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Gait initiation of persons with below-knee amputation: The characterization and comparison of force profiles

Stephen A. Rossi, BSME; William Doyle, MS; Harry B. Skinner, MD, PhD
Rehabilitation Research and Development Service and Preservation-Amputation, Care and Treatment, VA Medical Center, San Francisco, CA 94121; Department of Orthopaedic Surgery, University of California, San Francisco, CA 94143; Joint Graduate Group in Bioengineering: University of California Berkeley, Berkeley, CA 94720; University of California San Francisco, San Francisco, CA 94143

Abstract—The purpose of this study was to characterize gait initiation of persons with leg amputation and determine whether prosthetic alignment was a critical parameter in the initiation process. Gait initiation was chosen for study because of the difficult neuromuscular demands placed on the body in negotiating the transition from stance to ambulation. In this investigation, ground reaction force data were collected on seven persons with below-knee amputation. These subjects underwent a series of gait initiation trials while varying prosthetic alignment. An analysis of the data demonstrated key elements in the gait initiation process, including the motion of the center of gravity in preparation for steady-state walking. Significant asymmetries in the force profiles of the residual and nonamputated limbs were also found; gait initiation forces were consistently higher for the prosthetic limb and the timings of maxima and minima were indicative of an intact limb preference. Relatively small changes in prosthesis alignment proved not to have statistically significant effects on generalized force parameters. This result is consistent with the findings of other studies that gait initiation is a motor program with certain invariant characteristics.

Key words: *amputation, biomechanics, gait, prosthesis, walking.*

Address all correspondence and requests for reprints to: Dr. Harry B. Skinner, UCI Medical Center, Department of Orthopaedic Surgery, 101 City Drive South, Building 29-A, Orange, CA 92668-5382.

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INTRODUCTION

Daily activities often require the transition from stance to ambulation, making the initiation of gait an important topic for investigation. Research has historically focused on the study of steady-state gait, leaving the complex process of initiating gait poorly understood. Studies of steady-state gait seem to be of limited usefulness in characterizing abnormalities, since many motor, sensory, and limb deficiencies can be hidden by the ability of individuals to produce similar level-walking kinematics with differing neuromuscular activity. Gait initiation, on the other hand, is a transient movement phenomenon and involves a more complicated integration of neural mechanisms, muscular activity, and biomechanical forces. The extensive involvement of the neuromuscular system implies that an understanding of gait initiation can potentially elucidate the etiology of different gait pathologies.

An early study of gait initiation conducted by Carlsoo (1) used electromyography (EMG) and force plate techniques to describe some of the major characteristics of the initiation process. This study first discovered that the initiation of walking is preceded by postural adjustments and muscle activity that direct the center of pressure (COP) over the swing limb before the transfer of weight to the stance limb. Subsequent work by other authors (2,3) described the neuromuscular aspects of gait initiation in attempts to understand the degree of central motor programming.

These studies suggested that central excitation of alphamotorneurons for both extensors and flexors during the gait initiation sequence remains largely unaffected by decreased spindle afferent output. This does not imply that fusimotor activity is not important, but that compensation may come from other somatic afferentation. A comprehensive study by Mann et al. (4) detailed the progression of events in gait initiation using force plate and EMG data. This study corroborated the findings of Carlsoo in demonstrating that the COP moves posteriorly and toward the swing limb in preparation for motion. Breniere et al. (5) introduced mathematical modeling of an inverted pendulum to explain the behavior of the COP, and to differentiate its behavior from that of the center of gravity (COG). It was this model that first demonstrated that the COP must move behind the COG in order for there to be forward motion. A recent article by Brunt et al. (6) used the invariance of the timing of events in the force and EMG profiles for gait initiation to postulate that gait initiation is a centrally programmed activity.

Persons with below-knee (BK) amputation can particularly benefit from research into gait initiation, for they must make the transition from stance to ambulation with diminished feedback from muscle spindle activity and ankle proprioceptive signals. Many studies have already investigated the steady-state walking characteristics of such persons (7–10). These studies used various combinations of force, EMG, and kinematic data to characterize the asymmetry found in their gait. Not surprisingly, persons with amputation exhibited lower average cadences, velocities, and stride lengths, while increasing the single limb support time on the intact limb. To date, the only work investigating gait initiation in persons with BK amputation, as well as in nondisabled persons is found in the studies by Nissan and Whittle (11) and Nissan (12). In these pilot studies, a protocol was established for comparing the gait initiation of subjects with amputation to that of nondisabled persons; it was found that the former loads the prosthetic limb during initiation less than the latter loads the analogous limb. It was emphasized that future work must be done to validate the use of generalized gait parameters as a clinical tool.

With their compromised muscular control and balance maintenance, persons with BK amputation must compensate for lost ankle function while ensuring safe, efficient gait initiation. Prosthetic alignment was hypothesized to be a critical parameter in this process. For this reason, this study was designed to demonstrate 1) the characteristic COP and ground reaction force (GRF) patterns of a person with BK amputation during gait initiation, and 2) the effect of prosthetic alignment on simple and interpretable force parameters. Understanding amputee gait initiation could potentially lead to amputee training techniques for improved negotiation of transient movement phenomena and to prosthesis modifications that could enhance stability during postural shifts.

METHODS

Subjects

Seven males with BK amputation were recruited for this study through the Special Team for Amputation and Mobility Preservation (**Table 1**). These subjects were selected based on reasonable ambulation skills and a willingness to participate in scientific research, were fully informed of the research protocol, and signed consent forms

Table 1.
Subject data.

Subject	Age	Weight (N)	Height (cm)	Amputation	Prosthesis Type	Socket Type
1	53	823	72	left	endoskeleton	PTB
2	68	1068	68	right	endoskeleton	PTS
3	82	912	72	left	endoskeleton	PTB
4	69	846	69	left	endoskeleton	Corset
5	50	765	69	right	endoskeleton	PTS
6	67	801	73	right	endoskeleton	PTB
7	62	1024	76	right	endoskeleton	PTB

Journal of Rehabilitation Research and Development Vol. 32 No. 2 1995

approved by the University of California San Francisco Committee on Human Research. For the purposes of the investigation, each subject was provided an adjustable prosthesis. This involved the duplication of the current socket, the alignment of the prosthesis by an experienced prosthetist, and the acclimation of the subject to the new prosthesis. These prostheses housed modular adapters that permitted angular adjustment in the coronal and sagittal planes (**Figure 1**). Length adjustments could be achieved with the use of 2 cm spacers.

Measurement Protocol

Before beginning an experimental session, the subject was instructed in the general procedure and allowed to practice until comfortable with the protocol. Subjects were instructed to stand upright and balanced with one foot on each of two side-by-side force plates (Kistler Instrument Corporation, Amherst, NY). At the trigger of a light, the subjects initiated gait, and the collection of force data commenced, continuing for a period of 2.5 seconds. A trial consisted of six runs, three with the left limb leading and three with the right limb leading. In order to establish normative data, each subject underwent trials at normal initiation speed with the prosthesis aligned to the neutral position. Subsequent trials were conducted for each of six prosthetic adjustments. These included the positive and negative adjustment of the following alignment parameters: foot flexion (±5°), foot version (±5°), and leg length (±2 cm).

These alignment parameters were selected because of their importance from a clinical perspective and their ability to move points on the foot in all three directions of coordinate space. The magnitudes of the variations were large enough to cause imbalance and discomfort in the subjects, but small enough to be of clinical significance for the prosthetist performing the final minor adjustments to a prosthesis.

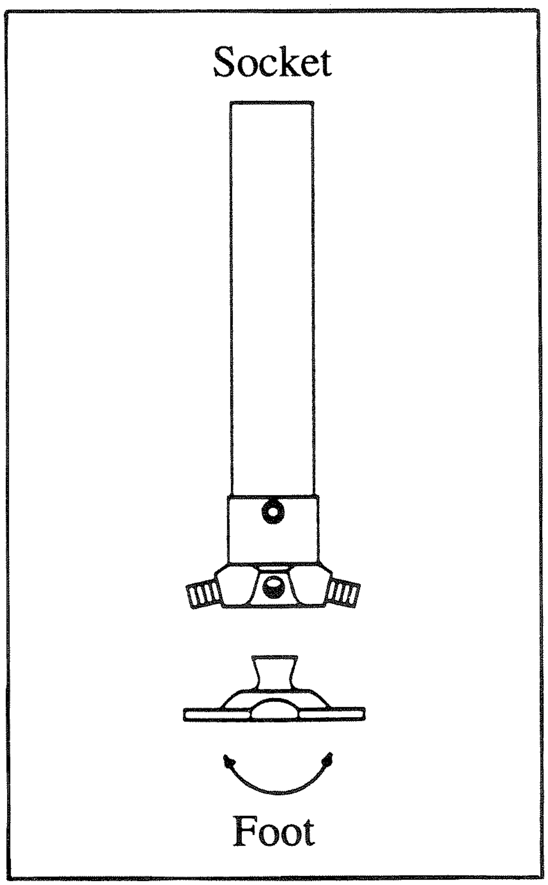

Figure 1.
The prosthetic limb pylon and the foot adaptor that allow for angular adjustment in the coronal and sagittal planes.

Data Collection and Analysis

The force data were collected at 200 Hz on a 486 IBM clone computer equipped with an analog-to-digital board and data acquisition software. The force plate transducer outputs were converted to force and COP data for each run. A FORTRAN program normalized the time to the period of initiation and the force to the body weight of the subject. The period of initiation was defined as the time between the onset of increased swing limb vertical forces and the termination of stance limb vertical forces, 1.25±0.15 (mean ± standard deviation) seconds in duration for the subjects studied. The program also identified the timings and magnitudes of the maxima and minima in the vertical, fore-aft, and medial-lateral force profiles for both the swing and stance limbs. The force magnitudes and their relative timings were tabulated for the 48 runs of each subject. The tabulated results became the data for a two factor ANOVA with repeated measures in order to identify differences produced by choice of leading limb (factor 1) and by alignment changes (factor 2). Significant effects were identified by p values less than 0.01.

RESULTS

The COP and the GRF profiles had similar shapes for all the subjects, regardless of the choice of leading limb. The excursion of the COP (**Figure 2**) began roughly centered between the two feet, usually favoring the side of the intact limb. It consistently moved posteriorly and toward

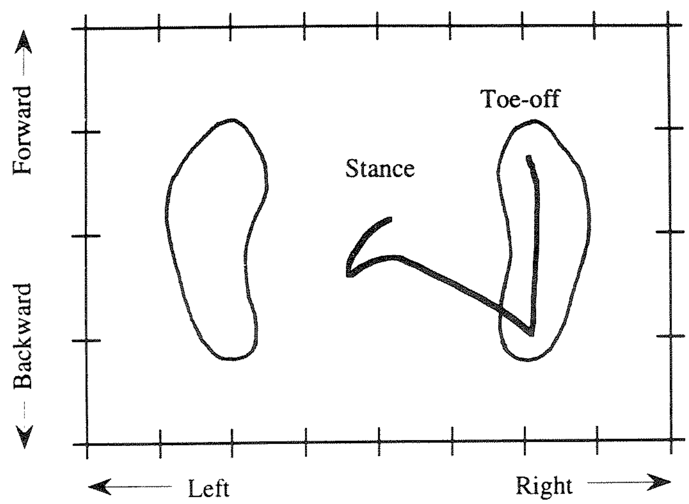

Figure 2.
The excursion of the COP during a typical gait initiation of a person with amputation. This particular example comes from subject 1 and demonstrates a left prosthetic limb lead.

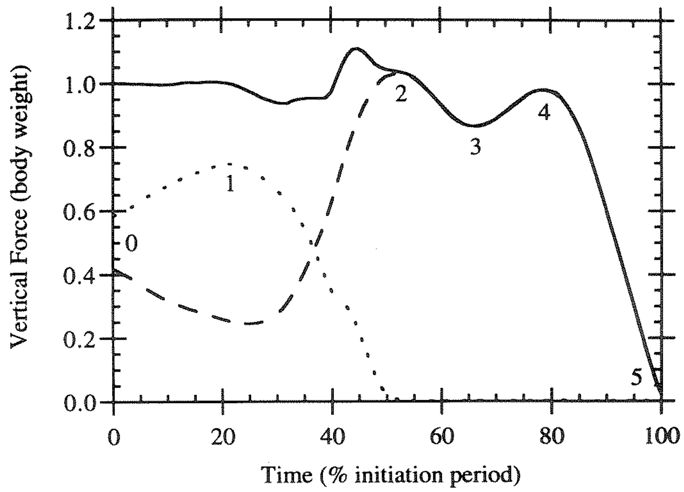

Figure 3.
The summation (solid line) of the contributions to the vertical force by the prosthetic swing (dotted line) and intact stance (dashed line) limbs during initiation for subject 2. The numbers 0 through 5 represent the significant maxima and minima in the gait initiation period, with 1 representing the simultaneous swing maximum and stance minimum.

the swing foot in a preparatory motion. It then reversed direction, moving laterally toward the stance foot heel. The COP finally traveled forward through the stance foot until toe-off. The vertical GRF profiles (**Figure 3**) consistently exhibited a shift in body weight to the swing limb prior to swing foot toe-off, the maximum shift occurring simultaneously with the COP direction reversal described above. The stance limb then began bearing the weight as it went into a single limb stance phase, which was characterized by a double-peaked force pattern also typical of steady-state gait. The summation of the vertical forces from the two limbs exhibited a profile that approximated body weight until single limb support and then followed the double-peaked pattern of single limb support. The fore-aft force profiles (**Figure 4**) demonstrated a net backward acting force exerted primarily by the swing limb prior to swing toe-off and by the stance limb during single limb support. The medial-lateral force profiles (**Figure 5**) demonstrated lateral acting forces, dominated initially by the swing limb and then by the stance limb.

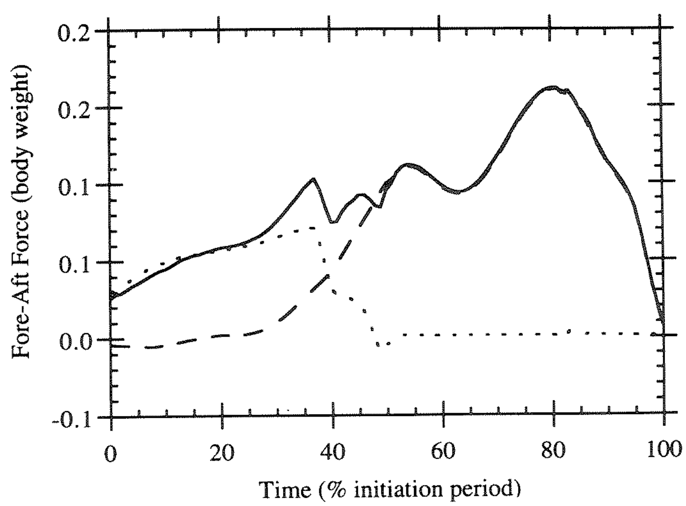

Figure 4.
The summation (solid line) of the contributions to the fore-aft force by the prosthetic swing (dotted line) and intact stance (dashed line) limbs during initiation for subject 2. Positive forces are in the backward direction.

The minima and maxima found on the swing and stance limb vertical GRFs (labeled in **Figure 3**) can be used to demonstrate quantitative differences between gait initiated by a prosthetic limb and gait initiated by an intact limb. Statistical analysis (**Table 2**) shows that the subject with amputation consistently loaded the intact limb more than the prosthetic limb, regardless of its role during initiation, whether as stance or swing limb. This tendency to load the intact limb more was true even during single limb support, a time when all the body weight was being supported by one limb. The timings of the force maxima and minima showed that the onset of prosthetic single limb support was delayed and that the second prosthetic stance

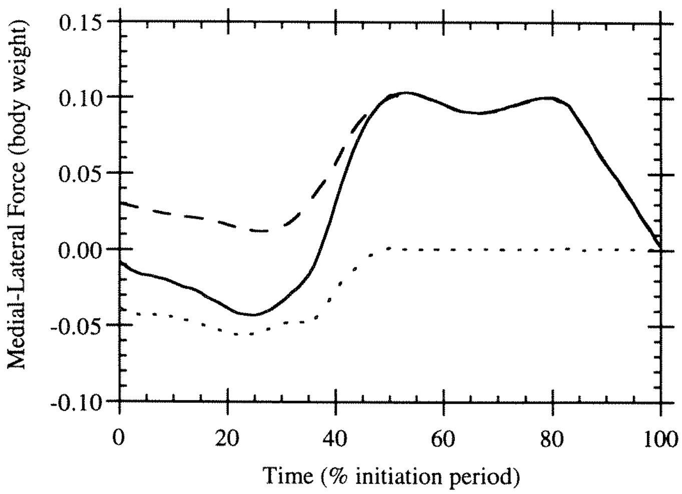

Figure 5.
The summation (solid line) of the contributions to the medial-lateral force by the prosthetic swing (dotted line) and intact stance (dashed line) limbs during initiation for subject 2. Positive forces are toward the left.

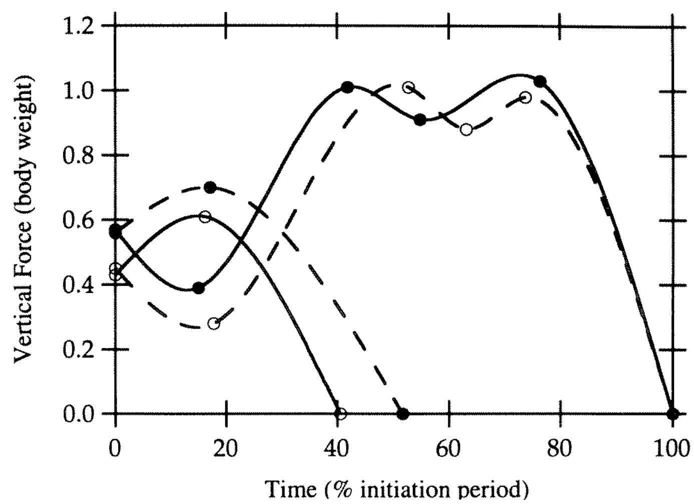

Figure 6.
A comparison between the vertical force profiles for the prosthetic (hollow dots) and intact (solid dots) limbs for the case when the prosthetic limb leads (solid line) and when the intact limb leads (dashed line). The data points are a statistical average of the maxima and minima taken from all subjects and all runs.

Table 2.
Comparison of vertical force maxima and minima (values in force normalized to body weight).

Event	Intact Limb	is	Prosthetic Limb	P Value
Stance 0	0.57 ± 0.09	>	0.45 ± 0.05	<0.0001
Stance 1	0.39 ± 0.16	>	0.28 ± 0.05	<0.0001
Stance 2	1.01 ± 0.03	>	1.01 ± 0.05	<0.1
Stance 3	0.91 ± 0.04	>	0.88 ± 0.05	<0.001
Stance 4	1.03 ± 0.07	>	0.98 ± 0.02	<0.0001
Swing 0	0.56 ± 0.05	>	0.43 ± 0.09	<0.0001
Swing 1	0.70 ± 0.05	>	0.61 ± 0.17	<0.001

Comparison of the timing of maxima and minima (values in percentage of gait initiation period).

Event	Intact Limb	is	Prosthetic Limb	P Value
Stance 1	15.0 ± 3.4	<	17.8 ± 3.3	<0.0001
Stance 2	41.8 ± 4.8	<	52.8 ± 5.9	<0.0001
Stance 3	54.8 ± 4.2	<	63.1 ± 4.8	<0.0001
Stance 4	76.4 ± 4.1	>	73.8 ± 4.2	<0.01
Swing 1	17.1 ± 3.2	>	16.2 ± 3.7	<0.01
Swing 2	51.7 ± 5.7	>	40.5 ± 5.0	<0.0001

peak occurred earlier, implying shorter prosthetic single limb support (**Figure 6**). Despite the inclusion of seven different patients and six prosthetic variations these statistics exhibited small variances, giving p values often much less than 0.0001.

Similar analysis was done for the fore-aft and medial-lateral GRFs (**Table 3**); only the absolute maximum was taken for each curve as the shapes were not as consistent between subjects. It was found that the swing limb peaks occurred earlier in the prosthetic than in the intact limb which is consistent with the vertical force findings. The swing limb fore-aft maximum was higher for the intact limb, and the stance limb medial-lateral force was higher for the prosthetic limb. Statistically significant findings were not as prevalent in the horizontal force profiles because of a higher degree of variability between subjects.

The variation of prosthetic alignment did not demonstrate significant effects on the generalized force parameters. The p values associated with these variables were often around 0.9, giving strong indication that no relationship existed for the selected alignment adjustments. With the seven subjects studied, there was an 80 percent chance of detecting an 8 percent change in either the force magnitudes or timings, assuming a 4 percent standard deviation of the test parameters and 95 percent confidence level for rejecting the null hypothesis. The 4 percent stan-

Table 3.
Comparison of horizontal force maxima (values in force normalized to body weight).

Event	Intact Limb	is	Prosthetic Limb	P Value
Stance - F/A	0.133 ± 0.045		0.127 ± 0.037	N.S.
Stance - M/L	0.078 ± 0.015	<	0.090 ± 0.021	<0.0001
Swing - F/A	0.065 ± 0.016	>	0.026 ± 0.024	<0.0001
Swing - M/L	0.075 ± 0.018		0.069 ± 0.023	N.S.

Comparison of the timing of maxima (values in percentage of gait initiation period)

Event	Intact Limb	is	Prosthetic Limb	P Value
Stance - F/A	78.0 ± 12.3		80.0 ± 6.0	N.S.
Stance - M/L	71.3 ± 7.7	>	66.9 ± 10.7	<0.02
Swing - F/A	25.3 ± 6.9	>	17.4 ± 6.8	<0.0001
Swing - M/L	23.1 ± 5.0	>	18.5 ± 4.3	<0.0001

dard deviation was determined using the within-subjects variation as an estimate of population standard deviation.

DISCUSSION

Center of Pressure Excursion

The fact that the COP moves posteriorly and toward the swing foot prior to swing toe-off has been well documented (1,4,5). Explanations for this phenomenon have been varied, but all lack completeness. The writings of Carlsoo and Mann et al. both postulated that COG follows COP over the swing foot in order to gather the momentum necessary to accelerate the body over the stance foot. They used the firing of the tibialis anterior and the suppression of the gastrocnemius as evidence that muscles are counteracting an initial fall backward, neglecting that these same muscles are involved in the acceleration of the body forward over the feet.

Mathematical analysis by Breniere et al. (5) used an inverted pendulum model to counter the explanations of previous authors by demonstrating that the COP must move backward in order for the COG to fall forward. This analysis was used to proclaim that a COP shift backward and toward the swing limb necessarily implies a fall forward and toward the stance limb.

Because the human body has multiple segments, however, we assert that the postural adjustment of body segments can result in motion in which COP and COG appear to move together. This would occur through a quasi-static process in which individual segments are accelerated and decelerated through small distances repeatedly over time. For example, shifting one's weight from one foot to the other through postural adjustments, provided it is done slowly, would not result in a dramatic oppositely moving COP shift and would in fact move both the COP and COG through the same distance.

Performing a simple force balance can help differentiate the behavior of COG from COP for gait initiation. While two plates registered forces independently in our experiment, the addition of these two forces indicated the net force acting on the body, which served to accelerate the COG. The summation of the fore-aft force profiles demonstrates that the net force acting on the body is always in the forward direction, dominated initially by the swing push-off and finally by stance push-off. As there is no part of the cycle which demonstrates negative fore-aft forces (posteriorly acting forces), there cannot be backward acceleration of the COG; even a quasi-static backward motion would have resulted in small negative initial forces followed by near zero fluctuating forces as the body moved in a controlled manner. Similarly, the summation of the medial-lateral force profiles showed swing limb-dominated lateral forces at the beginning of initiation, indicating that the COG was being pushed toward the stance limb from the start, to be redirected later in the cycle toward the swing limb. It is this oscillatory motion that keeps the COG controlled during bipedal forward motion.

The most interesting aspect of this COG motion is that it had always been previously explained by a falling mechanism, such as in the case of the inverted pendulum model. This concept is countered by the vertical force composite, which showed that prior to single limb support, the net vertical force approximates body weight fluctuating between values above and below body weight. In many of the subjects, this oscillation was even dominated by forces greater than body weight, indicating that there is no clear free-fall. If anything, there was a tendency to accelerate the body upward by the action of the stance foot on the floor. It is not until single limb support that one finds the characteristic dip of falling accelerations. It seems likely that preparatory gait initiation is a controlled forward/lateral push by the swing foot that puts COG and COP in the positions required to have a stance foot-assisted free fall.

Journal of Rehabilitation Research and Development Vol. 32 No. 2 1995

Gait Initiation Asymmetry

A preliminary study by Nissan and Whittle (11) provided normative data on gait initiation force profiles for nondisabled subjects. A comparison of the intact limb vertical force maxima of our subjects to the maxima of their nondisabled subjects reveals that persons with amputation tend to have less pronounced peaks during single limb support (**Table 4**). In a subsequent study by Nissan (12), gait initiation of persons with amputation was compared to nondisabled gait initiation, and this difference in peak vertical force during single limb support was reported.

Our study, however, was the first to compare the profiles from both limbs on the same patient, gaining the statistical power of having paired data and enabling significant asymmetries to be found. It was found that persons with amputation consistently bear more weight on the intact limb whether it serves as the swing or stance limb, which has interesting implications on the postural adjustments necessary to keep the COG moving straight forward. Higher forces were found for the intact limb even during the single limb stance phase of initiation; this implies that the muscles of the intact limb facilitated higher upward accelerations, both in the deceleration of the falling COG and the acceleration of the upward moving COG. Possible reasons for not loading the prosthetic limb include a natural consequence of decreased proprioceptive feedback, a compensation for muscular deficiencies, and a result of pain and discomfort experienced during training.

Asymmetries were also found in the timing of events in the gait initiation cycle of persons with amputation. The loading of the prosthetic stance limb was significantly delayed when compared to the loading of the intact limb. It seemed as though gait initiation of persons with amputation was designed to minimize time spent on the prosthetic limb, which corroborates the findings in steady-state gait studies. The existence of gait asymmetries raises the question whether such a person should ideally be exhibiting symmetrical characteristics when his or her body is necessarily asymmetrical. Design can restore geometric symmetry, but because muscle function cannot be completely replaced by a prosthesis, compromises are necessary. One indication that symmetry might be a reasonable goal of design and training is that the work of Lewallen et al. (13) showed that children with amputation achieved good symmetry in joint kinematics despite physical asymmetry through slower velocities, smaller steps, and increased double limb support.

Prosthetic Alignment

After identifying the force characteristics of gait initiation in subjects with BK amputation, a goal of this research was to identify the effects of prosthetic alignment on these characteristics. Because gait initiation force profiles were remarkably reproducible, they could be almost completely characterized by their maxima and minima. A statistical analysis showed us that our prosthetic alignment variations produced no statistically significant effect on the generalized force parameters. This result is partially explained by the finding by Zahedi et al. (14) that persons with BK amputation can tolerate a great deal of variability in the "optimal" alignment. These results may also indirectly tie into the work of other authors (6,15) who found that gait initiation seems to be part of a central program and contains certain invariant characteristics, such as the timing of muscle and force events. Our experienced prosthesis wearers may have such patterned initiation processes that anything short of drastic alterations would have minimal effects. This does not imply that alignment is not important for the training of patients for efficient gait.

Table 4.
Comparison of the vertical force maxima found in this study and in the literature (values in force normalized to body weight).

Event	Nissan & Whittle (Normals)	Nissan (Intact / Prosthetic Limb)	Our Study (Intact / Prosthetic Limb)
Swing 1	0.76 ± 0.09	0.66 ± 0.08 / 0.60 ± 0.15	0.70 ± 0.05 / 0.61 ± 0.17
Stance 2	1.11 ± 0.05	0.98 ± 0.02 / 1.03 ± 0.04	1.01 ± 0.03 / 1.01 ± 0.05
Stance 3	0.89 ± 0.04	0.91 ± 0.04 / 0.90 ± 0.02	0.91 ± 0.04 / 0.88 ± 0.05
Stance 4	1.10 ± 0.06	0.98 ± 0.02 / 0.99 ± 0.03	1.03 ± 0.07 / 0.98 ± 0.02

CONCLUSIONS

The study of the forces involved in gait initiation has shown that walking is precipitated by a forward lateral push of the COG by the swing limb in preparation for an assisted fall by the stance foot. Persons with amputation perform this process asymmetrically, mostly in their tendency to weight the intact limb as much and as long as possible. The force characteristics of the gait initiation period seem not to be sensitive to small but significant alignment variations, especially with subjects whose gait patterns have long been established. These results do seem to indicate that the majority of effort should be placed at the early stages of gait training during which the asymmetries are established and walking patterns are set.

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