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Use of a Dynamic Balance System to Quantify Postural Steadiness and Stability of Individuals with Lower-Limb Amputation: A Pilot Study

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Abstract

Introduction

Despite rehabilitation and gait training, the gait of individuals with lower-limb amputation is often asymmetric and falls and/or fear of falling are common. Clinical assessments of balance and stability include the Berg Balance Scale and the Dynamic Gait Index. Biomechanical assessments, conducted largely in research laboratories, are more objective, quantitative, and may provide greater resolution. These biomechanical measures include postural sway during both unilateral and bilateral standing tasks and the dynamic postural response to applied or volitional perturbations. The objective of this study was to investigate the utility of a dynamic balance system, a relatively new clinical tool incorporating dual force plates similar to that used in research laboratories, to assess the postural steadiness and stability of a small, diverse population of persons with lower-limb amputation. The specific aim was to investigate whether differences in balance of persons with amputation due to changes in prosthetic componentry were reflected in the resultant data.

Materials and Methods

Dynamic balance testing was conducted using the Bertec Balance Advantage–Dynamic CDP system on five adult subjects with varying levels of lower-limb amputation. Trials were conducted in both the subjects' current prosthesis and alternative prosthetic componentry after a 1-week acclimation period. Specific tasks included limits of stability, weight-bearing squats, and unilateral stance.

Results

Subjects had difficulty shifting their weight during the limits of stability task; both the maximum excursion and anteroposterior directional control varied with prosthetic componentry. Load sharing also varied with prosthetic componentry. Load sharing became more asymmetric as knee flexion increased during the weight-bearing bilateral squat tasks, with less weight supported on the prosthetic limb. Finally, the metrics for the unilateral stance task varied with prosthetic componentry.

Conclusions

The dynamic balance system tasks and related metrics demonstrated the potential to discern differences in balance in persons with amputation due to changes in prosthetic componentry. Further study is needed to investigate these parameters, their correlation with clinical measures of balance, and the effects of both prosthetic componentry and alignment.

There are nearly 2 million persons with amputation in the United States, with 113,000 lower-limb amputations performed each year.[1] Individuals with lower-limb amputation require an artificial limb or prosthesis for ambulation, transfers, and activities of daily living. Despite rehabilitation and gait training that emphasize weight shifting, balancing tasks, postural transitions, and walking exercises, the gait of individuals with lower-limb amputation is often asymmetric with preferential loading on their intact or sound limb.[2] Falls and fear of falling are common,[3] in part due to this asymmetric loading, weight shifting challenges, loss of ankle motor control and proprioception with the amputated limb, and mobility restrictions of the prosthetic foot and ankle.[2,3]

Common clinical assessments of balance and stability include the 14-task Berg Balance Scale (BBS)[4,5] and the 8-task Dynamic Gait Index (DGI).[6] Both assessments are typically conducted by physical therapists in a clinical setting to subjectively characterize balance and gait, as well as potential fall risk, in older adults. The BBS protocol includes both static and dynamic tasks that incorporate typical activities of daily living, such as sitting to standing, standing to sitting, standing with eyes closed, and standing on one foot.[4] Each task is rated on a 5-

point scale indicating functional level, “0” representing the lowest and “4” the highest level. The resulting overall score is then used to classify the individual's fall risk as low, medium, or high. The DGI protocol subjectively characterizes the ability to conduct dynamic tasks such as steady-state walking, walking with changing speeds, walking while avoiding obstacles, pivoting while walking, and stair climbing.[6] Each task is rated on a 4-point scale indicating impairment level, “0” representing severe impairment and “3” reflecting normal functional level. A total score of 19 or less (maximum of 24) is indicative of increased fall risk.

In contrast to these clinical assessments of balance that require only simple tools and provide subjective, qualitative results that have been validated for various populations of older adults, biomechanical assessment of balance and stability requires one or two force plates to quantify the ground reaction forces and their respective location, the center of pressure (COP). Balance may then be assessed using analysis of the time-varying coordinates of the COP or the stabilogram.[7] The related objective, quantitative measures of postural steadiness or performance of the postural control system under static conditions include postural sway during both unilateral and bilateral standing tasks for any individual. Specific measures of postural steadiness include:

- COP or sway path[7]: total excursion, average distance or root-mean-square distance from the geometric mean COP, peak anteroposterior and mediolateral displacement of the COP over the plane of support
- Sway area[7]: cumulative circumscribed area, enclosed area, area based on mean distance, confidence ellipse area, enclosed area relative to the base of support
- COP velocity[7]: mean velocity

Postural stability investigates the dynamic postural response to applied or volitional perturbations. Postural stability therefore examines motion of the COP of the human body in response to various volitional or external perturbations (e.g., sudden translational movement in the anteroposterior direction and rotational displacement of the ankle via sudden dorsiflexion or plantarflexion). In addition to these time-domain measures, frequency domain or spectral analysis has also been conducted to investigate both postural steadiness and stability.[7] Finally, as postural control is dependent on visual, vestibular, and proprioceptive feedback, the effects of visual feedback postural steadiness are often investigated by including both eyes open and eyes closed conditions.[7]

Dynamic balance systems incorporating force plates have been developed to facilitate quantitative biomechanical assessment of balance and postural stability in clinical environments (e.g., NeuroCom Smart Balance Master systems,[8] Balance Advantage systems[9]). These systems include hardware and software to enable postural stability assessment during a variety of protocols, such as sensory organization, motor control adaptation, limits of stability, weight-bearing squat, rhythmic weight shifting, and unilateral stance. A potential advantage of the Bertec system is the incorporation of an immersive virtual reality environment for high-resolution quantification of postural steadiness and stability, measures that could previously only be assessed in a research environment. The availability of dynamic balance systems in rehabilitation clinics therefore has the potential to enhance clinical assessment beyond that of subjective scoring of activities of daily living.

The purpose of this study was to investigate the utility of a new clinical tool to assess balance in persons with lower-limb amputation. Specifically, the Bertec Balance Advantage–Dynamic CDP system and related clinical protocols were used to investigate whether the resultant data could discern differences in balance in persons with amputation due to changes in prosthetic componentry.

METHODS

Subject Recruitment

A convenience sample of five adult subjects (one female; mean age of 51.6 years; Table 1) with varying levels of lower-limb amputation were recruited from the client database of a local prosthetist. Subject selection criteria were as follows: lower-limb amputation, K2 to K3 (Medicare Functional Classification Level, Centers for Medicare & Medicaid Services: K2 indicates community ambulatory; K3, community ambulatory with variable cadence) activity level (e.g., community ambulators), and use of a definitive prosthesis for at least 6 months. Subjects using an assistive device were not excluded from participation. Subject 4 was originally classified as a K3 ambulator; her prosthesis therefore incorporated K3 components. However, recent health issues necessitated use of walker and potential reclassification as a K2 ambulator. Written informed consent was solicited before the subject participation.

Table 1. Subject anthropometry, prosthetic history, and test prostheses

Subject	Amputation Level	Cause of Amputation	Time post-amputation, yr	Age, yr	Height, m	Weight, kg	Activity Level	Initial Prosthesis	Modified Prosthesis
1	L transtibia	Vascular	2	61	1.880	99.79	K3	Catalyst foot[a]	Seattle LightFoot[b]
2	R transtibial[c]	Trauma	12	45	1.854	124.2	K3	BiOM foot[d]	Seattle LightFoot[b]
3	L transfemoral	Vascular	6	31	1.752	71.2	K3	Total Knee,[c] Flex foot[e]	Safety knee,[f] single-axis foot[f]
4	R transtibial[g]	Vascular	9	71	1.635	116.75	K3/K2	Catalyst foot[a]	Seattle LightFoot[b]
5	Bilateral transtibial	Infection	6	50	1.780	77.8	K3	Catalyst foot[a]	Endolite Avalon hydraulic multi-axis foot[h]

[a] Steeper USA, San Antonio, TX.
[b] Trulife, Poulsbo, WA.
[c] Also left transhumeral amputation.
[d] BionX, Bedford, MA.
[e] Össur Americas, Foothill Ranch, CA.
[f] Ottobock, Austin, TX.
[g] Originally K3 ambulator; postcancer diagnosis and treatment, now uses walker for ambulation.
[h] Endolite USA, Miamisburg, OH.

Test Protocol

Dynamic balance testing was conducted with the subjects' current prosthesis. After testing, alternative prosthetic componentry (e.g., K2 foot/knee replaced with K3 foot/knee or vice versa) was integrated with the subject's original socket. After a 1-week acclimation period,[10] testing was repeated. During all trials, the subjects were secured in an overhead fall-arrest harness. All procedures were reviewed and approved by the University of Wisconsin-Milwaukee Institutional Review Board.

In addition to motion analyses conducted for a separate study,[11,12] dynamic balance assessment was conducted using the Balance Advantage–Dynamic CDP system (Bertec, Columbus, OH). Tasks included a subset of standard clinical protocols for dynamic balance systems: limits of stability (postural stability in response to volitional dynamic movement), weight-bearing squats, and unilateral stance (postural steadiness). During the limits of stability task, the subject shifted his or her center of mass (COM) anteriorly, laterally, posteriorly, and medially based on a visual projection of the COM in the current and target locations. As per the clinical protocol, eight weight shifts were conducted at 45° increments (e.g., 0°, forward [F]; 45°, right-forward [RF]; 90°, right [R]; 135°, right-backward [RB]; 180°, backward [B]; 225°, left-backward [LB]; 270°, left [L]; 315°, left-forward [LF]), see Figure 1A; the duration of each weight shifting trial was 10 seconds. The displayed projection of the subject's COM in the visual field was scaled based on the subject's height to achieve lean angles of approximately 8° to the front and sides and 4.5° to the back.[13] Subjects were instructed to shift their weight by moving their pelvis and torso without flexing their hips or knees, keeping their heels in contact with the force plates and their arms/hands at their sides. For the weight-bearing squat task, the subjects stood on both limbs with their hands on their hips. Trials were conducted with the knees in extension, followed by trials at 30°, 60°, and 90° knee flexion; three 5-second trials were performed for each position. Unilateral stance trials were performed on the prosthetic limb only and were conducted with the eyes open and eyes closed. Three 10-second trials were conducted in both conditions, with the eyes open condition preceding the eyes closed condition.

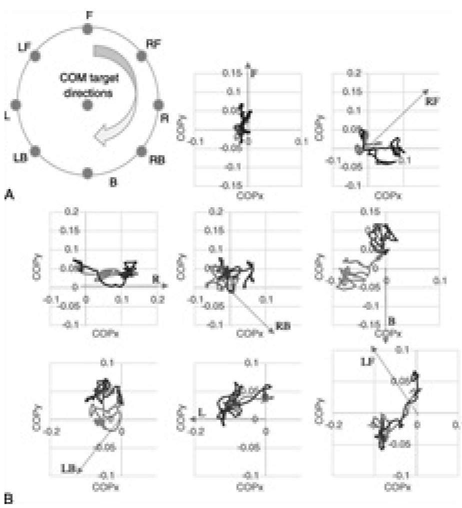

Figure 1: A, Limits of stability target directions. B, Sample stabilograms during the limits of stability task (subject 2). K3 foot trials are shown in gray; K2 foot trials are shown in black; +x = right, +y = anterior; the arrow illustrates the intended direction. F indicates forward; RF, right forward; R, right; RB, right backward; B, backward; LB, left backward; L, left; LF, left forward.

Data Analysis

Stabilograms were generated for the limits of stability trials, as illustrated in Figure 1A, contrasting the two prosthetic components. These stabilograms were characterized in terms of the maximum excursion (percentage of motion in the intended direction, toward the projected COM target) and directional control (ratio of movement in the intended direction relative to extraneous movement).[9] The weight-bearing squat trials were assessed in terms of the relative load on the prosthetic limb (e.g., 50% reflects balanced weight sharing and values exceeding 50% reflect greater load supported by the prosthetic limb). The unilateral stance trials were assessed in terms of the percentage of the trial duration during, which the subject successfully balanced solely on his or her prosthetic limb, as well as the number of sound limb toe touches during the various trials. With the exception of the limits of stability task for which only one trial was performed for each weight shift, the mean and standard deviation across trials were calculated for all other tasks for a given subject.

Results

The metrics associated with all dynamic balance system tasks reflected differences due to prosthetic componentry.

Stabilograms were plotted for each of the eight target directions tested in the limits of stability task (Figure 1B, representative subject). These plots were characterized in terms of the maximum excursion and directional control[7] (Figure 2); both measures varied with prosthetic componentry. For example, the patient with the right transtibial amputation (subject 2: Figures 1B and 2) demonstrated improved maximum excursion in five (F, RB, B, LB, L) of the eight directions with his K3 foot. His directional control was also improved with the K3 foot for all weight shifts except those in the forward and backward directions. The limits of stability data are summarized for all subjects in Figure 2. In general, subjects with lower-limb amputation often demonstrated difficulty weight shifting toward their prosthetic side as well as posteriorly.

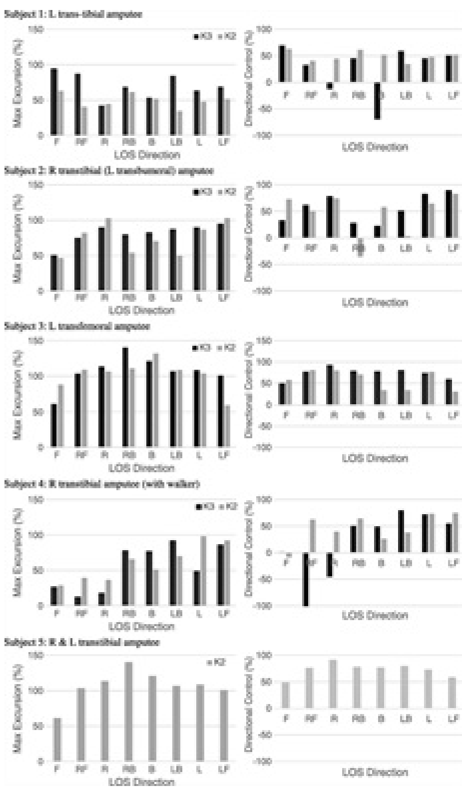

Figure 2: Summary limits of stability results for all subjects; limits of stability tasks were performed by subject 5 with his K2 feet only. Maximum excursion in the intended direction (left) and directional control (right). F indicates forward; RF, right forward; R, right; RB, right backward; B, backward; LB, left backward; L, left; LF, left forward.

The prosthetic limb loading during weight-bearing squat trials are summarized in Figure 3 for all subjects. Weight-bearing symmetry was also affected by prosthetic componentry. As knee flexion increased, for example, deeper squat approaching a seated position, less weight was supported on the prosthetic limb (e.g., prosthetic limb loading <50%). Note that subject 3, a person with a transfemoral amputation, was largely unable to support weight in his prosthetic leg when the knee was flexed–despite the inclusion of stance flexion in both his total knee and safety knee.

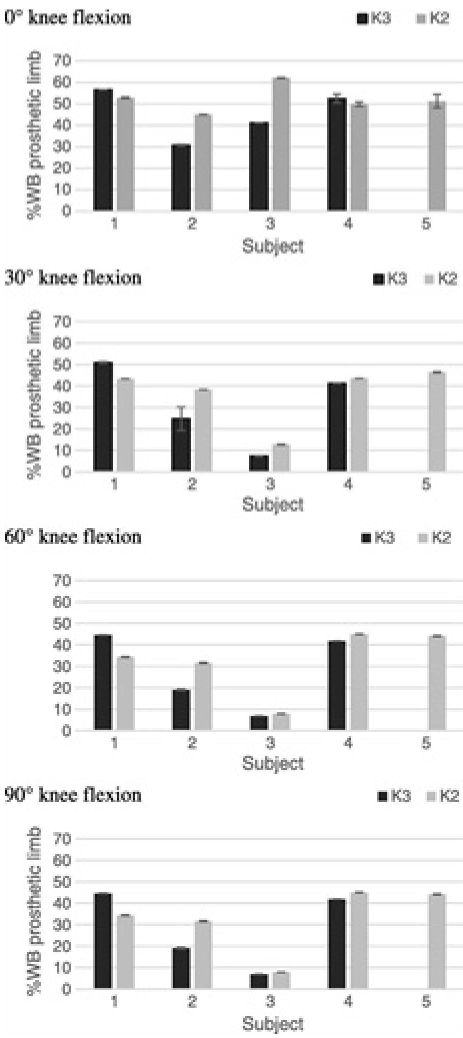

Figure 3: Summary of weight bearing on the prosthetic limb during squat trials at varying degrees of knee flexion for each subject for both K3 (black) and K2 (gray) prosthetic components. Note that weight-bearing squat trials were not conducted for subject 5 in his K3 prosthetic limbs.

Mean data over three trials for unilateral stance on the prosthetic limb are presented in Figure 4 for subjects 1 to 3; this task was not performed by subject 4 (walker-assisted gait) and subject 5 (individual with bilateral amputations). Unilateral stance performance varied with prosthetic componentry, although no specific trends were observed due to the large standard deviations between trials. In general, subjects were more stable in the eyes open condition, as demonstrated by the higher percentage of trial duration sustained on the prosthetic limb and/or fewer sound limb touches.

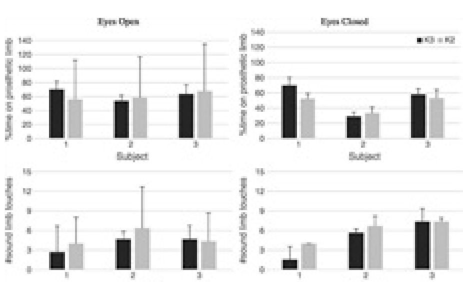

Figure 4: Summary of unilateral stance trials on the prosthetic limb. Performance was assessed in terms of the percentage of the trial duration during which the sound limb did not contact the floor or force plate (e.g., %time of prosthetic limb) and the number times during the trial duration in which the sound limb touched the floor or force plate. Data means and

standard deviations are reported for three trials in the eyes open (left) and eyes closed (right) conditions for each subject for both K3 and K2 prosthetic components. Note that unilateral stance trials were not performed by subject 4 (required a walker) nor subject 5 (individual with bilateral transtibial amputations).

Discussion

This pilot study involved the application of various clinical balance and stability protocols using the Bertec dynamic balance system for a diverse population of individuals with lower-limb amputation. Overall, the dynamic balance system proved successful for quantifying balance and stability in individuals with varying levels of lower-limb amputation. More importantly, the system and related standard test protocols were able to detect differences in the corresponding balance metrics as prosthetic components were varied. These protocols could be performed by persons with lower-limb amputation with minimal instruction and training. The time to complete these test protocols was comparable to that required to administer the BBS and DGI protocols.

The limits of stability protocol has direct relevance to traditional rehabilitation training, during which physical therapists emphasize weight shifting both mediolaterally and anteroposteriorly in preparation for walking, reaching, dressing, and stand-to-sit and sit-to-stand activities; weight shifting is also important for stumble recovery and fall prevention. Limits of stability assessment may also aid prosthetists regarding prosthetic alignment and/or prescription if peak excursion in a particular target direction and/or directional control parameters indicates that current prosthetic components prohibit or impair weight shifting in a specific direction.

As illustrated in the sample stabilograms (Figure 1B), these subjects with lower-limb amputation demonstrated difficulty shifting their weight in the various target directions, frequently exhibiting motion of their COM in nontarget directions (e.g., anterior or left instead of posterior; anterior and forward instead of right posterior). These extraneous movements in the nontarget direction are reflected in the relatively modest values for directional control (Figure 2). All subjects demonstrated poor anteroposterior directional control (e.g., poor backward directional control for subjects 1, 2, and 4; poor forward directional control for subjects 1, 3, and 5). These challenges are consistent with the loss of ankle motor control and proprioception with the amputated limb, as well as mobility restrictions of the prosthetic foot.[14,15] These balance impairments may be addressed, at least in part, by an alternative prosthetic foot with increased/decreased ankle rotational stiffness and/or prosthetic alignment variations (e.g., foot forward/backward, plantarflexed or dorsiflexed foot).

The weight-bearing squat protocol also assesses anteroposterior balance and has direct relevance to the stand-to-sit and sit-to-stand mobility tasks. The weight-bearing squat protocol was developed because increased knee flexion stresses the lower-limb joints; increased knee flexion often results in asymmetric loading due to sensory, proprioception, and/or strength loss, and is also affected by pain.[13] With the exception subject 5, a person with bilateral transtibial amputations, who demonstrated symmetric loading for all positions, load sharing was most symmetric during bilateral stance when the knee was extended. As knee flexion increased, for example, deeper squat approaching a seated position, less weight was supported on the prosthetic limb. Although the K3 and K2 feet for subjects 1 and 4 were the Catalyst and Seattle LightFoot foot, respectively, only subject 1 demonstrated greater limb loading on the prosthetic limb with the K3 versus K2 foot. Subjects 2, 3, and 4 consistently demonstrated increased prosthetic limb loading with the K2 foot when the knee was flexed.

For many subjects, despite instruction to squat to 30°, 60°, and 90°, these target positions were not attained, although knee flexion increased relative to the initial orientation. This discrepancy was likely influenced by the Bertec's projected instruction figures that illustrated 10°, 20°, and 30° knee flexion, contrary to the verbal instructions and demonstrations. Regardless, the trial progression corresponded to increased knee flexion. As prosthetic components were tested on separate days, knee flexion positions may have differed between sessions, and comparison of prosthetic limb loading between prosthetic components may therefore be suspect.

Future study should include Bertec system modification to reflect more accurate instructional pictures, consistent with the verbal instructions; kinematic sensors might also be used to document the actual knee flexion angle.

Prosthetic components affect weight-bearing symmetry and prosthetic loading. While a squat may be performed solely with hip and knee flexion, some ankle dorsiflexion is usually observed. Ankle dorsiflexion is affected by the type of prosthetic foot as well as the foot's rotational stiffness. For individuals with transfemoral amputations, stability during prosthetic knee flexion is affected by prosthetic knee design (e.g., stance flexion) and alignment. Prosthetic components and alignment are therefore expected to affect prosthetic loading during the weight-bearing squat protocol.

Unilateral stance tasks are highly sensitive but not specific.[13] This lack of specificity was demonstrated in this pilot study for which the percentage of trial duration on the prosthetic limb and the number of sound limb toe touches varied nonspecifically with prosthetic componentry. In general, uUnilateral stance was the most challenging task performed during this study. Because of their dependence on a walker and bilateral amputation, respectively, subjects 4 and 5 were not asked to perform this task. Performance during unilateral stance is affected by lower-limb strength, weight-bearing and sensory balance control, and movement strategies, as well as prior practice.[16–18] In fact, the younger subjects expressed interest in practicing this task to improve their performance. Future studies might include unilateral stance trials on the sound limb before performing such trials on the prosthetic limb to provide some initial training, identify potential balance impairments, establish an individual-specific measure of comparison, and help develop subject confidence. These protocols might also be used to assess the effects of repetition, exercise, and rehabilitative therapy on the related performance metrics.

Conclusions

This pilot study used a dynamic balance system to quantify the balance and stability of a small, diverse population of five individuals with lower-limb amputations who are community ambulators capable of variable cadence. Limits of stability, weight-bearing squats, and unilateral stance tasks were conducted using a fall arrest harness. Both the maximum excursion in the target direction and the directional control parameters demonstrated impairments for these subjects with amputation, particularly in the anteroposterior directions, and appeared to vary with prosthetic componentry. The loading on the prosthetic limb decreased with increasing knee flexion as the subject squatted. This measure also varied with prosthetic componentry, although the magnitude of knee flexion was poorly controlled between test sessions. The percentage of trial duration with unilateral support on the prosthetic limb, as well as the number of sound limb toe touches during the respective trials, was also measured for a subset of this population with lower-limb amputation and demonstrated potential dependence on prosthetic componentry. In summary, the dynamic balance system and limits of stability, weight-bearing squats, and unilateral stance tasks demonstrated the potential to discern differences in balance of persons with amputation due to changes in prosthetic componentry. Further study of an expanded, more homogeneous population of individuals with lower-limb amputation is needed to formally investigate these parameters, their correlation with clinical measures of balance such as the BBS and DGI, and the effects of both prosthetic componentry and alignment.

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