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Static versus dynamic prosthetic weight bearing in elderly trans-tibial amputees

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Abstract

The purpose of this study was to compare prosthetic weight-bearing tolerance in the standing position to the dynamic vertical ground reaction forces (VGRF) experienced during walking in elderly dysvascular trans-tibial amputees. Ten unilateral trans-tibial amputees attending an amputee clinic (mean age =67±6.5 years) were selected as subjects. Selection criteria were the level of amputation, age, medical fitness to participate and informed consent. Each participant completed five trials of standing (static) weight bearing measurement followed by 10 walking (dynamic) trials on a 10m level walkway, five trials for each limb. Static weight bearing (SWB) was measured using standard bathroom scales. Dynamic weight bearing (DWB) was measured during gait using a Kistler multichannel force platform. T-tests for dependent means indicated that the forces borne in prosthetic single limb stance (mean=0.97±0.03 times body weight (BW)) were significantly lower than the forces borne by the prosthetic limb during the first peak (weight acceptance) VGRF (mean = 1.08±0.08 BW; t = -4.999; p = 0.001) and significantly higher than the midstance VGRF (mean = 0.82±0.07 BW; t = 5.401; p<0.001). However, there was no significant difference between SWB and the second peak (push-off) VGRF generated by the prosthetic limb during walking (mean = 0.96±0.03 BW). It was concluded that clinical gait training may utilise SWB as a guide to an amputees' prosthetic weight bearing tolerance and requirements during walking.

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Introduction

Weight bearing, the act of supporting body weight during standing (static) and walking (dynamic) conditions, has been investigated in amputees using a variety of quantitative methods. Instrumentation has included strain gauges (Tibarewala and Ganguli, 1982), shoe-borne load cells (Lord and Smith, 1984; Ranu and Eng, 1987), force platforms (Hurley *et al,* 1990; Hermodsson *et al.,* 1994; Thorburn *et al,* 1990), and bathroom scales (Stolov *et al.,* 1971). Quantifying weight bearing provides an objective measure and numeric feedback of an amputees' prosthetic weight bearing tolerance.

That portion of weight not borne through the prosthesis during the prosthetic stance phase must be taken through the amputee's upper limbs and a walking aid. Assistive devices such as a cane (Winter *et al.* 1993) or a frame (Crosbie, 1992; Pardo *et al.,* 1993) have been used for many reasons: to reduce pressure on sensitive structures, improve balance or assist proprioception (Lord and Clark, 1991). Pardo *et al.* (1993) concluded that walking frames, crutches and canes were used to assist in supporting body mass vertically. While the cane assisted propulsion via vertical force transmitted to the upper limb, the walking frame helped to restrain ambulatory progression with no significant role in propulsion.

Clinically, it is possible to measure weight bearing using bathroom scales. Stolov *et al.* (1971) demonstrated clinical and research utilisation of static standing weight bearing measurements using bathroom scales. The authors reported the importance of percentage body weight over absolute weight as a criterion when reporting progression of weight bearing after immediate prosthesis fitting following trans-tibial amputation. Summers *et al.* (1978; 1988) used a Double Video Forceplate (DVF) to measure biomechanical parameters of stance and balance. They concluded that simple measurements of weight distribution between the feet and maximal weight-bearing while leaning on the prosthesis could provide objective confirmation of clinical improvement and assist in retraining amputees to stand, balance, and walk.

Assessing weight bearing during standing or balancing tasks, however, is limited to relatively quiet standing situations. Measurements in stance do not assess foot placement, fluctuations in ground reaction forces, nor progression of the centre of pressure experienced during dynamic gait (Winter, 1991; Wintra and Sienko, 1988; Himann and Cunningham, 1988; Engstrom and Van de Ven, 1985). Thorburn *et al.* (1990) assessed ground reaction forces, using a Kistler force platform, generated by five trans-tibial amputees while wearing the SACH, Flex-foot, STEN, SEATTLE, and CCII prosthetic feet. The authors reported that the various prosthetic feet resulted in similar force patterns mediolateral shear force, anteroposterior shear force, and vertical ground reaction force (VGRF) during the stance phase of gait.. The second peak of the VGRF in terminal stance ranged from 97.6% body weight (BW) with the Flex-foot to 99.5% BWwith the SACH foot. Andriacchi *et al.* (1977) reported the minimum amplitude of the VGRF (at midstance) to have the largest rate of change with walking speed. Peak VGRF at heel strike and propulsion were considered to exhibit a meaningful velocity dependence. Force amplitudes were found to vary linearly with velocity and were not as sensitive an indicator of gait abnormalities as temporal measurements.

Although assessment of dynamic weight bearing tolerance during gait is preferable to static assessment of weight bearing tolerance, it requires expensive equipment, specialist technicians, and expert interpretation. In contrast, tools to assess static weight bearing tolerance, such as bathroom scales, are inexpensive, readily available to the clinician and provide immediate quantitative data. However, no study was identified in the literature which related single limb static weight bearing tolerance, assessed using bathroom scales, to the forces generated by amputees during gait. Therefore, the purpose of the present study was to compare standing prosthetic weight bearing tolerance to the forces experienced during walking in elderly dysvascular trans-tibial amputees.

Methods

Subjects

Subjects included 10 elderly (mean age = 67±6.5 years) male unilateral trans-tibial amputees, secondary to vascular etiology, recruited from the Illawarra Regional Hospital Amputee Clinic. The subjects reported a mean of 2.8±2.2 years since amputation and had worn their existing prosthesis for an average of 11.8±4.4 months, approximately 12.8±2.4 hours per day. Prosthetic suspension in nine subjects was supracondylar after the fashion of Kegel (1986) and one was by a thigh lacer. Prosthetic foot componentry included seven SACH, two Seattle and one Greissinger. Concurrent disease processes reported by the subjects included six with diabetes mellitus, four had suffered a CVA, one had hypertension, two experienced myocardial infarction, two had osteoarthritis, one gout and one emphysema.

Written informed consent was completed by each subject before testing. All testing was conducted with the knowledge and consent of medical personnel familiar with the subjects' medical history and according to the University of Wollongong Human Research Ethics Committee requirements.

Static weight bearing assessment

Each subject's total body mass was measured (kg) while the subject stood motionless in the anatomical position for 2 seconds on a set of calibrated bathroom scales (130 kg capacity). Prosthetic weight bearing (kg) was then measured while the subjects stood on the scales in single limb stance on their prosthetic limb for 2 seconds. During total body weight and prosthetic weight bearing assessment the subjects were instructed to focus on an eye-level focal point to help them maintain balance. Five trials for each assessment were completed. Due to the need to bear weight through the prosthetic limb, most subjects required assistance to stand in single limb stance. Assistance was provided by the subjects leaning on a stable fixture placed adjacent to the scale. That upper limb force placed on the support was attributed to weight-bearing intolerance or to diminish balance (Isakov *et al.,* 1992). Differentiating the contribution of weight-bearing intolerance and balance was beyond the scope of this study. The numeric value in kilograms recorded from the scale during unilateral prosthetic stance was therefore defined as the prosthetic weight bearing tolerance. The mean scores for total body weight and prosthetic weight bearing were then substituted into equation (1) to calculate static weight bearing (SWB) tolerance:

$$SWB = \frac{\text{prosthetic weight bearing}}{\text{total body weight}} \quad (1)$$

SWB was expressed in body weight units (BW) for ease of comparison with dynamic weight bearing values.

Dynamic weight bearing assessment

Dynamic weight bearing (DWB) was assessed by quantifying the VGRF generated as the subjects walked at a self-selected velocity over a calibrated Kistler multichannel force platform (type 928IB, 600 mm x 400 mm) embedded midway along a 10 m wooden walkway. The force platform was mounted on a concrete pedestal and covered with a sample of granulated rubber sports surface to be level with the surrounding walkway. Forces from the eight output channels of the force platform were passed through a Kistler multichannel Charge Amplifier (type 9865AO) and recorded using a data acquisition board and a personal computer. The VGRF generated by the subjects were sampled (1000 Hz) over the time required for each subject to complete the stance phase on the test limb for five trials per limb. Adequate rest was provided between each trial to minimise fatigue. Before DWB assessment, the weight of each subject was recorded while they stood motionless on the force platform to enable later normalisation of force data relative to each subject's body weight.

Throughout both SWB and DWB assessment, subjects wore their usual walking shoes (eight wore athletic flexible shoes with a deep tread; two wore thin, rigid soled leather shoes.

From each VGRF force-time curve, three values were recorded as representative of DWB (see Figure 1):

(i) the first passive impact peak (1st Peak);

(ii) the dip at midstance (Midstance); and

(iii) the active propulsive peak of push-off (2nd Peak).

These values were expressed in units of body weight. Stance duration was then derived (ms) from initiation of weight acceptance to the completion of push-off.

Kinematics of the gait cycles

The gait of each subject was filmed from anterior and lateral views (25 Hz) using two Panasonic M7 VHS video cameras. The cameras were levelled on tripods 4.8 m (anterior view) and 2.3 m (lateral view) from the subject. The focal axis of each lens was perpendicular to the relevant plane of the force platform to minimise perspective errors (Miller and Nelson, 1973). A known scale was filmed during each trial to enable later conversion of photographic images to actual distance in metres.

Each gait trial was replayed using a Panasonic VHS video cassette recorder and monitor and visually inspected to select one representative prosthetic and normal trial for each subject based on even cadence and lack of

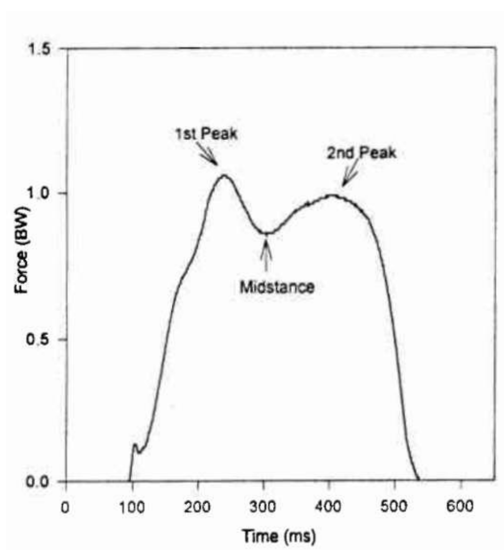

Fig. 1. Typical VGRF force-time curve generated during DWB on the prosthetic limb.

targeting. The video data were transferred from the video recorder to a 496DX-66 personal computer using a video grabber board (Creative Labs VideoSpigot) and Video/Cap software (Video for Windows 1.1). Data were captured (25 Hz) at least 10 frames before heel strike until 10 frames after toe-off to the test limb. The x and y coordinates for 15 anterior markers (vertex, clavicular acromion, humeral lateral epicondyle, radial styloid process, mid-inguinal crease, patellar superior pole, midway between lateral and medial malleoli, anterior tip of shoe; left and right sides) and 9 lateral markers (greater tronchanter, fibular head, lateral malleolus, fifth metatarsal head and tip of shoe on facing leg, femoral adductor tubercle, medial malleolus, first metatarsal head and tip of shoe on contralateral leg) representing the anatomical link system were digitised throughout the stance phase of the test limb. The digitised data were then smoothed using a fourth order low pass Butterworth digital filter (6 Hz cut-off frequency) (Winter, 1990).

The following variables were then calculated during the stance phase of the test limb from the smoothed digitised data using Digital Signal Processing (DSP) software (Andrews, 1994):

(i) velocity of forward progression, represented as the average linear velocity of the greater tronchanter marker in the sagittal plane from weight acceptance to push-off (stance phase);

(ii) lateral (weight) transference of the total body centre of gravity (COG). The total body COG was calculated utilising custom software based on Dempster's linked segment model (Winter, 1990). Lateral weight transference was defined as the horizontal displacement (cm) in the frontal plane of the COG relative to the base of support at midstance. A smaller displacement indicated less lateral weight transference of the subject's COG toward the base of support.

Statistics

Means, standard deviations and ranges were calculated for each kinetic and kinematic variable for the gait trials performed on each subject's sound and prosthetic limb and for SWB. T-tests for dependent means were then conducted (SYSTAT 5.01 for Windows) to determine any significant differences ($p<0.05$) between the forces generated during SWB and DWB. Pearson product moment correlations were also calculated to identify any significant relationships among the forces generated during SWB and gait and the gait descriptions.

Results

SWB versus DWB

The mean, standard deviation and range of values recorded during SWB and DWB are presented in Table 1. Forces borne through the prosthesis during SWB[1] (mean = 0.97±0.03 BW) were significantly lower than the forces borne by the prosthetic limb during the first passive peak VGRF (mean = 1.08±0.08; *df* = 9; p = 0.001) and were significantly higher than the forces borne during midstance (mean = 0.82±0.07; *df* = 9; t = 4.999; p<0.0001). However, there was no significant difference between SWB forces and the second propulsive peak VGRF (mean = 0.96±0.03 BW; *df*= 9; t =

[1]SWB forces of less than **1** indicated that the subjects required external support while standing on their prosthetic limb.

Table 1 Forces (BW) recorded during SWB and DWB (n=10).

	SWB*	Prosthetic limb DWB			Sound limb DWB		
		1st Peak	Midstance	2nd Peak	1st Peak	Midstance	2nd Peak
Mean	0.97	1.08'	0.82'	0.96	1.19	0 81	1.03
SD	0.03	0.08	0.07	0.03	0.17	0.09	0.06
Min	0.92	0.99	0.68	0.93	0,99	0.66	0.96
Max	0.99	1.25	0.90	1 01	1,48	0.93	1.15

*SWB was calculated for the prosthetic limb only. Therefore, comparisons of SWB and sound limb DWB were not conducted

indicates a significant difference (p<0.001) between SWB and DWB values on the prosthetic limb.

Table 2, Horizontal velocity, stance duration and lateral weight transference during the stance phase on the prosthetic and sound limbs (n=10).

	Velocity (m.s.')		Stance duration (ms)		Lateral weight transfer (cm)	
	Prosthetic	Sound	Prosthetic	Sound	Prosthetic	Sound
Mean	0.52	0.51	483	499	3.82	4.49
SD	0.18	0.15	214	224	0 96	1.33
Min	0.19	0.20	353	366	2.83	2.03
Max	0.74	0 67	1064	1113	5.60	6 06

0.629; p = 0.545) generated by the prosthetic limb during walking.

Strong significant negative correlations were found between the first passive peak VGRF and midstance VGRF for both the sound limb (r = -0.950) and the prosthetic limb (r = 0.860). This result indicated that a high first VGRF peak was associated with a low midstance VGRF.

Gait Descriptors

The mean, standard deviation and range of values calculated for the three kinematic gait descriptor are presented in Table 2. There was no significant difference between the prosthetic and sound limbs for the velocity of forward progression during the stance phase, stance duration or lateral weight transference onto the stance limb.

For SWB to be used as a predictor of walking ability, the relationship between SWB and dynamic gait descriptors and must be established. Correlation coefficients calculated between the gait descriptors and the forces recorded during SWB and DWB are presented in Table 3. Velocity of forward progression during the stance phase of the sound limb was significantly correlated to SWB (r = 0.680), indicating that the higher forces borne through the prosthesis during SWB were associated with greater velocity in the sound limb during gait. This relationship was not significant during the stance phase on the prosthetic limb (see Table 3). Velocity was also found to be significantly correlated to DWB at midstance for the prosthetic limb (r = -0.699) and the sound limb (r = -0.781). Therefore, greater velocities were associated with decreased VGRF at midstance. Although velocity was significantly correlated to DWB at the first peak VGRF on the sound limb (r = 0.735), it exhibited only a low positive relationship with DWB on the prosthetic limb at the first peak VGRF (r = 0.450). Velocity exhibited only a low negative relationship with the DWB at the second peak on the prosthetic and the sound limb (see Table 3). Stance

Table 3. Correlation coefficients between the three gait descriptors and SWB and DWB.

	Velocity (m.s.)		Stance duration (ms)		Lateral weight transfer	
	Prosthetic (n=10)	Sound (n=IO)	Prosthetic (n=10)	Sound (n=10)	Prosthetic (n=9)	Sound (n=10)
SWB'	0.472	0 680*	-0.649*	-0,624	0.241	0.237
DWB						
1st Peak	0 450	0.735*	-0.513	-0.565	0.132	-0.004
Midstance	-0 699*	-0.781	0.509	0.585	-0.193	-0.091
2nd Peak	-0.220	0 353	-0,233	-0.182	0 060	0.455

SWB was correlated to the three gait descriptors to establish the relationship between the static weight bearing ability and dynamic gait.
"indicates a significant correlation at p<0 05 (r>0.632; *df*= 8)

duration was found to be significantly correlated (r = -0.649) with SWB on the prosthetic limb, indicating longer stance duration on the prosthetic limb was associated with lower forces borne through the prosthesis during SWB. No other significant correlations were found between SWB or DWB and either stance duration or lateral weight transference during stance for the prosthetic or sound limb.

Discussion

The VGRF recorded during free walking in the present study were consistent with those reported previously for trans-tibial amputee patients (Thorburn *et al,* 1990). The bi-peaked vertical force-time curve characteristic during gait indicated a weight acceptance force greater than 1 BW, a VGRF less than 1 BW at midstance, followed by a second peak VGRF approximately 1 BW at push-off of the stance phase.

The mean velocity of forward progression recorded in the present study was less than mean velocities reported in previous studies of amputee gait (Hurley *et al,* 1990; Thorburn *et al,* 1990; Winter, 1991). However, there was a large range of velocities demonstrated in the present study (Table 2). Those subjects displaying lower velocities of forward progression during stance also exhibited less defined peaks and troughs in their VGRF force-time curves. The weight bearing tolerance of the high peaks were held for short durations whereas the lesser forces were tolerated longer.

The mean values of velocity, stance duration and lateral weight transference were not significantly different when comparing the prosthetic limb to the sound limb in the small sample of subjects in the present study. In contrast, Thorburn *et al.* (1990) found stance duration to be significantly longer for the sound limb than the prosthetic limb.

The SWB test provided a practical quantitative assessment of the weight bearing tolerance of the amputee subjects on their prosthetic limb, a test suitable for use in clinical settings. Static weight bearing in single limb stance can be measured quickly and easily using bathroom scales and can be expressed as a percentage of total body weight. The functional task of shifting the amputee's weight onto the prosthesis to step onto the scale demonstrated an objective task reflecting patient confidence and comfort. Walking aid selection and gait training goal setting may be guided by the percentage of prosthetic weight bearing as a predictor for further training.

Progression of weight bearing from dual limb to single limb stance can prepare an amputee for the VGRF experienced during midstance and push-off phases of gait. In the present study, SWB was shown to be less than the VGRF generated dynamically at weight acceptance by the prosthetic limb. It would therefore be insufficient to train tolerances equivalent to SWB when attempting to prepare patients to meet demands anticipated for weight acceptance at self-selected walking velocity. The SWB test provided more prosthetic weight bearing than that exhibited dynamically during midstance, particularly in the faster walkers. Therefore, training weight bearing tolerance in unilateral stance to full weight bearing would be adequate to enable patients to cope with the VGRF generated during distance in self-selected walking velocity. Furthermore, the SWB test demonstrated equivalent VGRF to that measured at push-off. Training amputees to tolerate full weight bearing in unilateral stance on the prosthesis would therefore prepare them for the push-off phase of gait.

The DWB first peak demonstrated a positive relationship with velocity. The greater the velocity, the greater was the VGRF at weight acceptance. This first peak was also greater than the subjects' SWB measurement. In the elderly population, however, velocity of gait diminishes with increasing age. Therefore, the need to train prosthetic weight bearing to cope with high first peaks will lessen with age. Others, whose concurrent medical problems slowed them further (Himann and Cunningham, 1988; Mueller *et al,* 1994), displayed a first peak equivalent to their body weight. In these slower individuals, the SWB value was equivalent to both the first and the second peaks of the force-time curve of the VGRF recorded during DWB.

Hermodsson *et al.* (1994) reported that the gait of vascular amputees differed from that of traumatic amputees, a difference that was not caused by reduced walking speed. The active forces during push-off on both the sound and the prosthetic leg in the trauma group were not found in the vascular group. This disparity could be an effect of the systemic disease. No distinction was made in the present study for the underlying pathology of weight bearing intolerance, such as stump sensitivity, ill fitting socket, habit; nor concurrent medical diagnoses, such as CVA, hip replacement, and myocardial infarction. All amputees in this study had vascular etiology.

Clinical application of the results of the present study lies in the hierarchy of the weight bearing demands that were identified. Midstance requires less than body weight tolerance, push-off requires full body weight tolerance though the prosthesis and weight acceptance requires tolerance of forces in excess of full body weight. Rehabilitation may proceed to increase velocity when weight acceptance forces have surpassed 100% body weight. It is therefore recommended that rehabilitation goals for new amputees should include techniques to train prosthetic weight bearing first in bilateral stance, progress to push-off from the prosthesis and finally to weight acceptance onto the prosthesis.

In conclusion, the SWB test provides a quantitative test of weight bearing tolerance of the trans-tibial amputee on the prosthetic limb. Weight acceptance during the stance phase of DWB required a greater force tolerance than SWB. The midstance phase, however, required less force tolerance than SWB and the propulsive phase required equivalent forces to those measured in SWB. Clinical gait training may therefore utilise SWB as a guide to an amputee's prosthetic weight bearing tolerance and requirements during walking.

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