds) for each support phase. DS = double limb support; SS = single limb support. DS 1 refers to the time from sway to the first toe-off. SS 1 refers to the time from the lead toe-off to the lead heel strike. DS 2 refers to the time from lead heel strike to trail toe-off. SS 2 refers to the time from trail toe-off to trail heel strike. DS 3 refers to the time from trail heel strike to the lead toe-off.

After the initial heel-strike, a second instance of double-limb support (DS 2) occurred. This interval was also found to be longer with the amputee group ($F_{1,20}$=13.75, p=0.001), but the difference was not as large as that found during the DS1 phase (0.38 sec. difference between the two groups during DS 1; 0.04 sec. difference during DS 2). No leg x group interactions were found.

Following the second instance of double-limb support, the trailing leg went into its initial swing phase (SS 2), which resulted in the leading limb to be responsible for single-limb support.

Not only did the amputee group show a longer support duration than the control group ($F_{1,20}$=15.46, p=0.001), differences between the four leg conditions ($F_{1,20}$=38.98, p<0.001 for the leg x group interaction) were found. This effect occurred due to a prolonged duration of single limb support when the amputees were leading with the intact limb.

Increases to the step length requirement resulted in systematic increases to the movement times for all three temporal measurements (p<0.008 for all variables).

3.4.2 Displacement of the COP

As shown in Figure 3.4, 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 new 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 A-P direction was determined. This point, as defined earlier in Chapter 2 (page 14), was 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 the amputee lead with the 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.

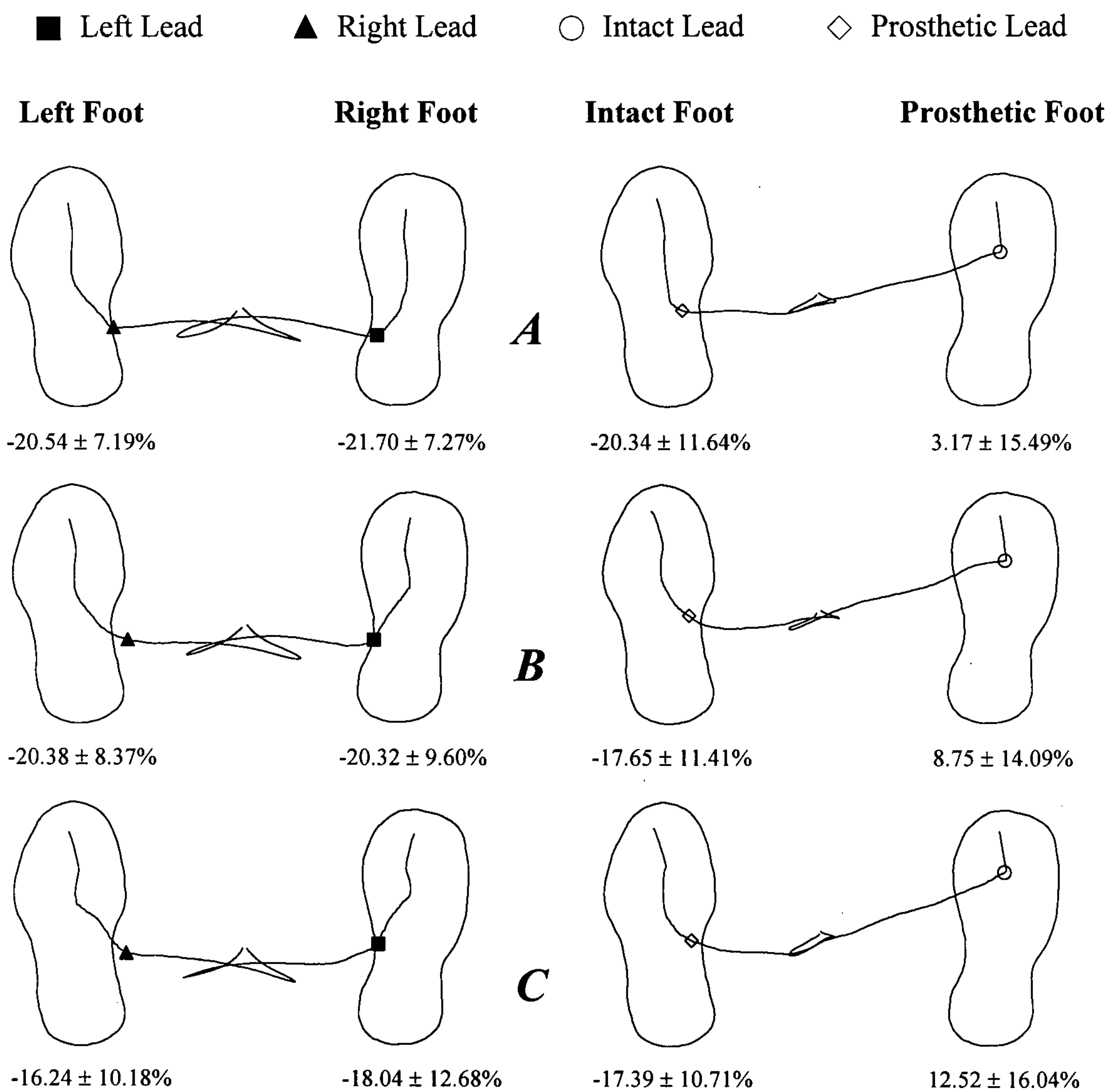


Figure 3.4: Mean displacement and mean (± SD) inflection point of the COP during gait initiation for *A*, the +0% step length condition; *B*, the +25% step length condition; and *C*, the +50% step length condition.

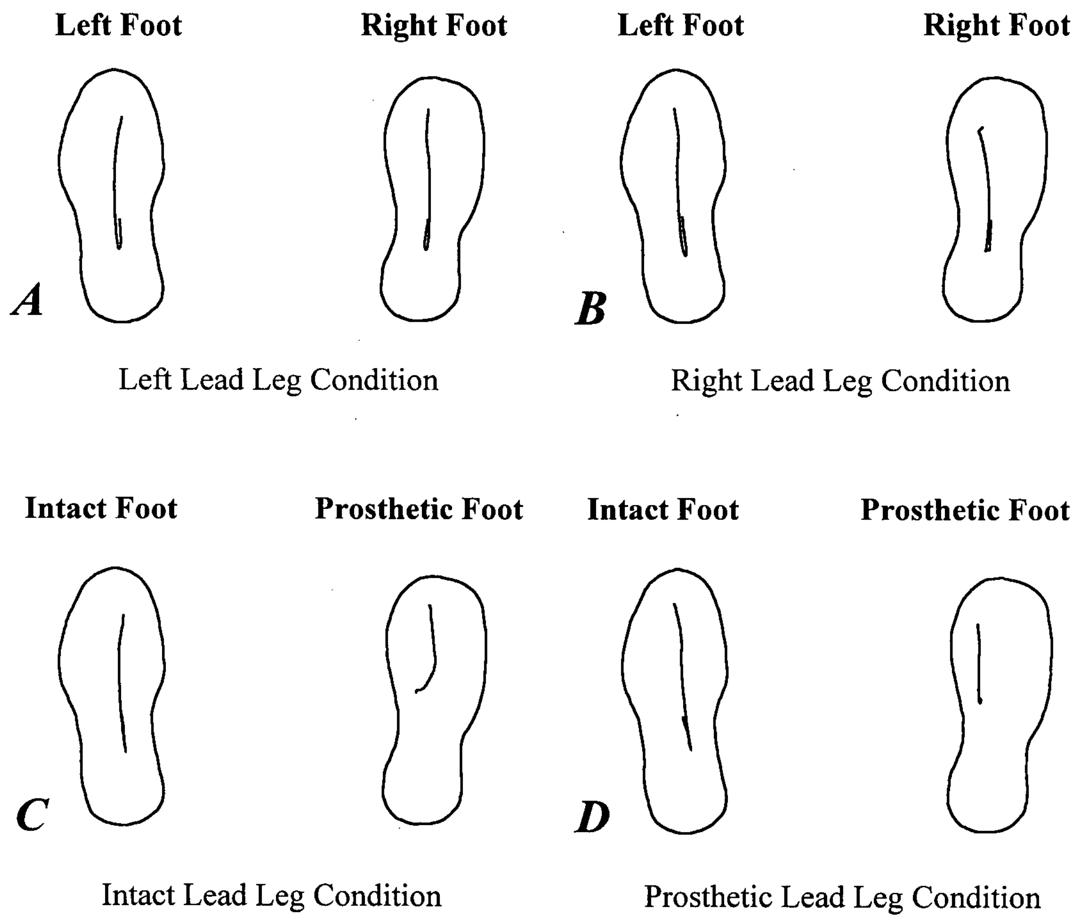
Since data were collected from two force platforms, the movement of the COP under each foot was also monitored (Figure 3.5). In the able-bodied, the magnitude of posterior COP displacement was similar regardless of the foot's responsibility. However, much like the net COP data obtained from both feet, there was significantly less posterior displacement under the prosthetic foot, as compared to the other three feet ($F_{1,20}=19.12$, $p<0.001$ for leg main effect, $F_{1,20}=17.60$, $p<0.001$ for the leg x group interaction).

3.4.3 Ground Reaction Forces

Previous studies have found differences in the application of force during gait initiation in unilateral below-knee amputees. Since the results from this study are in agreement with those obtained from the two previous experiments, most of the force data were not analyzed in detail. However, particular components (the A-P forces from the starting position, the A-P and vertical forces during the landing, and subsequent push-off, of the first step) were examined, as they determined the amount of forward propulsion that was generated by each individual.

a) Forces from the leading limb: Differences were found in the magnitude of peak force that was applied by the leading limb (Figure 3.6). The amputees generated a smaller peak A-P force ($F_{1,20}=4.56$, $p<0.05$) due to the decrease in force application from the prosthetic foot, as compared to the intact, left, and right lead limbs ($F_{1,20}=26.73$, $p<0.001$ for the leg main effect, $F_{1,20}=13.60$, $p=0.001$ for the leg x group interaction). As step length 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).

b) Forces from the trailing limb: The trailing leg also applied differing amounts of force depending on the limb condition (Figure 3.7). Individuals with a unilateral BKA employed a smaller peak magnitude of force ($F_{1,20}=9.08$, $p=0.007$) in the A-P direction than the control

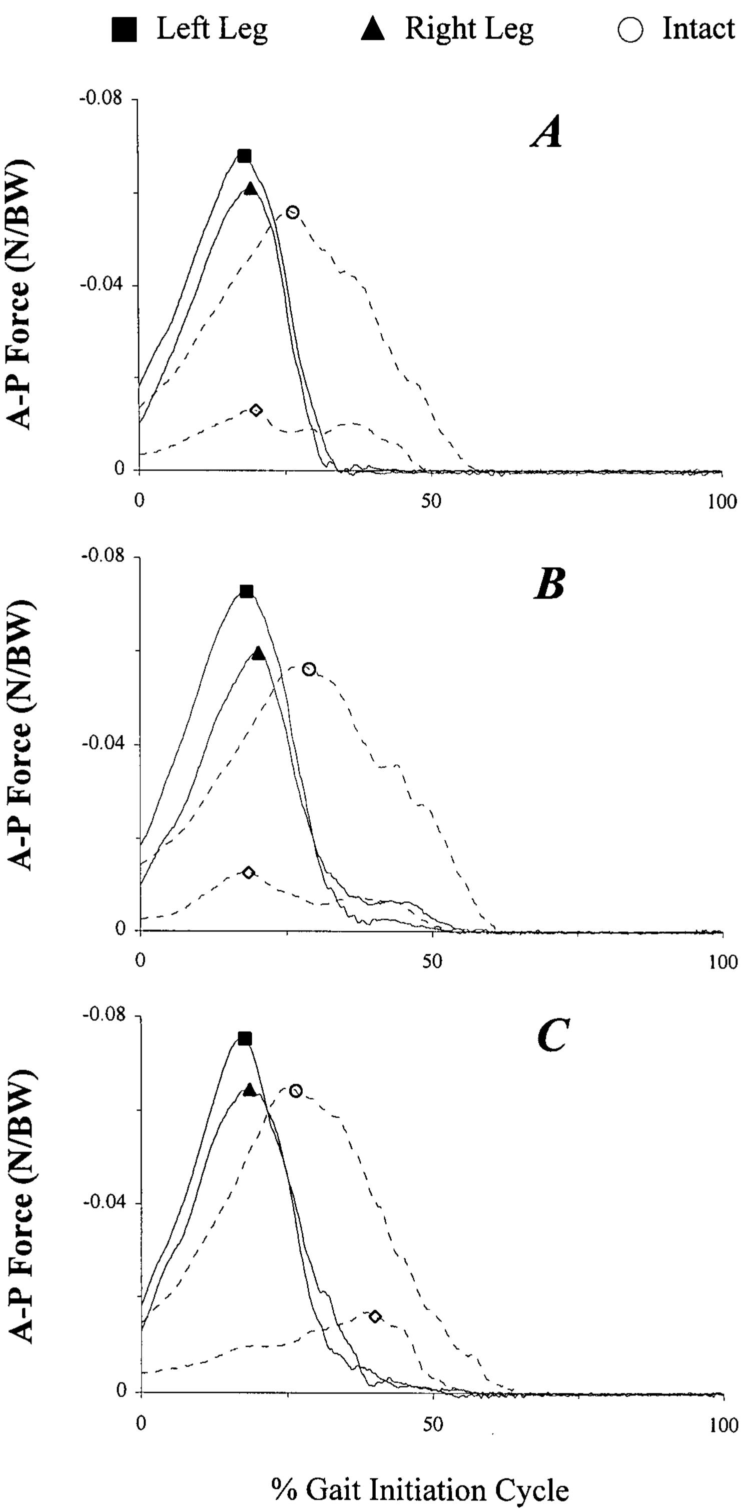


E

WHEN LEADING			
LEG	SL1	SL2	SL3
Left	-23.12 (6.21)	-23.24 (5.57)	-24.11 (6.35)
Right	-23.36 (5.68)	-25.23 (6.00)	-26.04 (5.86)
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			
LEG	SL1	SL2	SL3
Left	-25.37 (4.76)	-24.23 (4.36)	-24.17 (6.40)
Right	-24.42 (5.10)	-24.73 (4.53)	-24.85 (6.81)
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)

Figure 3.5: Mean displacement of the COP under each foot during the +0% step length condition for *A*, the left lead limb condition; *B*, the right lead limb condition; *C*, the intact lead limb condition; and *D*, the prosthetic lead limb condition. The mean (± SD) minimum value (expressed as % foot length) of the vertical displacement of the COP is presented in *E*.

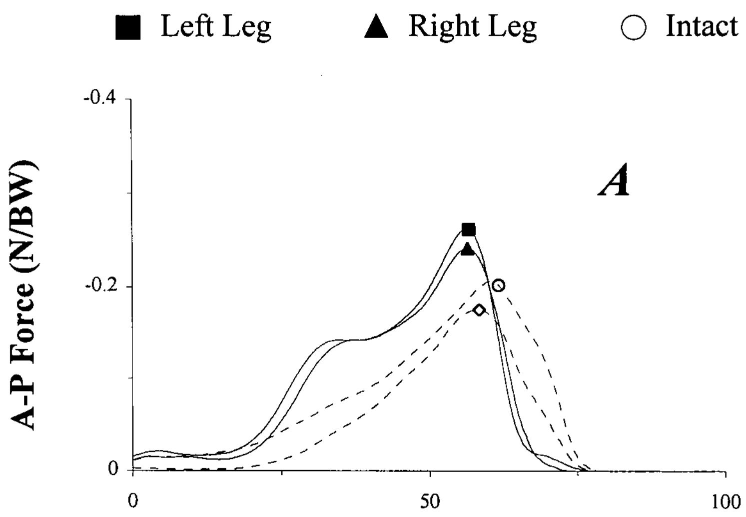


Leg	Mean (± SD) Peak Values
Left	-0.07 (0.02)
Right	-0.06 (0.02)
Intact	-0.07 (0.03)
Prosthetic	-0.02 (0.01)

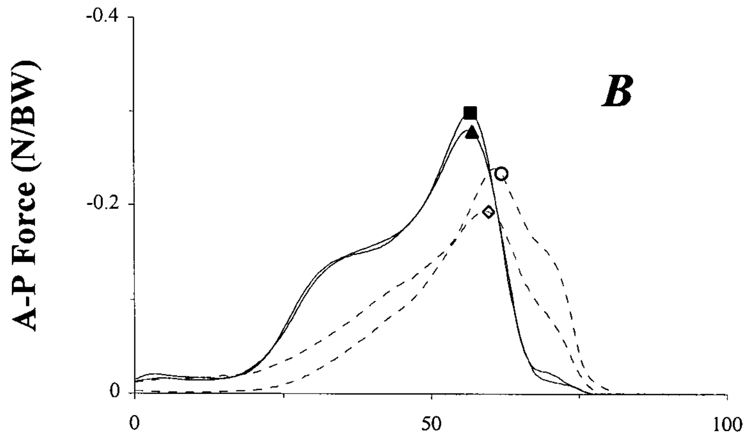
Leg	Mean (± SD) Peak Values
Left	-0.08 (0.03)
Right	-0.06 (0.02)
Intact	-0.07 (0.03)
Prosthetic	-0.02 (0.02)

Leg	Mean (± SD) Peak Values
Left	-0.08 (0.03)
Right	-0.07 (0.03)
Intact	-0.08 (0.03)
Prosthetic	-0.03 (0.02)

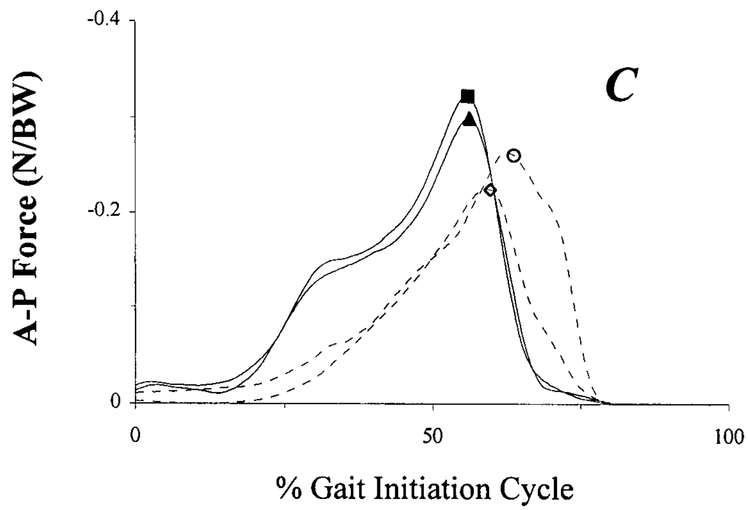
Figure 3.6: Mean (± SD) peak A-P force from the leading limb during *A*, the +0% SL condition; *B*, the +25% SL condition; and *C*, the +50% SL condition. Note that the scale of the y-axes is reversed in direction.



Leg	Mean (± SD) Peak Values
Left	-0.28 (0.03)
Right	-0.26 (0.05)
Intact	-0.23 (0.06)
Prosthetic	-0.20 (0.04)



Leg	Mean (± SD) Peak Values
Left	-0.31 (0.05)
Right	-0.30 (0.06)
Intact	-0.31 (0.08)
Prosthetic	-0.23 (0.04)



Leg	Mean (± SD) Peak Values
Left	-0.35 (0.07)
Right	-0.33 (0.06)
Intact	-0.34 (0.07)
Prosthetic	-0.25 (0.06)

Figure 3.7: Mean (± SD) peak A-P force from the trailing limb during *A*, the +0% SL condition; *B*, the +25% SL condition; and *C*, the +50% SL condition. Note that the scale of the y-axes is reversed in direction.

group. A leg x group interaction did not occur, suggesting that there were no differences between the intact and prosthetic limbs. As step length increased, the applied force significantly increased ($F_{2,40}$=65.59, p<0.001) by an equal amount by the two groups of subjects.

c) Forces after the first step: Mean values of the peak A-P and vertical ground reaction forces from the landing, and subsequent push-off, of the first step are shown in Figures 3.8 and 3.9. Although there were no significant differences between the two groups in the magnitude of braking and propulsion forces, a leg x group interaction was found ($F_{1,20}$=17.83, p<0.001 for peak braking; $F_{1,20}$=45.985, p<0.001 for peak propulsion). This interaction occurred due to a decrease in force application by the prosthetic limb, as compared to the intact, left and right limbs. Both the 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) as the step length requirement became longer.

A significant group effect was not found when analyzing the peak magnitudes of the vertical ground reaction force. However, a leg x group interaction was measured when examining the magnitude of 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 to the other three limb conditions.

3.4.4 Horizontal Impulse

The horizontal impulses, as defined previously in Chapter 2 (page 14), were calculated from both the leading and trailing limbs. For both of these conditions, it was found that the

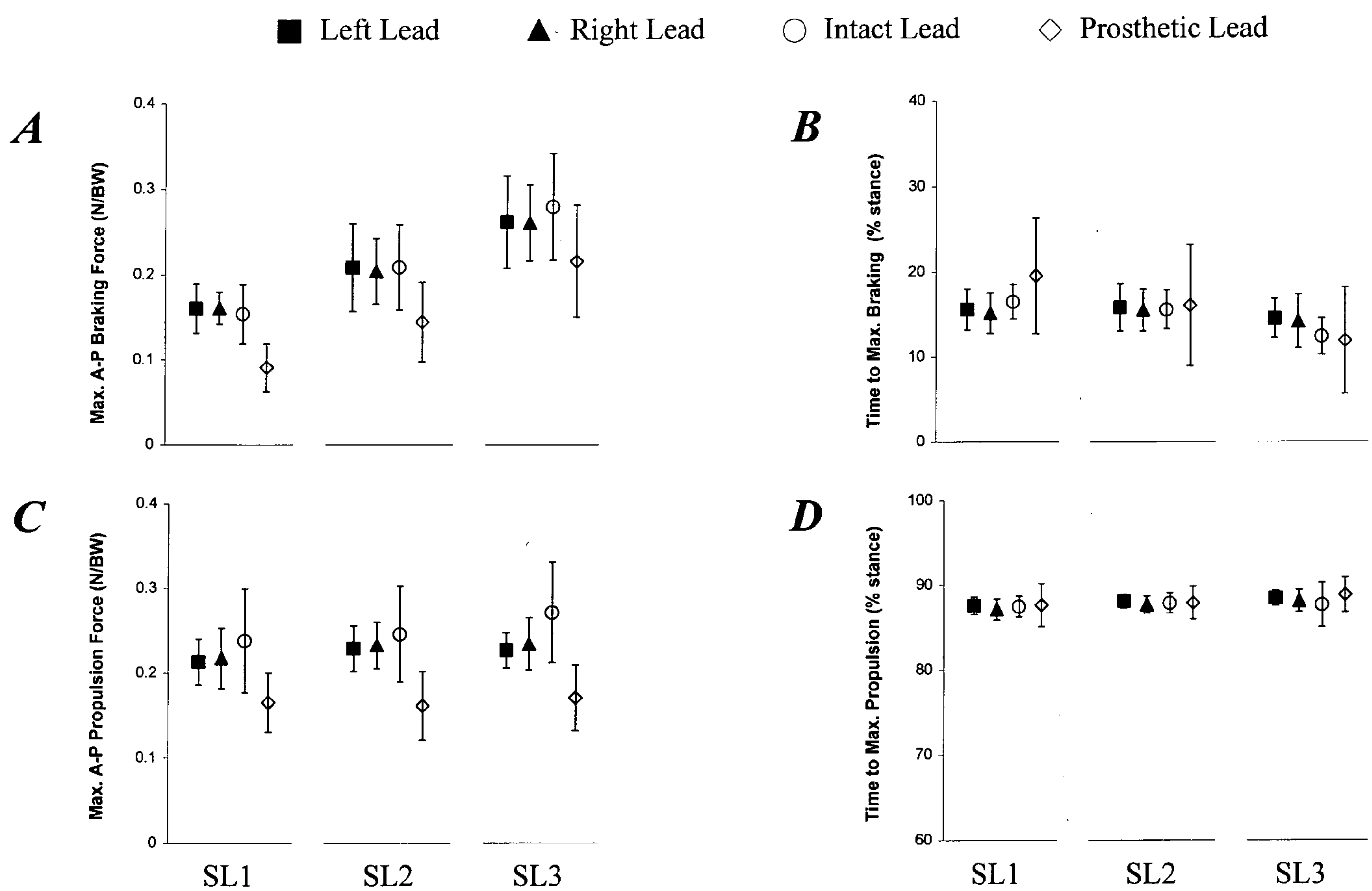


Figure 3.8: Mean (± SD) peak A-P force for each leg and step length conditions during *A*, braking; and *C*, propulsion. Mean (± SD) time when the peak A-P force occurred for *B*, braking; and *D*, propulsion.

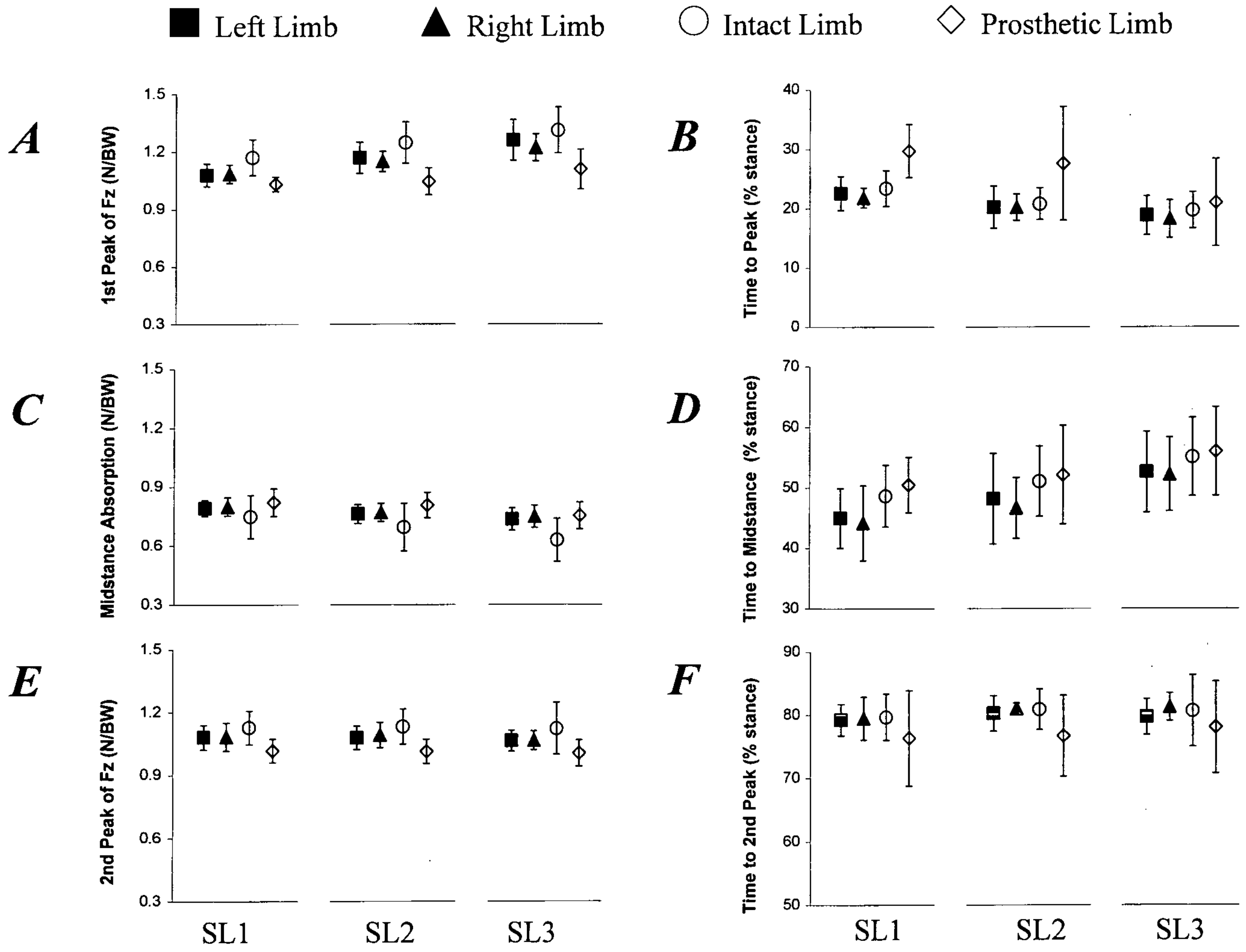


Figure 3.9: Mean (± SD) peak vertical reaction force for the three step length conditions during *A*, weight acceptance; C, midstance absorption; and E, push-off. *B*, *D*, and *F*, show the mean (± SD) timing of these events.

difference in the applied horizontal impulse between the intact and prosthetic limbs was greater than the difference between the left and right control limbs ($F_{1,20}$=12.91, p=0.002 for the lead limb x group interaction; $F_{1,20}$=12.89, p=0.002 for the trail limb x group interaction) (Figure 3.10). This occurred because the impulse generated from the prosthetic limb was less than the control conditions, while the intact limb generated a greater impulse than the controls. A step length effect was found for both the lead ($F_{2,40}$=45.60, p<0.001) and trail ($F_{2,40}$=256.71, p<0.001) limb conditions.

For each trial, a sum of the horizontal impulses was also calculated. This involved the addition of the impulse generated by the leading limb with that applied by the trailing limb. Despite the differences found in the lead and/or trail limbs between the four (left, right, intact and prosthetic) limb conditions, the results indicated that 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 a similar amount ($F_{2,40}$=281.38, p<0.001 for step main effect; step x group interaction was not significant).

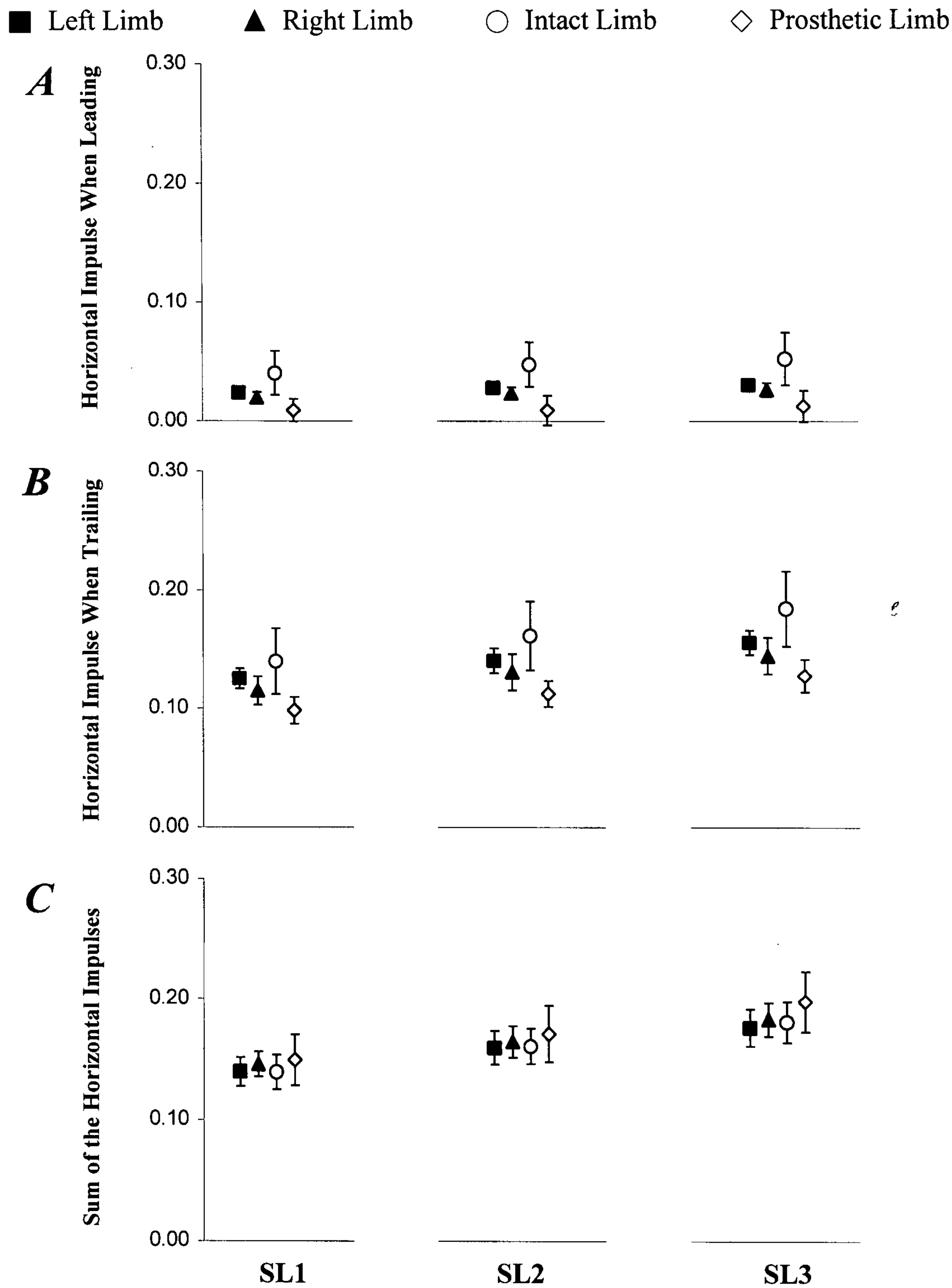


Figure 3.10: Mean (± SD) horizontal impulse, expressed as Ns/N, from *A*, the leading limb; and *B*, the trailing limb. The mean (± SD) sum of the horizontal impulses (i.e. the impulse from the leading leg plus the trailing leg) is shown in *C*.

CHAPTER 4: DISCUSSION

4.1 Overview:

The present study examined the postural and movement strategies of persons with a unilateral below-knee amputation during gait initiation. Because these individuals have structurally and functionally asymmetrical limbs, it was hypothesized that two unique strategies would appear, and that they would be dependent upon the roles and abilities of each limb. It was also hypothesized that with increasing step length requirements, the amputees would exhibit greater asymmetries. After careful analysis of the results, the findings did indeed support the hypotheses.

In general, the duration of the entire task was found to be significantly longer with the amputee group, as compared to the controls. This increase was primarily attributed to a prolonged initial double limb support phase (i.e. the period until the lead limb started its initial swing phase). This interval is believed to lengthen as it a) allowed for a greater impulse to be generated by both the lead and trail limbs and b) did not create as rapid of a disturbance to the amputee. Both are thought to be required in order to compensate for the decreased ability to generate forward propulsion and the diminished stability at the stump-socket interface of the prosthetic limb.

4.2 Standing Posture:

Individuals with a unilateral BKA did not exhibit significant differences in many of the variables examined during upright stance as compared to their able-bodied counterparts. The lone exception was the within-trial variability of the COP, as the amputees underwent smaller oscillations in the medio-lateral direction on their prosthetic limb. A likely reason for this occurrence is simply due to the properties of the prosthetic ankle, as this device cannot actively invert and evert using physiological muscle (Vittas et al., 1984). Thus, it would be reasonable not to expect much change in this foot's COP.

Other measurements of interest, such as the position of the COP and the limb load asymmetry, were not statistically different between group and/or limb conditions. This is in contrast with much of the current literature, which suggests that persons with a unilateral BKA tend to favour the intact limb during upright stance (Summers et al., 1987; Friberg, 1984; Lord and Smith, 1984; Engsberg et al., 1992), experience greater amounts of body sway (Fernie and Holliday, 1978; Geurts et al., 1992; Isakov et al., 1992), rely more heavily on the forefoot aspect of the prosthesis (Engsberg et al., 1989), and in some cases, stand with a kinematically different posture (Engsberg et al., 1991). However, similar to Geurts et al.'s (1992) findings, it is quite probable that the high variability observed between the amputee subjects in this study resulted in the lack of statistical significance for the majority of these measurements.

It was not entirely known why these amputees chose to adopt a wide variety of postural strategies. It was speculated that for some, it was a result of a poorly fitted prosthetic limb. Friberg (1984) has found that approximately 85% of unilateral below-knee amputees are fitted with a prosthesis of an incorrect leg length and so, it was likely that at least one of these subjects had experienced a leg length discrepancy, causing an atypical standing posture. It would be of utmost importance to determine in future studies whether an ill fitting prosthetic limb is a key determinant of postural strategy, as this phenomenon can have negative consequences, such as chronic hip and back pain that can then lead to decreased levels of physical activity (Friberg, 1984; Ehde et al., 2001).

Subject characteristics such as the type of prosthesis (Culham et al., 1986; Hsu et al., 1999; Torburn et al., 1990 and 1995), age of the subject (Vittas et al., 1986), level of physical activity (Pinzur et al., 1991) and the reason for amputation (Hermodsson et al., 1994) can have enormous effects on gait and posture. Because these variables differed greatly among the amputee subjects, it was quite likely that the inclusion criterion (i.e. any unilateral below-knee amputee who could walk independently) was much too broad. To resolve this problem, creating sub-groups would help to distinguish between amputees of different backgrounds. The major difficulty with this protocol is that a significantly larger sample size would need to be recruited.

Another alternative would be to calculate a correlation between the level of asymmetry and the individual's "walking ability", since poor amputee walkers are known to have greater asymmetries and a different positioning of the COP than those with better locomotor skills (Summers et al, 1987). A questionnaire, which would need to be much more detailed that the one used for this study, could certainly help in differentiating between postural compensations employed by amputees of varying walking abilities.

4.3 Gait Initiation:

During the gait initiation trials, the most global adaptation occurred with respect to task duration. Similar to Mouchnino et al.'s (1998) study, where individuals with a unilateral BKA could not laterally raise their limbs as fast as the able-bodied, the present study found that amputees required more time to achieve steady state gait than subjects belonging to the control group. Although temporal differences were found during each instance of heel strike and toe off, much of the total difference in task duration was attributed to an increase in the initial phase of double limb support. For example, during the preferred step length condition, 72% of the 0.53-second difference observed in the overall task duration resulted from the increase in time needed for amputees to get the leading toe off of the ground.

It was proposed that individuals with a unilateral BKA adopted this "movement time" strategy in order to counteract the *stability* and *propulsion* deficiencies during gait initiation.

4.3.1 Stability:

Proprioceptive information from the plantar sole plays an important role in the control of posture and gait (Do et al., 1990; Kavounoudias et al., 1998 & 1999). Unfortunately, individuals with a unilateral BKA lose this specific form of sensory feedback from one of their limbs, resulting in a decreased ability to maintain and regulate body movements. This has lead Viton et al. (2000) to recently suggest that amputees need to be more cautious when creating body disequilibrium, as the loss of lower limb musculature and sensory feedback around an ankle joint makes re-adjusting to slight errors difficult. This notion can be used to explain the differences found in task duration between the amputee and control groups.

When individuals with a unilateral BKA lead with the intact limb, they had to completely rely on the prosthesis for the initial phase of single limb support. Because re-adjustments to postural errors are difficult to complete when lacking sensory feedback from the foot, amputees would have needed to be certain that their postural shift was accurate before actually lifting the intact limb off of the ground. Thus during this lead limb condition, the first double limb support phase was found to take the longest to complete. Conversely, when the prosthetic limb was used for the initial step, the immediate body shift towards the prosthetic side was still required. However, during this situation, the amputee was 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 to the control conditions.

The idea of instability during the initial double limb support phase may then explain the compensatory strategy found with respect to the movement of the COP during gait initiation. When amputees led with their prosthetic limb, there were no significant changes 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 not to move as much in the posterior direction and the inflection point, considered the time when forward movement became an important factor, consistently occurred near the fore part of the prosthetic foot. This is distinctly different from the control condition, where this inflection point occurred towards the heel. This change in strategy was also apparent when examining the COP under each foot. Regardless of the lead limb requirement, the amputees exhibited little posterior displacement on the prosthetic side, while the COP on the intact side exhibited the typical range of movement. Despite this obvious difference observed between the limb conditions, these results are in disagreement with those found by Rossi et al. (1995), who found that regardless of the leading limb, the COP in individuals 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 a method to limit the body's disequilibrium. By moving the COP backwards, forward movement and consequently, displacement of the center of gravity occurs (Breniere and Do, 1991). By limiting this displacement, the amputee would have experienced less disequilibrium and hence, diminished the chance of falling or stumbling during gait initiation. Another reason for the change in the movement of the COP may simply be due to the prosthetic foot itself. Depending on the characteristics and model of the prosthetic foot, the properties of the prosthetic ankle may have limited the posterior displacement of the COP. Because of the rigidity in these devices, amputees may not have been able to rock back on their prosthetic heel without having the toes lift off the ground. Since a heel-only stance would be undesirable, the amputees may have simply chosen to rock back using only their intact limb. In three of the amputee subjects, the prosthetic heel was actually not on the platform. Rather, these individuals stood with their knee in a slightly flexed position, thereby lifting the heel off of the ground. 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 stump of the residual leg.

4.3.2 Forward Propulsion:

In agreement with the results obtained by Nissan (1991) and Rossi et al. (1995), 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 produced 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 the control conditions.

To compensate for these considerable reductions, individuals with a BKA would have needed to rely 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 the amputees 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 indicated 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.

To determine whether these compensatory values were sufficient enough to counteract the deficiencies found in the prosthetic limb, a summed impulse was calculated. Magnitudes from the leading and trailing limbs were added, such that this summed value would refer to the total impulse 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 on average, similar preferred step lengths, 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.

When 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 indicated that for all three step length conditions, the amputees were successful in generating large enough impulses with the intact limb to completely compensate for the deficiencies 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 limitations of the prosthetic limb.

4.3.3 Ground Reaction Forces After the First Step

Unlike the two previous studies examining gait initiation in individuals with a unilateral

BKA, the present study incorporated 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 that was applied 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 were also similar to those found during steady state walking (Hermodsson et al., 1994; Sanderson and Martin, 1997).

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 amputees applied a significantly greater load during weight acceptance and push-off, as well as a significantly increased unweighting during mid-stance, as compared to 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 were intriguing and not entirely clear.

When leading with the intact limb, the amputee would have been relying on the prosthetic limb for the initial body support. Unlike the other three trail limb conditions, the prosthetic limb lacks the ankle musculature to actively control ankle dorsiflexion and thus, during these trials, the amputee would likely have been stepping in a "free-fall" like state. Hence, as the initial heel strike of the leading limb occurred, an 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 is likely that this phenomenon is strongly linked to the greater incidence of lower back pain found in the amputee population (Ehde et al., 2001).

As the amputee 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 stump-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 the storage of potential energy in the intact limb. This, in turn, could be used to generate more force and therefore, more forward movement, for its upcoming push-off. Surprisingly, this was not reflected in the joint kinematics, where there were no noticeable differences between the various limb conditions at any of the lower limb joints. It was unfortunate that joint moment data could not be collected during this study, as it could uncover a drastically different hip or knee flexor moment in the intact limb.

As the amputee 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 (Sanderson and Martin, 1997).

CHAPTER 5: CONCLUSIONS AND FUTURE DIRECTIONS

5.1 Conclusions:

The purpose of this study was to investigate the postural and movement adaptations in unilateral below-knee amputees during gait initiation. It was hypothesized that the amputees 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 attributed primarily to one interval: the initial double-limb support phase, where the individual had to, from an upright standing position, lift the stepping foot off the ground. The prolongation of this double limb support phase was thought to occur for two reasons. First, the instability and discomfort of the prosthesis forced the amputees to be more cautious in displacing the COP, as it would be difficult for these individuals to immediately adjust to any errors in motor programming. Second, the increase in stance duration allowed a greater time for force to be applied in the posterior direction. Since the prosthesis could not generate as much force, the intact limb was found to over-compensate, by applying a greater amount of horizontal impulse. This allowed the sum of the impulses for the amputees to be similar to their able-bodied counterparts, which enabled them to successfully step far enough to satisfy the three step length requirements.

Although the amputees were able to initiate gait successfully with either limb, it is important to note that greater changes were made when leading with the intact limb. Task duration was the longest, the movement of the COP was noticeably different, greater impulse compensations were required during the initial step, and a greater peak vertical force was found during the initial heel strike. These results suggest that although being able to lead with either limb may be an important goal, leading with the unsteady prosthetic limb may be a better alternative.

Employing a prosthetic lead limb strategy appeared to be more effective as body support could be maintained with the very stable intact limb during the initial single limb support phase. Second, the impact of a decreased ability to create propulsion on the prosthetic side could be lessened as proportionately, the leading limb was not required to produce as much of a horizontal impulse as the trailing limb. Finally, as the initial heel strike occurred, the amputee was able to control the rate of downward movement with the trailing limb, thereby reducing the impact of landing.

5.2 Recommendations for Future Work:

Should a follow-up study be undertaken, a few changes to the experimental protocol should be implemented. First and foremost, the group characteristics need to be addressed. Due to time constraints, all unilateral below-knee amputees were allowed to volunteer. However, this introduced a high degree of variability in many of the results. Hence, it is strongly suggested to implement criteria, such as foot type, walking ability, reason for amputation, and time since amputation. If, however, the number of individuals increases dramatically, it would be of interest to sub-divide the amputees based on the above criteria. Some of these sub-groups, such as vascular patients with little walking experience, may exhibit greater compensations than those observed in this study.

Another area of concern was the random and unspecified presentation of the lead limb requirement. Although the subjects were instructed not to anticipate the onset and selection of the visual cue, many individuals inevitably did so. As a result, there were trials in which an individual had hesitated. This then caused a highly variable onset of sway, EMG activity and force application. To counteract this problem, it is suggested that the lead limb requirement be presented to the individual in a "blocked" fashion, prior to the start of each trial. It may also be desirable to implement a "reaction time" component to this task, as this may further distinguish the temporal adaptations found in this study.

Lastly, the strategy to prolong task duration warrants further investigation. For a task like gait initiation, the ability and need to walk immediately from a standing position may not be a common occurrence in daily life. However, there will be instances when an unexpected perturbation occurs, such that changes in posture and/or movement need to be completed as rapidly as possible. It would therefore be of interest to examine whether and how individuals with a unilateral below-knee amputation could successfully react to this type of disturbance.

CHAPTER 6: REFERENCES

1. Alberta Amputee Sport and Recreation Association (2001). Frequently asked questions. Retrieved December 20, 2001, from http://www.aasra.ab.ca/faq.htm.

2. Aruin AS, Nicholas JJ, Latash ML. Anticipatory postural adjustments during standing in below-the-knee amputees. Clinical Biomechanics. 12: 52-59, 1997.

3. Breniere Y, Bril B. Development of postural control of gravity forces in children during the first 5 years of walking. Experimental Brain Research. 121: 255-262, 1998.

4. Breniere Y, Dietrich G. Heel-off perturbation during gait initiation: biomechanical analysis using triaxial accelerometry and a force plate. Journal of Biomechanics. 25: 121-127, 1992.

5. Breniere Y, Do MC. Control of gait initiation. Journal of Motor Behavior. 23: 235-240, 1991.

6. Breniere Y, Do MC. When and how does steady state gait movement induced from upright posture begin? Journal of Biomechanics. 19: 1035-1040, 1986.

7. Breniere Y, Do MC, Bouisset S. Are dynamic phenomena prior to stepping essential to walking? Journal of Motor Behavior. 19: 62-76, 1987.

8. Brunt D, Lafferty MJ, Mckeon A, Goode B, Mulhausen C, Polk P. Invariant characteristics of gait initiation. American Journal of Physical Medicine and Rehabilitation. 70: 206-212, 1991.

9. Brunt D, Liu SM, Trimble M, Bauer J, Short M. Principles underlying the organization of movement initiation from quiet stance. Gait and Posture. 10: 121-128, 1999.

10. Brunt D, Short M, Trimble M, Liu SM. Control strategies for initiation of human gait are influenced by accuracy constraints. Neuroscience Letters. 285: 228-230, 2000.

11. Carlsoo S. The initiation of walking. Acta Anatomica. 65: 1-9, 1966.

12. Crenna P, Frigo C. A motor programme for the initiation of forward-oriented movements in humans. Journal of Physiology. 437: 635-653, 1991.

13. Couillandre A, Breniere Y, Maton B. Is human gait initiation program affected by a reduction of the postural basis? Neuroscience Letters. 285: 150-154, 2000.

14. Culham EG, Peat M, Newell E. Below-knee amputation: a comparison of the effect of the SACH foot and single axis foot on electromyographic patterns during locomotion. Prosthetics and Orthotics. 10: 15-22, 1986.

15. Czerniecki JM, Gitter A. Insights into amputee running: a muscle work analysis. American Journal of Physical Medicine and Rehabilitation. 71: 209-218, 1992.

16. Do MC, Bussel B, Breniere Y. Influence of plantar cutaneous afferents on early compensatory reactions to forward fall. Experimental Brain Research. 79: 319-324, 1990.

17. Ehde DM, Smith DG, Czerniecki JM, Campbell KM, Malchow DM, Robinson LR. Back pain as a secondary disability in persons with lower limb amputations. Archives of Physical Medicine and Rehabilitation. 82: 731-734, 2001.

18. Engsberg JR, Allinger TL, Harder JA. Standing pressue distribution for normal and below-knee amputee children. Prosthetics and Orthotics International. 13: 152-155, 1989.

19. Engsberg JR, Aldridge KC, Harder JA. Lower limb intersegmental forces for below-knee amputee children during standing. Prosthetics and Orthotics International. 15: 185-191, 1991.

20. Engsberg JR, Tedford KG, Springer MJN, Harder JA. Weight distribution of below-knee amputee and able-bodied children during standing. Prosthetics and Orthotics International. 16: 200-202, 1992.

21. Fernie GR, Holliday PJ. Postural sway in amputees and normal subjects. The Journal of Bone and Joint Surgery. 60-A: 895-898, 1978.

22. Friberg O. Biomechanical significance of the correct length of lower limb prostheses: a clinical and radiological study. Prosthetics and Orthotics International. 8: 124-129, 1984.

23. Geurts ACH, Mulder TW, Nienhuis B, Rijken RAJ. Postural reorganization following lower limb amputation. Scandinavian Journal of Rehabilitation Medicine. 24: 83-90, 1992.

24. Geurts ACH, Mulder TW, Nienhuis B, Rijken RAJ. Dual-Task assessment of reorganization of postural control in persons with lower limb amputation. Archives of Physical Medicine and Rehabilitation. 72: 1059-1064, 1991.

25. Hermodsson Y, Ekdahl C, Persson BM, Roxendal G. Gait in male trans-tibial amputees: a comparative study with healthy subjects in relation to walking speed. Prosthetics and Orthotics International. 18: 68-77, 1994.

26. Hermodsson Y, Ekdahl C, Persson BM, Roxendal G. Standing balance in trans-tibial amputees following vascular disease or trauma: a comparative study with healthy subjects. Prosthetics and Orthotics International. 18: 150-158, 1994.

27. Hsu MJ, Nielsen D, Yack HJ, Shurr D. Physiological measurements of walking and running in people with transtibial amputations with 3 different prostheses. Journal of Orthopaedic and Sports Physical Therapy. 29: 526-533, 1999.

28. Isakov E, Mizrahi J, Susak Z, Onna I. A Swedish knee-cage for stabilizing short below-knee stumps. Prosthetics and Orthotics International. 16: 114-117, 1992.

29. Isakov E, Burger H, Krajnik J, Gregoric M, Marincek C. Double-limb support and step-length asymmetry in below-knee amputees. Scandinavian Journal of Rehabilitation Medicine. 29: 75-79, 1997.

30. Isakov E, Keren O, Benjuya N. Trans-tibial amputee gait: time-distance parameters and EMG activity. Prosthetics and Orthotics International. 24: 216-220, 2000.

31. Kavounoudias A, Roll R, Roll JP. The plantar sole is a 'dynamometric map' for human balance control. NeuroReport. 9: 3247-3252, 1998.

32. Kavounoudias A, Roll R, Roll JP. Specific whole-body shifts induced by frequency-modulated vibrations of human plantar soles. Neuroscience Letters. 266: 181-184, 1999.

33. Ledebt A, Bril B, Breniere Y. The build-up of anticipatory behavior: An analysis of the development of gait initiation in children. Experimental Brain Research. 120: 9-17, 1998.

34. Lepers R, Breniere Y, Maton B. Changes to the gait initiation programme following a running exercise in human subjects. Neuroscience Letters. 260: 69-73, 1999.

35. Lord M, Smith DM. Foot loading in amputee stance. Prosthetics and Orthotics International. 8: 159-164, 1984.

36. Madeleine P, Voigt M, Arendt-Nielsen L. Reorganisation of human step initiation during acute experimental muscle pain. Gait and Posture. 10: 240-247, 1999.

37. Maki BE, McIlroy WE. The role of limb movements in maintaining upright stance: the "change-in-support" strategy. Physical Therapy. 77: 488-507, 1997.

38. Malouin F, Richards CL. Preparatory adjustments during gait initiation in 4-6 year-old children. Gait and Posture. 11: 239-253, 2000.

39. Mann R, Hagy JL, White V, Liddell D. The initiation of gait. Journal of Bone and Joint Surgery. 61-A: 232-239, 1979

40. Mille ML, Mouchnino L. Are human anticipatory postural adjustments affected by a modification of the initial position of the center of gravity? Neuroscience Letters. 242: 61-64, 1998.

41. Miller CA, Verstraete MC. A mechanical energy analysis of gait initiation. Gait and Posture. 9: 158-166, 1999.

42. Mouchnino L, Mille ML, Cincera M, Bardot A, Delarque A, Pedotti A, Massion J. Postural reorganization of weight-shifting in below-knee amputees during leg raising. Experimental Brain Research. 121: 205-214, 1998.

43. Nissan M. The initiation of gait in lower limb amputees: some related data. Journal of Rehabilitation Research and Development. 28: 1-12, 1991.

44. Nissan M, Whittle MW. Initiation of gait in normal subjects: a preliminary study. Journal of Biomedical Engineering. 12: 165-171, 1990.

45. Nolan L, Lees A. The functional demands on the intact limb during walking for active trans-femoral and trans-tibial amputees. Prosthetics and Orthotics International. 24: 117-125, 2000.

46. Northwestern University Prosthetics-Orthotics Center (2001). Prosthetics information page. Retrieved December 20, 2001, from http://www.nupoc.northwestern.edu/pros_info.shtml.

47. Pinzur MS, Asselmeier M, Smith D. Dynamic electromyography in active and limited walking below-knee amputees. Orthopedics. 14: 535-538, 1991.

48. Rossi SA, Doyle W, Skinner HB. Gait initiation of person with below-knee amputation: the characterization and comparison of force profiles. Journal of Rehabilitation Research and Development. 32: 120-127, 1995.

49. Sadeghi H, Allard P, Duhaime M. Muscle power compensatory mechanisms in below-knee amputee gait. American Journal of Physical Medicine and Rehabilitation. 80: 25-32, 2001.

50. Sanderson DJ, Martin PE. Lower extremity kinematic and kinetic adaptations in unilateral below-knee amputees during walking. Gait and Posture. 6: 126-136, 1997.

51. Summers GD, Morrison JD, Cochrane GM. Amputee walking training: a preliminary study of biomechanical measurements of stance and balance. International Disability Studies. 10: 1-5, 1987.

52. Tokuno CD, Sanderson DJ, Inglis JT. Center of pressure profiles during gait initiation in unilateral below-knee amputees. In: Control of Posture and Gait 2001, pp.314-317. Maastricht, Netherlands: International Society of Postural and Gait Research.

53. Torburn L, Perry J, Ayyappa E, Shanfield S. Below-knee amputee gait with dynamic elastic response prosthetic feet: a pilot study. Journal of Rehabilitation Research and Development. 27: 369-384, 1990.

54. Torburn L, Powers CM, Guiterrez R, Perry J. Energy expenditure during ambulation in dysvascular and traumatic below-knee amputees: a comparison of five prosthetic feet. Journal of Rehabilitation Research and Development. 32: 111-119, 1995.

55. Viton JM, Timsit M, Mesure S, Massion J, Franceschi JP, Delarque A. Asymmetry of gait initiation in patients with unilateral knee arthritis. Archives of Physical Medicine and Rehabilitation. 81: 194-200, 2000.

56. Vittas D, Larsen TK, Jansen EC. Body sway in below-knee amputees. Prosthetics and Orthotics International. 10: 139-141, 1986.

57. Winter DA. Biomechanics and motor control of human gait: normal, elderly, pathological. 2nd edition. Univ. Waterloo Press.

APPENDIX A: LITERATURE REVIEW

While gait initiation has been thoroughly examined in the able-bodied population, data from individuals with a unilateral below-knee amputation have been practically non-existent. This is surprising since the amputee's asymmetrical limbs will undoubtedly change the biomechanics of this movement. Thus, to gain insight into this particular task, a detailed review of the literature is presented below. Studies involving unilateral below-knee amputees during upright standing, steady-state gait, and obstacle avoidance have also been incorporated as they may provide further information about the characteristics of amputee locomotion.

A.1 Definition of Gait Initiation:

Gait initiation is comprised of the transitional period between upright standing and steady-state gait. Unfortunately, this is a fairly vague definition and thus, it has resulted in a wide variety of different criteria when quantifying the exact duration of this movement.

To signal the start of initiation, most (Brunt et al., 1999; Breniere and Bril, 1998; Ledebt et al., 1998; Nissan and Whittle, 1990; Mann et al., 1979) have simply used the stimulus signal as their point of reference, while others have used one aspect of the kinematic or kinetic data to more accurately define this starting point. The use of the stimulus as the onset of movement is beneficial as it is very easy to detect. However, the subsequent analysis of the data relies heavily on the individual's reaction time to this signal, resulting in a large variability in the timing of events across trials.

On the other hand, when relying on a kinematic or kinetic event, the start of sway has typically referred to a criterion such as the start of lateral malleolus marker movement (Mille and Mouchnino, 1998), the velocity of the center of pressure (COP) being greater than 0.01 m/s

(Madeleine et al, 1999.), a change in baseline COP greater than 4 mm (Maki and McIlroy, 1997) or a change in COP that is 3 standard deviations from the average baseline value (Malouin and Richards, 2000). These standards are much more accurate and consistent from a trial-to-trial basis than the simple use of a stimulus signal. Although a slight discrepancy between studies may occur at the beginning of the movement depending upon the definition used, the majority of problems do not occur here.

Instead, significant concerns arise when defining the end of gait initiation, the time when steady-state gait is thought to occur. Some have set the end of initiation at toe-off of the stance limb (Nissan & Whittle, 1990; Malouin and Richards, 2000) or even a bit later, at heel-strike of the trailing limb (Mann et al., 1979; Madeleine et al., 1999). Although no arguments were made as to why these specific cut-off points were used, a recent study does support the use of these criteria. Miller and Verstraete (1999) examined the mechanical energy of the body during gait initiation and found that the pattern of energy components became similar by the end of the second step, while magnitudes of peak energy levels became comparable to amounts found in steady-state gait by the third step following upright stance. Therefore, to accurately identify the entire gait initiation cycle, it appears that the sampling duration should occur until at least the end of the second step, and possibly until the third.

However, it is much more common for researchers to use the first heel-strike of the swing leg (i.e. end of the first step) to define the completion of gait initiation (Breniere and Dietrich, 1992; Ledebt et al., 1998; Lepers et al., 1999). This is primarily based on the work by Breniere and Do (1986), who found that the velocity of the body's center of mass (COM) after the first step did not differ from the velocity during the second and third steps. Further, these velocities were found to be remarkably similar to those obtained in studies examining steady-state gait.

Although this criterion may be acceptable when testing the general population, this assumption may not hold true when examining special groups of interest. Children learning to walk, for example, do not achieve steady-state velocity by the end of their initial step, but do so sometime during their second and third steps (Ledebt et al., 1998; Malouin and Richards, 2000). By limiting data analysis only to the end of the first step, components unique to children may possibly be overlooked.

Similar problems may arise when examining persons with a unilateral BKA. Since these individuals are known to have slightly slower steady-state velocities (Hermodsson et al., 1994; Hsu et al., 1999), problems associated with prosthesis stability and/or forward propulsion may also cause these individuals to take a slightly longer time to achieve steady-state velocity. Hence, gait initiation may not actually finish until the second or third steps. Since the present study focused on the below-knee amputee population, a longer time interval was used to ensure that any potential adaptations could be observed.

A.2 Movement Description:

Gait initiation is made up of two major phases: an anticipatory postural phase and a movement execution phase.

A.2.1 Postural Phase:

When an individual is standing upright, small oscillations in EMG activity, body kinematics and/or joint kinetics may be observed. But for the most part, these measures will remain relatively stable since the person is at rest and in equilibrium. However, once the decision to initiate gait occurs, a complex but extremely co-ordinated set of neuromuscular adjustments are made. These are needed to minimize the amount of disturbance that will be induced by the subsequent movement, but at the same time, enhance the objectives of gait initiation (Crenna and Frigo, 1991).

An anticipatory postural adjustment (APA) is first observed in the muscles located in the lower limbs. The soleus of both limbs initially deactivate (Carlsoo, 1966; Crenna and Frigo, 1991), while a small amount of activity is observed in the hip abductors and quadriceps of the swinging limb (Mann, 1979). These muscles help create a change in the direction and magnitude of the ground reaction force, where a small increase in force is applied in the posterior direction, as well as a minor thrust laterally towards the swing limb (Nissan and Whittle, 1990). Not only do these changes act to separate COP from the COM, the medial-lateral thrust (M-L) helps move the COM towards the stance limb, thereby promoting body stability in anticipation of forward movement (Maki and McIlroy, 1997).

EMG activity of the same muscles (i.e. the hip abductors and quadriceps) found within the stance limb are observed next, followed by the activation of the tibialis anterior from both the stance and swing limbs (Mann, 1979). This results in a greater amount of force being applied laterally towards the stance limb and again, towards the heels. The extra push in the anterior-posterior (A-P) direction generates a forward lean of the upper body, and consequently, the COM. Because this anterior movement of the COM greatly affects the amount of initial forward movement, in situations when a greater amount of forward movement is needed, a greater amount of posterior displacement of the COP will be observed (Breniere and Do, 1991).

The typical movement of the COP can be found in Figure A1.

A.2.2. Movement Execution Phase:

Once a sufficient amount of body stability has been achieved on the foot of the stance limb, heel-off and subsequently, toe-off of the swing limb can begin to occur. Continued muscle

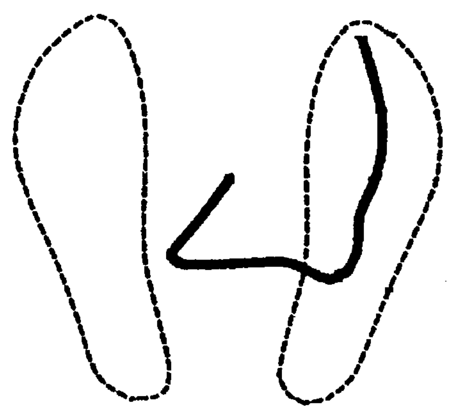


Figure A.1: A typical COP pattern measured during a left lead limb condition. (Reproduced from Mann et al., 1979)

activity is observed in the soleus, hip abductors, hamstrings, and quadriceps of the stance limb (Mann et al., 1979; Carlsoo, 1966), helping to stabilize the body for the upcoming single-limb support phase. This pattern of EMG activity is in agreement with the external kinematics, where the hip and knee joints of the stance limb remain extended, while the ankle begins to dorsiflex (Nissan and Whittle, 1990). This strategy will undoubtedly help to store the potential energy necessary for stance limb push-off, while still maintaining overall body support. Meanwhile, in the lead limb, muscle contractions occur mainly in the soleus (to promote movement via plantarflexion), but are followed by EMG activity in the rectus femoris and tibialis anterior (Mann et al., 1979). The latter two muscles help facilitate knee extension, hip flexion and ankle dorsiflexion during the swing phase, allowing the individual to take as large a step length as deemed necessary.

Once the first step (i.e. heel contact of the lead limb) is complete, double limb support is regained. Now, the stability and movement responsibilities become alternated, such that the trailing limb is used for forward propulsion and the leading limb is relied upon for body support. Changes in muscle activation depict this exchanging of roles. The tibialis anterior, peroneals, hip abductors and quadriceps of the trailing limb increases its EMG activity in a characteristic sequence such that the individual is able to increase his velocity, but at the same time, prepare for the limb's eventual heel strike, where body stability must again be of concern (Mann et al., 1979). It is at around this point in the gait initiation cycle, where steady-state gait is achieved.

A.2.3. The Invariance of the Phases:

Many have attempted to change this characteristic APA sequence by implementing different speed, step length and posture requirements. However, in many cases, the postural strategy was kept fairly constant. For example, Breniere and Do (1991) imposed different step length conditions during initiation, as this distance can be considered one of the constraints acting on the inverted-pendulum model. Although the results showed that there was a significant increase in velocity of the COM at the end of the gait initiation cycle, the duration of the movement remained invariant. Similarly, Brunt et al. (1991) tried to alter the APA by introducing three different speed conditions. Again, the results showed many incidents of task invariance. The occurrence of tibialis anterior activity, the time to toe-off and/or heel strike of the swing limb, and the timing of weight bearing were all found to remain consistent across speed requirements, when expressed as a percentage of the gait initiation cycle. Finally, Lepers et al. (1999) attempted to determine the strength of these invariant features. Subjects were initially instructed to perform a set of running exercises. Immediately following this pre-test, they were measured on the manner in which they initiated gait. Although the final COM velocity increased significantly after the exercise protocol, many of the invariant features (e.g. duration of the postural and execution phases, the amount of posterior displacement in the COP) still remained.

Because of these findings, many have concluded that this movement, especially the postural phase component, is centrally programmed through the use of a motor program. This would suggest that gait initiation could be performed in the absence of sensory feedback and the program remains fairly rigid. Therefore, if needed, individuals with a unilateral BKA should be able to initiate gait using their original motor program.

A.3 Gait initiation in Unilateral Below-Knee Amputees:

Unfortunately, only two published studies have examined the gait initiation movement in persons with a unilateral BKA. The first, conducted by Nissan (1991), examined both the kinematics and kinetics during the initiation movement. The results showed that these individuals undergo similar compensations during the movement execution phase of gait initiation as they do during steady state locomotion. That is, amputees exhibited smaller joint ranges of motion and produce smaller ground reaction forces during the initial push-off phase. These may have been an indication of maintaining a more vertically oriented prosthetic limb position in order to decrease the chance of knee collapse (Sanderson and Martin, 1996) as well as the inability to adequately produce forward propulsion using the prosthetic limb (Menard, McBride and Sanderson, 1992).

Although Nissan's results are interesting, very little information can be garnered about the postural strategy. This is important since several deviations have been found during upright stance. In many cases, a large amount of limb loading asymmetry (ranging from 32% to 45% of total body weight on the prosthetic limb) can be found between the amputee's two limbs (Summers et al., 1987; Lord and Smith, 1984, Engsberg et al., 1992). Those that did not find any significant asymmetries simply attributed it to the large variability between amputee subjects (Geurts et al., 1992). Furthermore, the loading of the feet was also distributed differently, where much of the weight was being applied in the anterior and medial areas of the prosthetic foot, as compared to the normal and intact feet (Engsberg et al, 1989 and 1992; Lord and Smith, 1984).

Together, these results suggest that amputees prefer to rely on the intact limb for standing balance but consequently, experience a decrease in body stability (Geurts et al., 1992; Fernie and Holliday, 1978; Isakov et al., 1992).

Another drawback of Nissan's work was his lack of temporal analysis. Much of his data seemed to indicate the presence of temporal asymmetries between the control, intact and prosthetic lead limb conditions. Had these results been investigated further, the data may have resulted in findings similar to those obtained during steady-state gait or stair ambulation, when amputees spend shorter durations of time in stance phase (Hermodsson et al., 1994; Sanderson and Martin, 1996 and 1997; Powers et al., 1997; Isakov et al., 2000) as well as double-limb support (Isakov et al., 1997) on the prosthetic limb. Since the lack of stability has been a widely accepted reason for these temporal asymmetries during gait, it would be of interest to find these differences during gait initiation as well.

The other study examining below-knee amputees during gait initiation was conducted by Rossi et al. (1995), where the ground reaction forces and the COP profiles were measured. Although they found timing and amplitude differences in the vertical ground reaction force between the intact and prosthetic limbs, the most surprising conclusion came regarding the COP. The results indicated that the path of the COP moved in a similar shape for all of their amputee subjects as compared to the able-bodied, regardless of the choice of leading limb. This would imply that the original version of the postural and movement strategies was still being used.

However, this is unlikely to occur. Keeping the original sequence of muscle activation would not result in the same effects post-amputation. Due to the inability to actively dorsi- and plantarflex the prosthetic ankle, the initial soleus-tibialis anterior EMG sequence would do little in displacing the COP in the anterior-posterior direction. If individuals with a unilateral BKA wanted to maintain their original COP profile in order to facilitate forward movement, compensation from the other residual muscles and/or body kinematics would need to occur. In fact, there is evidence supporting this hypothesis. In a lateral leg raising task, Viton et al. (2000) found increased EMG activity in the proximal muscles of the prosthetic limb, in lieu of the gastrocnemius, to successfully lift the limb to an angle of 45°. Had these compensatory muscles not have activated, task success would not have been possible.

Second, there are known issues regarding the stability and comfort of the prosthesis at the stump-socket interface (Isakov et al., 1992). Hence, it would be unlikely that the COP could move as smoothly and/or as quickly as normal, especially when the prosthetic limb was responsible for the initial single limb support phase. It would be much more plausible to see slight differences not only between an amputee's prosthetic and intact limbs, but especially when compared to a leg belonging to an able-bodied individual. Again, the findings of Viton et al. (2000) support this claim. During the leg extension task, there were significantly longer weight transfer and movement phases in the amputee subjects as compared to the controls. Furthermore, since the RMS of the COP displacement was also significantly less in the amputee population, the results suggest that these individuals are more cautious about the impending movement.

Doubts about the maintenance of the original APA during gait initiation are also supported by a preliminary study (Tokuno et al., 2001). Significant changes in the COP profile prior to, and during the movement execution phase were found in the amputee group, as compared to the able-bodied individuals. It appeared that the movement of the COP, and therefore the APA, were also highly dependent upon the choice of lead limb. Posterior displacement was significantly smaller than the control condition when the prosthetic limb was leading, while there was an absence of posterior movement when the intact limb was leading. This strongly suggests that the amputees had created two unique strategies based on the limitations and abilities of their two distinctly different limbs.

Support for the use of two distinct strategies can also be found during other related tasks, such as obstacle avoidance. During the "step over" phase, the amputees showed a decreased amount of knee flexion on the prosthetic limb, compared to both the intact and control limbs (Hill et al., 1997). As this would limit the individual's ability to successfully avoid the obstacle, there was a concurrent increase in the amount of ankle plantarflexion from the body-supporting intact limb as well as a significant increase in the amount of rotational work produced by the hip joint of the prosthetic limb (Hill et al., 1997). The combined effect was a toe clearance large enough to step over the obstacle, despite the decrease in knee flexion of the prosthetic limb.

Aruin et al. (1997) examined the postural strategies of individuals with a unilateral BKA while subjects performed bilateral shoulder movements during standing. An asymmetry postural strategy was found within the amputee population, where greater EMG activity was observed on the intact side of the body, as compared to the amputated side. This uneven recruitment of muscles resulted in a noticeable lateral shift in the COP towards the intact limb, reinforcing the notion that amputees prefer to bear more weight on the steadier biological limb (Summers et al., 1987; Lord and Smith, 1984, Engsberg et al., 1992). It was suggested that these changes in postural strategy were made based on the unique properties of each limb.

If the results found in the preliminary gait initiation study, as well as the obstacle avoidance and the bilateral shoulder extension studies are indeed accurate, it should indicate that persons with a unilateral BKA would undergo significant changes in their neuromuscular organization, such that the opposite side of the body automatically compensates for the limitations of the prosthesis. Since these postural changes should reflect upon the kinematic and kinetic changes during gross body movements, this information will indeed be valuable resource to those working in the rehabilitation setting.

APPENDIX B: SUBJECT DATA

The following 44 pages contain data from each of the subjects. The presented data include:

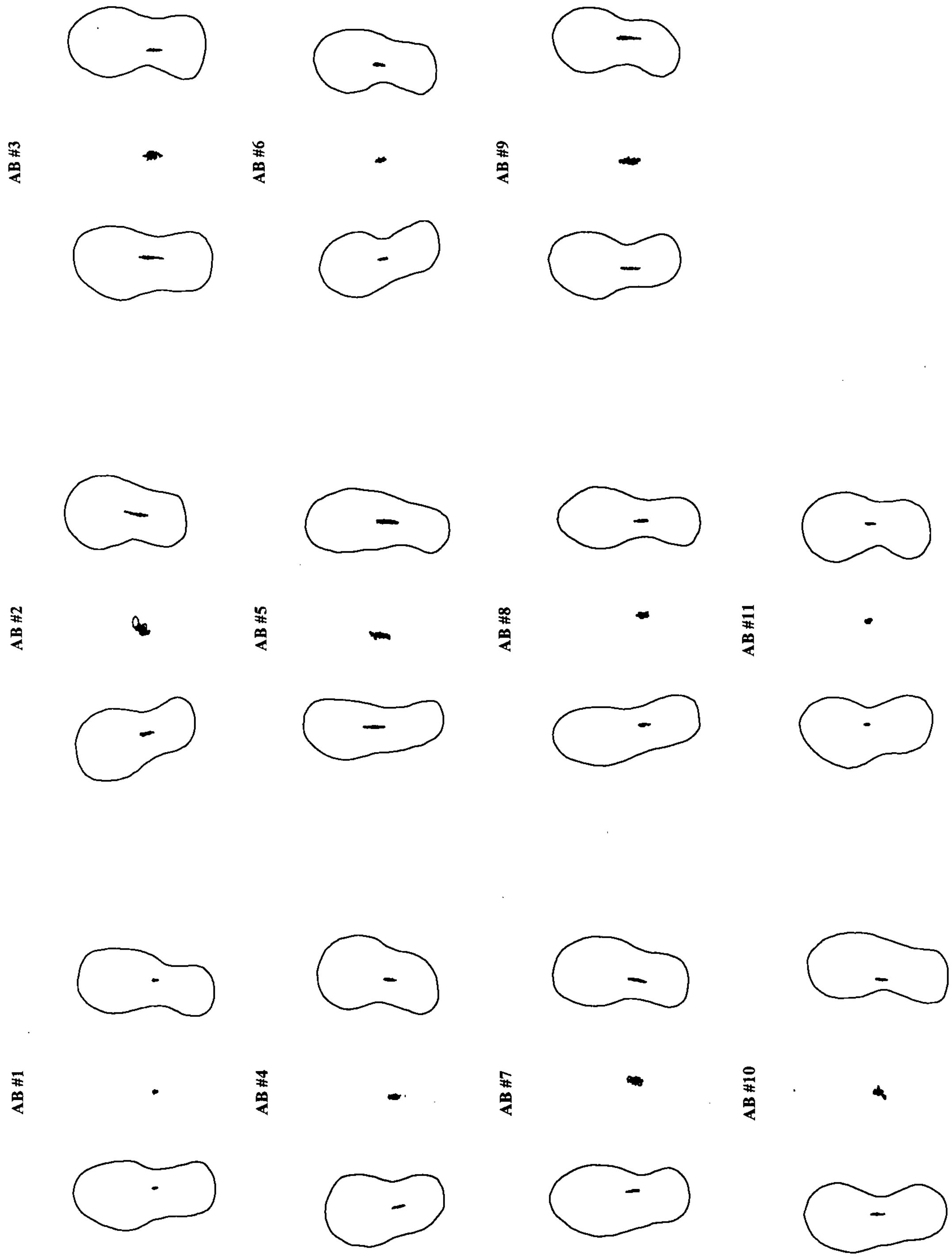
- the center of pressure during the standing trial
- the task duration of the gait initiation trials
- the center of pressure during the gait initiation trials
- the A-P ground reaction forces from the leading and trailing limbs
- the A-P and vertical ground reaction forces from the stepping limb
- the horizontal impulse from the starting position

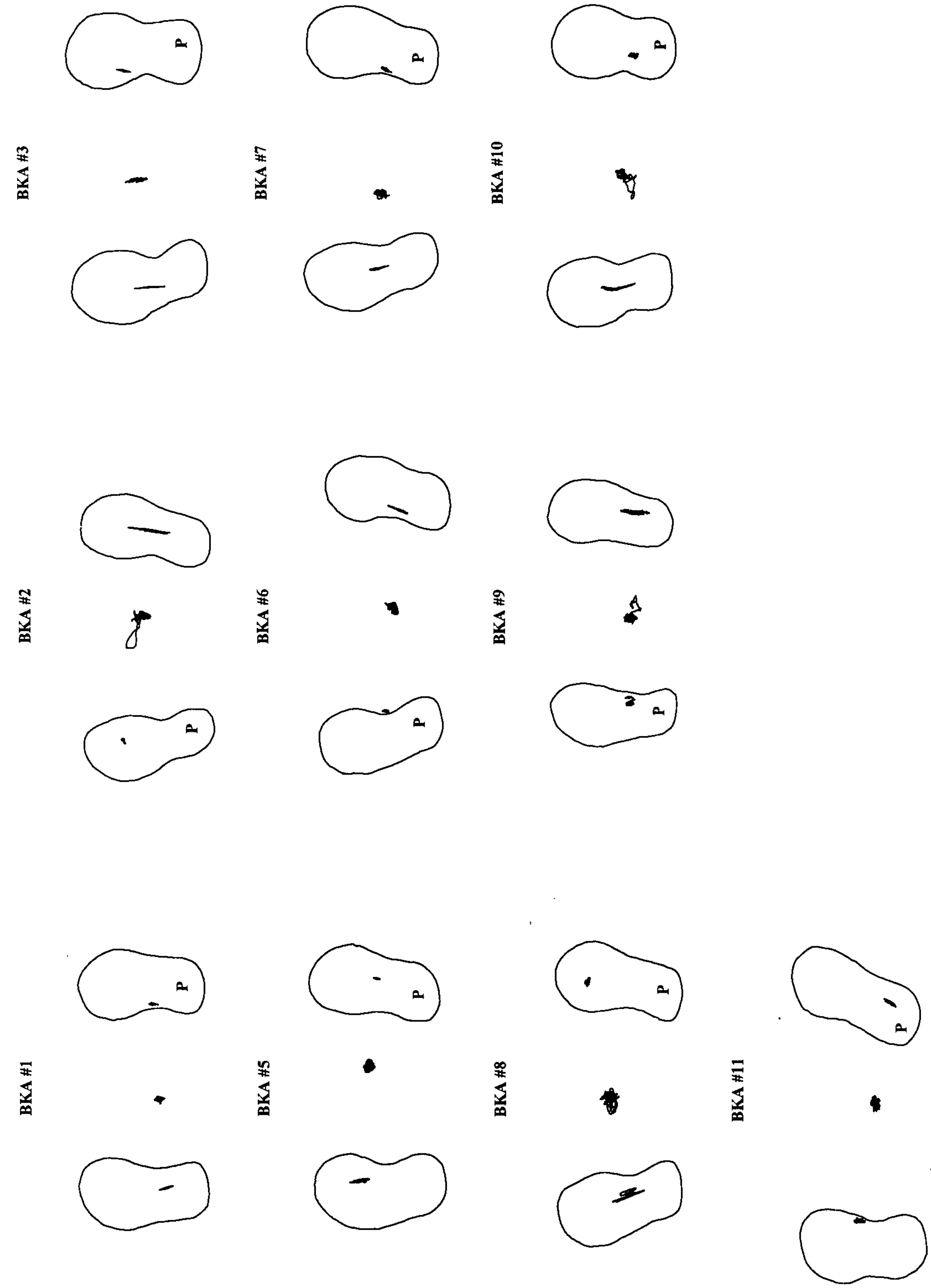
APPENDIX B-1: COP DURING UPRIGHT STANCE

The following two pages contain the position of the center of pressure (from the left and right feet, as well as the combined) for each individual during the one-minute standing trial. Note that BKA #4 did not undergo the standing trial, as the test was implemented at a later date and that for the amputee subjects, the prosthetic side has *not* been converted.

Abbreviations:

- AB = able-bodied (control)
- BKA = unilateral below-knee amputee
- P = prosthetic side


AB #1
AB #2
AB #3
AB #4
AB #5
AB #6
AB #7
AB #8
AB #9
AB #10
AB #11


BKA #1
BKA #2
BKA #3
BKA #5
BKA #6
BKA #7
BKA #8
BKA #9
BKA #10
BKA #11
P

APPENDIX B-2: TASK DURATION

The following six pages contain the mean intervals for each individual at each of the step length conditions, as well as the mean (± SD) interval for each group, during the gait initiation trials.

Abbreviations:

- AB = able-bodied (control)
- BKA = unilateral below-knee amputee
- Sway = start of sway (>4mm shift in the baseline center of pressure)
- TO 1 = initial toe-off of the leading limb
- HS 1 = initial heel-strike of the leading limb
- TO 2 = initial toe-off of the trailing limb
- HS 2 = initial heel-strike of the trailing limb
- TO 3 = second toe-off of the leading limb
- DS 1 = 1st double limb support phase (time from sway to TO 1)
- SS 1 = 1st single limb support phase (time from TO1 to HS 1)
- DS 2 = 2nd double limb support phase (time from HS 1 to TO 2)
- SS 2 = 2nd single limb support phase (time from TO 2 to HS 2)
- DS 3 = 3rd double limb support phase (time from HS 2 to TO 3)

STEP LENGTH 1

Left Leg Leading

	AB #1	AB #2	AB #3	AB #4	AB #5	AB #6	AB #7	AB #8	AB #9	AB #10	AB #11		Average
Sway	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	Sway	0.00
TO 1	0.70	0.56	0.68	0.57	0.47	0.39	0.45	0.56	0.69	0.56	0.54	TO 1	0.56
HS 1	1.21	1.04	1.10	0.94	0.91	0.75	0.91	1.03	1.11	0.98	1.01	HS 1	1.00
TO 2	1.34	1.16	1.21	1.08	1.01	0.88	1.02	1.13	1.23	1.14	1.11	TO 2	1.12
HS 2	1.79	1.59	1.60	1.50	1.40	1.27	1.42	1.52	1.64	1.56	1.55	HS 2	1.53
TO 3	2.02	1.76	1.81	1.72	1.56	1.42	1.69	1.70	1.81	1.80	1.72	TO 3	1.73
DS 1	0.70	0.56	0.68	0.57	0.47	0.39	0.45	0.56	0.69	0.56	0.54	DS 1	0.56
SS 1	0.52	0.48	0.41	0.37	0.44	0.36	0.47	0.47	0.43	0.42	0.47	SS 1	0.44
DS 2	0.12	0.12	0.12	0.14	0.10	0.13	0.11	0.11	0.11	0.17	0.10	DS 2	0.12
SS 2	0.45	0.43	0.39	0.42	0.39	0.40	0.40	0.39	0.41	0.41	0.44	SS 2	0.41
DS 3	0.24	0.17	0.21	0.22	0.16	0.15	0.27	0.17	0.16	0.25	0.17	DS 3	0.20

Right Leg Leading

	AB #1	AB #2	AB #3	AB #4	AB #5	AB #6	AB #7	AB #8	AB #9	AB #10	AB #11		Average
Sway	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	Sway	0.00
TO 1	0.57	0.59	0.57	0.62	0.44	0.39	0.46	0.50	0.64	0.59	0.53	TO 1	0.54
HS 1	1.07	1.13	1.02	0.99	0.90	0.75	0.94	0.97	1.14	1.05	1.02	HS 1	1.00
TO 2	1.19	1.26	1.11	1.11	1.00	0.87	1.06	1.07	1.25	1.21	1.13	TO 2	1.11
HS 2	1.67	1.74	1.55	1.59	1.40	1.24	1.51	1.43	1.67	1.62	1.57	HS 2	1.54
TO 3	1.85	1.91	1.72	1.77	1.56	1.41	1.74	1.64	1.85	1.88	1.76	TO 3	1.74
DS 1	0.57	0.59	0.57	0.62	0.44	0.39	0.46	0.50	0.64	0.59	0.53	DS 1	0.54
SS 1	0.50	0.54	0.45	0.37	0.46	0.36	0.48	0.47	0.50	0.46	0.49	SS 1	0.46
DS 2	0.12	0.13	0.09	0.12	0.10	0.11	0.12	0.10	0.11	0.16	0.11	DS 2	0.12
SS 2	0.48	0.49	0.44	0.48	0.40	0.38	0.45	0.36	0.42	0.40	0.44	SS 2	0.43
DS 3	0.18	0.17	0.17	0.18	0.17	0.17	0.23	0.21	0.18	0.27	0.19	DS 3	0.19

STEP LENGTH 2

Left Leg Leading

	AB #1	AB #2	AB #3	AB #4	AB #5	AB #6	AB #7	AB #8	AB #9	AB #10	AB #11		Average
Sway	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	Sway	0.00
TO 1	0.55	0.52	0.72	0.64	0.54	0.45	0.47	0.48	0.87	0.48	0.58	TO 1	0.57
HS 1	1.12	1.01	1.13	0.99	0.98	0.82	1.00	0.98	1.37	0.87	1.12	HS 1	1.03
TO 2	1.22	1.12	1.24	1.13	1.07	0.94	1.10	1.09	1.47	1.03	1.21	TO 2	1.15
HS 2	1.71	1.52	1.62	1.53	1.47	1.32	1.54	1.46	1.91	1.43	1.67	HS 2	1.56
TO 3	1.95	1.70	1.83	1.77	1.64	1.48	1.81	1.66	2.08	1.63	1.83	TO 3	1.76
DS 1	0.55	0.52	0.72	0.64	0.54	0.45	0.47	0.48	0.87	0.48	0.58	DS 1	0.57
SS 1	0.57	0.49	0.41	0.35	0.44	0.37	0.54	0.50	0.50	0.40	0.53	SS 1	0.46
DS 2	0.11	0.11	0.11	0.14	0.09	0.11	0.10	0.11	0.11	0.15	0.09	DS 2	0.11
SS 2	0.49	0.41	0.38	0.40	0.41	0.39	0.43	0.37	0.44	0.40	0.47	SS 2	0.42
DS 3	0.24	0.18	0.21	0.24	0.17	0.15	0.27	0.19	0.17	0.20	0.15	DS 3	0.20

Right Leg Leading

	AB #1	AB #2	AB #3	AB #4	AB #5	AB #6	AB #7	AB #8	AB #9	AB #10	AB #11		Average
Sway	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	Sway	0.00
TO 1	0.93	0.54	0.70	0.65	0.48	0.48	0.47	0.51	0.64	0.55	0.53	TO 1	0.59
HS 1	1.51	1.05	1.13	1.02	0.98	0.85	1.00	1.00	1.13	0.95	1.05	HS 1	1.06
TO 2	1.61	1.16	1.22	1.14	1.06	0.98	1.10	1.10	1.24	1.08	1.16	TO 2	1.17
HS 2	2.19	1.62	1.63	1.60	1.48	1.36	1.60	1.46	1.69	1.47	1.61	HS 2	1.61
TO 3	2.37	1.78	1.84	1.78	1.65	1.53	1.79	1.68	1.86	1.71	1.80	TO 3	1.80
DS 1	0.93	0.54	0.70	0.65	0.48	0.48	0.47	0.51	0.64	0.55	0.53	DS 1	0.59
SS 1	0.58	0.51	0.44	0.37	0.50	0.37	0.53	0.49	0.49	0.40	0.53	SS 1	0.47
DS 2	0.09	0.11	0.09	0.12	0.09	0.13	0.10	0.10	0.12	0.13	0.11	DS 2	0.11
SS 2	0.58	0.46	0.40	0.46	0.42	0.39	0.50	0.35	0.45	0.39	0.45	SS 2	0.44
DS 3	0.18	0.16	0.21	0.18	0.17	0.17	0.19	0.22	0.17	0.25	0.19	DS 3	0.19

STEP LENGTH 3

Left Leg Leading

	AB #1	AB #2	AB #3	AB #4	AB #5	AB #6	AB #7	AB #8	AB #9	AB #10	AB #11		Average
Sway	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	Sway	0.00
TO 1	0.62	1.00	0.61	0.57	0.44	0.42	0.43	0.47	1.05	0.53	0.66	TO 1	0.62
HS 1	1.17	1.63	1.05	0.97	0.96	0.82	1.04	1.05	1.56	1.02	1.17	HS 1	1.13
TO 2	1.27	1.70	1.14	1.08	1.03	0.94	1.14	1.14	1.67	1.14	1.26	TO 2	1.23
HS 2	1.80	2.11	1.55	1.49	1.48	1.36	1.64	1.54	2.10	1.59	1.75	HS 2	1.67
TO 3	1.98	2.29	1.73	1.72	1.65	1.50	1.87	1.72	2.27	1.83	1.90	TO 3	1.86
DS 1	0.62	1.00	0.61	0.57	0.44	0.42	0.43	0.47	1.05	0.53	0.66	DS 1	0.62
SS 1	0.55	0.63	0.44	0.40	0.52	0.41	0.61	0.57	0.51	0.49	0.50	SS 1	0.51
DS 2	0.10	0.08	0.09	0.12	0.07	0.12	0.10	0.09	0.11	0.13	0.09	DS 2	0.10
SS 2	0.53	0.41	0.41	0.41	0.45	0.41	0.50	0.41	0.44	0.44	0.49	SS 2	0.45
DS 3	0.18	0.18	0.18	0.23	0.17	0.15	0.23	0.18	0.17	0.25	0.15	DS 3	0.19

Right Leg Leading

	AB #1	AB #2	AB #3	AB #4	AB #5	AB #6	AB #7	AB #8	AB #9	AB #10	AB #11		Average
Sway	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	Sway	0.00
TO 1	0.65	0.90	0.66	0.64	0.45	0.38	0.48	0.57	0.53	0.57	0.63	TO 1	0.59
HS 1	1.20	1.44	1.14	1.06	0.95	0.79	1.08	1.05	1.07	1.03	1.19	HS 1	1.09
TO 2	1.30	1.53	1.22	1.17	1.03	0.91	1.18	1.14	1.18	1.16	1.27	TO 2	1.19
HS 2	1.85	2.02	1.65	1.66	1.48	1.33	1.76	1.49	1.62	1.61	1.72	HS 2	1.65
TO 3	2.01	2.19	1.83	1.83	1.64	1.49	1.94	1.74	1.79	1.85	1.93	TO 3	1.84
DS 1	0.65	0.90	0.66	0.64	0.45	0.38	0.48	0.57	0.53	0.57	0.63	DS 1	0.59
SS 1	0.55	0.54	0.48	0.43	0.50	0.41	0.60	0.49	0.55	0.46	0.56	SS 1	0.50
DS 2	0.10	0.09	0.08	0.10	0.08	0.12	0.10	0.09	0.10	0.14	0.08	DS 2	0.10
SS 2	0.55	0.48	0.42	0.49	0.45	0.42	0.58	0.35	0.45	0.45	0.45	SS 2	0.46
DS 3	0.16	0.17	0.18	0.17	0.16	0.16	0.18	0.24	0.17	0.24	0.21	DS 3	0.19

STEP LENGTH 1

Intact Leg Leading

	BKA #1	BKA #2	BKA #3	BKA #4	BKA #5	BKA #6	BKA #7	BKA #8	BKA #9	BKA #10	BKA #11		Average
Sway	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	Sway	0.00
TO 1	0.81	0.99	0.98	0.86	1.00	0.83	1.31	1.50	0.63	0.75	1.15	TO 1	0.98
HS 1	1.15	1.35	1.34	1.31	1.45	1.29	1.73	1.90	1.04	1.27	1.48	HS 1	1.39
TO 2	1.29	1.52	1.46	1.45	1.66	1.47	1.89	2.05	1.18	1.47	1.62	TO 2	1.55
HS 2	1.72	2.20	2.04	1.94	2.24	2.05	2.45	2.57	1.63	2.06	2.15	HS 2	2.10
TO 3	1.89	2.48	2.21	2.11	2.45	2.24	2.62	2.74	1.80	2.24	2.33	TO 3	2.28
DS 1	0.81	0.99	0.98	0.86	1.00	0.83	1.31	1.50	0.63	0.75	1.15	DS 1	0.98
SS 1	0.34	0.36	0.36	0.44	0.45	0.46	0.42	0.41	0.41	0.52	0.34	SS 1	0.41
DS 2	0.14	0.17	0.12	0.15	0.21	0.18	0.16	0.15	0.14	0.20	0.14	DS 2	0.16
SS 2	0.44	0.68	0.58	0.49	0.58	0.58	0.56	0.52	0.45	0.59	0.53	SS 2	0.55
DS 3	0.16	0.28	0.17	0.17	0.21	0.19	0.17	0.17	0.17	0.18	0.18	DS 3	0.19

Prosthetic Leg Leading

	BKA #1	BKA #2	BKA #3	BKA #4	BKA #5	BKA #6	BKA #7	BKA #8	BKA #9	BKA #10	BKA #11		Average
Sway	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	Sway	0.00
TO 1	0.88	0.75	1.06	0.94	1.11	1.13	0.91	0.97	0.82	0.76	0.35	TO 1	0.88
HS 1	1.29	1.33	1.53	1.47	1.64	1.72	1.48	1.56	1.30	1.34	0.87	HS 1	1.41
TO 2	1.44	1.60	1.68	1.62	1.81	1.87	1.66	1.71	1.43	1.57	1.02	TO 2	1.58
HS 2	1.82	2.11	2.15	2.07	2.35	2.29	2.15	2.18	1.88	2.04	1.42	HS 2	2.04
TO 3	1.98	2.38	2.34	2.23	2.52	2.52	2.34	2.36	2.05	2.27	1.60	TO 3	2.24
DS 1	0.88	0.75	1.06	0.94	1.11	1.13	0.91	0.97	0.82	0.76	0.35	DS 1	0.88
SS 1	0.40	0.57	0.47	0.53	0.53	0.59	0.56	0.59	0.48	0.58	0.52	SS 1	0.53
DS 2	0.15	0.28	0.16	0.14	0.17	0.16	0.18	0.15	0.13	0.23	0.16	DS 2	0.17
SS 2	0.38	0.51	0.47	0.45	0.54	0.41	0.49	0.47	0.45	0.47	0.40	SS 2	0.46
DS 3	0.17	0.26	0.19	0.16	0.18	0.23	0.19	0.18	0.17	0.23	0.17	DS 3	0.19

STEP LENGTH 2

Intact Leg Leading

	BKA #1	BKA #2	BKA #3	BKA #4	BKA #5	BKA #6	BKA #7	BKA #8	BKA #9	BKA #10	BKA #11		Average
Sway	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	Sway	0.00
TO 1	0.79	0.94	1.10	0.85	0.86	1.33	1.19	1.74	0.82	0.82	0.94	TO 1	1.04
HS 1	1.16	1.38	1.55	1.33	1.32	1.83	1.59	2.16	1.27	1.39	1.32	HS 1	1.48
TO 2	1.30	1.54	1.65	1.44	1.49	1.99	1.73	2.31	1.40	1.56	1.45	TO 2	1.62
HS 2	1.78	2.33	2.27	1.85	2.06	2.58	2.30	2.84	1.86	2.16	2.01	HS 2	2.18
TO 3	1.95	2.59	2.44	2.11	2.22	2.74	2.50	3.01	2.03	2.33	2.19	TO 3	2.37
DS 1	0.79	0.94	1.10	0.85	0.86	1.33	1.19	1.74	0.82	0.82	0.94	DS 1	1.04
SS 1	0.37	0.43	0.45	0.48	0.45	0.50	0.40	0.42	0.45	0.57	0.38	SS 1	0.45
DS 2	0.13	0.16	0.11	0.11	0.17	0.16	0.13	0.15	0.13	0.17	0.13	DS 2	0.14
SS 2	0.48	0.79	0.62	0.41	0.57	0.59	0.57	0.54	0.46	0.60	0.55	SS 2	0.56
DS 3	0.17	0.26	0.17	0.26	0.17	0.16	0.21	0.17	0.17	0.17	0.18	DS 3	0.19

Prosthetic Leg Leading

	BKA #1	BKA #2	BKA #3	BKA #4	BKA #5	BKA #6	BKA #7	BKA #8	BKA #9	BKA #10	BKA #11		Average
Sway	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	Sway	0.00
TO 1	0.83	0.83	0.75	0.72	1.52	0.88	1.04	0.94	0.75	0.63	1.26	TO 1	0.92
HS 1	1.29	1.69	1.24	1.30	2.12	1.53	1.59	1.57	1.28	1.28	1.96	HS 1	1.53
TO 2	1.43	1.95	1.37	1.40	2.27	1.65	1.74	1.67	1.38	1.45	2.13	TO 2	1.68
HS 2	1.84	2.55	1.88	1.84	2.77	2.13	2.21	2.14	1.78	1.94	2.62	HS 2	2.15
TO 3	2.01	2.75	2.08	2.01	2.95	2.37	2.38	2.30	1.98	2.18	2.77	TO 3	2.34
DS 1	0.83	0.83	0.75	0.72	1.52	0.88	1.04	0.94	0.75	0.63	1.26	DS 1	0.92
SS 1	0.45	0.86	0.49	0.58	0.60	0.65	0.54	0.62	0.53	0.65	0.70	SS 1	0.61
DS 2	0.15	0.26	0.13	0.10	0.15	0.12	0.15	0.11	0.10	0.17	0.17	DS 2	0.15
SS 2	0.41	0.60	0.52	0.44	0.50	0.48	0.47	0.46	0.40	0.49	0.49	SS 2	0.48
DS 3	0.17	0.20	0.20	0.17	0.18	0.24	0.17	0.16	0.20	0.24	0.15	DS 3	0.19

STEP LENGTH 3

Intact Leg Leading

	BKA #1	BKA #2	BKA #3	BKA #4	BKA #5	BKA #6	BKA #7	BKA #8	BKA #9	BKA #10	BKA #11		Average
Sway	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	Sway	0.00
TO 1	0.89	0.92	1.11	0.70	1.27	1.19	1.65	1.34	0.77	0.87	0.93	TO 1	1.06
HS 1	1.28	1.34	1.59	1.22	1.89	1.72	2.07	1.83	1.29	1.48	1.31	HS 1	1.55
TO 2	1.39	1.50	1.67	1.31	2.05	1.86	2.20	1.94	1.41	1.62	1.44	TO 2	1.67
HS 2	1.92	2.29	2.39	1.79	2.64	2.52	2.80	2.53	1.91	2.28	1.98	HS 2	2.28
TO 3	2.08	2.52	2.56	2.04	2.81	2.69	2.98	2.70	2.08	2.42	2.15	TO 3	2.46
DS 1	0.89	0.92	1.11	0.70	1.27	1.19	1.65	1.34	0.77	0.87	0.93	DS 1	1.06
SS 1	0.39	0.43	0.47	0.52	0.61	0.53	0.43	0.48	0.52	0.61	0.39	SS 1	0.49
DS 2	0.11	0.15	0.08	0.09	0.16	0.15	0.12	0.11	0.12	0.14	0.13	DS 2	0.12
SS 2	0.53	0.80	0.72	0.48	0.60	0.66	0.61	0.59	0.51	0.65	0.54	SS 2	0.61
DS 3	0.17	0.22	0.17	0.24	0.17	0.17	0.17	0.17	0.17	0.15	0.17	DS 3	0.18

Prosthetic Leg Leading

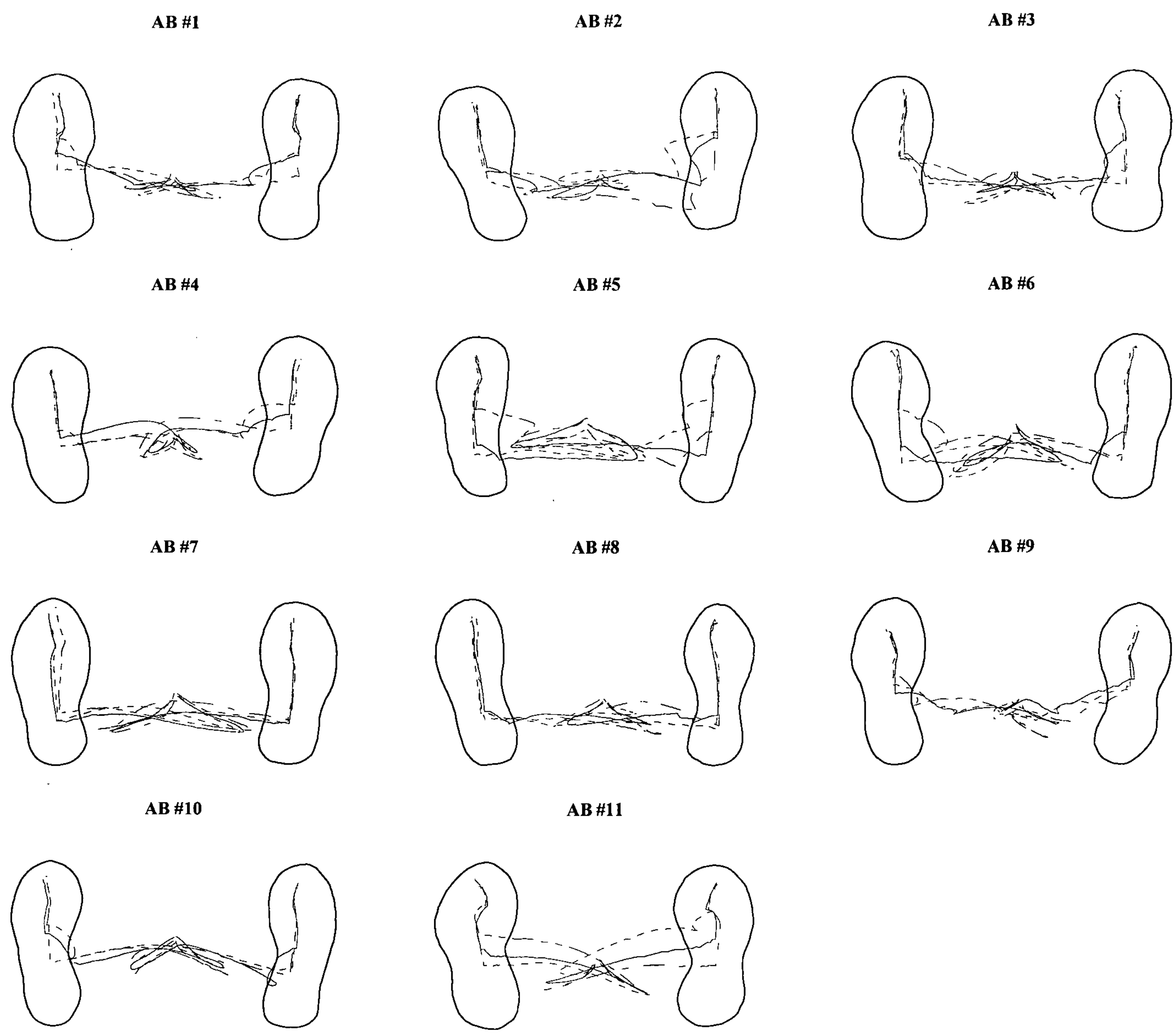
	BKA #1	BKA #2	BKA #3	BKA #4	BKA #5	BKA #6	BKA #7	BKA #8	BKA #9	BKA #10	BKA #11		Average
Sway	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	Sway	0.00
TO 1	0.87	1.14	1.02	0.71	1.36	1.40	1.41	1.34	1.08	0.68	0.81	TO 1	1.07
HS 1	1.32	1.96	1.55	1.35	1.98	2.12	2.00	2.04	1.64	1.41	1.46	HS 1	1.71
TO 2	1.45	2.17	1.67	1.45	2.10	2.23	2.14	2.11	1.72	1.56	1.58	TO 2	1.83
HS 2	1.89	2.79	2.26	1.91	2.59	2.73	2.67	2.57	2.17	2.13	2.03	HS 2	2.34
TO 3	2.05	2.98	2.46	2.08	2.75	2.95	2.84	2.74	2.33	2.29	2.18	TO 3	2.51
DS 1	0.87	1.14	1.02	0.71	1.36	1.40	1.41	1.34	1.08	0.68	0.81	DS 1	1.07
SS 1	0.45	0.81	0.53	0.64	0.62	0.72	0.59	0.70	0.56	0.73	0.64	SS 1	0.64
DS 2	0.13	0.21	0.12	0.10	0.12	0.11	0.14	0.07	0.09	0.15	0.13	DS 2	0.12
SS 2	0.44	0.62	0.60	0.46	0.49	0.49	0.53	0.46	0.44	0.57	0.44	SS 2	0.50
DS 3	0.17	0.19	0.19	0.17	0.17	0.22	0.17	0.17	0.17	0.17	0.15	DS 3	0.17

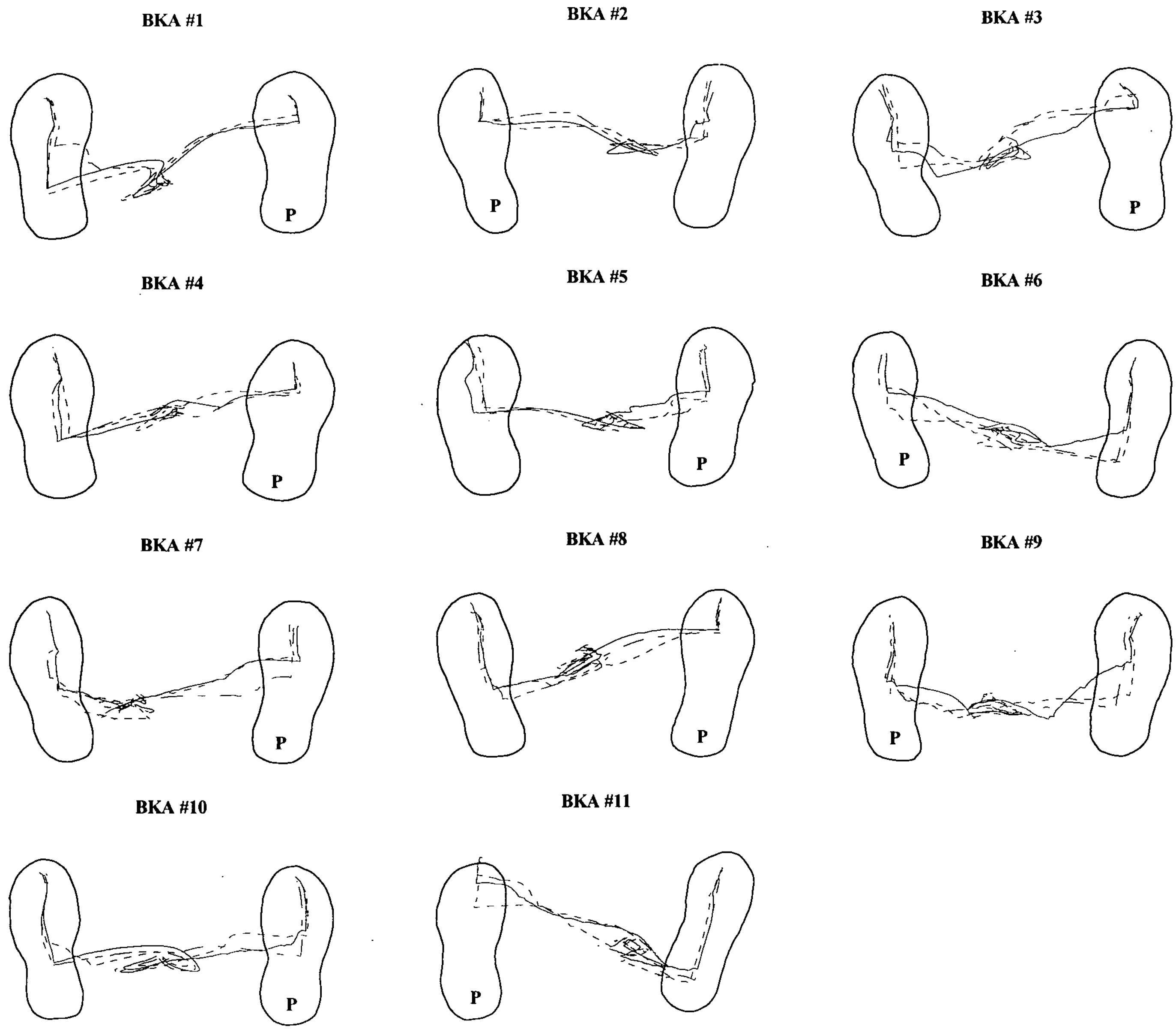
APPENDIX B-3: COP DURING GAIT INTIATION

The following two pages contain the mean movement of the center of pressure for each individual during the gait initiation trials. The dashed line represents the preferred step length condition, the solid line represents the +25% step length condition, and the broken line represents the +50% step length condition.

Abbreviations:

- AB = able-bodied (control)
- BKA = unilateral below-knee amputee
- P = prosthetic limb


AB #1
AB #2
AB #3
AB #4
AB #5
AB #6
AB #7
AB #8
AB #9
AB #10
AB #11

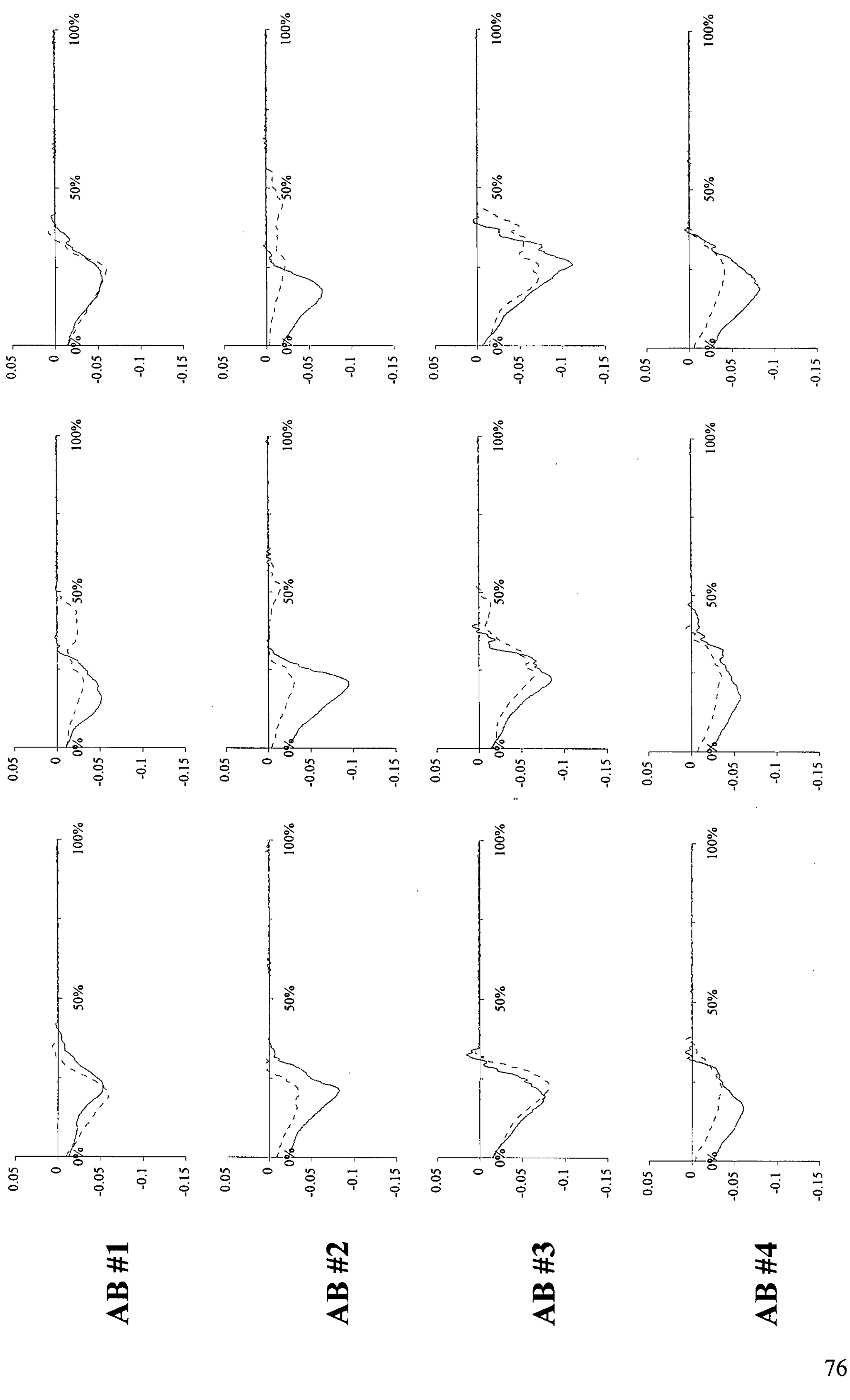

BKA #1
BKA #2
BKA #3
BKA #4
BKA #5
BKA #6
BKA #7
BKA #8
BKA #9
BKA #10
BKA #11
P

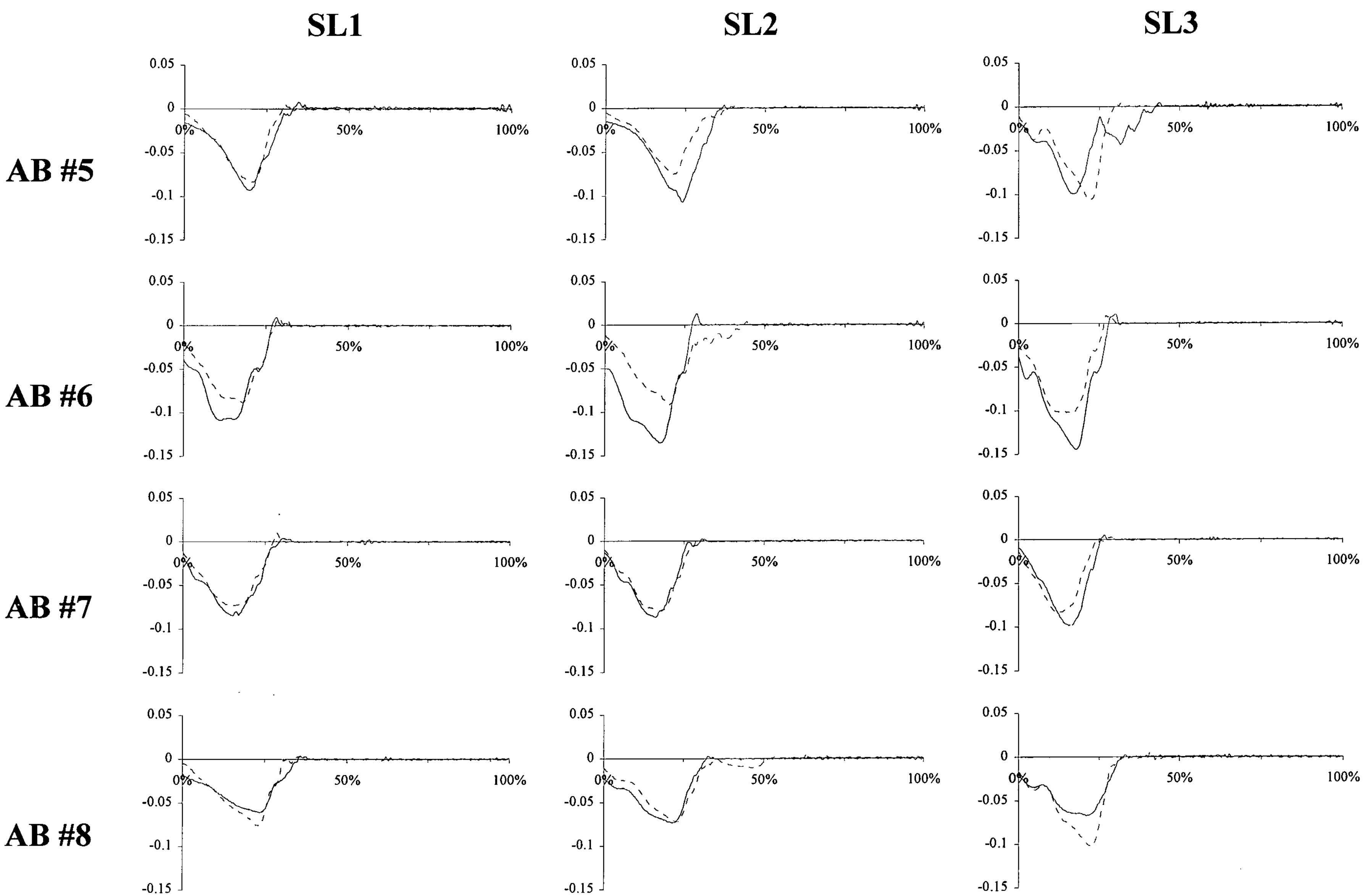
APPENDIX B-4: FORCES FROM THE LEADING LIMB

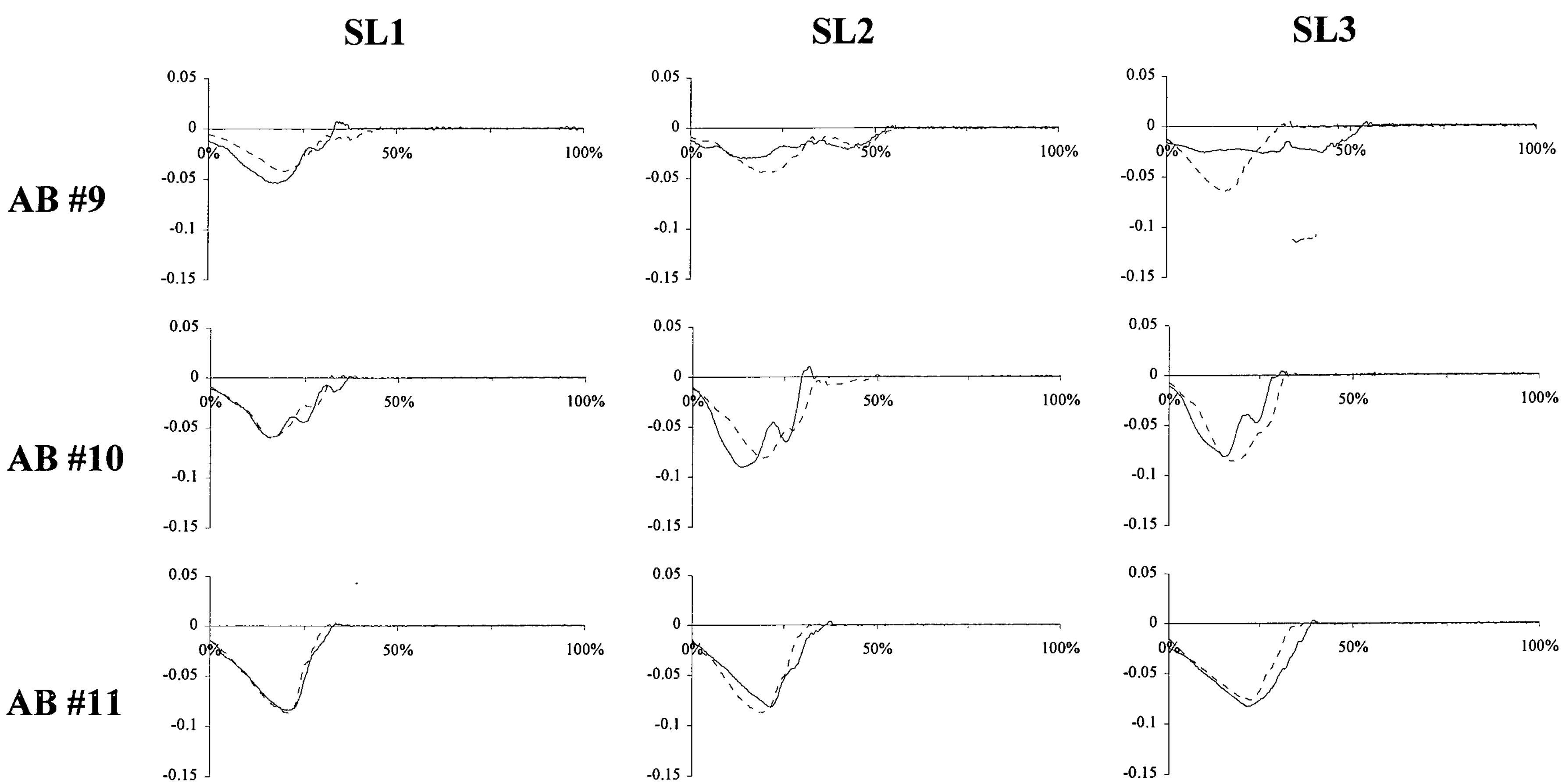
The following six pages contain the mean A-P force-time curve from the leading limb for each individual during the gait initiation trials. The solid line represents trials with a left (or intact) lead limb condition while the dashed line represents trials with a right (or prosthetic) lead limb condition. Units are in N/BW for the vertical axis and % gait cycle (start of sway to second toe-off of the leading limb) for the horizontal axis.

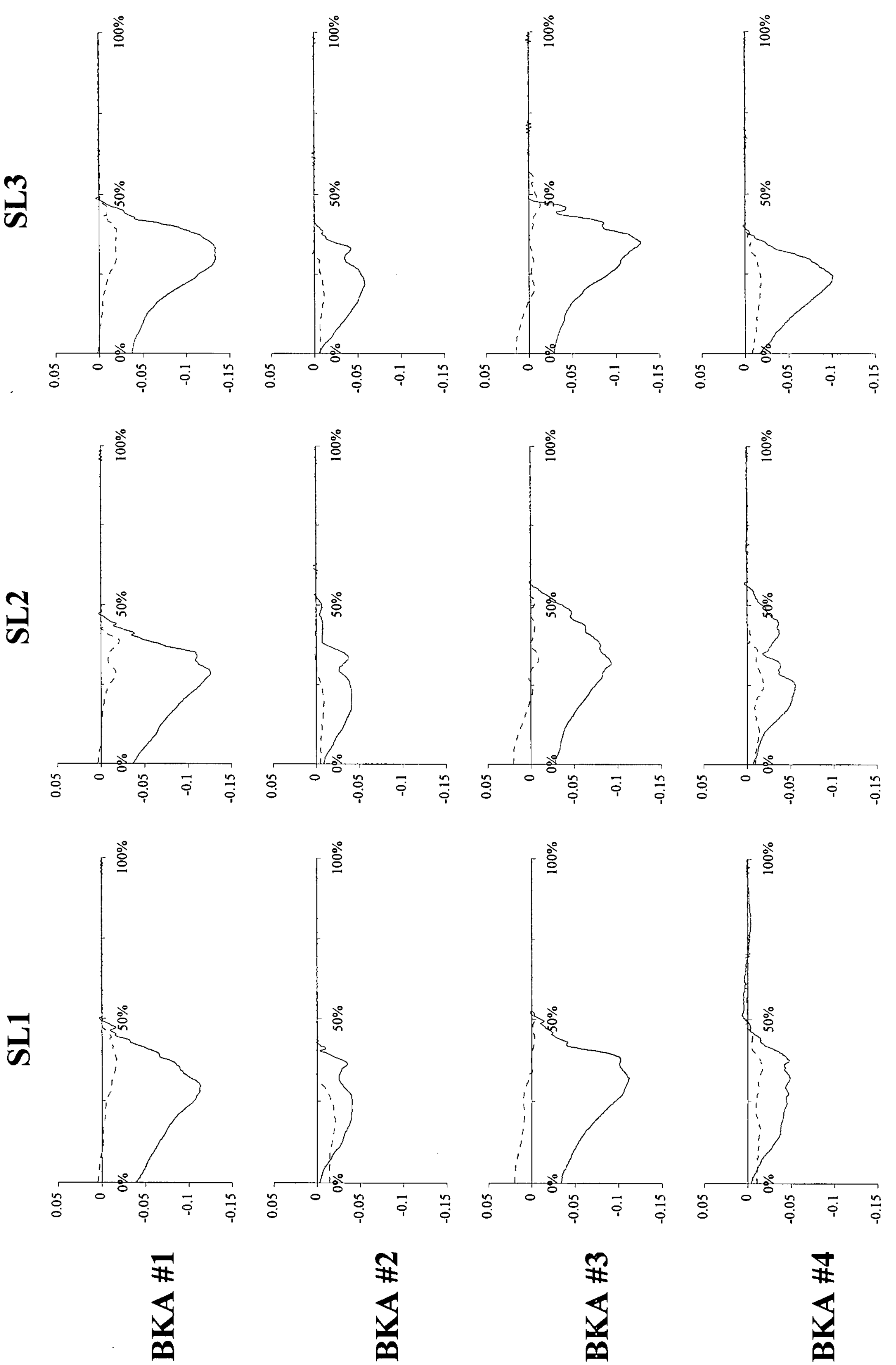
Abbreviations:

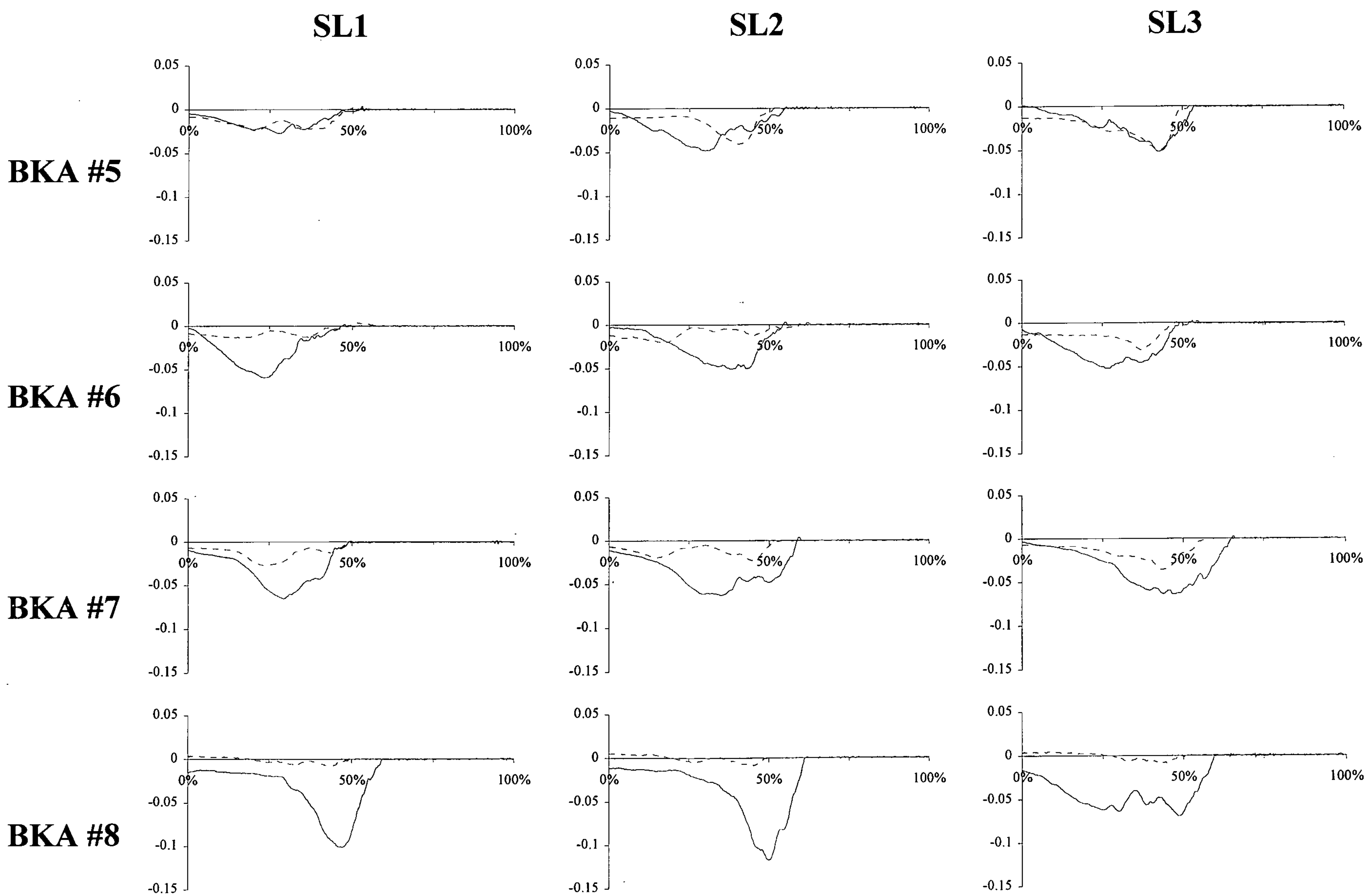
- AB = able-bodied (control)
- BKA = unilateral below-knee amputee
- SL = step length condition


SL1
SL2
SL3
AB #1
AB #2
AB #3
AB #4
0.05
0
-0.05
-0.1
-0.15
0%
50%
100%

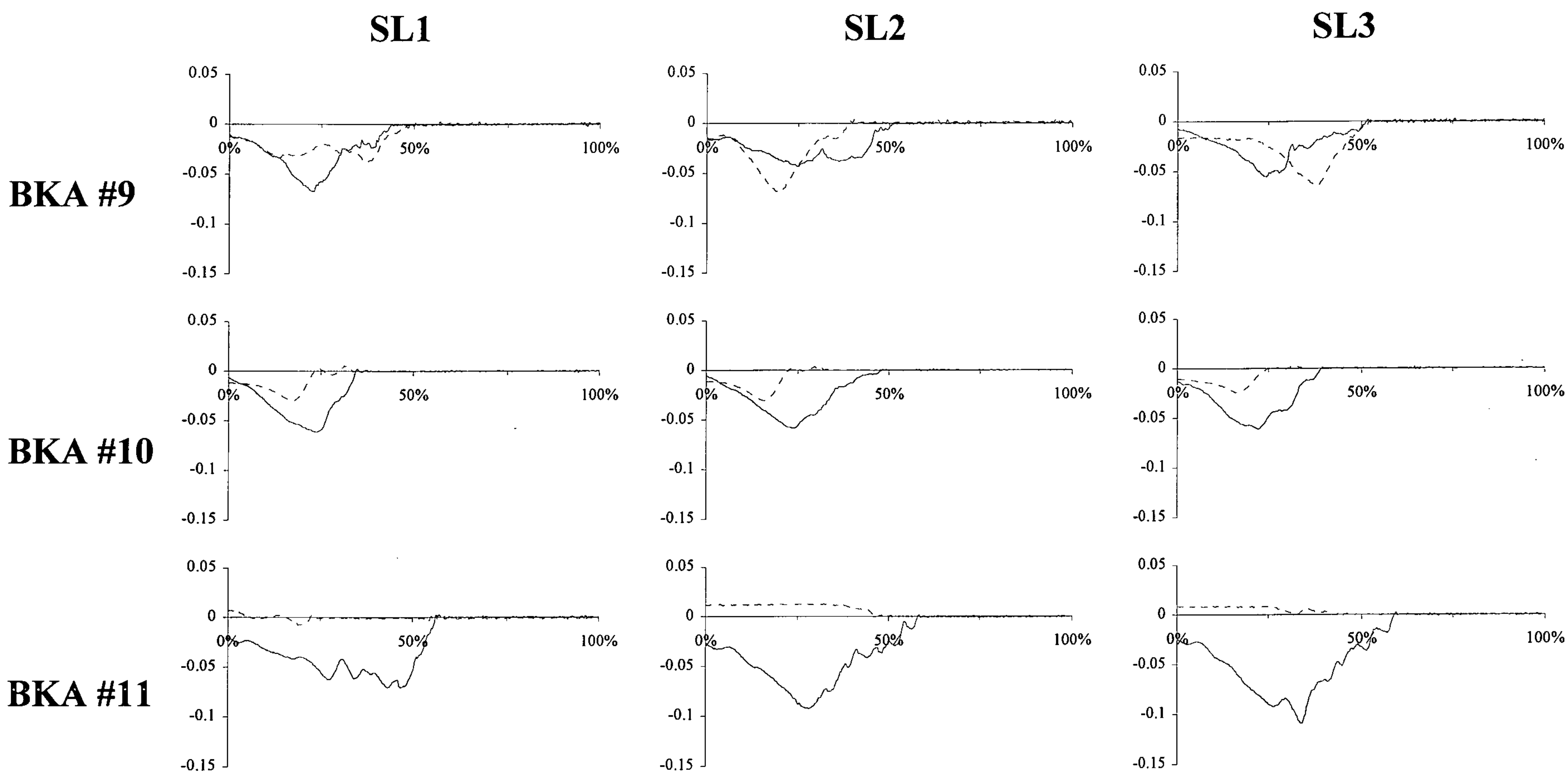

SL1
SL2
SL3
AB #5
AB #6
AB #7
AB #8
0.05
0
-0.05
-0.1
-0.15
0%
50%
100%


SL1
SL2
SL3
AB #9
AB #10
AB #11
0.05
0
-0.05
-0.1
-0.15
0%
50%
100%


SL1
SL2
SL3
BKA #1
BKA #2
BKA #3
BKA #4
0.05
0
-0.05
-0.1
-0.15
0%
50%
100%



SL1
SL2
SL3
BKA #5
BKA #6
BKA #7
BKA #8
0.05
0
-0.05
-0.1
-0.15
0%
50%
100%

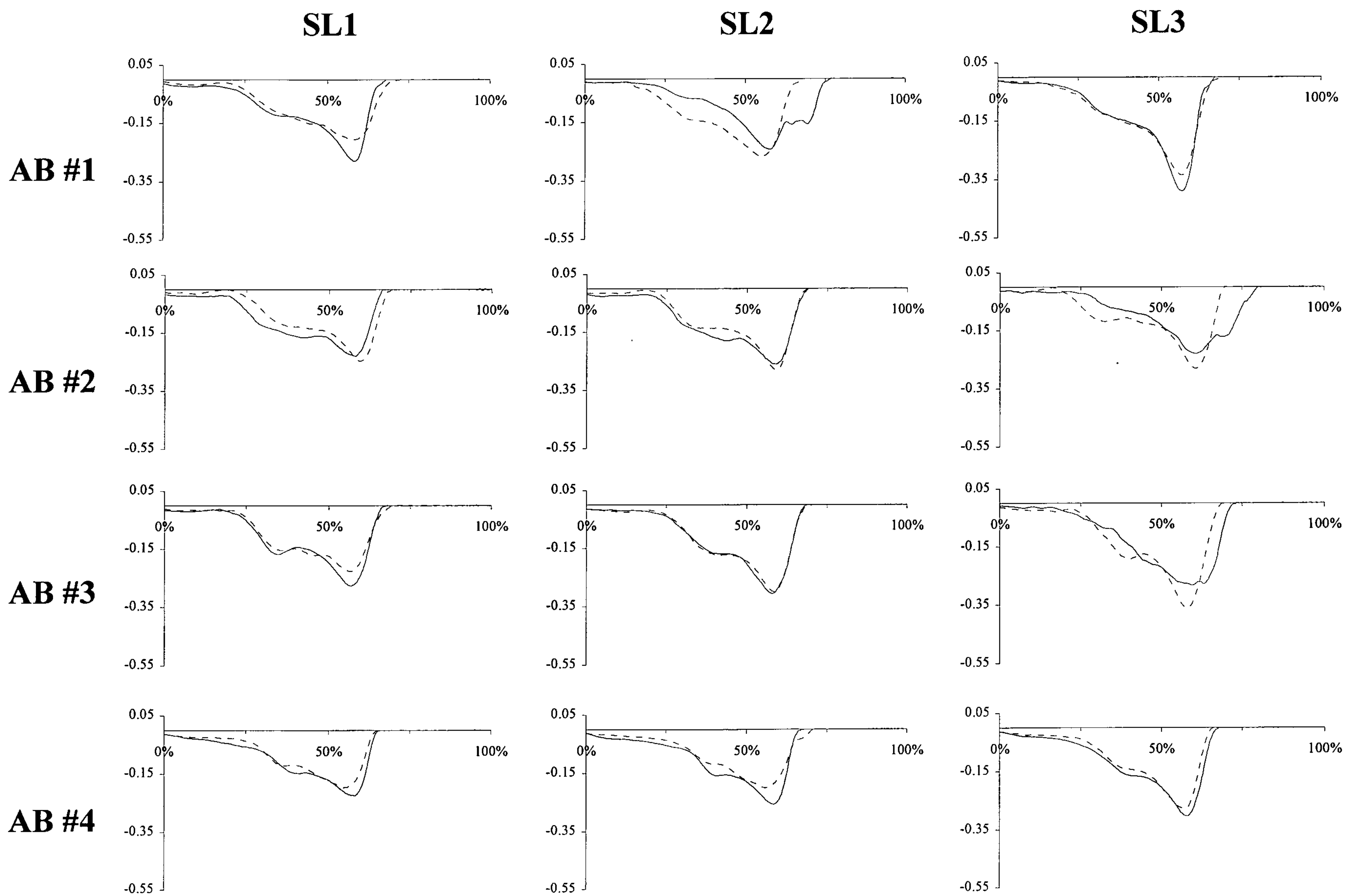

SL1
SL2
SL3
BKA #9
BKA #10
BKA #11
0.05
0
-0.05
-0.1
-0.15
0%
50%
100%

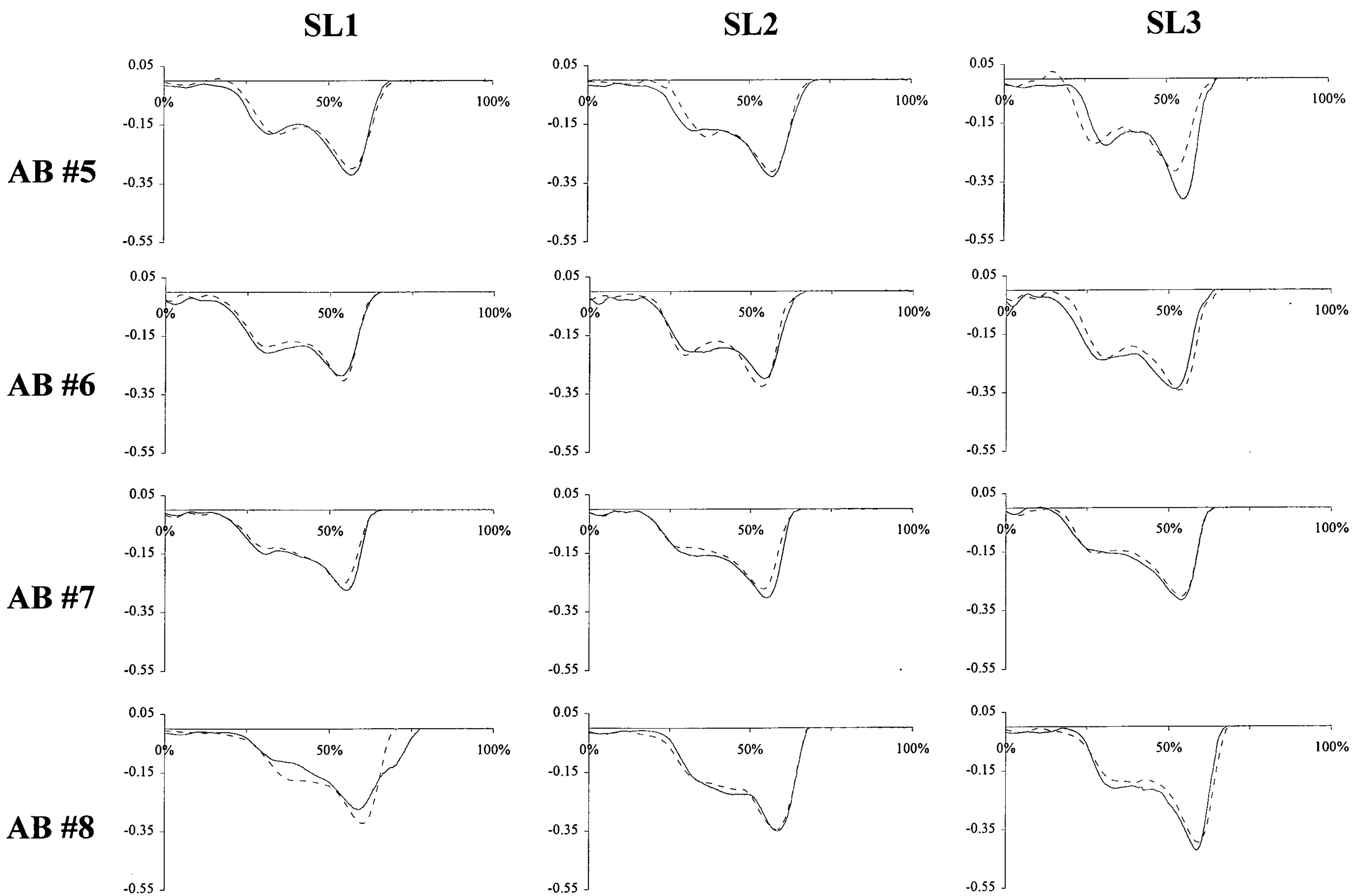
APPENDIX B-5: FORCES FROM THE TRAILING LIMB

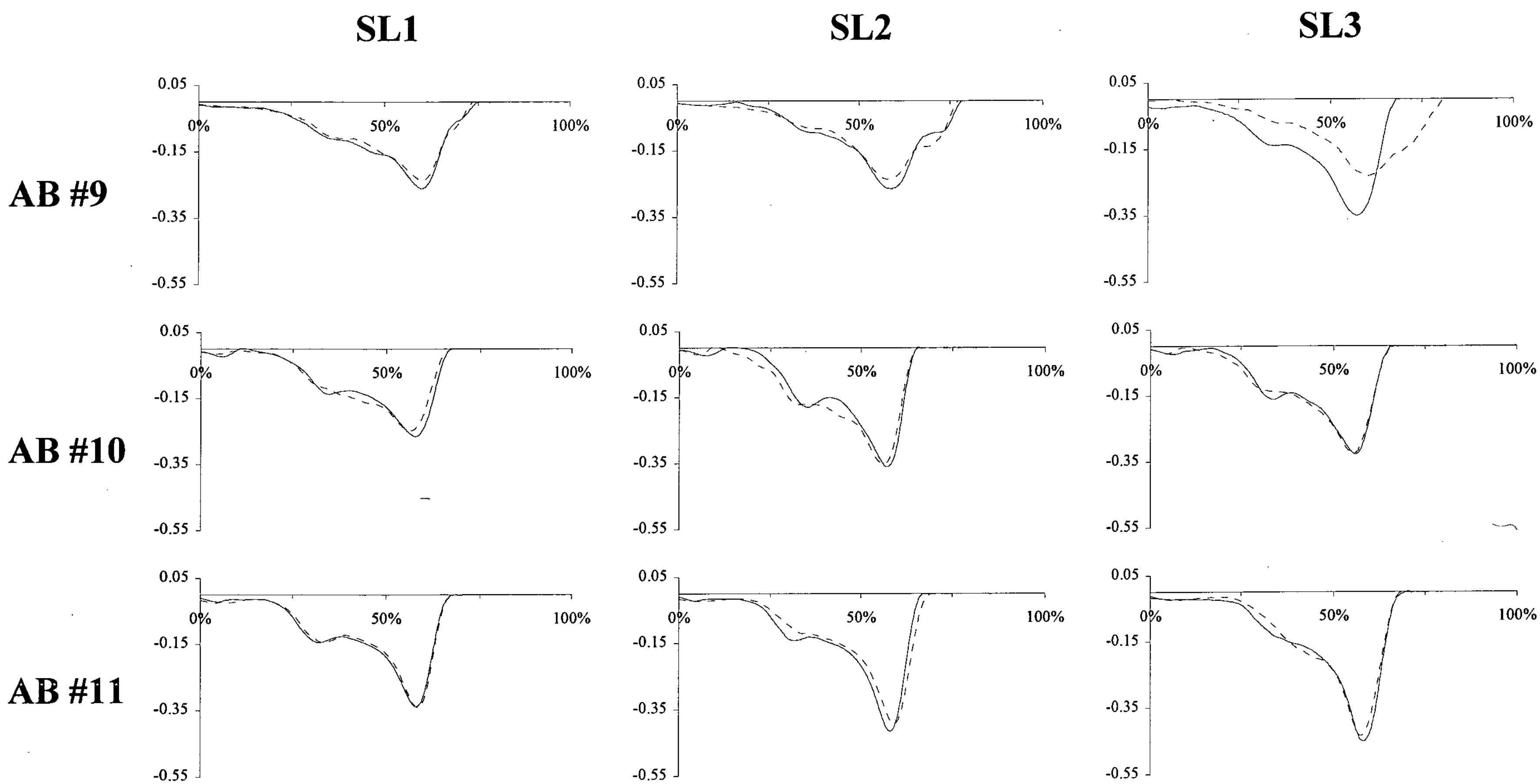
The following six pages contain the mean A-P force-time curve from the trailing limb for each individual during the gait initiation trials. The solid line represents trials with a left (or intact) trail limb condition while the dashed line represents trials with a right (or prosthetic) trail limb condition. Units are in N/BW for the vertical axis and % gait cycle (start of sway to second toe-off of the leading limb) for the horizontal axis.

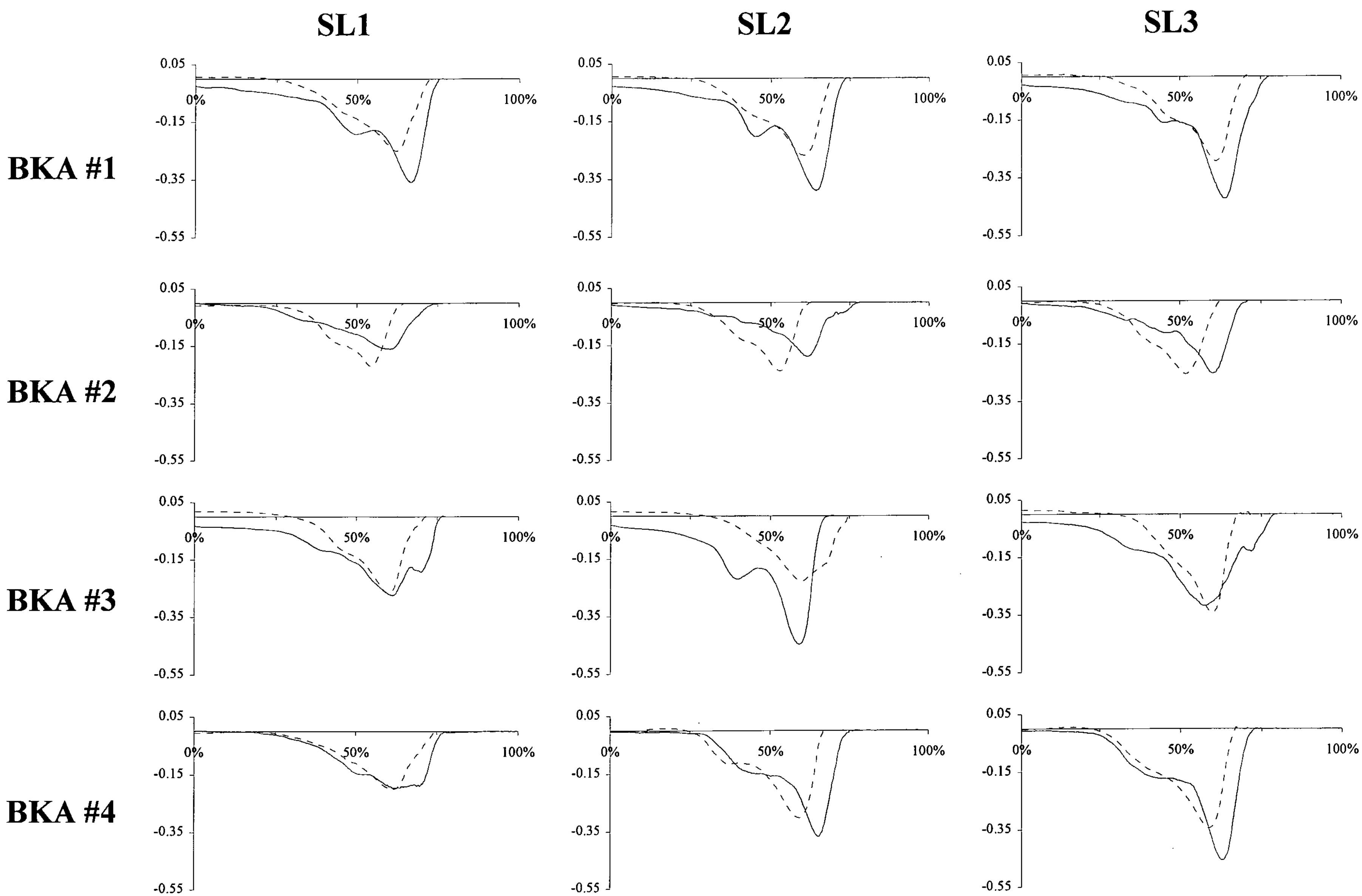
Abbreviations:

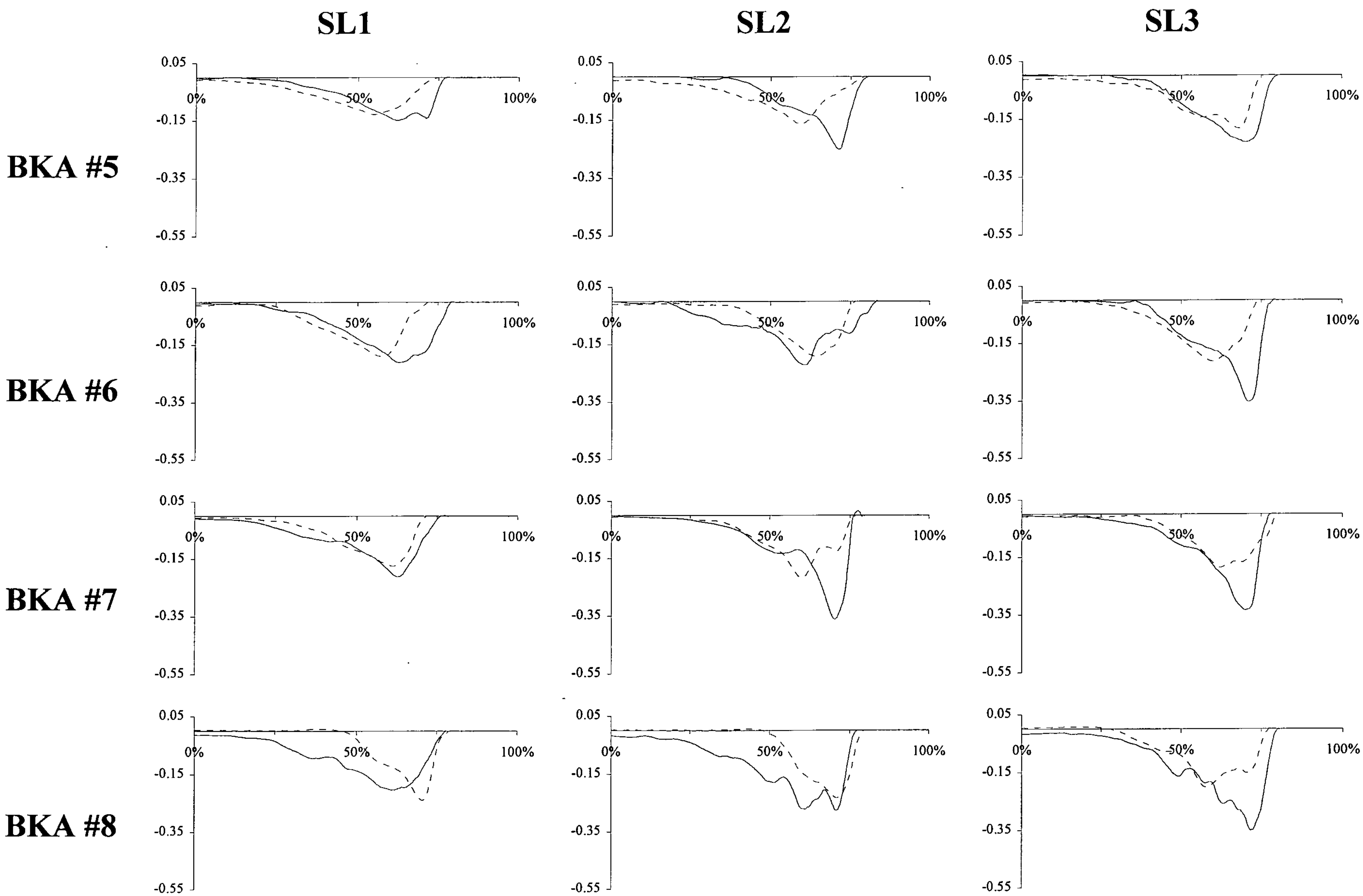
- AB = able-bodied (control)
- BKA = unilateral below-knee amputee
- SL = step length condition

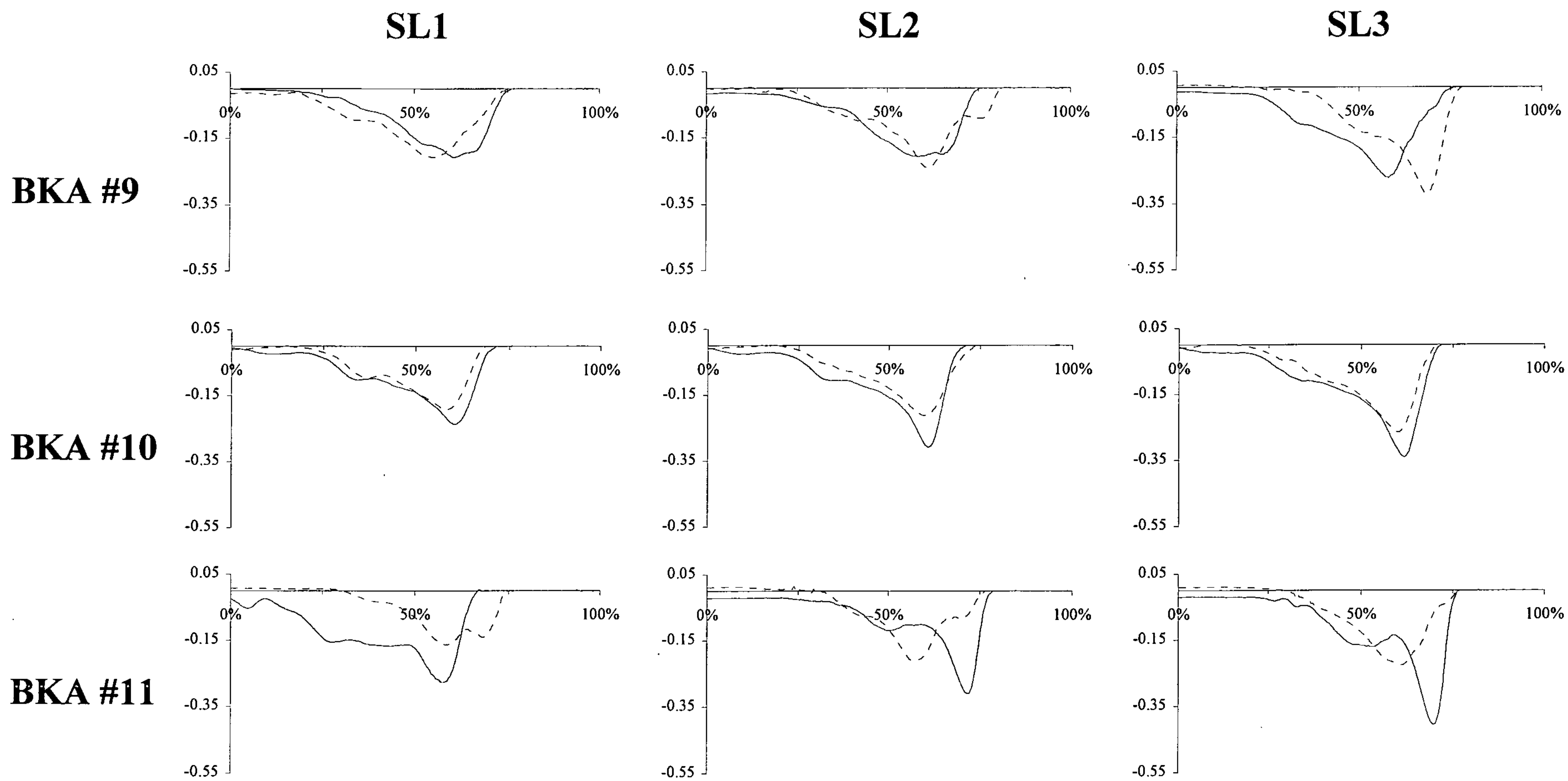

SL1
SL2
SL3
AB #1
AB #2
AB #3
AB #4
0.05
-0.15
-0.35
-0.55
0%
50%
100%


SL1
SL2
SL3
AB #5
AB #6
AB #7
AB #8
0.05
-0.15
-0.35
-0.55
0%
50%
100%


SL1
SL2
SL3
AB #9
AB #10
AB #11
0.05
-0.15
-0.35
-0.55
0%
50%
100%


SL1
SL2
SL3
BKA #1
BKA #2
BKA #3
BKA #4
0.05
-0.15
-0.35
-0.55
0%
50%
100%


SL1
SL2
SL3
BKA #5
BKA #6
BKA #7
BKA #8
0.05
-0.15
-0.35
-0.55
0%
50%
100%

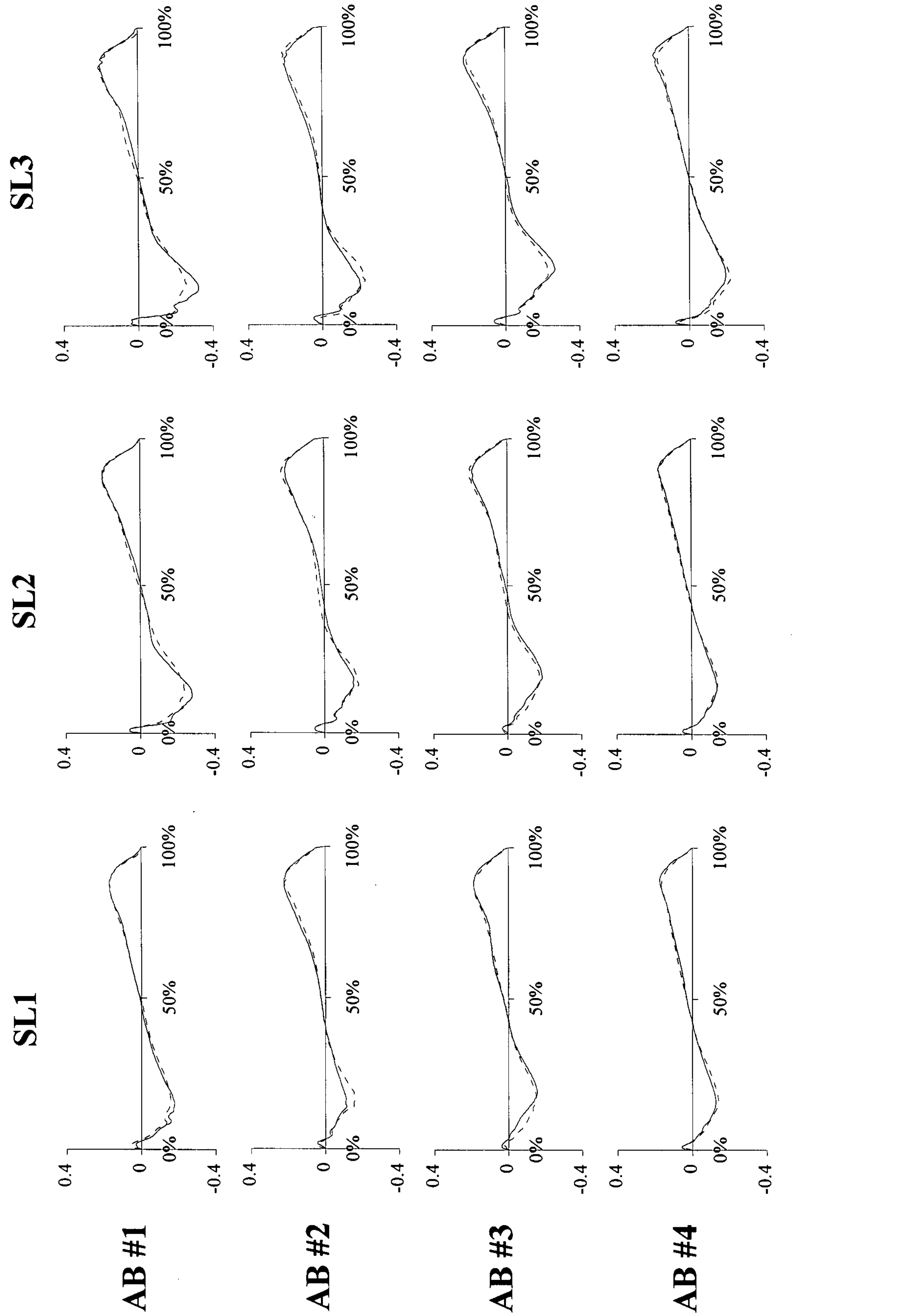

SL1
SL2
SL3
BKA #9
BKA #10
BKA #11
0.05
-0.15
-0.35
-0.55
0%
50%
100%

APPENDIX B-6: A-P FORCE FROM THE FIRST STEP

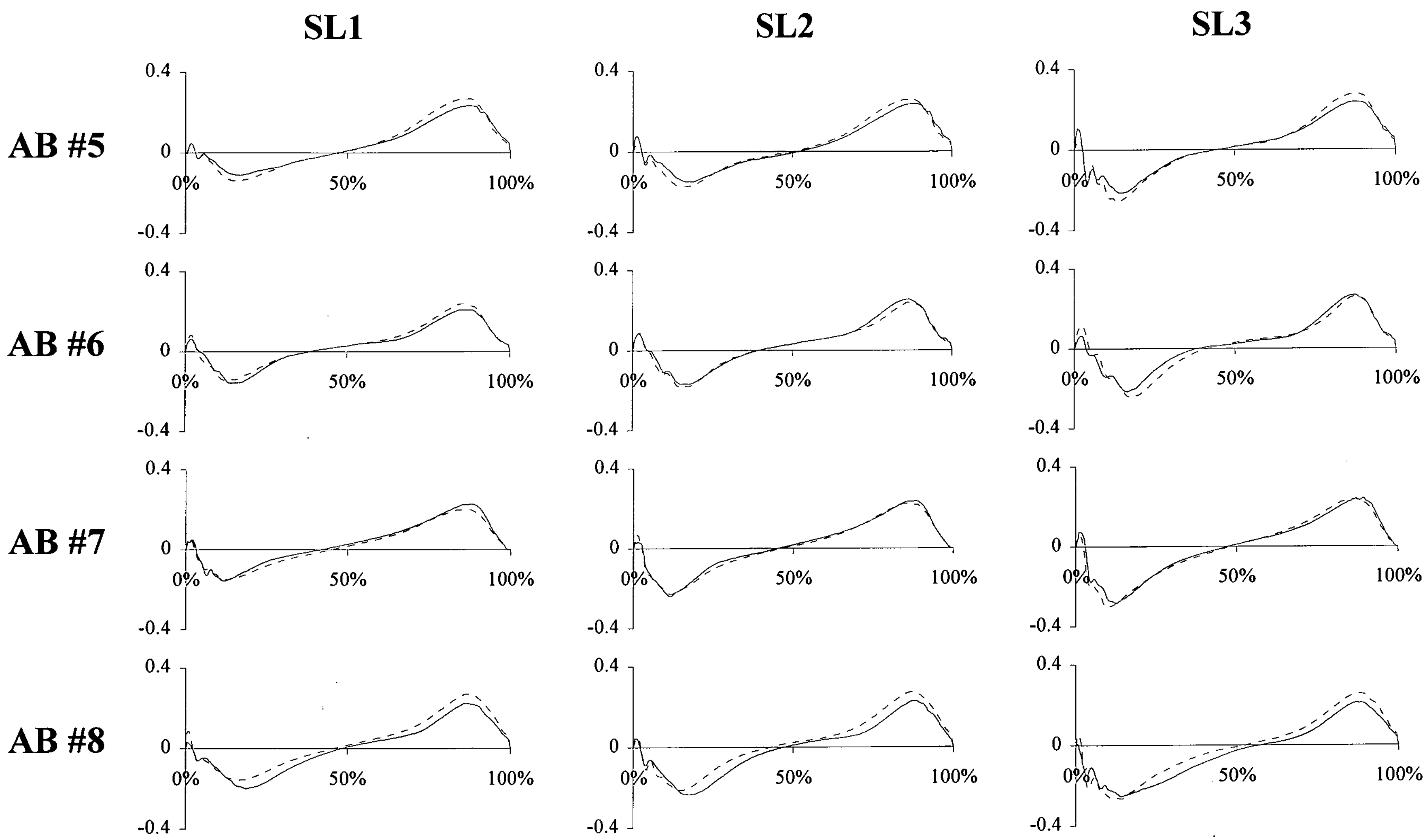
The following six pages contain the mean A-P force-time curve from the initial step for each individual during the gait initiation trials. The solid line represents trials with a left (or intact) lead limb condition while the dashed line represents trials with a right (or prosthetic) lead limb condition. Units are in N/BW for the vertical axis and % gait cycle (1^{st} heel strike to 2^{nd} toe-off of the leading limb) for the horizontal axis.

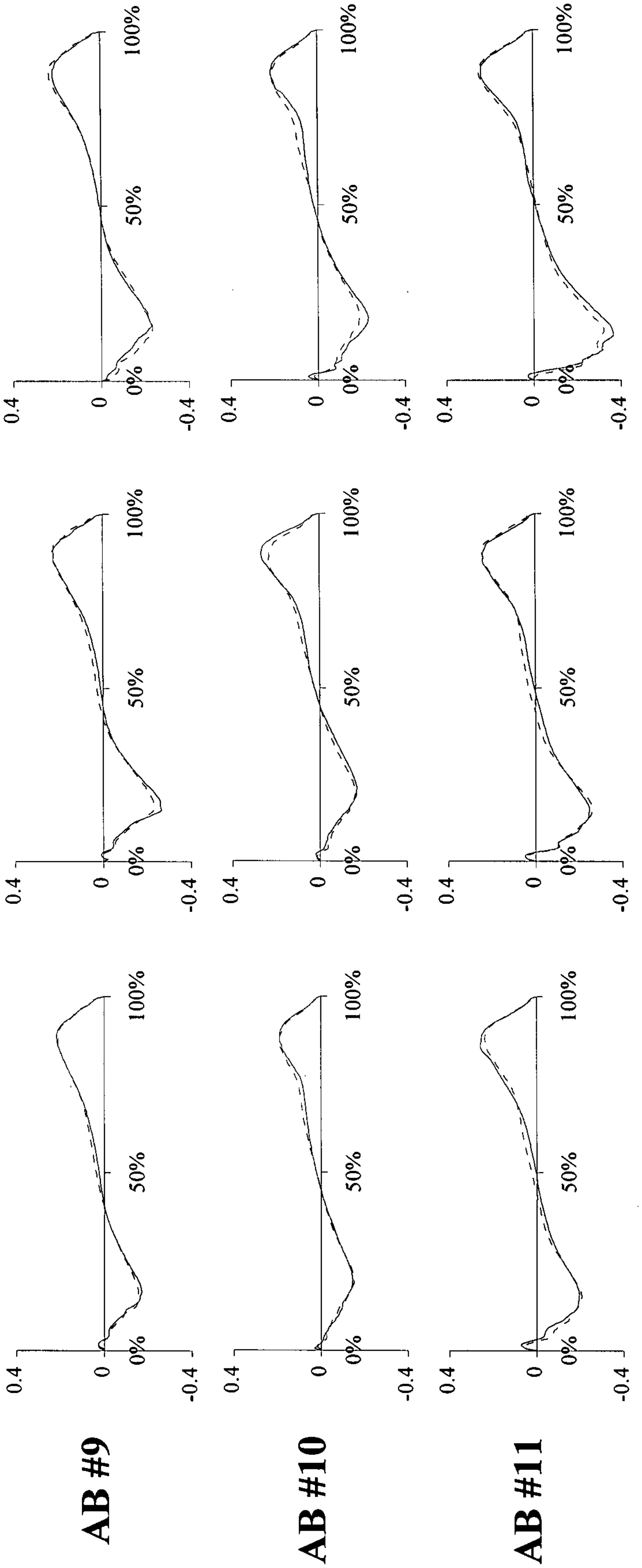
Abbreviations:

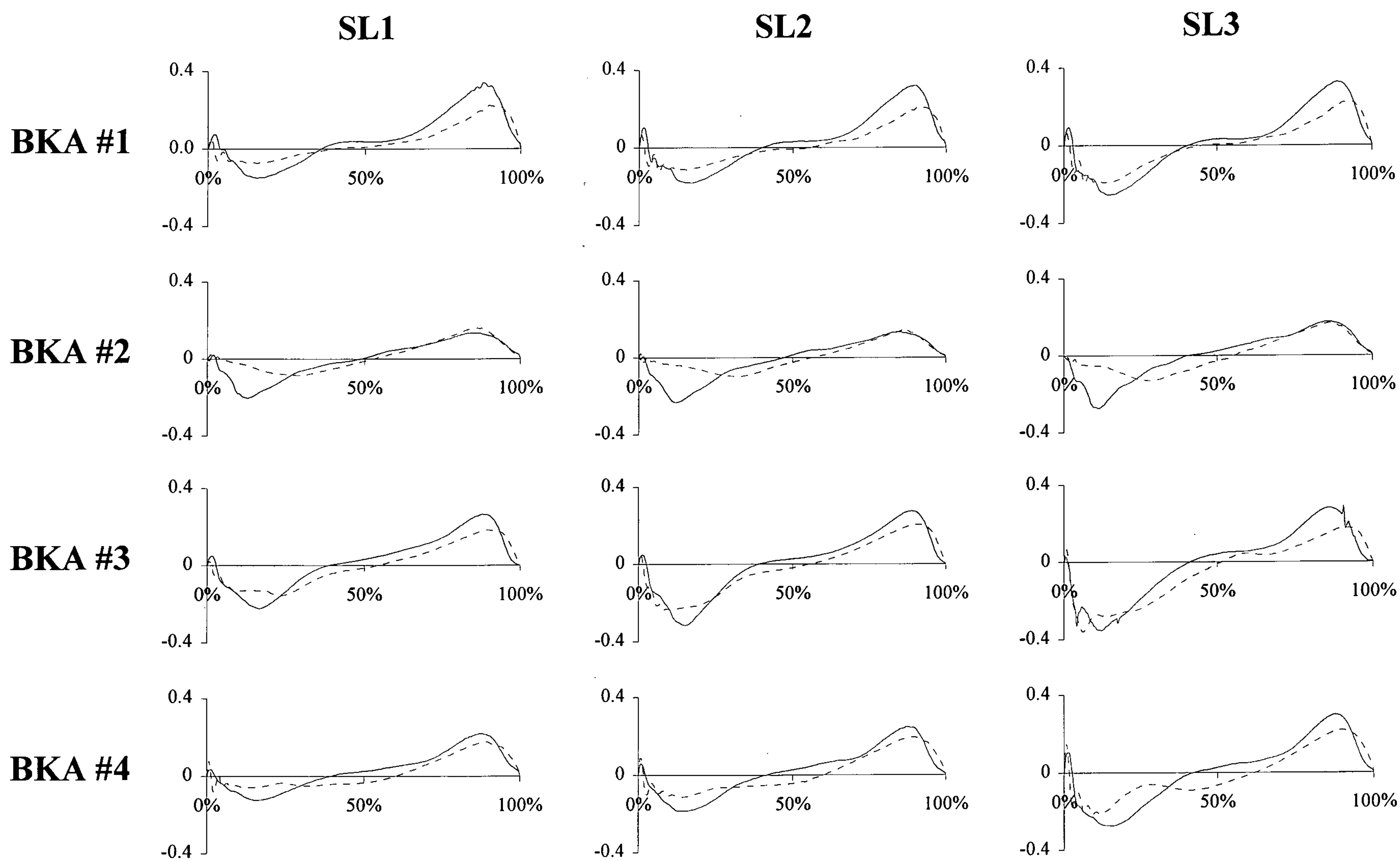
- AB = able-bodied (control)
- BKA = unilateral below-knee amputee
- SL = step length condition



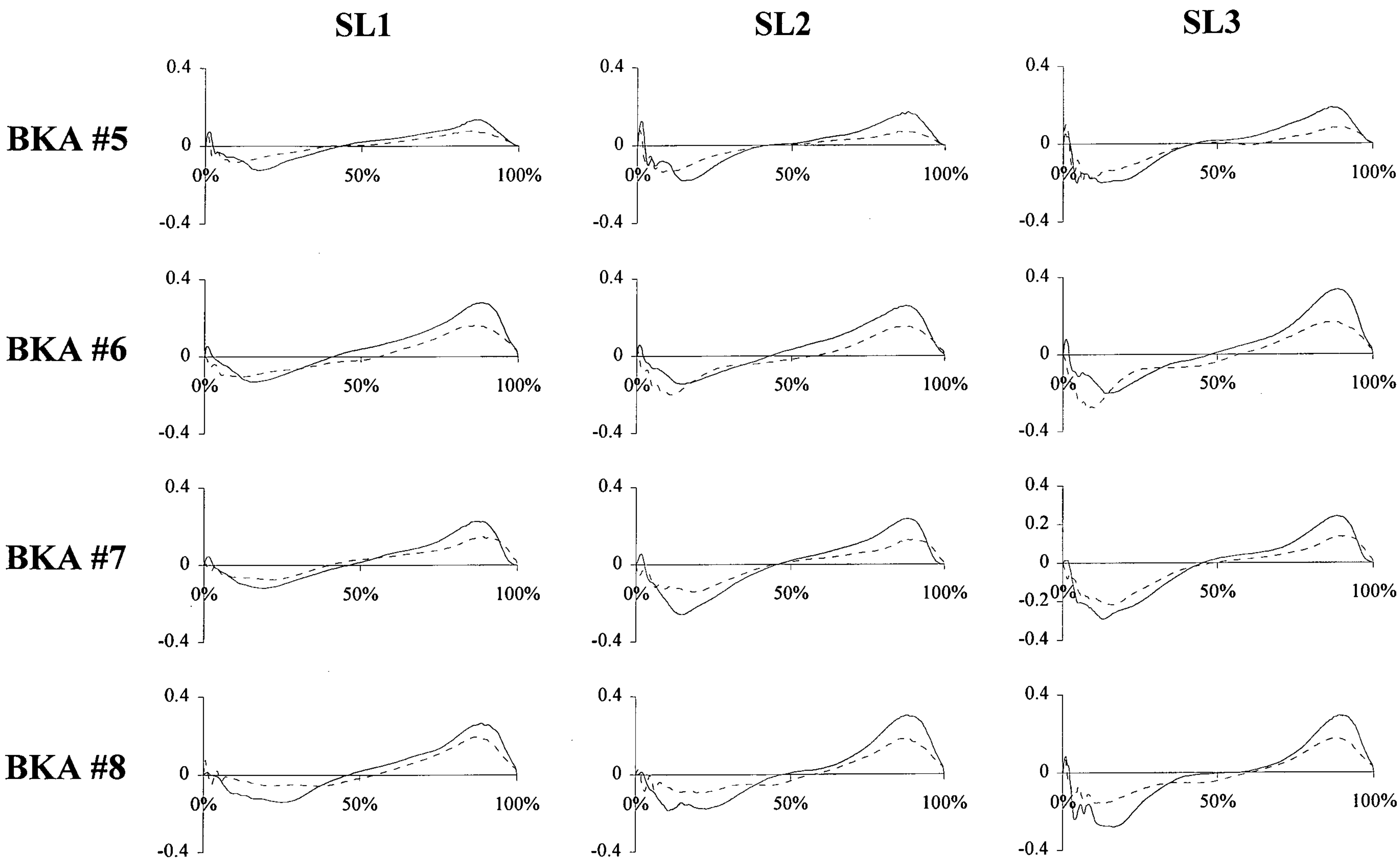
SL1
SL2
SL3
AB #1
AB #2
AB #3
AB #4
0.4
0
-0.4
0%
50%
100%

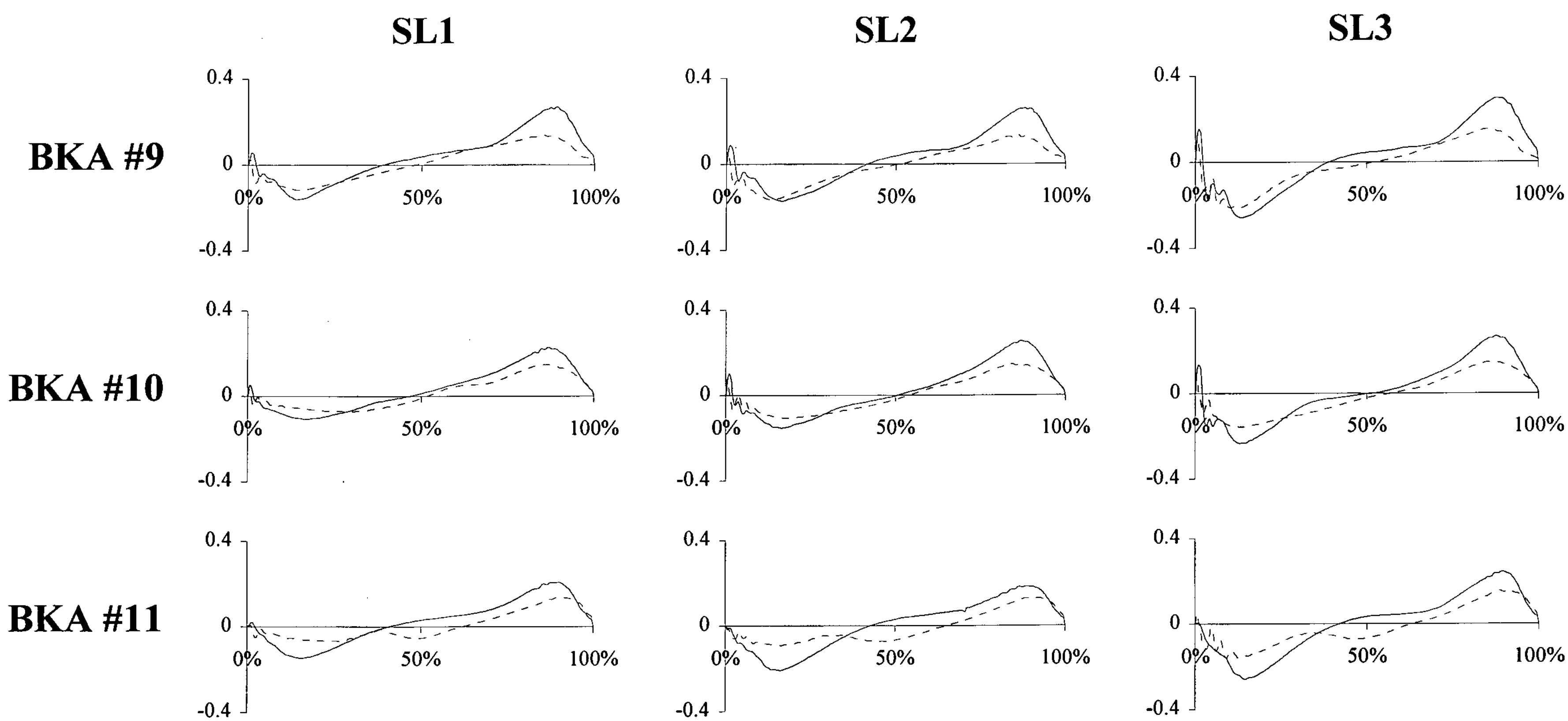

SL1
SL2
SL3
AB #5
AB #6
AB #7
AB #8
0.4
0
-0.4
0%
50%
100%


SL1
SL2
SL3
AB #9
AB #10
AB #11
0.4
0
-0.4
0%
50%
100%



SL1
SL2
SL3
BKA #1
BKA #2
BKA #3
BKA #4
0.4
0.0
0
-0.4
0%
50%
100%


SL1
SL2
SL3
BKA #5
BKA #6
BKA #7
BKA #8
0.4
0.2
0
-0.2
-0.4
0%
50%
100%

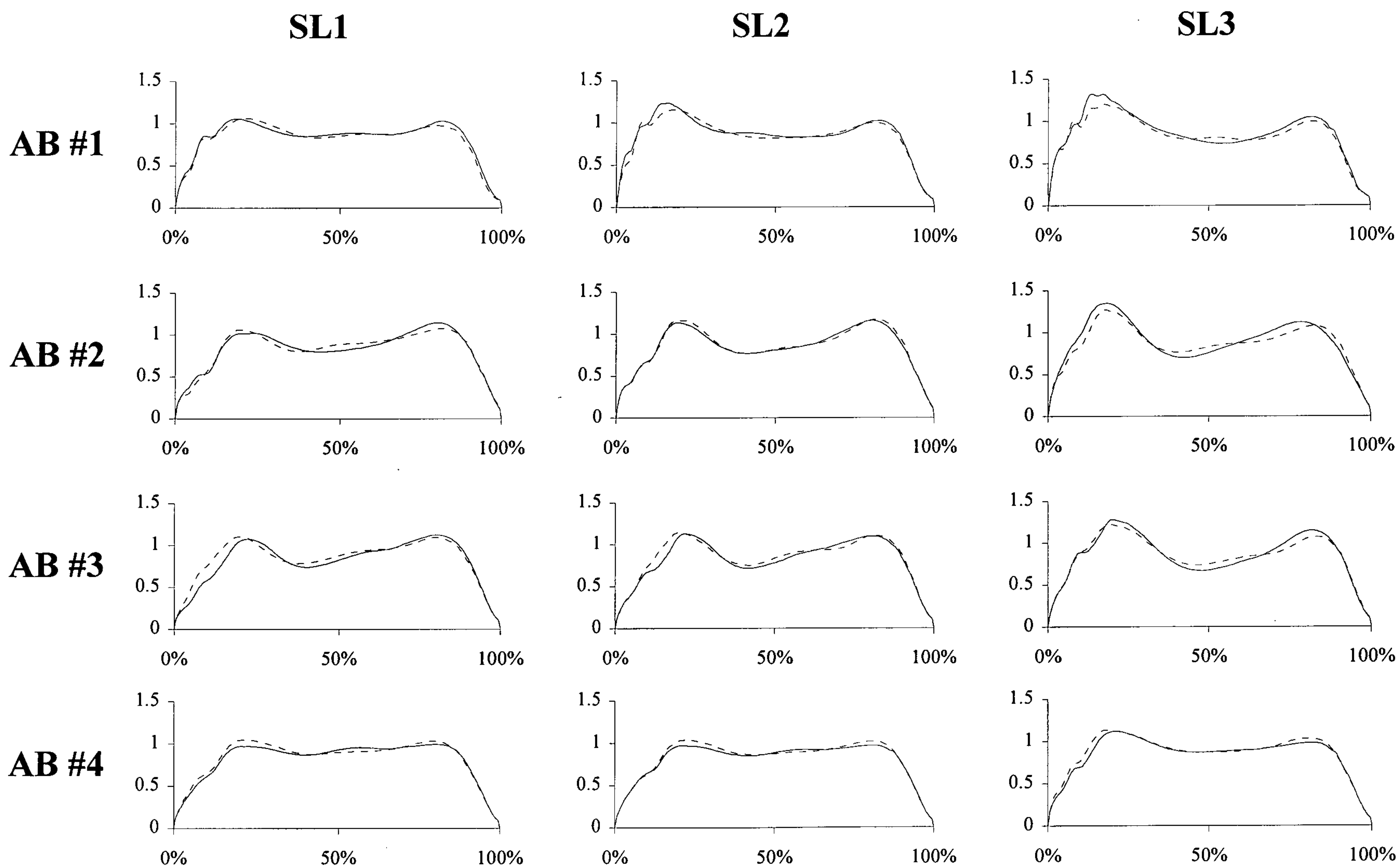

SL1
SL2
SL3
BKA #9
BKA #10
BKA #11
0.4
0
-0.4
0%
50%
100%

APPENDIX B-7: VERTICAL FORCE FROM THE FIRST STEP

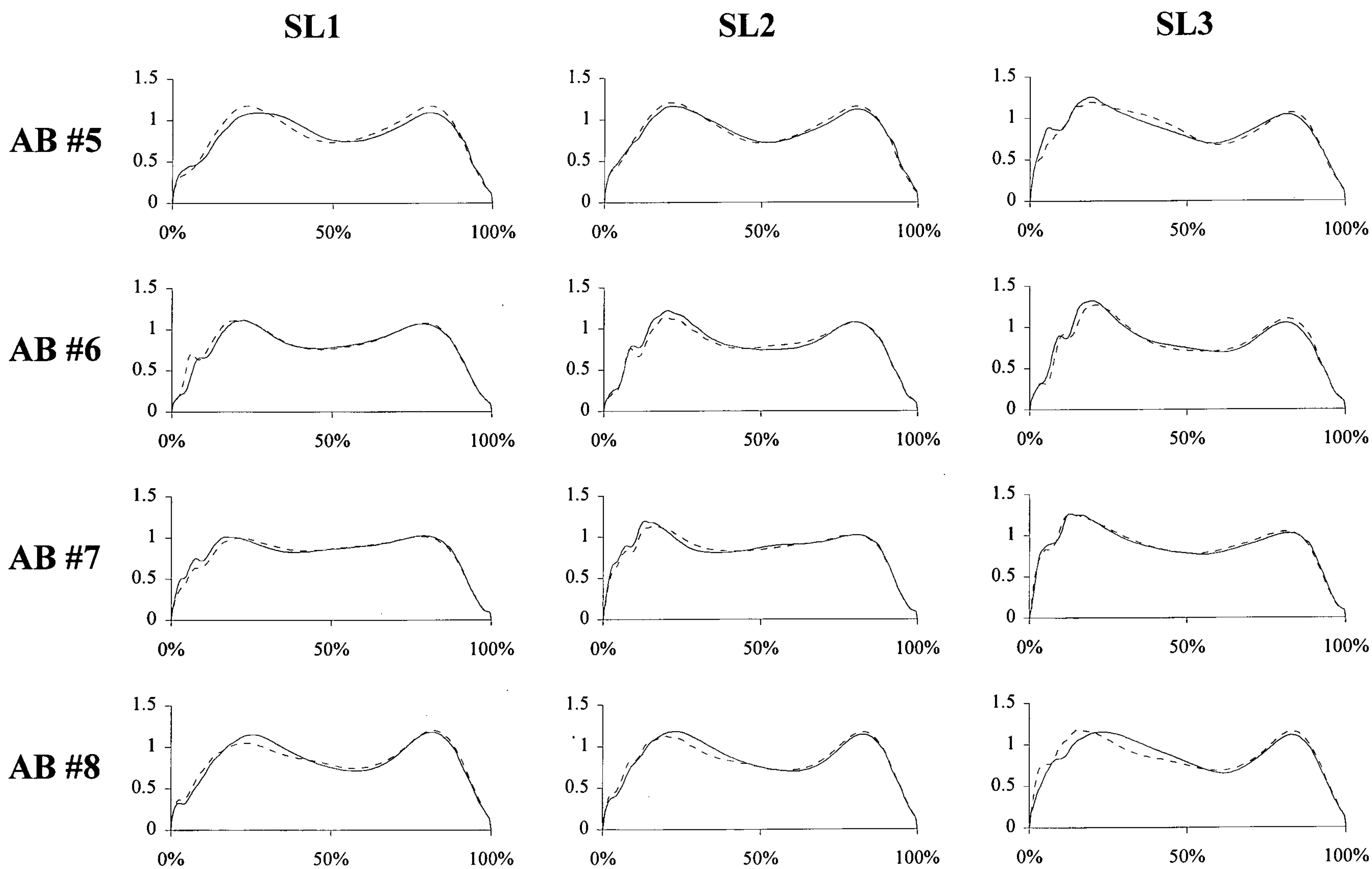
The following six pages contain the mean vertical force-time curve from the initial step for each individual during the gait initiation trials. The solid line represents trials with a left (or intact) lead limb condition while the dashed line represents trials with a right (or prosthetic) lead limb condition. Units are in N/BW for the vertical axis and % gait cycle (1st heel strike to 2nd toe-off of the leading limb) for the horizontal axis.

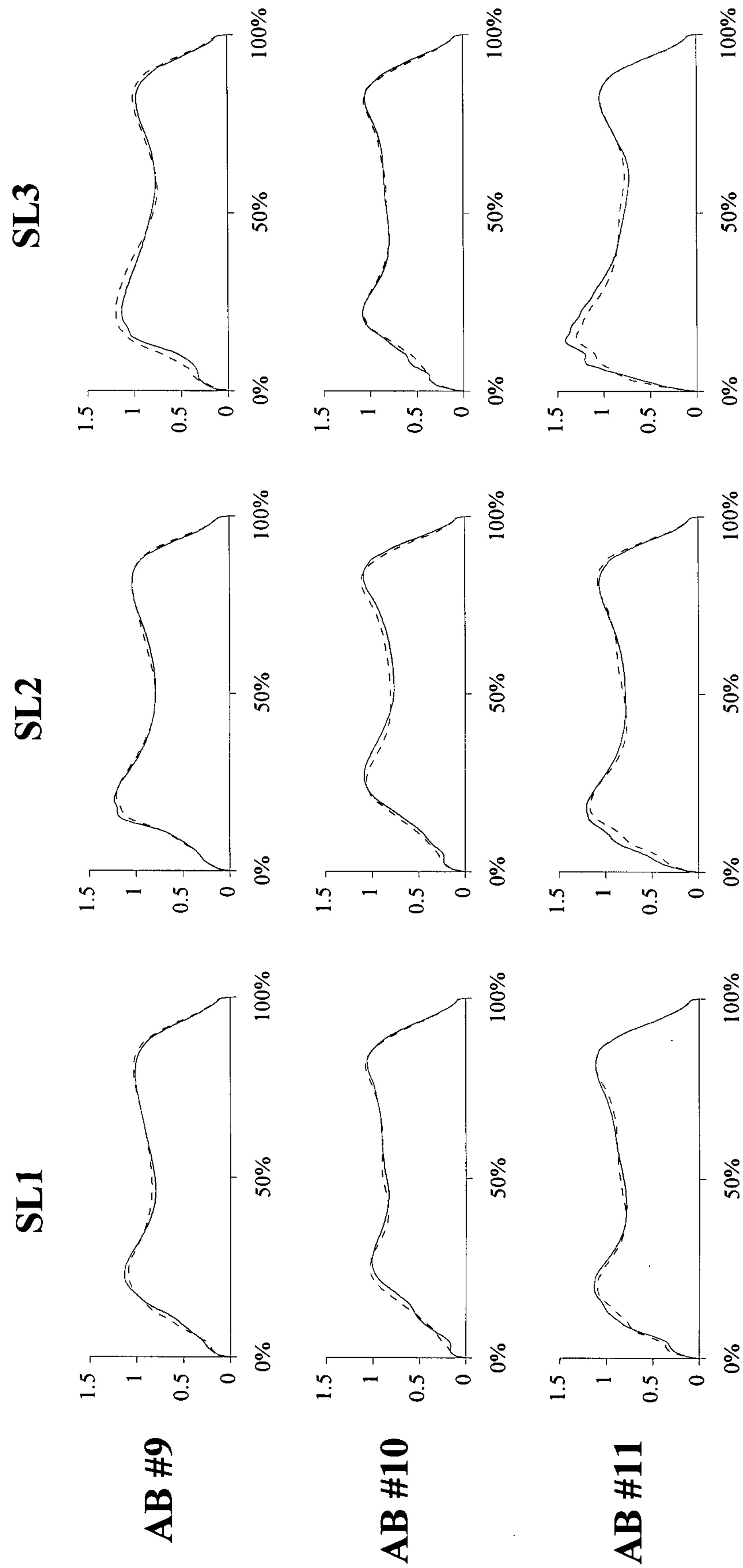
Abbreviations:

- AB = able-bodied (control)
- BKA = unilateral below-knee amputee
- SL = step length condition

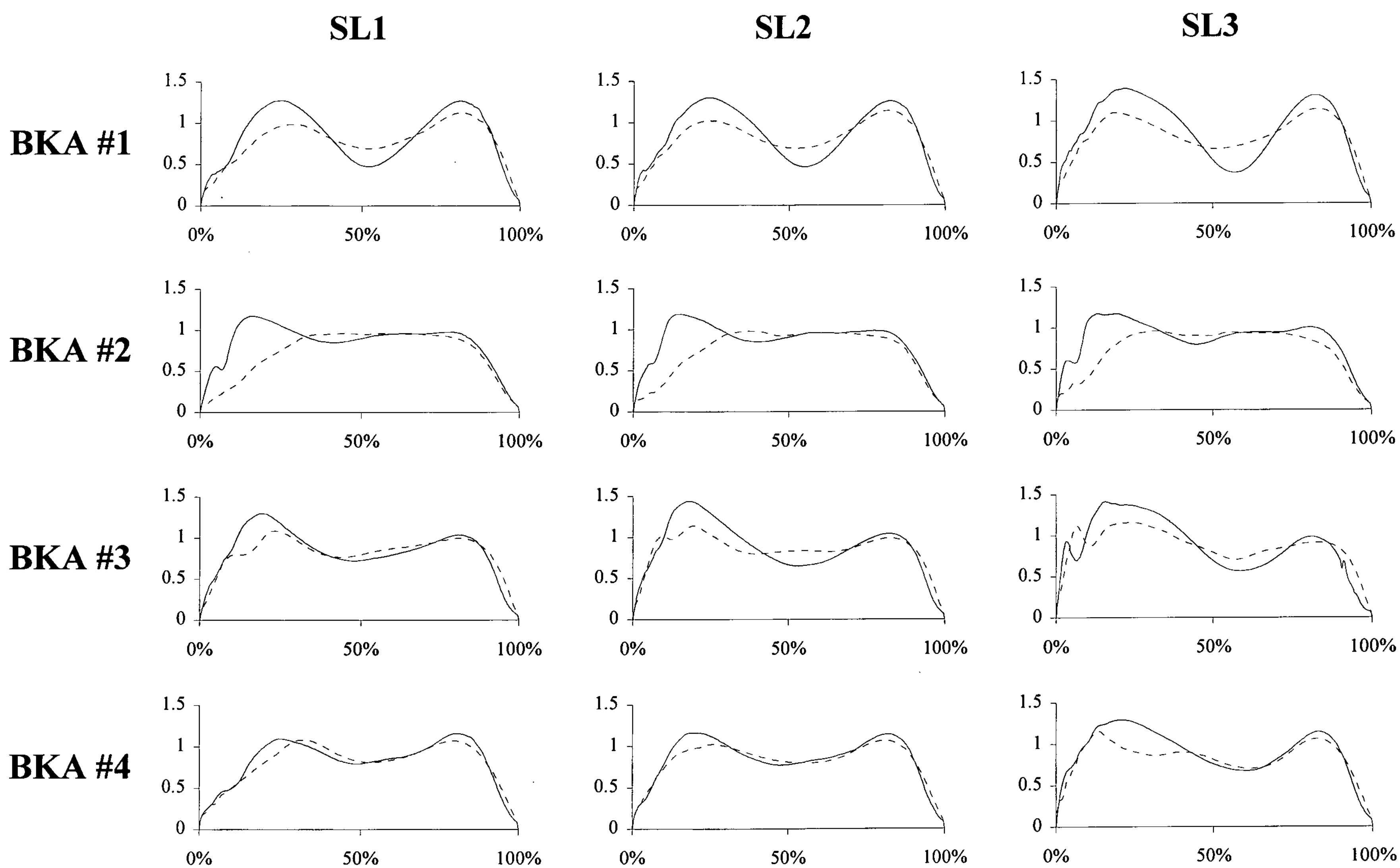


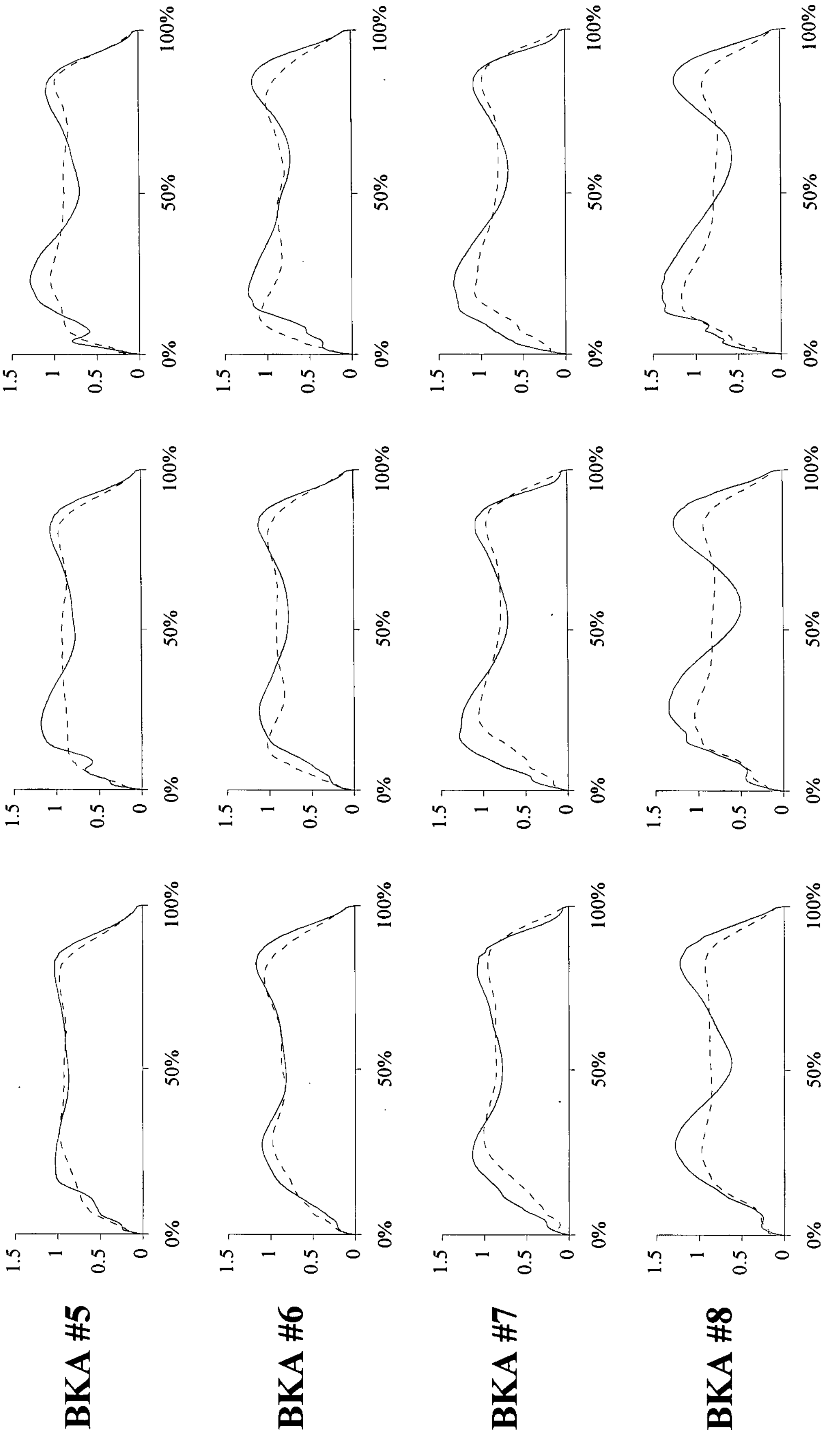
SL1
SL2
SL3
AB #1
AB #2
AB #3
AB #4
1.5
1
0.5
0
0%
50%
100%

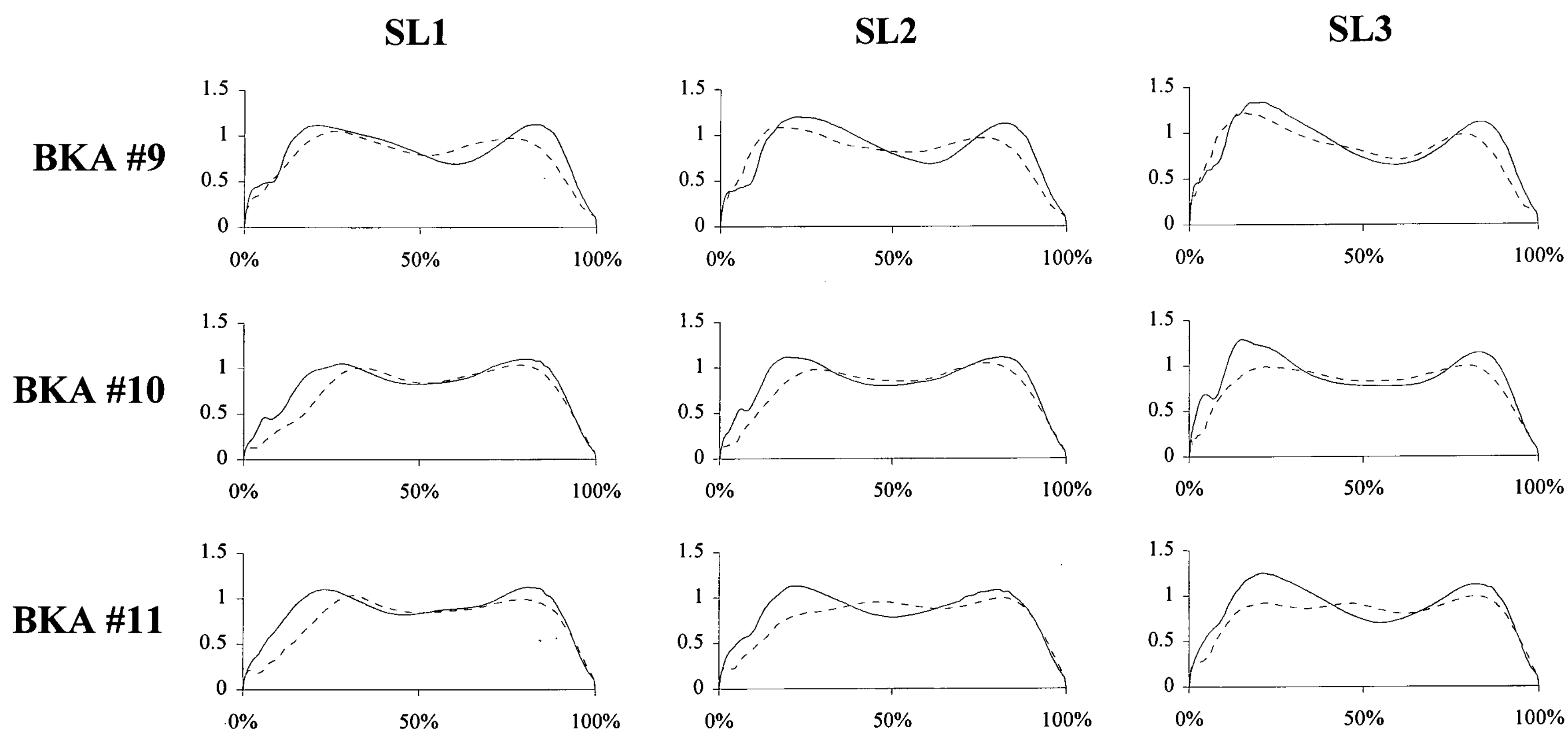

SL1
SL2
SL3
AB #5
AB #6
AB #7
AB #8
1.5
1
0.5
0
0%
50%
100%



SL1
SL2
SL3
AB #9
AB #10
AB #11
1.5
1
0.5
0
0%
50%
100%


SL1
SL2
SL3
BKA #1
BKA #2
BKA #3
BKA #4
1.5
1
0.5
0
0%
50%
100%


SL1
SL2
SL3
BKA #5
BKA #6
BKA #7
BKA #8
1.5
1
0.5
0
0%
50%
100%


SL1
SL2
SL3
BKA #9
BKA #10
BKA #11
1.5
1
0.5
0
0%
50%
100%

APPENDIX B-8: HORIZONTAL IMPULSE

The following two pages contain the mean horizontal impulses (as defined in Chapter 2, page 14) for each individual during the gait initiation trials. Units are in Ns/BW.

Abbreviations:

- AB = able-bodied (control)
- BKA = unilateral below-knee amputee
- SL = step length condition

As Lead Limb	SL 1 Left	SL 1 Right	SL 2 Left	SL 2 Right	SL 3 Left	SL 3 Right
AB #1	-0.02	-0.02	-0.02	-0.02	-0.03	-0.03
AB #2	-0.03	-0.01	-0.03	-0.01	-0.03	-0.02
AB #3	-0.02	-0.02	-0.03	-0.03	-0.04	-0.03
AB #4	-0.02	-0.01	-0.03	-0.02	-0.04	-0.02
AB #5	-0.02	-0.02	-0.03	-0.02	-0.03	-0.03
AB #6	-0.03	-0.02	-0.04	-0.03	-0.04	-0.03
AB #7	-0.03	-0.02	-0.03	-0.02	-0.03	-0.03
AB #8	-0.02	-0.02	-0.02	-0.02	-0.02	-0.03
AB #9	-0.02	-0.02	-0.02	-0.02	-0.03	-0.02
AB #10	-0.02	-0.02	-0.03	-0.03	-0.03	-0.03
AB #11	-0.03	-0.03	-0.03	-0.03	-0.04	-0.03
Group Mean	-0.02	-0.02	-0.03	-0.02	-0.03	-0.03

As Trail Limb	SL 1 Left	SL 1 Right	SL 2 Left	SL 2 Right	SL 3 Left	SL 3 Right
AB #1	-0.12	-0.12	-0.15	-0.14	-0.16	-0.16
AB #2	-0.13	-0.11	-0.14	-0.12	-0.15	-0.13
AB #3	-0.12	-0.11	-0.13	-0.13	-0.16	-0.15
AB #4	-0.11	-0.09	-0.12	-0.10	-0.14	-0.12
AB #5	-0.13	-0.12	-0.14	-0.13	-0.16	-0.14
AB #6	-0.12	-0.11	-0.13	-0.12	-0.15	-0.14
AB #7	-0.12	-0.11	-0.14	-0.13	-0.16	-0.14
AB #8	-0.15	-0.14	-0.16	-0.15	-0.16	-0.17
AB #9	-0.13	-0.12	-0.14	-0.14	-0.16	-0.15
AB #10	-0.12	-0.11	-0.13	-0.14	-0.14	-0.14
AB #11	-0.13	-0.13	-0.15	-0.15	-0.18	-0.17
Group Mean	-0.13	-0.12	-0.14	-0.13	-0.16	-0.15

Summed Impulse	SL 1 Left	SL 1 Right	SL 2 Left	SL 2 Right	SL 3 Left	SL 3 Right
AB #1	-0.14	-0.14	-0.16	-0.18	-0.18	-0.19
AB #2	-0.13	-0.14	-0.15	-0.15	-0.16	-0.17
AB #3	-0.13	-0.14	-0.16	-0.16	-0.18	-0.19
AB #4	-0.11	-0.13	-0.13	-0.14	-0.15	-0.16
AB #5	-0.14	-0.15	-0.16	-0.16	-0.17	-0.19
AB #6	-0.14	-0.14	-0.16	-0.16	-0.18	-0.18
AB #7	-0.14	-0.14	-0.15	-0.16	-0.17	-0.18
AB #8	-0.16	-0.17	-0.18	-0.18	-0.19	-0.20
AB #9	-0.14	-0.14	-0.16	-0.17	-0.17	-0.18
AB #10	-0.13	-0.14	-0.17	-0.16	-0.17	-0.17
AB #11	-0.16	-0.16	-0.18	-0.18	-0.21	-0.21
Group Mean	-0.14	-0.15	-0.16	-0.16	-0.18	-0.18

As Lead Limb	SL 1 Intact	SL 1 Prosthetic	SL 2 Intact	SL 2 Prosthetic	SL 3 Intact	SL 3 Prosthetic
BKA #1	-0.06	-0.01	-0.07	-0.01	-0.08	-0.01
BKA #2	-0.03	-0.01	-0.03	-0.01	-0.03	-0.01
BKA #3	-0.07	0.01	-0.08	0.00	-0.09	0.00
BKA #4	-0.03	-0.01	-0.04	-0.01	-0.04	-0.01
BKA #5	-0.02	-0.02	-0.03	-0.02	-0.03	-0.03
BKA #6	-0.03	-0.01	-0.04	-0.02	-0.04	-0.02
BKA #7	-0.03	-0.01	-0.06	-0.01	-0.06	-0.02
BKA #8	-0.06	0.00	-0.07	0.00	-0.07	0.00
BKA #9	-0.02	-0.02	-0.03	-0.03	-0.03	-0.03
BKA #10	-0.03	-0.01	-0.03	-0.01	-0.03	-0.01
BKA #11	-0.06	0.00	-0.06	0.01	-0.07	0.00
Group Mean	-0.04	-0.01	-0.05	-0.01	-0.05	-0.01

As Trail Limb	SL 1 Intact	SL 1 Prosthetic	SL 2 Intact	SL 2 Prosthetic	SL 3 Intact	SL 3 Prosthetic
BKA #1	-0.18	-0.09	-0.19	-0.10	-0.21	-0.11
BKA #2	-0.10	-0.10	-0.12	-0.11	-0.14	-0.12
BKA #3	-0.20	-0.09	-0.22	-0.11	-0.24	-0.13
BKA #4	-0.12	-0.09	-0.15	-0.11	-0.19	-0.14
BKA #5	-0.11	-0.10	-0.13	-0.11	-0.15	-0.13
BKA #6	-0.15	-0.11	-0.16	-0.12	-0.19	-0.14
BKA #7	-0.13	-0.10	-0.17	-0.12	-0.18	-0.13
BKA #8	-0.15	-0.09	-0.19	-0.11	-0.22	-0.12
BKA #9	-0.12	-0.12	-0.13	-0.13	-0.15	-0.15
BKA #10	-0.13	-0.10	-0.15	-0.12	-0.18	-0.14
BKA #11	-0.14	-0.08	-0.17	-0.09	-0.18	-0.10
Group Mean	-0.14	-0.10	-0.16	-0.11	-0.18	-0.13

Summed Impulse	SL 1 Intact	SL 1 Prosthetic	SL 2 Intact	SL 2 Prosthetic	SL 3 Intact	SL 3 Prosthetic
BKA #1	-0.16	-0.18	-0.17	-0.19	-0.19	-0.22
BKA #2	-0.13	-0.12	-0.14	-0.13	-0.16	-0.15
BKA #3	-0.16	-0.19	-0.19	-0.21	-0.21	-0.24
BKA #4	-0.12	-0.13	-0.15	-0.16	-0.18	-0.20
BKA #5	-0.12	-0.13	-0.14	-0.15	-0.16	-0.18
BKA #6	-0.14	-0.16	-0.16	-0.18	-0.18	-0.21
BKA #7	-0.14	-0.15	-0.18	-0.19	-0.19	-0.21
BKA #8	-0.15	-0.15	-0.18	-0.19	-0.19	-0.22
BKA #9	-0.14	-0.14	-0.16	-0.16	-0.18	-0.18
BKA #10	-0.13	-0.15	-0.15	-0.16	-0.17	-0.19
BKA #11	-0.14	-0.14	-0.15	-0.15	-0.17	-0.18
Group Mean	-0.14	-0.15	-0.16	-0.17	-0.18	-0.20

APPENDIX C: INFORMED CONSENT FORM

THE UNIVERSITY OF BRITISH COLUMBIA

School of Human Kinetics
210, War Memorial Gymnasium
6081 University Boulevard
Vancouver, B.C. Canada V6T 1Z1
Tel: (604) 822-3838 Fax: (604) 822-6842

POSTURAL AND MOVEMENT ADAPTATIONS DURING GAIT INITIATION IN UNILATERAL BELOW-KNEE AMPUTEES

Principal Investigator: David J. Sanderson, Ph. D., Associate Director for Graduate Affairs and Research, 822-4361.

Co-investigator: Craig Tokuno, M.Sc. student, School of Human Kinetics, Biomechanics Lab, room 28 War Memorial Gym, 822-0941.

Purpose:

The goal is to record the movements (kinematic), forces that cause the movements (kinetic) and electrical activity of the muscles (electromyographic) for unilateral below-knee amputees during the initiation of gait. The patterns of three measures will be compared between unilateral, or one sided, below-knee amputees and able-bodied individuals.

This research project is for a Master of Science graduate degree and will be submitted to a journal for publication.

Procedures:

Upon arrival to the biomechanics lab, subjects will change into a close-fitting shirt and shorts. Reflective markers will be placed over the prominence of the hip, knee, ankle, heel and 5th metatarsal head (toe). Small surface electrodes, which are metal plates used to record the electrical activity of the muscle, will be placed on several muscles: quadriceps, hamstrings, calf muscles, shin muscles, and gluteus medius (buttocks). The use of surface electrodes is similar to that used for the measurement of heart activity. Proper footwear will be provided.

Part (a) - Gait Initiation Stage: Participants will be instructed to stand on a force platform (a device, similar to a bathroom scale, that measures the force between the feet and the ground). They will be instructed to listen for an auditory signal. When the signal occurs, the participant will take a single step forward with either their preferred (7 trials) or non-preferred (7 trials) limb at their most comfortable stride length. Following this set of 14 trials, stride length will be manipulated, resulting in another 2 sets (i.e. 2 different stride lengths) of 14 trials. Video recordings, from cameras placed at the sides of the participant (called the sagittal plane), force platform data, and the electromyographic patterns during the initiation step will be recorded.

The total time for each subject's involvement will be about two hours.

Data analysis: The video recordings obtained from the gait initiation trials will be digitized using the reflective markers. This will reveal the angular displacements and velocities for the hip, knee and ankle joints. Force platform data will reveal, through the use of the inverse dynamics approach, the net joint forces and moments across the three lower limb joints. Finally, the EMG data will reveal muscle activity for each of the recorded muscles. These data will be synchronised in time and normalised to the duration of a stride cycle. All data will be compared between the three limb conditions (the normal, intact and prosthetic limbs).

Confidentiality: Any information regarding subject identification resulting from this study will be kept strictly confidential. All data files will be coded such that only the principal investigator will have subject identification information. All such documents will be identified only a code number and kept in a locked filing cabinet. Participants will not be identified by name in any reports of the completed study.

Contact: If I have any questions or require further information regarding this study, I may contact Dr. David Sanderson at 822-4361 or Craig Tokuno at 822-0941.

If I have concerns about my treatment as a research subject, I may contact the Director of Research Services at the University of British Columbia, Dr. Richard Spratley at 822-8596.

Consent: I understand that my participation in this study is entirely voluntary and I may refuse to participate or withdraw from the study at any time without jeopardy.

I have received a copy of this consent form for my records.

I consent to participate in this study.

______________________________ ______________________
Subject Signature: Date

______________________________ ______________________
Signature of Witness: Date

APPENDIX D: PHYSICAL ACTIVITY QUESTIONNAIRE

1. Considering a **7-day period** (a week), how many times on the average do you do the following kinds of exercise for **more than 15 minutes** (write on each line the appropriate number)?

Times per Week

   a. STRENUOUS EXERCISE (HEART BEATS RAPIDLY) ________
   (i.e. running, jogging, hockey, football, soccer, squash, basketball, cross-country skiing, judo, roller skating, vigorous swimming, vigorous long distance bicycling)

   b. MODERATE EXERCISE (NOT EXHAUSTING) ________
   (i.e. fast walking, baseball, tennis, easy bicycling, volleyball, badminton, easy swimming, alpine skiing, popular and folk dancing)

   c. MILD EXERCISE (MINIMAL EFFORT) ________
   (i.e. yoga, archery, fishing from river band, bowling, horseshoes, golf, snow-mobiling, easy walking)

2. Considering a **7-day period** (a week), how often do you engage in any regular activity long enough to work up a sweat (heart beats rapidly)?

OFTEN	SOMETIMES	NEVER/RARELY
☐	☐	☐

Subject #: ________