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SHOCK ABSORPTION DURING TRANSTIBIAL AMPUTEE GAIT: DOES LONGITUDINAL PROSTHETIC STIFFNESS PLAY A ROLE?

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Abstract

Background—Reduced-stiffness components are often prescribed in lower-limb prostheses, but their efficacy in augmenting shock absorption has been inconclusive.

Objectives—To perform a systematic variation of longitudinal prosthetic stiffness over a wide range of values and evaluate its effect on shock absorption during gait.

Study Design—Repeated-measures crossover experiment.

Methods—12 subjects with a unilateral transtibial amputation walked at normal and fast self-selected speeds. Longitudinal prosthetic stiffness was modified by springs within a shock absorbing pylon: NORMAL (manufacturer-recommended), 75% of normal (MEDIUM), 50% of normal (SOFT), and RIGID (displacement blocked). The variables of interest were kinematic (stance-phase knee flexion and pelvic obliquity) and kinetic (prosthetic-side ground reaction force (GRF) loading peak magnitude and timing).

Results—No changes were observed in kinematic measures during gait. A significant difference in peak GRF magnitudes between MEDIUM and NORMAL ($p = 0.001$) during freely selected walking was attributed to modified walking speed ($p = 0.008$). GRF peaks were found to be statistically different during fast walking, but only between isolated stiffness conditions. Thus, altering longitudinal prosthesis stiffness produced no appreciable change in gait biomechanics.

Conclusions—Prosthesis stiffness does not appear to substantially influence shock absorption in transtibial prosthesis users.

1. Background

Lower-limb shock-absorbing prosthetic components, including feet, pylons, and gel liners, often serve to reduce prosthesis stiffness. The effects of this reduced stiffness are not well

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Declaration of Conflicting Interests

The Authors declare that there is no conflict of interest.

understood, particularly during walking, but are thought to relate to shock absorption. Shock absorption is a particular concern in persons with lower-limb amputation because many anatomical mechanisms that provide shock absorption are no longer present. However, prosthetic components have the potential to compensate for this loss and protect the residual limb¹. Reduced-stiffness prosthetic components are typically composed of elastic elements that are designed to compress during weight bearing, storing impact energy within the deformed material rather than transferring it to the anatomical structures of the residual limb. Thus, the total energy within an ideal spring system remains the same, but the peak force is reduced by distributing the force event over a longer period of time. Decreasing the stiffness of a prosthesis should theoretically reduce the magnitude of the ground reaction force (GRF) loading peak and increase the time to reach this peak from force onset, commonly used metrics for characterizing shock absorption.

Previous studies have investigated the effects of reduced-stiffness components in comparison with more traditional components (e.g., a shock-absorbing pylon vs. a rigid pylon, or an energy-storing foot vs. a conventional SACH foot). However, these studies have reported few differences in GRF loading peaks during level walking at a self-selected speed^{2–10}. Further, very few studies have accompanied *in vivo* testing of these components with mechanical testing to establish the mechanical differences between different prosthetic stiffness conditions^{2, 3, 7, 11}. An amputee independent investigation of mechanical properties is essential to quantify experimental conditions, as differently constructed prosthetic components might be described by similar net stiffness values^{7, 12}. Thus, whether the previously tested components were actually characterized by distinct stiffness values that were substantially different between conditions is unknown. If not, this may explain the lack of significantly altered GRFs reported in previous gait analyses. Further investigation is needed to determine the appropriate stiffness properties required to provide adequate shock absorption to prosthesis users during load bearing.

The purpose of this study was to quantify the effect of changes in prosthetic stiffness on force generation during level walking among transtibial prosthesis users. The longitudinal prosthetic stiffness was varied systematically and independently from any other prosthesis characteristic over an independently verified range of values considered to be clinically relevant. The effect of longitudinal stiffness was investigated for several gait parameters related to shock absorption. It was hypothesized that changes in longitudinal prosthetic stiffness would result in kinetic changes demonstrated by altered GRF peaks and timing, although it was recognized that significant changes would most likely occur only at the lowest level of prosthesis stiffness.

2. Methods

2.1 Data Collection

2.1.1 Subject Recruitment—Subjects were recruited from the Northwestern University Prosthetics-Orthotics Center (NUPOC) and the Jesse Brown Department of Veterans Affairs Medical Center (JBVAMC). Inclusion criteria were:

- Age of 18 – 80 years

- Unilateral transtibial amputation
- Ability to walk at normal and fast speeds without undue fatigue
- Adequate space between socket and ground to fit the experimental components (~25 cm)
- At least six months of experience with a definitive prosthesis

Exclusion criteria were:

- Poor socket fit that reduced control or limited activity
- Use of an assistive device to walk distances < 10 m
- Body mass exceeding specified limit of prosthetic components (> 125 kg)

All subjects provided written informed consent prior to participation in this study, and experimental procedures were approved by the Northwestern University Institutional Review Board.

2.1.2 Experimental Prosthesis—The experimental prosthesis consisted of the subjects' own socket and suspension system, an Endolite TT (Telescopic-Torsion) Pro shock-absorbing pylon (SAP) (Endolite, Miamisburg, OH), a Seattle LightFoot (Trulife, Dublin, Ireland) of moderate keel stiffness, and a standardized shoe (Target Corporation, Minneapolis, MN). The TT Pro was selected because it allows easy modification of the longitudinal stiffness of the prosthesis through the substitution of different springs (Figure 1). The transverse plane rotation feature of the TT Pro pylon was disabled for the purpose of this study. The Seattle LightFoot was chosen because of its low profile, comparatively high longitudinal stiffness, and popularity as a commercial prosthetic foot component. A certified prosthetist assembled the experimental prosthesis and performed a dynamic alignment such that both the prosthetist and subject were satisfied with the comfort and function of the prosthesis during walking. Alignments were performed with the manufacturer-recommended spring for each subject prior to data collection. Subjects were blinded to the stiffness condition and given a brief accommodation period of approximately 5 minutes, or until the subject indicated that they felt comfortable walking with the device.

2.1.3 Procedure—Data collection was performed in the JBVAMC Motion Analysis Research Laboratory (MARL), equipped with a digital real-time camera system (Motion Analysis Corporation (MAC), Santa Rosa, CA) and six force platforms (AMTI, Watertown, MA) embedded within a 10m walkway. Four prosthetic stiffness conditions were evaluated: the manufacturer-recommended stiffness (NORMAL), 50% of NORMAL stiffness (SOFT), 75% of NORMAL stiffness (MEDIUM), and a rigid condition (RIGID). The NORMAL stiffness served as the baseline condition, and was selected from one of three springs available with the TT Pro based on body weight ("impact level" was set as moderate for all subjects). Further, the SOFT and RIGID conditions represented the 'extremes' of prosthetic stiffness. A steel cylinder insert for the RIGID condition is a good approximation of a rigid pylon; the SOFT stiffness was considered the least stiff condition that would not result in a functional limb length discrepancy during prosthetic stance phase. Two spring sets were used for the experiments, and their stiffnesses were evaluated using a mechanical testing

machine. Stiffness values for Spring Set #1 were 68.2 kN/m (SOFT), 89.3 kN/m (MEDIUM), and 111.8 kN/m (NORMAL). Stiffness values for Spring Set #2 were 85.6 kN/m (SOFT), 111.8 kN/m (MEDIUM), and 153.8 kN/m (NORMAL). The RIGID condition was characterized by a stiffness of 3556.9 kN/m. Stiffness modifications were achieved by the removal of the spring within the housing of the TT Pro pylon and replacement with the appropriate stiffness spring or RIGID steel cylinder. A position sensor was attached to the pylon to ensure that the softest springs were not exceeding the allowable travel within the pylon. The NORMAL condition was tested first to establish baseline self-selected walking speeds. Randomization of the remaining three stiffness conditions was performed by means of a random integer generator¹³.

Reflective markers were placed on the subjects in a modified Helen Hayes arrangement¹⁴. Subjects were instructed to walk with the NORMAL spring condition at both a 'normal' and a 'fast' self-selected walking speed. The normal (i.e., freely selected) speed was evaluated as a clinically significant walking task, while fast walking was evaluated because previous studies report changes in force transmission during higher-impact activities such as fast walking^{7,9}.

Subjects walked until five clean force plate strikes were acquired for the prosthetic-side foot. Marker data were sampled at 240 Hz, and force data were sampled at 1920 Hz. Data from three NORMAL trials were processed and averaged to determine a baseline for each speed condition. Walking trials for each of the remaining three stiffness conditions were retained only if the subjects were within 10% of their baseline speed⁸.

2.2 Data Processing

Marker position data were post-processed in Cortex (MAC, v 4.0.0.1387) and were filtered using a 4th order, low-pass Butterworth filter with a cutoff frequency of 6.0 Hz. GRF data were not filtered, but were decimated by a factor of eight to match the kinematic sampling frequency. OrthoTrak (MAC, v 6.6.4) was used to calculate kinematic and kinetic variables. The data were analyzed using custom programs written in MATLAB (MathWorks, Natick, MA). Peak vertical force during loading response (i.e., initial contact through contralateral toe-off) and the timing of this vertical peak force event were measured. Stance-phase knee flexion and pelvic obliquity, mechanisms that provide shock absorption during the loading response phase in able-bodied individuals¹⁵, were also analyzed. Finally, walking speed was calculated for each trial, despite the $\pm 10\%$ speed control, to identify any systematic speed effects.

2.3 Statistical Analysis

Statistical analysis was performed using IBM SPSS Statistics 22 (IBM Corp., Armonk, NY). Normality of the data distribution was tested using the Shapiro-Wilk test. Friedman's test, a nonparametric repeated-measures two-way analysis of variance (ANOVA) for ranks was selected to identify statistically significant differences when evidence suggested that some data were not from a normally distributed population. Post hoc pairwise comparisons with a Bonferroni correction were performed to determine differences between prosthetic stiffness conditions at a significance level of $\alpha = 0.05$.

3 Results

Twelve subjects (49 ± 18 years, 84.8 ± 21 kg) with a unilateral transtibial amputation participated in this study (Table 1). The average self-selected walking speed was 1.22 ± 0.18 m/s for the normal condition and 1.57 ± 0.28 m/s for the fast condition across all stiffness conditions (Tables 2 and 3). A significant change in freely selected walking speed was observed between the NORMAL and MEDIUM conditions. A summary of the kinematic data is presented in Table 2, along with the corresponding p-values. No statistically significant differences were found between prosthetic stiffness conditions for either knee flexion or pelvic obliquity (Figure 2).

Table 3 presents the kinetic results by stiffness condition and walking speed. These results include the GRF loading peak normalized to body weight as well as the timing of this peak. A statistically significant difference in the peak GRF magnitudes between the NORMAL and MEDIUM stiffness conditions was identified during freely selected walking. No statistically significant changes in time to peak loading force were observed across conditions during freely selected walking. For fast walking, the NORMAL stiffness condition demonstrated a decreased peak force magnitude compared to the SOFT condition. In addition, a statistically significant decrease in time to peak was found with the RIGID condition compared to the SOFT and MEDIUM conditions. Figure 3 illustrates the average force profile for a single subject across all five trials for all four stiffness conditions during the normal walking speed condition.

4 Discussion

This study aimed to determine the influence of prosthesis stiffness on kinetic and kinematic variables related to shock absorption during the gait of transtibial prosthesis users. Kinematic shock-absorbing mechanisms—stance-phase knee flexion and pelvic obliquity—were investigated during loading response, but no statistically significant differences were found. Thus, it seems that subjects maintained consistent kinematic patterns despite substantial changes in prosthesis stiffness.

The mean GRF loading response peaks ranged from 98.2 – 104.5% BW at normal walking speeds, which are comparable to prosthetic-side peak forces reported in the literature¹⁶. The NORMAL and MEDIUM stiffness conditions were found to have different GRF loading peak magnitudes, but this difference is likely attributable to the significant change in walking speed between the NORMAL and MEDIUM conditions (1.18 vs 1.25 m/s). Perhaps more tellingly, the extreme stiffness conditions (SOFT vs. RIGID) were characterized by almost identical GRF magnitudes and walking speeds. No statistically significant changes were found in time to peak loading force across conditions during freely selected walking. These results corroborate the unchanged impact forces reported in the studies of running shoe insole stiffness¹⁷. Thus, it appears that the levels of longitudinal prosthesis stiffness that were tested are unable to meaningfully influence shock absorption during loading response at normal self-selected walking speeds.

In contrast, the fast walking speed condition demonstrated statistically significant changes in GRF variables that cannot be attributed to differences in speed. Changes in GRFs have been previously reported during fast walking and stepping down^{7, 9, 18}. It has been speculated that the lower stiffness components may provide more substantial benefits during higher impact activities (such as fast walking, landing, and running) than during normal walking. However, in the current study, slightly *higher* forces were observed for lower stiffness conditions (SOFT and MEDIUM) compared with the NORMAL stiffness condition, contrary to the hypothesized effect. This finding is inconsistent with the behavior of a purely mechanical system, and may reflect a deliberate “stepping hard” behavior by the research subjects to facilitate somatosensory feedback to their residual limbs¹⁹.

Though the decreased time to peak found with the RIGID condition compared to the SOFT and MEDIUM conditions was relatively small during fast walking (2.4% of stance phase), it indicates a faster loading rate in the RIGID condition and supports the study hypothesis. On the other hand, no differences were found between other stiffness conditions in time to loading peak, indicating that the amount by which stiffness is reduced appears to be unimportant at fast self-selected walking speeds. Therefore, evidence that longitudinal prosthetic stiffness begins to affect GRF loading characteristics as walking speed increases is limited, a result supported in the literature^{7, 9}. Nonetheless, the majority of individuals with transtibial amputations are likely to ambulate primarily at speeds less than 1.57 m/s (the average fast walking speed). Therefore, the data suggest that the levels of prosthetic stiffness evaluated in this study do not substantially affect the GRF loading peaks.

The observed kinetic invariance may demonstrate a flawed conceptual model of the prosthetic-side limb system. While the mechanical characteristics of the prosthetic components were well-controlled, the stiffness characteristics of the residual limb/socket interfaces were unknown. A previous study of the leg stiffness of able-bodied runners reported leg stiffness values ranging from approximately 8–16 kN/m²⁰. If the soft tissue and residual anatomical structures of the amputated limb have a comparable stiffness to the able-bodied limb—an order of magnitude lower than the tested prosthesis stiffness values—then these elements could dominate the behavior of this system. In this case, *all* of the tested prosthetic stiffness values could be too high to influence the total limb stiffness.

A limiting factor in this experiment is the control of walking speed. Subjects were required to maintain a speed within 10% (above or below) of their baseline, which allowed for some variability in walking speed. Interestingly, subjects walked slowest in the NORMAL condition, which may be attributable to their lack of familiarity with the experimental prosthesis. The walking speeds for the other three conditions were approximately 4–6% faster on average than for the NORMAL condition. Even though these speeds were within the 10% threshold, the consistently increased speed compared to the initial testing condition (NORMAL) indicates the presence of an order effect. However, the consistent walking speed between the remaining conditions indicates that the order effect was unlikely to have substantially affected the results.

It is also important to consider whether the benefits of reduced stiffness components outweigh their potential consequences, especially since little conclusive evidence of such

biomechanical advantages exists. Previous work in which the stiffness of a prototypical foot was systematically varied concluded that the intermediate stiffness conditions might be preferable with respect to mechanical energy requirements²¹ and muscular work²². Zelik et al. hypothesized that softer spring conditions produced greater energy loss due to increased excursion of the body center of mass during gait, while a more rigid spring may have caused faster loading rates²¹. Thus, there may be other factors besides shock absorption that should be considered when determining an optimal prosthesis stiffness prescription. Reducing the stiffness of a prosthesis excessively, for example, could result in improved shock absorption but require the adoption of undesirable gait compensations. Optimization of the prosthesis stiffness could provide enormous benefits in the reduction of energy expenditure during locomotion and may increase the comfort of the prosthesis user²³. The present study indicates that, as currently prescribed within a transtibial prosthesis, adding longitudinal stiffness in isolation may not provide the anticipated shock absorption benefits.

5 Conclusion

A systematic variation of longitudinal prosthetic stiffness was performed to assess the influence of stiffness on both kinetic and kinematic variables related to shock absorption. The broad range of stiffness conditions compared to previous studies led us to hypothesize that kinetic variables (i.e., GRF peak magnitude and timing) would be affected by changes in stiffness. Negligible kinetic variance was detected between conditions, particularly at freely selected walking speeds. This result demands further investigation, as gait biomechanics should change predictably if there has been a substantial modification in the limb properties. Exploration of the passive stiffness properties of the transtibial residual limb is required to evaluate the assumption that commercially available levels of prosthesis stiffness are sufficient to create a meaningful change in the total prosthetic-side limb stiffness characteristics.

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Clinical Relevance

Varying the level of longitudinal prosthesis stiffness did not meaningfully influence gait biomechanics at self-selected walking speeds. Thus, as currently prescribed within a transtibial prosthesis, adding longitudinal stiffness in isolation may not provide the anticipated shock absorption benefits. Further research into residual limb properties and compensatory mechanisms is needed.

Figure 1.
The Endolite TT Pro Pylon. An “exploded” view of the experimental SAP including the location of the spring, which was switched out to create varying longitudinal stiffness characteristics within the prosthesis.

Figure 2.

Prosthetic-side pelvic obliquity and knee flexion during stance phase for a single subject. For the pelvic obliquity curve, 0° represents a neutral pelvis and positive values denote a higher pelvis on the prosthetic-side limb. For the knee flexion curve, 0° represents a straight knee and positive values denote flexion. The shaded regions represent one standard deviation from the mean of five steps. The dotted vertical line denotes contralateral toe-off.

Figure 3.

Prosthetic-side vertical GRF profile for a single subject. The shaded regions represent one standard deviation from the mean of five steps. The dotted vertical line denotes contralateral toe-off.

Table 1

Subject characteristics.

Subject	Gender	Age (years)	Height (m)	Mass (kg)	Time since Amputation (years)	Cause of Amputation	Spring Set
1	M	70	1.85	83	37	trauma	1
2	F	68	1.765	123	10	vascular	2
3	M	29	1.84	78.5	4	trauma	2
4	M	31	1.805	78.5	11	cancer	2
5	M	73	1.67	73.25	20	trauma	1
6	M	54	1.675	74.5	14	trauma	1
7	F	37	1.68	52.2	10	trauma	2
8	F	64	1.625	99.5	12	trauma	2
9	M	55	1.81	119	9	vascular	2
10	F	52	1.685	64.5	16	trauma	1
11	M	24	1.765	75.5	5	trauma	1
12	M	30	1.88	96	6	trauma	2
Mean		49 ± 18	1.75 ± 0.08	84.8 ± 21.0	13 ± 9		

Table 2

Kinematic variables and walking speed. Mean $\pm$ standard deviation for each experimental stiffness condition at both normal and fast self-selected walking speeds. Knee flexion angle and pelvic obliquity range of motion were calculated during the period from heel contact to contralateral toe-off.

Spring Condition	Normal Walking			Fast Walking		
	Stance-Phase Knee Flexion (deg)	Pelvic Obliquity Range of Motion (deg)	Speed (m/s)	Stance-Phase Knee Flexion (deg)	Pelvic Obliquity Range of Motion (deg)	Speed (m/s)
SOFT	9.5 $\pm$ 6.0	2.1 $\pm$ 1.0	1.23 $\pm$ 0.19	10.1 $\pm$ 6.6	2.3 $\pm$ 1.3	1.57 $\pm$ 0.29
MEDIUM	10.7 $\pm$ 5.9	2.4 $\pm$ 1.1	1.25 $\pm$ 0.18*	11.1 $\pm$ 5.9	2.5 $\pm$ 1.4	1.57 $\pm$ 0.30
NORMAL	10.1 $\pm$ 5.8	2.2 $\pm$ 1.0	1.18 $\pm$ 0.17*	11.5 $\pm$ 5.6	2.5 $\pm$ 1.1	1.55 $\pm$ 0.29
RIGID	11.1 $\pm$ 5.8	2.1 $\pm$ 1.1	1.23 $\pm$ 0.20	11.7 $\pm$ 5.8	2.4 $\pm$ 1.4	1.59 $\pm$ 0.29
p-value	$p = 0.202$	$p = 0.388$	$p = 0.008$ (*MED-NORM $p = 0.005$)	$p = 0.132$	$p = 0.754$	$p = 0.194$

Table 3

Kinetic variables and walking speed. Mean $\pm$ standard deviation for each experimental stiffness condition at both normal and fast self-selected walking speeds. The magnitude and timing of peak force was identified during loading response phase (i.e., prior to contralateral toe-off).

Spring Condition	Normal Walking			Fast Walking		
	Peak Force (% BW)	Time to Peak (% Stance Phase)	Speed (m/s)	Peak Force (% BW)	Time to Peak (% Stance Phase)	Speed (m/s)
SOFT	103.0 $\pm$ 12.6	22.3 $\pm$ 4.0	1.23 $\pm$ 0.19	126.3 $\pm$ 24.3*	20.8 $\pm$ 4.4*	1.57 $\pm$ 0.29
MEDIUM	104.5 $\pm$ 11.3*	22.5 $\pm$ 3.8	1.25 $\pm$ 0.18*	121.3 $\pm$ 22.5	20.7 $\pm$ 3.8*	1.57 $\pm$ 0.30
NORMAL	98.2 $\pm$ 11.1*	22.5 $\pm$ 4.1	1.18 $\pm$ 0.17*	118.7 $\pm$ 22.6*	20.4 $\pm$ 3.9	1.55 $\pm$ 0.29
RIGID	103.3 $\pm$ 19.0	21.7 $\pm$ 3.7	1.23 $\pm$ 0.20	123.4 $\pm$ 24.3	18.3 $\pm$ 4.7*	1.59 $\pm$ 0.29
p-value	p = 0.001 (*MED-NORM p > 0.001)	p = 0.257	p = 0.008 (*MED-NORM p = 0.005)	p = 0.006 (*SOFT-NORM p = 0.021)	p = 0.001 (*MED,SOFT-RIGID p = 0.004)	p = 0.194