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The effect of a shock-absorbing pylon on the gait of persons with unilateral transtibial amputation

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Abstract—Shock-absorbing pylons (SAPs) are components that increase prosthetic compliance and provide shock absorption during walking, running, and other high-impact activities in persons with leg amputations. This study investigated the effect of SAPs on the gaits of persons who walk with transtibial prostheses. Two gait analyses were performed on 10 subjects walking with and without an Endolite TT (Telescopic-Torsion) Pylon. Comparison of kinematic and kinetic gait parameters indicated that few quantitative changes were found in the way people walked with and without the SAPs. The most consistent change among subjects was a reduction in the magnitude of an isolated-force transient that occurred during the prosthetic loading response phase, an effect that was more evident at higher speeds. Results from a questionnaire that was administered to subjects indicated they generally preferred walking with the SAP for reasons related to comfort. We conclude that SAPs may provide significant benefit for persons with transtibial amputations who are able to routinely walk at speeds above approximately 1.3 m/s.

Key words: gait, prosthetics, shock-absorbing pylon, shock absorption, transtibial amputee.

INTRODUCTION

Shock-absorbing pylons (SAPs) are prosthetic components that are claimed to reduce shock forces during walking and other high-impact activities. SAPs are spring-like mechanisms that are fitted in the shank section of transtibial prostheses. They shorten telescopically with applied axial loads, thereby decreasing the stiffness of the prosthesis and presumably reducing the impact forces associated with walking. Persons who walk with SAPs in their prostheses often express subjective preference for the devices, for reasons such as increased comfort during walking, decreased force applied to the residual limb, and decreased pain. However, quantitative gait studies have yet to provide objective evidence suggesting that persons fitted with SAPs benefit significantly.

Abbreviations: BW = body weight, GRF = ground reaction force, MAC = Motion Analysis Corporation, SAP = shock-absorbing pylon, SACH = Solid Ankle Cushion Heel, SL = step length, TT = Telescopic-Torsion, VACMARL = VA Chicago Motion Analysis Research Laboratory, VSP = vertical shock pylon.

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Shock absorption to reduce floor impact forces at the time of initial contact has been identified as one of the primary locomotor functions during normal walking [1,2]. Weight transfer onto the accepting limb during gait is rapid and fairly abrupt, which creates the challenge of accepting rapidly moving body weight (BW) in a manner that both absorbs the shock of floor contact and creates a stable limb over which the body can advance [3]. Cushioning of impact forces generated during normal locomotion is achieved through the physical properties of biological tissues, footwear, and walking surfaces [4] and through actions of the lower limb and pelvis, such as stance-phase knee flexion and pelvic obliquity [1,5–7]. Insufficient shock absorption during gait has been attributed to low back pain [8,9]; cartilage degeneration at the joints [10]; and osteoarthritis in the ankle, knee, and spine [9,11].

In able-bodied persons, shock absorption during gait is believed to occur mainly in the legs and, to a lesser degree, in the spine [12]. The kinematic responses of the locomotor system during the loading response phase of walking actively attenuate impact forces, while passive attenuation is provided by the structure and viscoelastic properties of tissues [13]. The viscoelastic components of the joints themselves are believed to have only a minor role in protecting the body from harmful impulsive loads. The most effective strategy the human locomotor system has to withstand and manage impact forces is by changing the form of the body through the rotation of the joints [9,14].

During the period of weight acceptance in normal walking, motions of the legs and pelvis provide a system to facilitate shock absorption [3]. At initial contact, ankle plantarflexion and knee flexion both serve as shock absorbers to lessen the impact of floor contact [1,7,11,15,16]. Pelvic obliquity has also been identified as having a role in reducing shock during gait [1,5,6]. Data suggest that eccentric muscular contraction enables the knee joint to absorb more energy during the loading response phase than either the hip or the ankle joints [17], suggesting that the knee is the major lower-limb joint for providing shock absorption [18]. Researchers have found that increasing knee flexion decreases the stiffness of the leg and diminishes transmission of mechanical shock from the foot to the skull [19–22]. The body appears to guard the head against potentially harmful impact shock [8,18,22,23]. Absorption of shock, especially its low frequency components, by deceleration of a moving limb segment through eccentric contraction is possibly the major function of most of the muscles in the lower limb [24–26].

The physiological mechanisms that normally provide shock absorption during walking may be significantly diminished or completely absent in persons with leg amputations. Effective shock absorption of transtibial prostheses is believed to be necessary to avoid proximal joint diseases [27]. The shock transmitted by the prosthesis should probably not be higher than the shock transmitted by a normal lower leg [27]. One of the reasons that persons with lower-limb amputations walk slower than able-bodied persons may be to decrease shock forces to the residual limb so that the forces are at comfortable, manageable levels. One investigator observed that a transfemoral amputee walking at his maximum speed (~1.5 m/s) exhibited a vertical acceleration similar to that of normal subjects at their maximum walking speeds (~2.4 m/s) [28]. This investigator suggested that the vertical acceleration—and the mechanical load associated with it—may be one of the limits to walking speed, both in normal and pathological gaits. Providing sufficient shock absorption to persons with amputation, either through gait training to restore lost function or through the addition of specialized prosthetic components, may enable them to walk significantly faster.

Many gait studies investigating shock absorption have compared the effect of different prosthetic components on the walking characteristics of persons with amputations. The inherent compliance of foot-ankle prosthetic components probably enable them to all absorb some degree of shock [29]. One study compared the shock-absorbing characteristics of prosthetic foot designs as a presumed measure of comfort during gait [30], and found that the SACH (Solid Ankle Cushion Heel) foot design attenuated the higher frequency components of acceleration more than the Seattle Ankle/Lite Foot. The SACH foot has a cushioned heel that provides shock absorption as it is compressed during initial contact, and it simulates the normal plantarflexion movement of the foot from initial contact to the foot-flat position [30]. Effectiveness of the cushioned heel as a shock absorber led to its continued use in most prosthetic foot designs [3]. Another group of investigators compared the gaits of five able-bodied subjects to the gaits of five unilateral transtibial amputees walking with three different feet [31]. They found that regardless of prosthetic foot type, there was a loss of the shock absorption function of the prosthetic-side knee during loading response. In another study, investigators compared five ankle-foot devices and found that transtibial amputees preferred walking with devices that developed less shock and had greater damping at heel contact [32]. It has been suggested that the declination of peak accelerations at heel strike is an important aspect of the foot-ankle performance in transtibial amputees [33].

Only a few studies have specifically looked at the effect of shock-absorbing prosthetic components on amputee gait. Researchers at University of Strathclyde designed a kneeless prosthesis for transfemoral amputees that shortened telescopically during swing phase [34,35], providing increased stability and an improved proprioceptive feedback of prosthetic foot position. One of the features enjoyed by some users of this telescoping leg was the increased compressibility during the loading response phase, a feature that is similar to the action of SAPs.

Mooney et al. examined the differences in the ground reaction forces (GRFs) and moments of force during gait between the sound and residual limbs of a single transtibial amputee using the Flex Foot Re-Flex Vertical Shock Pylon (VSP) [36], a type of SAP that uses a pistoning pylon assembly along with a leaf spring. The subject was tested at his freely selected walking speed with the VSP and then with the shock absorber immobilized. The investigators found that when the VSP was disabled, the vertical GRF during prosthetic stance exhibited a third peak in midstance and slightly decreased forces during the loading response phase and the fore-aft GRFs were slightly decreased. No changes were found in the magnitudes of the prosthetic joint moments between testing conditions.

Miller and Childress analyzed the Flex Foot Re-Flex VSP and used gait analysis to determine its effect on transtibial amputee gait for a variety of activities [37]. Experimental gait analysis was performed on two transtibial amputees who were using the VSP while walking, jogging in place, and descending a curb. The GRFs, vertical trunk motion, event timing, and pylon compression were measured during the analyses. To determine the effect of the VSP, the investigators immobilized the pylon and repeated the experiment. Few biomechanical differences were found for walking when comparing trials with and without vertical compliance; however, differences were noted during fast walking and jogging.

Hsu et al. investigated the effect that the Flex Foot Re-Flex VSP had on energy expenditure during gait [38]. They compared the energy cost for five men with unilateral transtibial amputations who walked and ran on a treadmill at various speeds while wearing a SACH foot, a Flex Foot, and a Re-Flex VSP. The investigators reported that during both walking and running, the Re-Flex VSP significantly reduced energy cost, increased gait efficiency, and decreased exercise intensity when compared with the Flex Foot or SACH foot. They suggested that the design benefits of the prosthetic mechanisms are speed dependent and the differences become more apparent at speeds above 67.05 m/min (1.12 m/s).

Michaud et al. reported that one subject with a transtibial amputation who walked with a Flex Foot Re-Flex VSP demonstrated a sustained pelvic obliquity offset toward the prosthetic side when he walked [39]. Although the subject's pelvis was level during standing, a 6° offset was present during walking. This offset was speculated to be the result of a "dynamic leg length discrepancy" in the prosthesis because of the compression of the SAP shortening the leg. No definitive conclusions were drawn, but further investigation into this issue was recommended.

This current study uses both quantitative and subjective analyses to investigate the effect that SAPs have on the walking performance of persons with unilateral transtibial amputation. SAPs are typically only fit on younger, more active amputees, but because shock absorption has been identified as a primary function of normal locomotion, all amputees may benefit from the increased prosthetic compliance provided by SAPs. Decreasing the shock forces during gait should allow amputees to walk greater distances, for longer periods of time, and with generally increased comfort.

METHODS

Subjects

Subject selection was based upon the following inclusion criteria:

- Unilateral transtibial amputation without serious complications.
- Six or more months experience with a definitive prosthesis.
- Able to walk unaided for a distance of at least 10 m.
- Able to walk at a constant, comfortable speed without undue fatigue.
- Not fitted previously with a SAP.

Ten persons (9 males and 1 female) with unilateral transtibial amputation were enrolled in the study. The subjects ranged in age from 31 to 79 years (mean = 54 years, SD = 17 years). Seven of the subjects had traumatic amputation, while the other three had amputation because of vascular disease. All subjects provided informed consent before participating in the study. Information on the research subjects and their prostheses is summarized in **Table 1**.

Data Collection and Processing

All gait analyses were performed in the Veterans Affairs (VA) Chicago Motion Analysis Research Laboratory (VACMARL). VACMARL is equipped with an eight-camera, 60/120/240Hz switch-selectable HiRes motion measurement system from the Motion Analysis Corporation (MAC)* that is used to measure the instantaneous positions of markers placed on the body. Reflective markers were placed on the body using the Helen Hayes arrangement. The locations of the markers are used to define the biomechanical model that is used to analyze the data. VACMARL has six force platforms from Advanced Mechanical Technology, Inc. (AMTI)†, embedded in the walkway to measure GRFs as subjects walk across them. These data indicate the magnitude and direction of forces exerted by the subjects on the floor during the different times of the gait cycle and tend to characterize different types of gait. Kinematic and kinetic data were acquired at 120 Hz and 960 Hz, respectively. OrthoTrak software (MAC) was used to process the marker position data and generate graphs for further analysis.

Initially, we performed baseline gait analyses on all subjects walking on their conventional prostheses (i.e., without SAPs). During the gait analyses, subjects were asked to walk at their normal speed, a speed slightly faster than their normal speed, their fastest comfortable speed, a speed slightly slower than their normal speed, and finally their slowest comfortable speed. Our goal was to record the patterns of motion across their range of comfortable walking speeds so that we could analyze changes that occurred in the gait parameters as a function of speed. All the speeds were self-selected by the subjects based upon verbal instructions by the investigators. A minimum of three trials of data was collected at each speed of walking.

Immediately after the first data acquisition session, subjects were fitted with an Endolite Telescopic-Torsion (TT) Pylon (**Figure 1**). Selection of the TT Pylon spring stiffness was based upon each subject's weight and activity level as recommended by the manufacturer. The Endolite TT Pylon offers several advantages over other SAPs currently on the market: it can be fit with a wide variety of prosthetic feet, it is relatively inexpensive,

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†AMTI, 176 Waltham Street, Watertown, MA 02172; 617-926-6700

Table 1.
Subject data.

Subject	Gender	Age (yr)	Height (m)	Weight (kg)	Amputation Etiology	Years Since Amputation	Prosthetic Foot Type	Prosthetic Socket/ Suspension
1	F	43	1.59	54.0	Traumatic	1	College Park	PTB/Gel liner & pin
2	M	69	1.81	110.5	Traumatic	50	Carbon Copy II	PTB-SC
3	M	37	1.87	98.0	Traumatic	5	Master Step	TSB/Gel liner & pin
4	M	42	1.82	121.0	Traumatic	30	Cirrus	PTB-SCSP
5	M	46	1.86	85.5	Traumatic	1	College Park	PTB-SC
6	M	57	1.88	78.3	Traumatic	25	Seattle	PTB-SC RMB
7	M	31	1.73	77.0	Traumatic	4	College Park	TSB/Gel liner & pin
8	M	73	1.80	91.0	Vascular	4	Carbon Copy II	TSB/Gel liner & pin
9	M	64	1.78	139.5	Vascular	2	College Park	TSB/Gel liner & pin
10	M	79	1.81	105.0	Vascular	1	College Park	PTB-SC

PTB = patellar tendon bearing, SC = supracondylar, TSB = total surface bearing, SP = suprapatellar, RMB = removable medial brim

Figure 1.
Endolite Telescopic-Torsion Pylon uses a helical spring to increase prosthetic compliance and provide shock absorption during walking.

damping is minimized through the use of helical springs, and the stiffness is set with one of the three available springs (73 kN/m, 108 kN/m, 137 kN/m) that can be interchanged with relative ease. When research subjects were fitted with the TT Pylon, the length of the prosthesis was set to approximately 1/4 in. (~6 mm) longer than that of the sound leg, per the manufacturer's recommendation, to compensate for the shortening of the SAP during stance phase. Without this leg length adjustment, the amputee would be at greater risk of tripping during sound limb swing phase. If, however, the prosthetic leg length were too long, the amputee would be at greater risk of tripping during prosthetic swing phase. A person walking with the Endolite TT Pylon should ideally get about 5 mm to 8 mm travel in the unit during prosthetic stance phase, with an absolute maximum of 13 mm compression.

After walking with the TT Pylon for a minimum of 4 weeks, the subjects returned to VACMARL for a second gait analysis. The protocol for this data acquisition session was identical to that protocol used during the first gait analysis. After the second session, subjects were administered a questionnaire to document their perceptions of walking with the SAP. Subjects were asked to rate a number of statements pertaining to comfort, stability, and prosthetic use with the SAP according to the following scale: strongly agree, agree, no change or not applicable, disagree, or strongly disagree. Subjects were permitted to keep the SAP in their prosthesis at the completion of the study if they desired.

Data Analysis

Kinematic and kinetic data from the two gait analyses were compared, and differences because of the increased prosthetic compliance provided by the SAP were noted. Gait parameters of particular interest included walking speed, step-length symmetry, pelvic obliquity, and vertical and fore-aft GRFs. When walking on rigid prostheses, amputees may unintentionally compromise their gait performance using compensatory actions to reduce impact forces on the prosthesis at initial contact and during the loading response phase of walking. SAPs may enable users to reduce or eliminate these compensatory actions, thereby improving walking performance and aesthetics while providing generally increased comfort.

It was hypothesized that, if SAPs increase prosthetic compliance and can reduce impact forces in the prosthesis during gait, then subjects may be able to increase their walking speeds. However, it was also thought that the increased compliance might inhibit forward progression of the body in late prosthetic stance phase because of compression of the SAP, thereby decreasing the speed compared with a more rigid prosthesis. Therefore, the data trials of the freely selected and fastest speeds of subjects walking with the SAPs were compared with those acquired when they walked on their conventional prostheses without the SAPs to determine if statistically significant differences existed.

The difference between prosthetic and sound step lengths (SLs) was analyzed for changes that may have resulted from increased prosthetic compliance provided by the SAP. Persons with unilateral amputation often have an asymmetry between their sound and prosthetic SLs, in which they take a slightly longer step with the prosthesis when they walk. If it is assumed that SAPs "improve" gait, they should presumably increase SL symmetry. However, it was speculated that SAPs may actually increase asymmetry; *prosthetic* SL may increase because of a reduction in the impact forces at initial contact, and *sound* SL may decrease because of a shortening of the prosthetic leg length as the SAP compresses during stance phase. The SL difference was calculated as (prosthetic SL – sound SL). We compared these differences between the two testing conditions when subjects first walked without and then with the SAP to determine what effect, if any, increased prosthetic compliance had on SL symmetry.

We analyzed pelvic obliquity to determine how sustained rotational offsets were affected when subjects walked with the SAP. Earlier work suggests that SAPs may increase this rotational offset toward the prosthetic side because of the compression of the unit during prosthetic stance phase. Therefore, the addition of the SAP was expected to significantly increase the observed offset in the pelvic obliquity data.

The vertical and fore-aft GRFs for the prosthetic limb were analyzed for changes reflecting the increased prosthetic limb compliance with the addition of the SAP. Expectations were that if we increased the compliance of the prosthesis, the initial peaks in the vertical and fore-aft GRFs during prosthetic stance phase would decrease and the time from initial contact to that peak would increase.

A technique was employed to isolate transient force events that occurred during the loading response phase of walking that would otherwise be obscured by the magnitudes of the peak forces in the GRF. This method involved taking each vertical GRF record as a function of time and fitting a line between the data points at the time of initial contact and at the first peak in the vertical force record (**Figure 2(a)**). This line can be considered to represent the average rate of loading of the BW onto the prosthetic limb as the load is transferred from the trailing sound leg. For each data sample during this corresponding time, the force values represented by the fitted line were subtracted from the original force record. This procedure effectively isolates transient peaks in the vertical GRF that occur during the loading response phase and would otherwise be obscured by forces associated with BW, thereby facilitating analysis of their peak magnitudes. A similar technique was employed by Whittle and Williams in a study of heel-strike transients within vertical GRF records during walking [40]. In the example shown in **Figure 2(b)** for subject 5 walking without a SAP, the transient force has a peak magnitude of approximately 40 percent BW above the average loading rate.

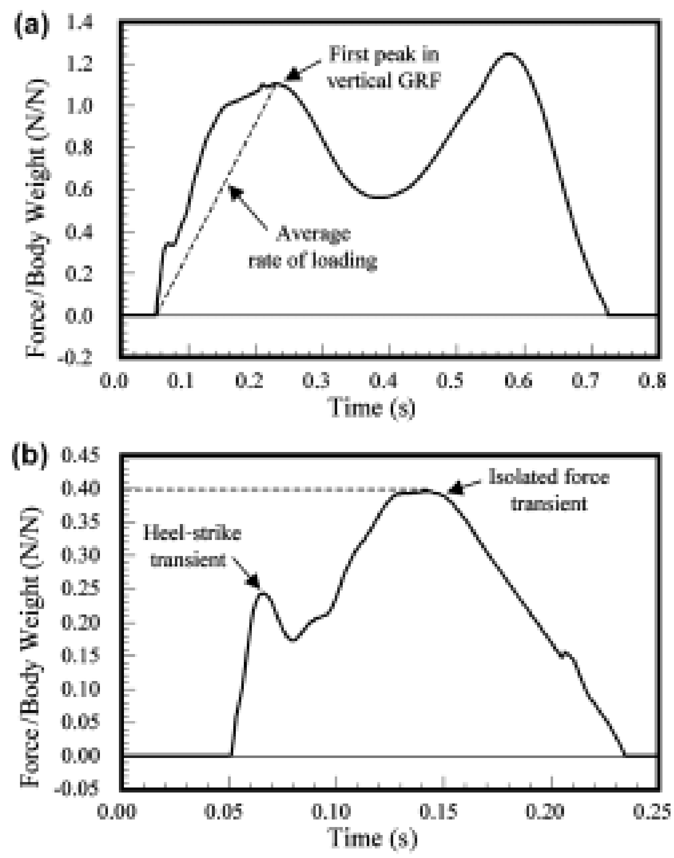

Figure 2.
Isolated-force transient method involves subtracting average rate of loading from vertical GRF record during loading response phase of walking. Average rate of loading is defined by fitting a line between end points at time of initial contact and when vertical force reaches a maximum about time of contralateral toe off. **(a)** Record of subject 5 walking on his everyday prosthesis without a SAP at a speed of 1.58 m/s. **(b)** Result after subtracting average rate of loading from vertical GRF. Heel-strike transient becomes much more apparent, and peak magnitude of force is observed to be about 40% of BW.

The heel-strike transient, which is the first peak in **Figure 2(b)** and has a magnitude of about 25 percent BW, also becomes more pronounced in the processed waveform.

We analyzed differences in the quantitative data for subjects walking with and without the SAPs using paired t-tests on each subjects' averaged data (three trials) at a level of significance of $\alpha = 0.05$. We also analyzed group data using t-tests to determine if the gait of all subjects who walked with the SAP had significant changes. The walking speed, SL symmetry, and pelvic obliquity were analyzed with two-tailed t-tests, while the other gait parameters were analyzed with one-tailed tests. Data from the questionnaires were summarized by subjects' responses; no statistical analyses were performed on these results.

RESULTS

Statistical analysis of the quantitative data from the gait analyses yielded inconsistent results between subjects. Two subjects significantly increased their freely selected speed when walking with the SAP, while two other subjects walked significantly slower (**Table 2**). Three subjects significantly increased their fastest walking speeds with the SAP, while two subjects decreased their fastest speed. No statistical differences in walking speed were observed across subjects.

The difference between the prosthetic and sound limb SLs was observed to change significantly in two subjects at their freely selected speed, resulting in an increased SL asymmetry. The change in the SL difference with the addition of the SAP was considered to be more symmetric when the difference with the SAP was closer to zero than when the subject walked without it. Though not statistically significant, 8 of the 10 subjects were found to have greater SL asymmetry when they walked with the SAP at their freely selected speed than when they walked without it, with the prosthetic SL being greater than the sound SL. Two subjects showed slight improvement in their symmetry, but the result was not statistically significant. At the fastest comfortable walking speed, the SL difference changed significantly in two subjects, with one of them

Table 2.
Temporal-spatial data.

Subject	Freely Selected Walking Speed				Fastest Walking Speed			
	Speed without SAP (m/s)	Speed with SAP (m/s)	Difference in SLs without SAP (cm)	Difference in SLs with SAP (cm)	Speed without SAP (m/s)	Speed with SAP (m/s)	Difference in SLs without SAP (cm)	Difference in SLs with SAP (cm)
1	0.98	1.24*	10.0	13.7	1.63	1.78*	13.8	16.9
2	1.23	1.22	0.3	2.3	1.42	1.49*	–1.7	1.5*
3	1.45	1.44	–0.4	4.1*	2.11	2.05	–2.8	–0.9
4	1.29	1.15*	1.3	1.7	1.71	1.65	–10.3	2.0
5	1.17	1.21	6.4	4.5	1.95	1.89*	7.2	5.3
6	1.24	1.13*	2.1	6.6	1.65	1.51*	3.3	7.9*
7	1.11	1.16	3.3	4.1	1.86	1.93	2.2	3.8
8	0.88	0.90	13.7	16.3	1.07	1.04	16.1	14.1
9	0.85	0.96*	1.5	7.6*	1.08	1.18*	6.4	4.2
10	0.95	0.96	–4.6	–3.8	1.36	1.31	1.4	1.8
All Subjects	1.11	1.14	3.3	5.7*	1.58	1.58	3.6	5.5
Traumatic (1–7)	1.21	1.22	3.2	5.3	1.76	1.77	1.7	4.9
Vascular (8–10)	0.89	0.94	3.5	6.7	1.17	1.18	7.9	7.0

*Significance at $\alpha = 0.05$, two-tailed paired t-test.
SL = step length

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improving his or her symmetry and the other becoming more asymmetric. Six of the subjects improved their symmetry at the fastest walking speed, but the result was significant in only one case. Analysis of the group data showed a significant increase in the SL difference among all subjects at the freely selected walking speed, leading to greater asymmetry. Similar changes were observed in the remainder of the group data at both speeds, though the results were not statistically significant. The vascular amputees demonstrated slightly greater SL symmetry at the fastest speed, but it was not significant.

Analysis of the pelvic obliquity data for sustained offsets yielded inconsistent results. When walking without the SAP, some subjects had an offset toward the prosthetic limb while others had an offset toward the sound leg. Addition of the SAP was found to have no significant effect on the magnitude or direction of the pelvic obliquity offset.

No subjects showed a statistically significant reduction in the magnitude of the first peak in the vertical GRF at the freely selected walking speed (**Table 3**). One subject demonstrated a significant increase in the time to reach the first peak in the vertical GRF from the time of initial contact when walking with the SAP. At their fastest comfortable walking speed, no subjects demonstrated significant decreases in the magnitude of the first peak of the vertical GRF with the SAP, but three subjects increased the time required to reach the first peak. Analysis of the fore-aft GRFs at the subjects' freely selected walking speeds indicated that one subject significantly decreased the magnitude of the first peak when walking with the SAP, and two increased the time to reach the first peak (**Table 4**). At their fastest walking speeds, no subjects decreased the magnitude of the first peak in the fore-aft GRF with the SAP, and four subjects increased the time required to reach the first peak.

The only data where statistically significant differences were noted with some consistency were in the magnitudes of the isolated-force transients occurring during the prosthetic loading response phase when subjects walked with the SAPs at the fastest walking speeds (**Table 5**). At the fastest walking speed, half of the subjects with traumatic amputations demonstrated a significant reduction in the force transient that occurred during the loading response phase of gait when they walked with the SAP. However, no statistically significant reductions were found

Table 3.
Vertical GRF data for prosthetic limb.

Subject	Freely Selected Walking Speed				Fastest Walking Speed			
	Peak Force without SAP (BW)	Peak Force with SAP (BW)	Time to Peak without SAP (ms)	Time to Peak with SAP (ms)	Peak Force without SAP (BW)	Peak Force with SAP (BW)	Time to Peak without SAP (ms)	Time to Peak with SAP (ms)
1	0.99	0.97	212.0	212.5	1.08	1.22	102.1	93.2
2	0.98	1.00	241.0	214.6	1.08	1.06	171.9	156.0
3	1.08	1.05	155.6	185.2	1.72	1.57	88.5	110.2*
4	0.99	0.98	188.2	221.3*	1.18	1.17	145.5	137.9
5	1.04	1.02	224.5	202.5	1.26	1.23	97.3	119.3*
6	0.99	0.97	217.0	220.1	1.10	1.06	145.1	123.4
7	1.08	1.11	183.3	202.1	1.51	1.48	88.3	102.8*
8	1.03	1.06	222.7	241.3	1.13	1.12	191.0	195.3
9	1.02	1.05	194.5	220.8	1.08	1.21	164.3	203.7
10	1.13	1.16	205.0	227.1	1.21	1.20	427.9	394.0
All Subjects	1.03	1.04	240.4	214.8	1.24	1.23	162.2	163.6
Traumatic (1–7)	1.02	1.01	203.1	208.3	1.28	1.26	119.8	120.4
Vascular (8–10)	1.06	1.09	207.4	229.8*	1.14	1.18	261.1	264.3

*Significance at $\alpha = 0.05$, one-tailed paired t-test.
BW = body weight

Table 4.
Fore-aft GRF data for prosthetic limb.

Subject	Freely Selected Walking Speed				Fastest Walking Speed			
	Peak Force without SAP (BW)	Peak Force with SAP (BW)	Time to Peak without SAP (ms)	Time to Peak with SAP (ms)	Peak Force without SAP (BW)	Peak Force with SAP (BW)	Time to Peak without SAP (ms)	Time to Peak with SAP (ms)
1	0.05	0.10	83.1	71.9	0.16	0.20	57.8	58.8
2	0.13	0.14	131.9	183.6	0.16	0.18	127.1	93.4
3	0.12	0.14	125.7	140.6	0.27	0.34	88.6	97.2*
4	0.11	0.10	137.0	157.3*	0.16	0.15	127.1	107.6
5	0.08	0.10	164.6	130.7	0.22	0.20	65.6	75.5*
6	0.10	0.09	72.4	88.7*	0.16	0.14	63.0	87.7*
7	0.13	0.14	143.4	134.7	0.30	0.26	83.9	90.3*
8	0.12	0.13	163.5	166.0	0.16	0.17	155.7	144.8
9	0.11	0.10*	167.7	159.6	0.14	0.15	150.0	140.1
10	0.11	0.10	167.5	177.5	0.13	0.13	123.7	146.2
All Subjects	0.11	0.11	135.7	141.1	0.19	0.19	104.3	104.2
Traumatic (1–7)	0.10	0.12	122.6	129.7	0.20	0.21	87.6	87.2
Vascular (8–10)	0.11	0.11	166.3	167.7	0.14	0.15	143.1	143.7

*Significance at $\alpha = 0.05$, one-tailed paired t-test.
BW = body weight

Table 5.
Isolated-force transient magnitude data from the vertical GRF for prosthetic limb.

Subject	Freely Selected Walking Speed		Fast Walking Speed		Fastest Walking Speed	
	Without SAP (BW)	With SAP (BW)	Without SAP (BW)	With SAP (BW)	Without SAP (BW)	With SAP (BW)
1	0.193	0.180	0.260	0.294	0.489	0.305*
2	0.390	0.240	0.274	0.422	0.274	0.307
3	0.355	0.156	0.267	0.094	0.933	0.607*
4	0.110	0.152	0.116	0.141	0.066	0.166
5	0.220	0.253	0.404	0.263	0.523	0.363*
6	0.285	0.338	0.334	0.315	0.404	0.293*
7	0.174	0.289	0.266	0.213	0.103	0.207
8	0.239	0.258	0.141	0.351	0.209	0.266
9	0.166	0.203	NA	0.220	NA	0.540
10	0.161	0.192	0.399	0.289	0.141	0.266
All Subjects	0.224	0.227	0.273	0.265	0.357	0.320
Traumatic (1–7)	0.240	0.230	0.274	0.249	0.409	0.335
Vascular (8–10)	0.189	0.218	0.270	0.320	0.175	0.267

*Significance at $\alpha = 0.05$, one-tailed paired t-test.

in the force transient at the freely selected or moderately fast walking speeds, though several subjects did demonstrate trends in this direction. The force record shown in **Figure 3** from subject 5 walking with the SAP at a moderately fast speed illustrates this force transient reduction. The transient has been reduced from about 40 percent BW (**Figure 2(b)**) to about 15 percent BW (**Figure 3(b)**). Note, however, that the magnitude of the heel strike transient is apparently unchanged with the addition of the SAP. Generally, the magnitudes of the GRFs vary directly with walking speed, with higher speeds corresponding to larger GRF magnitudes. The subject was walking faster with the SAP in **Figure 3(b)** (1.73 m/s) than he was without the SAP in **Figure 2b** (1.58 m/s); yet the results indicate that a significant reduction was found in the isolated-force transient magnitude with the SAP. **Figure 4** summarizes the force transient magnitude data for all subjects from both testing conditions. Significant reductions in the force transient magnitudes only become evident at speeds above about 1.3 m/s.

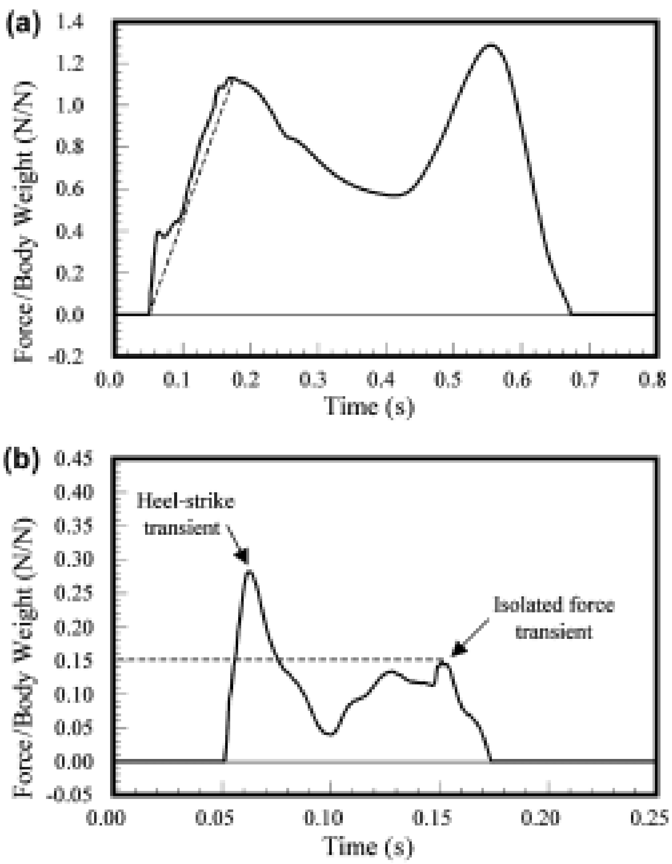

Figure 3.
(a) Vertical GRF for a single data trial from subject 5 walking with SAP at a speed of 1.73 m/s. **(b)** Isolated-force transient method indicates a significant reduction in magnitude of force during loading response phase, from about 40% of BW previously to about 15% of BW with SAP. Notice, however, that magnitude of heel-strike transient is apparently unaffected by SAP.

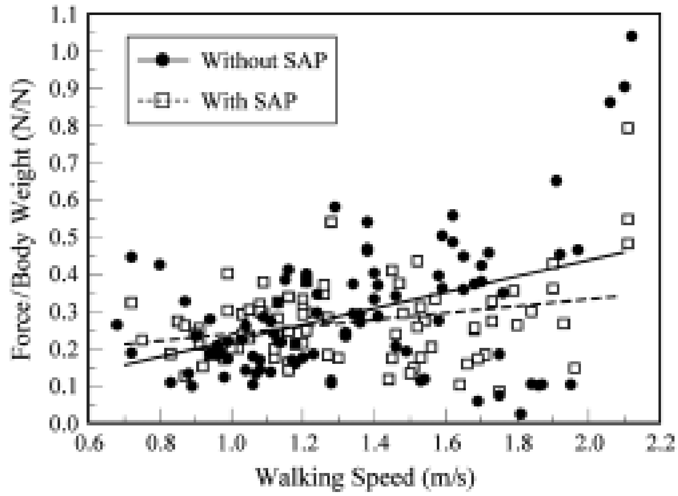

Figure 4.
Data from all subjects showing how magnitude of isolated-force transient changed with walking speed and with addition of SAP. Reductions in force magnitudes that are attributable to subjects walking with SAP only become apparent at speeds greater than about 1.3 m/s.

The questionnaire revealed that the majority of subjects preferred walking with the SAP (**Table 6**). Nearly all subjects reported that the SAP increased comfort during walking, made walking feel smoother, made walking downstairs or stepping off of a curb easier, and made the prosthesis feel less stiff. Most of the subjects neither noticed an increase in the weight of the prosthesis because of the addition of the SAP nor found the prosthesis harder to swing (because of increased moment of inertia). There were, however, one or two subjects who consistently rated the SAP poorly on all points, but in almost all cases, they were in the vascular amputee group. Clear differences of opinion on the benefit of the SAP were expressed in the questionnaires between the traumatic and the vascular amputee subjects.

DISCUSSION

This current study revealed that few significant differences were found in the temporal-spatial and GRF gait parameters with the addition of the SAP. Inconsistent changes were observed in the gait parameters when subjects

Table 6.
Summary of questionnaire results. Values represent number of subjects indicating this response.

Questionnaire Statements	Strongly Agree	Agree	No Change or Not Applicable	Disagree	Strongly Disagree
The SAP increases comfort during standing.	—	5	1	3	1
The SAP increases comfort during walking.	1	6	2	—	1
Overall, the SAP provides greater comfort.	1	6	2	—	1
Pain in my residual limb is reduced by the SAP.	—	—	9	1	—
The SAP does NOT add noticeable weight to my prosthesis.	2	4	2	1	2
The SAP does NOT make my prosthesis harder to swing as I walk.	—	6	2	1	1
The longer prosthetic length due to the SAP does not bother me.	—	3	3	2	2
I do NOT catch my prosthetic toe more with the SAP.	—	3	3	3	1
I do NOT catch my sound limb toe more with the SAP.	1	5	3	—	1
I do NOT feel like I am stepping into a hole with the SAP.	2	5	1	1	1
I do NOT feel less stable with the SAP.	2	5	1	—	2
The SAP enables me to walk longer distances.	1	2	5	—	2
I am able to walk faster with the SAP.	1	2	5	—	2
The SAP enables me to turn easier.	1	6	1	1	1
Walking feels smoother with the SAP.	1	7	—	—	2
The SAP does NOT increase the amount of energy needed to walk.	2	4	3	1	2
It is easier to go upstairs with the SAP.	—	4	5	—	1
It is easier to go downstairs or step off a curb with the SAP.	—	8	1	—	1
It is easier to go up an incline with the SAP.	1	3	4	—	2
It is easier to go down an incline with the SAP.	2	4	2	—	2
The SAP allows me to be more active.	2	2	4	—	2
The SAP makes my prosthesis feel less stiff.	—	8	1	1	—
The SAP reduces twisting between the socket and my residual limb.	2	3	4	1	—
I like having the SAP in my prosthesis.	3	5	—	2	—
I want to keep the SAP in my prosthesis.	3	6	—	1	1

walked with the SAP, with few statistically significant differences either within or across subjects. Some of the differences that were analyzed demonstrated clear trends, such as in the difference between the sound and prosthetic SLs. Statistical significance for a few of the gait parameters where trends were evident may have been improved had more data trials been acquired and analyzed for each subject at each of the walking speeds and had more than 10 subjects been included in the study.

Both the group and the individual subject's data indicated that there were greater asymmetries between the sound and prosthetic SLs when the subjects walked with the SAP, but the statistical analyses did not indicate the differences were significant except in a few cases. The increased SL asymmetry occurred because of a relative lengthening of the prosthetic SL compared with that of the sound limb. Scrutiny of the data indicated that the sound-limb SL decreased when subjects walked with the SAP, apparently because of compression of the SAP in late prosthetic stance phase as the body accelerated forward. However, this increased SL asymmetry observed in our unilateral amputee subjects walking with the SAP should probably not be of concern to prosthetists, since no other evidence was found indicating that walking performance was adversely affected.

Subject 1 was the only subject who showed statistically and clinically significant increases in both her freely selected and maximum walking speeds with the

SAP. This particular subject was a relatively new amputee who, while she met the inclusion criteria for this study for time since amputation, was undergoing intensive gait training by a physical therapist during the time she participated in this study. Her results may be misleading; the increases observed in her walking speed may not have actually been from the SAP, but could be attributable to improved walking technique or increased confidence she developed while walking on her prosthesis.

Interestingly, the results from the questionnaires indicated that most subjects preferred walking with the increased prosthetic compliance and that many of them believed that their gait performance improved with the SAP, although the gait analyses did not necessarily support this belief. Subjects may perceive a significant difference in the "feel" of the prosthesis during ambulation even when quantitative gait analysis fails to measure a significant change in gait parameters. This suggests that the changes that occur in the prosthetic performance may be too subtle to be discerned by looking at conventional measures of walking or that the typical gait parameters may not directly relate to the function the SAPs provide during subjects walking. Apparently, from the subjective ratings on the questionnaires, the effect of SAPs on gait was associated with the perception of comfort during prosthetic loading. Therefore, other objective, quantitative measures need to be identified that can relate better to users' perceptions of standing and walking on their prosthesis, such as the isolated-force transient method described here. The isolated-force transient method is a technique that is used to determine the magnitude of force associated with the prosthetic loading response phase of gait, a transient event that is ultimately transmitted to the residual limb and felt by the ambulator. SAPs appear to reduce the magnitude of this force by reducing prosthetic limb stiffness, thereby decreasing the impact force on the prosthesis and increasing comfort for the user.

Our quantitative data indicate that SAPs reduce force transients in the residual limb as the prosthesis is being loaded during walking by up to 60 percent of the original force magnitude in some users. This reduction is not readily evident by visual inspection of the GRFs, since the peak magnitude of force that is reached at the end of the loading response phase (i.e., first peak of vertical GRF) is virtually unaffected. The force transients that occur in the prosthesis during the loading response phase become more evident when the average loading rate on the prosthetic limb is subtracted from the GRF record. We believe that these force transients may be associated with momentum transfer of the residual limb mass as the prosthesis is loaded. As demonstrated by the examples provided in **Figures 2 and 3**, the forces that are generated in the prosthesis and residual limb as BW is rapidly transferred during the loading response phase are much less when total limb stiffness is reduced by the SAP. The research subjects' responses to statements in the questionnaire pertaining to the increased comfort, decreased pain, and increased ease of ambulation when walking with SAPs (**Table 6**) support the notion that force transients applied to the residuum are reduced and that this reduction is discernable by users. While the quantitative data indicate the effect of SAPs on walking may be subtle, they do appear to provide significant benefit nonetheless when users walk at speeds above about 1.3 m/s.

Some of the subjects appeared to have lower force transient magnitudes at their faster speeds when walking without the SAP than when walking with it (**Table 5** and **Figure 4**). No logical explanation exists for why this would have occurred based upon the changes that were made to the subjects' prostheses, since this result is contrary to expectations for persons walking on prostheses that are more compliant. One possibility for this result could be that the subjects readily adapted to the prosthetic compliance and modified their gaits accordingly. Engineering analyses make it possible to model purely mechanical systems as series of masses, springs, and dampers, and predicting the effects of changing the compliance on overall system behavior is relatively easy. However, when the compliance in a person's prosthesis is changed, the human neuromuscular system may adapt to the modification in such a way that the result may not be what was expected.

Another reason why the magnitudes of the force transients did not decrease when subjects walked with the SAP could be due to measurement error, because the force transient isolation method becomes more difficult to apply as walking speeds get faster and the impact forces increase. The vertical GRF records obtained from transtibial amputees during their freely selected gait are often similar to those observed in able-bodied ambulators, having a characteristic bimodal appearance with dominant peaks of approximately equal amplitude. The first peak in the vertical GRF is approximately coincident in time with contralateral toe off. At moderately fast walking speeds in transtibial amputees, the magnitudes of the force transients occurring during the prosthetic loading response phase become greater, and the vertical GRF may even acquire a trimodal appearance. In this case, it is the second peak that corresponds with contralateral toe off and not the first. When the force records demonstrated this trimodal characteristic in the data obtained for this study, the end point selected for the force transient isolation method was not the first peak in the vertical GRF, but the second one. The results obtained for these records were good and usually demonstrated the expected decrease in the force transient magnitude when subjects walked with the SAP compared to when they walked without it. However, when the subjects walked at their fastest speeds, the force transients during the prosthetic loading response phase would often surpass the magnitude of the first peak in the vertical GRF and even obscure it. Since the force transient isolation method relies upon defining the magnitude of the vertical GRF peak corresponding with contralateral toe off as an end point, the analysis of the GRF data in these cases became more subjective and less accurate because choosing the second end point became difficult. Therefore, some of the results from the force transient isolation method that were obtained when subjects were walking at the fastest speeds are questionable. We are working to refine our isolated-force transient method so that it can be applied more objectively and accurately in future studies of shock-absorbing prosthetic components.

Heel-strike transients are apparently unaffected by the addition of SAPs, as was demonstrated with the example data shown in **Figures 2** and **3**. These forces have been associated with momentum transfer as the mass of the foot rapidly decelerates at the time of, and immediately following, initial contact during walking [14]. Because the prosthetic foot is distal to the SAP, the momentum transfer associated with initial contact of the foot is unaffected and thus the magnitude of the heel-strike transient is unchanged. Increasing the compliance between the foot mass and the walking surface, such as with the use of softer shoe heels, shoe inserts, and prosthetic heels, can reduce heel-strike transients. A thorough examination of the heel-strike transients was not performed in this study, and further investigation of this matter is recommended before definitive conclusions can be drawn.

The results from the questionnaires indicated that 8 of the 10 subjects liked walking with the SAPs in their prostheses and wanted to keep the units indefinitely (**Table 6**). They responded favorably to statements pertaining to comfort, ease of use, foot clearance, and ascending and descending stairs and ramps with the SAP. In spite of the few significant differences that were observed in quantitative gait data, these subjects readily perceived a difference in their prosthesis when the SAP was added to it, and they liked the sensation it provided during both standing and walking. The SAP undoubtedly reduced the stiffness of their prosthesis and added some degree of shock absorption during walking. The combination of the subjective and quantitative results from this study suggests that when prosthetists fit their clients with SAPs, they should probably rely more on user feedback to ascertain benefit than on their own visual gait assessment.

Only two subjects did not like walking with the SAPs, claiming that they felt unstable on the units. These subjects preferred the rigidity of their prostheses to the compliance provided by the SAPs. At the completion of the study, they chose to return the SAPs to the investigators and have their prostheses restored to their original configurations. Coincidentally, both of these subjects had amputation secondary to vascular disease, and they both walked at freely selected speeds of about 1.0 m/s and maximum speeds of about 1.2 m/s. The results of this study suggest that these two subjects walked at speeds that are probably too slow to derive significant benefit from SAPs. Persons who have sustained a traumatic amputation are often younger and more physically fit than are those individuals who have undergone leg amputations because of vascular disease. Therefore, persons with traumatic amputations will tend to walk faster, and their faster walking speeds suggest that they likely will derive measurable benefit from SAPs. However, vascular amputees may still benefit from using SAPs, provided that they are able to walk at speeds above approximately 1.3 m/s. Another factor that may have contributed to these subjects' decisions to remove the SAPs from their prostheses could have been decreased tactile sensation in their residual limbs. Both of these subjects had known sensory deprivation in their sound limbs because of diabetes, and they may have also had decreased sensation in their residual limbs. To compensate, they may have relied on impact forces in their prostheses to provide feedback regarding initial contact and limb loading, and therefore preferred to walk with a stiffer prosthesis instead of one having increased compliance provided by SAPs.

Several factors may have affected the results in this study. The study did not control for prosthetic foot type in the methodology. Our goal was to make a relative comparison between the gait characteristics of each subject when walking on his or her conventional prosthesis and when walking with the SAP. One particular advantage of the Endolite TT Pylon is that it can be fit to virtually any prosthetic foot. However, a few prospective subjects were excluded from the study because they were walking with a Flex Foot, which cannot be fitted with the Endolite TT Pylon. Regardless, in all cases, the compliance of each subject's prosthesis was increased (i.e., stiffness was decreased) with the addition of the SAP, independent of foot type. Some types of prosthetic feet are more compliant than other types, so the addition of the SAP to the prosthesis may have had different effects, depending upon which foot was being used. Had foot type been controlled, the subjects' conventional prostheses may have initially had a more comparable stiffness across subjects and the group results may have demonstrated more consistent trends.

Another factor that was not considered in this study was the transverse rotation provided by the Endolite TT Pylon. The TT Pylon is designed to provide ±30° of internal-external rotation during walking. This rotational feature was not expected to adversely affect quantitative or subjective measures of prosthetic compliance, so the investigators did not attempt to control for it or to eliminate it. However, many of the research subjects provided unsolicited comments in favor of this attribute. While we do not believe necessarily that the transverse rotation affected gait performance to the extent that it obscured the effects of increased prosthetic compliance, some subjects' opinions of the SAPs may have been highly influenced by this feature. The transverse rotation provided by the TT Pylon may have reduced the magnitude of the shear forces felt by the subjects between their residual limb and the prosthetic socket, making their gait more comfortable. In addition, some of the subjects specifically commented that this feature improved their golf swing. The quantitative gait data that were acquired for this study did not permit accurate measurement of the transverse rotation in the SAPs. Further gait studies should be conducted on amputee subjects that specifically investigate whether the provision for transverse rotation improves walking performance.

CONCLUSIONS

Our results indicate that a significant benefit may exist to fitting transtibial amputees with SAPs, provided that users routinely walk at speeds above approximately 1.3 m/s. The SAPs were found to decrease forces applied to the residual limb during the loading response phase of gait by up to 60 percent of BW for some subjects. Preference for the SAPs was fairly divided between research subjects who had sustained traumatic amputation and those who had undergone amputation because of vascular disease. Traumatic amputees tend to be younger and more active than vascular amputees, so they will typically walk at higher speeds and thus will be able to take greater advantage of the shock-absorption capabilities offered by the SAPs. Most of the research subjects reported that having the SAPs in their prostheses increased comfort during walking, made walking feel smoother, made walking downstairs or stepping off of a curb easier, and made the prosthesis feel less stiff. Short-term benefits for amputees walking with SAPs may include comfort and better prosthetic performance during gait, whereas long-term use of these devices may prevent joint and back problems that can arise from walking on a rigid prosthesis.

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REFERENCES

1. Perry J. Gait analysis: normal and pathological function. Thorofare (NJ): Charles B. Slack, Inc.; 1992.
2. Winter DA. The biomechanics and motor control of human gait: Normal, elderly and pathological. 2nd ed. Waterloo (ON): University of Waterloo Press; 1991.
3. Perry J, Boyd LA, Rao SS, Mulroy SJ. Prosthetic weight acceptance mechanics in transtibial amputees wearing the single axis, Seattle Lite, and Flex Foot. IEEE Trans Rehabil Eng 1997;5(4):283–89.
4. Lafortune MA, Lake MJ. Human pendulum approach to simulate and quantify locomotor impact loading. J Biomech 1995;28(9):1111–14.
5. Sutherland DH, Kaufman KR, Moitoza JR. Kinematics of normal human walking. In: Rose J, Gamble E, editors.

Human walking. 2nd ed. Baltimore (MD): Raven Press Ltd.; 1994. p. 414–18.
6. Gard SA, Childress DS. The effect of pelvic list on the vertical displacement of the trunk during normal walking. Gait Posture 1997;5:233–38.
7. Gard SA, Childress DS. The influence of stance-phase knee flexion on the vertical displacement of the trunk during normal walking. Arch Phys Med Rehabil 1999;80(1): 26–32.
8. Voloshin A, Wosk J. An in vivo study of low back pain and shock absorption in the human locomotor system. J Biomech 1982;15(1):21–27.
9. Collins JJ, Whittle MW. Impulsive forces during walking and their clinical implications. Clin Biomech 1989;4(3): 179–87.
10. Radin EL, Parker HG, Pugh JW, Steinberg RS, Paul IL, Rose RM. Response of joints to impact loading—III: Relationship between trabecular microfractures and cartilage degeneration. J Biomech 1973;6(1):51–57.
11. Nack JD, Phillips RD. Shock absorption. Clin Podiatr Med Surg 1990;7(2):391–97.
12. Smeathers JE. Transient vibrations caused by heel strike. Proceedings of the Institution of Mechanical Engineers. Part H, J Eng Med 1989;203(4):181–86.
13. Lafortune MA, Lake MJ, Wilson R. Shock transmissibility of the human body. Proceedings of the Eighth Biennial Conference of the Canadian Society for Biomechanics; 1994 Aug 18–20; Calgary, AB, Canada; p. 168–69.
14. Whittle MW. Generation and attenuation of transient impulsive forces beneath the foot: A review. Gait Posture 1999;10:264–75.
15. Perry J. Kinesiology of lower extremity bracing. Clin Orthop 1974;102:18–31.
16. Snyder RD, Powers CM, Fontaine C, Perry J. The effect of five prosthetic feet on the gait and loading of the sound limb in dysvascular below-knee amputees. J Rehabil Res Dev 1995;32(4):309–15.
17. Winter DA. Energy generation and absorption at the ankle and knee during fast, natural, and slow cadences. Clin Orthop 1983;175:147–54.
18. Ratcliffe RJ, Holt KG. Low frequency shock absorption in human walking. Gait Posture 1997;5:93–100.
19. Greene PR, McMahon TA. Reflex stiffness of man's antigravity muscles during kneebends while carrying extra weights. J Biomech 1979;12(12):881–91.
20. McMahon TA, Valiant G, Frederick EC. Groucho running. J Appl Physiol 1987;62(6):2326–37.
21. Lafortune MA, Hennig EM, Lake MJ. Dominant role of interface over knee angle for cushioning impact loading and regulating initial leg stiffness. J Biomech 1996a; 29(12):1523–29.
22. Lafortune MA, Lake MJ, Hennig EM. Differential shock transmission response of the human body to impact severity and lower limb posture. J Biomech 1996b;29(12): 1531–37.
23. Wosk J, Voloshin A. Wave attenuation in skeletons of young healthy persons. J Biomech 1981;14(4):261–67.
24. Bowker JH, Hall CB. Normal human gait. In: Atlas of Orthotics: Biomechanical principles and application (American Academy of Orthopaedic Surgeons), St. Louis (MO): C.V. Mosby Company; 1975. p. 133–43.
25. Bowker JH. Kinesiology and Functional Characteristics of the Lower Limb. In: Atlas of limb prosthetics: Surgical and prosthetic principles (American Academy of Orthopaedic Surgeons), St. Louis (MO): C.V Mosby Company; 1981. p. 261–71.
26. Pratt DJ. Mechanisms of shock attenuation via the lower extremity during running. Clin Biomech 1989;4:51–57.
27. Leeuwen JL van, Speth LAWM, Daanen HAM. Shock absorption of below-knee prostheses: A comparison between the SACH and the Multiflex foot. J Biomech 1990;23(5):441–46.
28. Cappozzo A. The mechanics of human walking. In: Patla AE, editor. Adaptability of human gait. North-Holland: Elsevier Science Publishers B.V.; 1991. p. 167–86.
29. Edelstein JE. Prosthetic feet: State of the art. Phys Ther 1988;68(12):1874–81.
30. Lehmann JF, Price R, Boswell-Bessette S, Dralle A, Questad K. Comprehensive analysis of dynamic elastic response feet: Seattle Ankle/Lite Foot versus SACH foot. Arch Phys Med Rehabil 1993;74:853–61.
31. Gitter A, Czerniecki JM, DeGroot DM. Biomechanical analysis of the influence of prosthetic feet on below-knee amputee walking. Am J Phys Med Rehabil 1991;70(3): 142–48.
32. Wirta RW, Mason R, Calvo K, Golbranson FL. Effect on gait using various prosthetic ankle-foot devices. J Rehabil Res Dev 1991;28(2):13–24.
33. Jaarsveld HWL van, Grootenboer HJ, De Vries J. Accelerations due to impact at heel strike using below-knee prosthesis. Prosthet Orthot Int 1990;14:63–66.
34. Seliktar R. Self energized power system for above knee prostheses. J Biomech 1971;4:431–35.
35. Seliktar R, Kenedi RM. A kneeless leg prosthesis for the elderly amputee, advanced version. Bull Prosth Res 1976; 10(25):97–119.
36. Mooney J, Hill S, Supan T, Barth D. Comparison of floor reaction and rotational forces in the gait of a transtibial amputee using a Re-Flex VSP Flex Foot design: A pilot study. Proceedings of the 21st Annual Meeting & Scientific Symposium of the AAOP (American Academy of Orthotists and Prosthetists); 1995 March 21–25; New Orleans (LA); p. 24.

37. Miller LA, Childress DS. Analysis of a vertical compliance prosthetic foot. J Rehabil Res Dev 1997;34(1):52–57.
38. Hsu MJ, Nielsen DH, Yack HJ, Shurr DG. Physiological measurements of walking and running in people with transtibial amputations with 3 different prostheses. J Orthop Sports Phys Ther 1999;29(9):526–33.
39. Michaud SB, Gard SA, Childress DS. A preliminary investigation of pelvic obliquity patterns during gait in transtibial and transfemoral amputees. J Rehabil Res Dev 2000;37(1):1–10.
40. Whittle MW, Williams CD. Reliability of force platforms in the estimation of insole shock attenuation. Proceedings, Eighth Biennial Conference, Canadian Society for Biomechanics; 1994 August 18–20; Calgary, Canada; 170–71.

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