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Dynamic balance changes within three weeks of fitting a new prosthetic foot component

Jenny A. Kent[a,*], Nicholas Stergiou[a,b], Shane R. Wurdeman[a,1]

[a] Department of Biomechanics and Center for Research in Human Movement Variability, University of Nebraska at Omaha, 6160 University Drive, Omaha, NE 68182-0860, USA
[b] College of Public Health, 984355 University of Nebraska Medical Center, Omaha, NE 68198-4355, USA

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ABSTRACT

Balance during walking is of high importance to prosthesis users and may affect walking during baseline observation and evaluation. The aim of this study was to determine whether changes in walking balance occurred during an adaptation period following the fitting of a new prosthetic component.

Margin of stability in the medial-lateral direction (MOS_{ML}) and an anterior instability margin (AIM) were used to quantify the dynamic balance of 21 unilateral transtibial amputees during overground walking. Participants trialled two prosthetic feet presenting contrasting movement/balance constraints; a Higher Activity foot similar to that of their own prosthesis, and a Lower Activity foot. Participants were assessed before (Visit 1) and after (Visit 2) a 3-week adaptation period on each foot.

With the Higher Activity component, MOS_{ML} decreased on the prosthetic side, and increased on the sound side from Visit 1 to Visit 2, eliminating a significant inter-limb difference apparent at Visit 1 (Visit 1–sound = 0.062 m, prosthetic = 0.075 m, $p = 0.018$; Visit 2–sound = 0.066 m, prosthetic = 0.074 m, $p = 0.084$). No such change was seen with the Lower Activity foot (Visit 1–sound = 0.064 m, prosthetic = 0.077 m, $p = 0.007$; Visit 2–sound = 0.063 m, prosthetic = 0.080 m, $p < 0.001$). Significant changes in AIM were observed at Visit 2 (Visit 1: −0.16 (0.08) m, Visit 2: −0.17 (0.08) m; $F = 23.396$, $p < 0.01$).

These findings suggest that changes in balance during walking can occur following the initial receipt of a device regardless of whether the component is of the same functional category as the one an individual is accustomed to using.

1. Introduction

Technological developments have led to an increase in prosthetic devices available on the market. Components that differ both subtly and markedly in structure and response, degrees of freedom (allowable movements), flexibility and mass theoretically increase the potential to successfully tailor prescription to the unique functional requirements of an individual. In practice, however, the prosthetist is faced with an overwhelming range of options to select from and limited objective measures to inform the decision [1].

While the process of building a custom prosthesis may take many appointments, the final delivery occurs in a single appointment with a prosthetist during which alignment of the device is finalized based on observational analysis of standing and walking and on verbal feedback from the patient. Assessment of the function of a prosthesis at the time of delivery is complicated by the ensuing adaptation to the device. With any functional change to a prosthesis, an individual must reorganize their movement to arrive at an optimal solution that effectively integrates the change into their walking patterns [2]. As this self-organization occurs, it should be expected that many endpoint variables will also change. At delivery, a prosthetist is required to optimize a device and its function based on limited information available prior to acclimation of the individual to the geometry, function and response of the device.

The consideration of balance during walking, or *dynamic balance*, is important in device selection for two reasons. First, the ability to maintain balance and control during walking is in itself of high importance for individuals with limb loss [3], and central to confidence, participation and autonomy in everyday life [4]. Second, a lack of balance may have implications for the quality of the movement produced during the assessment [5]. Compensations, secondary movements and the inability to exploit functional features of a prosthetic foot

* Corresponding author.
E-mail addresses: jkent@unomaha.edu (J.A. Kent), nstergiou@unomaha.edu (N. Stergiou), swurdeman@hanger.com (S.R. Wurdeman).
[1] Present address: Department of Clinical and Scientific Affairs, Hanger Clinic, 11155 S. Main St., Houston, TX 77025, USA.

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may simply be a reflection of a lack of experience exploring the new extremity, and a lack of confidence in or appreciation of its boundaries or response. It is therefore important to be mindful of changes that may be expected to take place as the individual adapts to a new component.

One might speculate that an individual would be more readily able to integrate and control components that are functionally similar to one he or she had been accustomed to using. This is of importance as, if true, a prosthetist prescribing a functionally similar component could be reasonably confident that the movement observed on delivery is a reflection of potential function.

The aim of this study was to explore the changes to dynamic balance over an adaptation period following receipt of two new devices; one of which, based on activity level, was more similar to the device the individual was accustomed to, and one markedly different, attempting to induce a greater need to adapt. It was hypothesised that individuals would readily adapt to the former, reflected in a lack of measurable change in dynamic balance over the adaptation period, and conversely, that greater changes to dynamic balance over time would be observed for the less familiar component.

2. Methods

2.1. Participants

Twenty-one unilateral transtibial amputees gave informed consent to participate in a randomized crossover trial, approved by the university institutional review board (Table 1). All were experienced prosthesis users, K3 or K4 level according to Medicare classification, and appropriately wore high activity feet with their personal prostheses.

2.2. Procedures

Participants were tested on two different foot components: one of an activity level similar to their own based on prescribing guidelines (Higher Activity foot) and one rated at a lower activity level (Lower Activity foot), with order of provision randomized. All Higher Activity feet were energy storage and return-type components. For the Lower Activity component, all participants received a Solid Ankle Cushioned Heel (SACH) foot with the exception of one participant who was provided a multi-articulated flexible keel foot which is similarly rated at a lower activity level. Participants maintained their own socket and suspension to reduce confounding elements related to fit. A certified prosthetist performed all fitting and alignment for the study.

The first trial device was fitted during Visit 1 (V1). Individuals donned a tight fitting uniform with retroreflective markers placed on the pelvis and feet. These markers were placed consistently on both sides; superficial to the anterior and posterior superior iliac spines of the pelvis, at the lateral malleoli, and the dorsum of the second metatarsal head. Markers were placed on the prosthetic foot at analogous locations to the sound limb.

Testing was performed following alignment of the new prosthetic setup, after approximately 10 min of walking. Kinematic data were collected at 60 Hz using a 12-camera motion capture system (Eagle, Motion Analysis Systems, Santa Rosa, CA) during 10 traverses of the laboratory at a self-selected walking speed.

Table 1
Demographics for 21 participants with unilateral, transtibial amputation. Mean (SD) values reported.

Age (yrs)	Time Since Amputation (yrs)	Height (m)	Mass (kg)	Etiology
53.4 (11.9)	7.7 (6.1)	1.78 (0.08)	100.6 (18.8)	Trauma (n = 13) Vascular (n = 5) Tumor (n = 1) Other (n = 2)

Participants wore the foot component for three weeks. After this period, following a repeat assessment (Visit 2: V2), the foot was swapped for the other (Higher/Lower Activity) trial component and the prosthesis setup was re-aligned for the new component. The three week process was then repeated.

Data were tracked in Cortex (Motion Analysis Systems, Santa Rosa, CA) and 10–15 strides from each participant were extracted for analysis. All computations were performed in Visual 3D (C-motion, Germantown, MD).

2.3. Margin of stability

Margin of stability is a measure of dynamic balance based on the relative motion of the center of mass of the body with respect to the base of support provided by the feet [6]. Whereas in quiet standing it is accepted that balance can be maintained as long as the vertical projection of the center of mass lies within an individual's base of support, this criterion is insufficient during dynamic activities such as walking. In order to account for this increased challenge to balance, the extrapolated center of mass (XcoM) [6] incorporates an inertial term based on inverted pendulum dynamics:

$$XcoM = x + v\sqrt{l/g}$$

where x and v *are* the instantaneous position and velocity of the center of mass in the horizontal plane respectively, l is leg length and g is acceleration due to gravity. The margin of stability is traditionally defined as the minimum distance between the XcoM and the boundary of the base of support during movement [6].

Margin of stability was calculated in the medial-lateral (MOS_{ML}) direction as defined by Hof et al. [6,7]. In the anterior direction, using a similar approach we calculated the distance between the XcoM and the anterior margin of the base of support of the stance limb at the point of initial contact of the contra-lateral swing limb. In other words, we measured the extent to which the XcoM travelled beyond the base of support of the stance foot before contralateral limb foot contact. Despite the consistency in calculation we refer to this as the Anterior Instability Margin (AIM) due to differences conceptually between its definition of this and the traditional MOS. MOS_{ML} and AIM were both defined such that a positive value implied that the XcoM remained within the boundary of the base of support, i.e. a positive margin of stability implies stability, and a negative value occurs with the movement of the XcoM outside the base of support during a step. A greater magnitude of AIM in the negative direction indicates that the XcoM moves further outside the base of support before the contralateral swing limb is placed.

Lateral base of support was defined by the vertical projection of the lateral ankle markers onto the horizontal [8]. The anterior border was defined using the toe marker. Both markers were selected based on the reliability of their placement across sessions and foot components. The center of mass of the body was approximated by a point half way between midpoints of the markers placed on the anterior superior iliac spines and posterior superior iliac spines. Medial-lateral marker profiles were detrended prior to the calculation of the XcoM and MOS_{ML} to remove the effect of slight deviations in walking direction during the traverse of the laboratory [9]. Prior to calculation of the variable, the coordinates of the ankle markers and center of pressure at each time point were corrected by subtracting from them the vector defined by the center of mass medial-lateral position between the start and end of the trial. Foot contact events were detected via a kinematic algorithm [10] and manually verified. The average speed for each traverse was additionally measured via the displacement of the sacral marker.